Published in final edited form as:

IEEE Trans Nucl Sci. 2011 October ; 58(5): . doi:10.1109/TNS.2011.2165296.

Resolution Properties of a Prototype Continuous Miniature Crystal Element (cMiCE) Scanner

Robert S. Miyaoka [Senior Member, IEEE],

University of Washington Department of Radiology, Seattle, WA USA (phone: 206-543-2084; fax: 206-543-8356; rmiyaoka@u.washington.edu).

Xiaoli Li [Student Member, IEEE],

University of Washington Department of Physics, Seattle, WA USA (lixiaoli@u.washington.edu).

William Hunter [Member, IEEE],

University of Washington Department of Radiology, Seattle, WA USA (wcjh@u.washington.edu).

Larry A. Pierce II,

University of Washington Department of Radiology, Seattle, WA USA (lapierce@u.washington.edu).

Wendy McDougald,

University of Washington Department of Radiology, Seattle, WA USA (wam2@u.washington.edu).

Paul E. Kinahan [Fellow, IEEE], and

Paul E. Kinahan is with the University of Washington Department of Radiology, Seattle, WA USA (kinahan@uw.edu).

Thomas K. Lewellen [Fellow, IEEE]

University of Washington Department of Radiology, Seattle, WA USA (tkldog@u.washington.edu).

Abstract

Continuous miniature crystal element (cMiCE) detectors are a potentially lower cost alternative for high resolution discrete crystal PET detector designs. We report on performance characteristics of a prototype PET scanner consisting of two cMiCE detector modules. Each cMiCE detector is comprised of a $50 \times 50 \times 8$ mm³ LYSO crystal coupled to a 64 channel multi-anode PMT. The cMiCE detectors use a statistics-based positioning method based upon maximum likelihood estimation for event positioning. By this method, cMiCE detectors can also provide some depth of interaction event positioning information. For the prototype scanner, the cMiCE detectors were positioned across from one another on a horizontal gantry with a detector spacing of 10.7 cm. Full tomographic data were collected and reconstructed using single slice rebinning and filtered back projection with no smoothing. The average image resolutions in X (radial), Y (transverse) and Z (axial) were 1.05 ± 0.08 mm, 0.99 ± 0.07 mm, 1.24 ± 0.31 mm FWHM. These initial imaging results from a prototype imaging system demonstrate the outstanding image resolution performance that can be achieved using the potentially lower cost cMiCE detectors.

I. Introduction

Small animal PET imaging offers researchers a noninvasive imaging technique to perform longitudinal animal studies. Since the development of the first small animal PET systems (i.e., the RAT-PET and the microPET systems [1,2]), there have been a number of small animal PET systems developed that achieve an image resolution of < 2.0 mm full width at half maximum (FWHM). However since mice are generally three orders of magnitude

smaller than a human, small animal PET systems require ~ 0.7 mm FWHM image resolution to provide species equivalent image resolution between whole body mouse imaging and whole body human imaging. While the HIDAC small animal PET system was able to achieve <1.0 mm image resolution using multi-wire proportional chamber technology combined with lead converter plates [3], it had a low intrinsic detection efficiency. Prototype research scintillator based PET systems have been able to achieve <1.0 mm FWHM image resolution [4-8]; however, due in part to high costs for fabricating ultra-high resolution discrete crystal detectors none of these systems were commercialized. Thus limitations in adoption are the fabrication complexity and costs for PET detector systems utilizing discrete crystals with cross-sections of less than 1.0 mm by 1.0 mm.

An alternative approach that can reduce both complexity and cost is based on a monolithic crystal PET detector design [7-15]. The continuous miniature crystal element (cMiCE) detector [9,16,17] is similar in concept to the modular gamma camera detector that had previously been developed at the University of Arizona [18]. A large useful imaging area for the cMiCE detector is achieved using a statistics based positioning (SBP) method based upon the principles of maximum likelihood [19,20]. A 46.6 by 46.6 mm² useful imaging area for a 50 by 50 mm² crystal is used in this work. To further improve the overall detection efficiency for a ring of cMiCE detectors a trapezoidal crystal geometry can be used as described and evaluated in [21]; however, for this work a rectangular crystal geometry was used.

The initial goals of the cMiCE project were to support a system configuration that could achieve an absolute detection efficiency of $>2.5\%$ and an image resolution of <1.2 mm FWHM using filtered back projection. The original cMiCE detector was comprised of a $25 \times 25 \times 4$ mm³ slab of LSO coupled to a 1 inch square Hamamatsu 6+6 cross-anode, metal dynode PMT. This detector module achieved ~ 1 mm FWHM intrinsic spatial resolution for a 22×22 mm² useful imaging area; however, it had limited detection efficiency. Thus efforts were made to design cMiCE detectors using thicker crystal slabs.

For the second version of the cMiCE detector we switched to the larger area (52 mm by $\times 52$ mm) H8500 flat panel PMT, which has 64 (8×8) channels (Hamamatsu, Japan). These were coupled with thicker crystal slabs (e.g., $50 \times 50 \times 8$ mm³) [16,17]. With this detector we developed techniques to extract depth of interaction information [17,22,23].

In this work, we constructed a prototype cMiCE detector system consisting of two 8 mm thick LYSO crystal cMiCE detector modules, a rotating stage, and a 128-channel data acquisition system. Data were collected in a step and acquire mode, saved to disk, and processed off-line. After event positioning within each crystal block, the coincidence data were reconstructed using single slice rebinning [24] and filtered back projection with no smoothing.

II. Detector design and methods

A. cMiCE Prototype: Detector

Figure 1 shows a photograph of a cMiCE detector module. It consists of a $50 \times 50 \times 8$ mm³ slab of LYSO (St. Gobain, Newbury, OH) coupled to a 64-channel multi-anode PMT (H8500, Hamamatsu, Japan). One of the 50 mm by 50 mm surfaces was polished. All other surfaces were lapped with 100 grit compound to provide a roughened surface. The surface treatment was provided by St. Gobain as part of the order. The polished surface was coupled to the PMT using Bicon BC-630 optical grease. The opposite face was painted white using a mixture of white latex paint and BC-620 (Bicon Direct, Solon, OH) white reflective paint.

The side surfaces of the crystal were painted with a flat black latex paint. Painting and roughening the side surfaces helped to reduce light reflections from them.

B. Statistics-based Positioning (SBP) Method [9]

Three-dimensional (3D) event positioning for the cMiCE module is achieved using maximum likelihood estimation [17]. We refer to our methodology as statistics based positioning (SBP) [9]. In this method, the distributions of observing signal outputs $M = M_1, M_2, \dots, M_n$ for scintillation position x , are assumed to be independent normal distributions with mean, $\mu(x)$, and standard deviation $\sigma(x)$.

The likelihood function for making any single observation m_i from distribution M_i given x is:

$$L[m_i|x] = \prod_{i=1}^n \frac{1}{\sigma_i(x) \sqrt{2\pi}} \exp\left(-\frac{(m_i - \mu_i(x))^2}{2\sigma_i^2(x)}\right) \quad (1)$$

Mathematically the maximum likelihood estimator of the event position x is given by:

$$\hat{x} = \arg \min_x \left[\sum_i \frac{(m_i - \mu_i(x))^2}{2\sigma_i^2(x)} + \sum_i \ln(\sigma_i(x)) \right] \quad (2)$$

The SBP method requires that statistics (e.g., mean and variance) of the detector response be characterized as a function of interaction location. The detector response at a given location is measured using a coincidence timing collimated point source with a spot size of ~0.6 mm full width at half maximum (FWHM). To calibrate the detector, the point source flux is raster scanned across the face of the detector at a user specified sampling interval. Two SBP lookup tables (LUTs) corresponding to the mean and variance of the light probability density function (PDF) versus (x,y) position are created during the characterization process. For this work, the detector was sampled at 1.013 mm intervals in both X and Y and the LUTs were interpolated to 0.253 mm bins. To speed up the detector characterization process we have built a multi-coincidence source apparatus [25]. The device allows us to characterize a cMiCE detector at 16 spatially separated locations simultaneously, thus reducing the time required for detector calibration.

To support 3D positioning in a cMiCE detector, we have developed a maximum-likelihood-based method for building 3D LUTs. The method used to separate the data into different depth regions is described elsewhere [17,22,23]. Initially the data are sorted to provide equal sensitivity per depth; however, during the calibration procedure the counts per DOI region are grouped according to the characteristics of the data. Therefore after depth separation, the heights of the regions and the sensitivity of each region may vary. Therefore the counts associated with each depth are recorded and used to determine the height of each DOI region. We assume the interaction probability with depth follows an exponential distribution. After the 3D LUTs are generated, the maximum likelihood estimator for event positioning is utilized as described above. For our 8 mm thick cMiCE detectors, we use four DOI LUTs for event positioning [21].

C. cMiCE Prototype: System Assembly

The prototype cMiCE detector imaging system consisted of two cMiCE detector modules using 8 mm thick LYSO monolithic detectors, a rotating stage, and a 128-channel data acquisition system. The cMiCE detectors were mounted onto a horizontal gantry plate with a detector separation of 10.7 cm, as pictured in Figure 2. The PMT was operated at -950 V.

For imaging, the source object was placed on the imaging platform. Data were collected for 30 view angles over 180 degrees and then processed off-line. The data acquisition electronics consisted of a custom made front-end amplifier board, NIM electronics; a VME-based acquisition system, and an Apple computer running the ORCA software package [26]. The front-end amplifier board amplified each of the 64 signal channels and the last dynode signal from each of the PMTs. The last dynode signals from the amplifier boards were fed to an ORTEC octal constant fraction discriminator (CFD) module (OCT CF8000, Oak Ridge, TN). The CFD outputs were fed to an ORTEC CO4020 quad 4-input logic unit for overlap coincidence determination. The output of the logic unit was routed to a LeCroy Model 222 dual gate generator (LeCroy Corporation, Chestnut Ridge, NY) and a custom VME board for control of the gate signal that was sent to the analog to digital converters (ADCs). The ADCs were CAEN V792 32-channel charge integrating ADCs (CAEN, Italy). Four units were used to support 128-signal channel acquisition. The data acquisition software was the ORCA package developed at the University of Washington [26].

D. cMiCE Prototype: Data Processing

The data were collected event by event, stored to disk and processed off-line. The first step of data processing was energy windowing the data. The individual PMT channels were normalized and summed and a 410-625 keV energy window was applied to the data. The anode uniformity map provided by the manufacturer of the PMTs was used to correct for the varying light sensitivity of the PMT anodes. The data were then positioned within the detector using our SBP algorithm. Data were positioned using four DOI SBP lookup tables and also lookup tables without DOI. An event was kept if it was positioned within the 46×46 mm² useful imaging area of our detectors, otherwise it was discarded. The fully-3D coincidence data were re-binned to a stack of 2D sinograms using single slice rebinning (SSRB) [24] and then reconstructed using filtered back projection with a ramp filter. The maximum axial acceptance angle for SSRB was set to 5.6 degrees. A sampling density correction was applied to the sinogram data before image reconstruction. No corrections were applied for attenuation, scattered or random coincidences, detector normalization or dead time.

The data acquisition and reconstruction parameters were as follows. For image resolution determination, the data were binned into 400×0.125 mm distance bins and 180×1 degree angle bins. The data were reconstructed using filtered back projection with a non-apodized ramp filter. The image resolution was determined by summing the profiles corresponding to the profile through the maximum voxel in the image plus the profiles ± 4 voxels along the radial and tangential dimensions of the three dimensional image volume. Therefore the profile used to determine the image resolution of the system is the sum of 81 profiles. This was done to reduce the effects of noise in the measurement. For image display, the data were binned into 200×0.25 mm distance bins and 180×1 degree angle bins.

III. Results

A. cMiCE Prototype: Detector Results

The intrinsic spatial resolution characteristics of our cMiCE detectors were determined during the detector calibration procedure. Contour plots representing the FWHM of the intrinsic spatial resolution characteristics for each of the detectors are illustrated in Figure 3. The indicated test locations are separated by 2.026 mm. The average X and Y spatial resolution is 1.28 ± 0.35 mm FWHM over a 46.6×46.6 mm² region of the first detector. The average energy resolution was 17.2% at 511 keV. For the second detector the average X and Y spatial resolution was 1.35 ± 0.34 mm FWHM. The average energy resolution was

18.4 %. The DOI spatial resolution for both 8 mm thick cMiCE detectors was ~ 3.5 mm FWHM.

B. cMiCE Prototype: Imaging Results

An image of the reconstructed point source repositioned to four radial displacements is shown in Figure 4. After energy windowing and single slice binning, there were approximately 40K to 50K events for each of the reconstructed point sources. To aid in viewing the source locations, the images were scaled to the same maximum voxel value and then added together for Figure 4. A profile through points at ~ 1 mm and ~ 5 mm offset is illustrated in Figure 5. The measured reconstructed image resolution for each of the separate point source images is listed in Table 1. In addition to positioning the data using four DOI bins, the data were also positioned and reconstructed without DOI information. The image resolution results without the use of DOI for positioning are listed in Table 2. The reconstructed image resolution was approximately 16% improved when using the DOI information for positioning. There was no benefit to DOI positioning in regards to the axial image resolution. The effect of using DOI information was not visually apparent in reconstructed images of the point sources. However, since there was a measured improvement in reconstructed image resolution, we believe the DOI information will improve quantitation when imaging small objects.

IV. Discussion and conclusion

A two detector prototype cMiCE PET imaging system has been constructed and initial point source images have been collected to demonstrate the image resolution capabilities of the system. Our results display the excellent intrinsic spatial resolution that can be achieved using monolithic crystal detectors. The average intrinsic spatial resolution achieved for our two cMiCE detectors were better than 1.35 mm FWHM for the useful imaging area of the detectors. Using our SBP algorithm, we were also able to use $\sim 85\%$ of the surface area of our cMiCE detectors for imaging, $46 \times 46 \text{ mm}^2$ out of $50 \times 50 \text{ mm}^2$. While we believe that an 85% useful imaging area for our cMiCE detectors is adequate for excellent system performance we have also evaluated and characterized a trapezoidal shaped cMiCE detector [27]. Using that design the useful imaging area increases to 93% and it also virtually eliminates the gaps between the crystals in the octagonal ring geometry. While we used rectangular crystals in this work, we envision using trapezoidal crystals in a final system design. We do not envision rotating the detector system; however, that is also another design possibility.

The cMiCE detector modules also provide DOI positioning information. In this work, we demonstrated the benefit of using four DOI bins on the reconstructed image resolution. Although the number of DOI bins can be increased, this approach also increases the memory resources required for a real-time implementation of the SBP method with marginal improvements to the DOI resolution. Using four DOI LUTs, we measured a DOI positioning resolution of ~ 3.5 mm [21]. The reconstructed image resolution was approximately 16% improved when using the DOI information. There was no benefit to DOI positioning in regards to the axial image resolution, probably because of the restricted axial acceptance angle that was used to limit the effects of SSRB blurring. The effect of using DOI information was not visually apparent in reconstructed images of the point sources. However, since there was a measured improvement in reconstructed image resolution, we believe the DOI information will improve quantitation when imaging small objects.

A third benefit of continuous crystal detectors is that they can provide detector sampling finer than their intrinsic spatial resolution. This helps to maintain the detector's spatial resolution characteristics during the image reconstruction process.

It is worth noting that when using large area detectors in an octagonal geometry there is a parallax problem for point sources very close to the center of the field of view. This was previously demonstrated by the PENNPET system [28]. This probably explains why the image resolution is better at the center of the field of view when DOI is taken into account. This is not the case in discrete crystal ring detector designs.

Overall the reconstructed in-plane image resolution approached 1.0 mm FWHM using filtered back projection with a ramp filter, without adjusting for the size of the source. The image resolution was fairly uniform for the central 30 mm of the imaging field of view that we tested and can support whole body imaging of a mouse at ~1.0 mm image resolution. The axial resolution was also ~1.0 mm FWHM near the center of the imaging field of view. The axial resolution showed some broadening as the point source was moved away from the center of the field of view due to the use of SSRB. Implementing fully 3D image reconstruction with point source modeling could lead to reconstructed image resolution below 1.0 mm FWHM, and can potentially achieve the 0.7 mm FWHM image resolution that would provide species equivalent image resolution between mice and men.

We note a slight asymmetry in the reconstructed point sources. The radial resolution was slightly broader than the transverse resolution. We believe that the asymmetry is caused by a slight misalignment of the detectors that is not accounted for during the image reconstruction. Methods to correct for detector misalignment for high resolution PET imaging systems are an active area of interest in our laboratory [29].

In addition, some of the limitations of this work were the slow data acquisition bandwidth provided by the VME-based data acquisition system that limited the count rate capabilities of the camera and needing to do the SBP positioning off-line. To address both of these issues we are currently designing a high-throughput data acquisition electronics board [30]. We have implemented both 2D and 3D SBP algorithms in our FPGA [31,32]. The 2D SBP implementation runs at over 360 KHz and the 3D SBP implementation runs at over 220 KHz. We believe that these bandwidths are suitable to handle the maximum expected coincidence qualified counting rate for a single detector.

Other limitations to this work include the use of only two detector modules and the use of single slice binning of the collected data. While the detector spacing was 10.7 cm, the maximum imaging field of view was 46.6 mm (i.e., equal to width of the useful imaging area of each cMiCE detector). Realistically even imaging at 20.0 mm off-center causes large normalization corrections due to sensitivity differences at the different projection angles. Furthermore there was significant single slice binning artifacts for a reconstructed image of a point source 20.0 mm off-center in the scanner and therefore we decided to not include that data point in this work. We would expect any artifacts due to single slice binning would not be present for images produced using FORE plus two-dimensional reconstruction or fully three-dimensional reconstruction methods. However, we believe using single slice binning and filtered back-projection was adequate for this proof of concept study to demonstrate the image resolution potential of our cMiCE detectors.

The current cMiCE design requires the acquisition and processing of many electronic channels. We believe that we have partially addressed this issue by designing compact data acquisition electronics that are capable of implementing our positioning algorithm in real time. Methods to reduce the number of positioning channels are to go to multiplexing schemes. We have investigated row-column summing and other multiplexing schemes (e.g., collecting individual signals from the edge and corner channels and multiplexing the inner channels, etc.); however, it degraded performance for an 8×8 sensor element photodetector [33]. On the other hand when we investigated row-column summing for 12×12 or 16×16

sensor element photodetectors, it performed very well [22,23]. While our current cMiCE detector is based upon PMT technology, we envision future designs using denser SiPM arrays and row-column signal multiplexing to reduce the number of channels that need to be acquired while also improving spatial resolution due to finer sampling of the crystal using smaller pixel SiPM arrays.

In addition to the 8 mm thick cMiCE detectors we described in this work, we have fabricated and characterized cMiCE detectors with LYSO crystals as thick as 15 mm. For the 15 mm thick detector we measured an intrinsic spatial resolution of 1.74 ± 0.35 mm FWHM in X and Y and 4.8 ± 0.36 mm FWHM in DOI using 7 DOI LUTs [21]. These results are consistent with our previous results that indicated an inherent trade-off in detection versus quantitation task performance due to changes in crystal thickness (i.e., sensitivity) versus intrinsic spatial resolution [34]. These results also demonstrate the scalability of our cMiCE detector design. For example, an 8 mm thick crystal may be ideal for a small animal imaging system, but a 15 mm thick crystal may be more suitable for a head or breast imaging system where one can trade-off some image resolution for detection efficiency.

In summary, these results demonstrate that monolithic crystal PET detectors are a potentially cost effective viable alternative to finely pixilated, discrete crystal detectors for the development of ultra-high resolution PET systems.

Acknowledgments

This work was supported in part by the NIH grants NIBIB EB001563 and EB002117.

REFERENCES

1. Cherry SR, et al. MicroPET: a high resolution PET scanner for imaging small animals. *IEEE TNS*. 1997; 44(3):1161–1166.
2. Bloomfield PM, Rajeswaran S, Spinks TJ, Hume SP, Myers R, Ashworth S, Clifford KM, Jones WF, Byars LG, Young J, Andreaco M, Williams CW, Lammertsma AA, Jones T. The design and physical characteristics of a small animal positron emission tomograph. *Phys Med Biol*. 1995; 40:1105–1126. [PubMed: 7659733]
3. Jeavons AP, Chandler RA, Dettmar CAR. A 3D HIDAC-PET camera with sub-millimetre resolution for imaging small animals. *IEEE TNS*. 1999; 46(3):468–473.
4. Tai YC, et al. MicroPET II: design, development and initial performance of an improved microPET scanner for small-animal imaging. *Phys Med Biol*. 2003; 48(11):1519–1537. [PubMed: 12817935]
5. Miyaoka RS, Janes ML, Lee K, Park B, Kinahan PE, Lewellen TK. Development of a single detector ring micro crystal element scanner: QuickPET II. *Molecular Imaging*. 2005; 4(2):117–127. [PubMed: 16105510]
6. Lee K, Kinahan PE, Fessler JA, Miyaoka RS, Janes M, Lewellen TK. Pragmatic fully 3D image reconstruction for the MiCES mouse imaging PET scanner. *Phys Med Biol*. 2004; 49(19):4563–4578. [PubMed: 15552417]
7. Miyaoka RS, Kohlmyer SG, Lewellen TK. Performance characteristics of micro crystal element (MiCE) detectors. *IEEE TNS*. 2001; 48(4):1403–1407.
8. Lee K, Miyaoka RS, Janes ML, Lewellen TK. Detector characteristics of the micro crystal element scanner (MiCES). *IEEE TNS*. 2005; 52(5):1428–1433.
9. Joung J, Miyaoka RS, Lewellen TK. cMiCE: a high resolution animal PET using continuous LSO with a statistics based positioning scheme. *NIM Section A*. 2002; 489(1-3):584–598.
10. Lewellen TK. Recent developments in PET detector technology. *Phys Med Biol*. 2008; 53:R287–R317. [PubMed: 18695301]
11. Tavernier S, Bruyndonckx P, Leonard S, Devroede O. A high-resolution PET detector based on continuous scintillators. *Nucl. Instrum. Meth. Phys. Res. A*. 2005; 537:321–325.

12. Lerche CW, et al. Depth of gamma-ray interaction within continuous crystals from the width of its scintillation light-distribution. *IEEE TNS*. 2005; 52(3):560–572.
13. Hunter WCJ, Barrett HH, Furenlid LR. Calibration method of ML estimation of 3D interaction position in a thick gamma-ray detector. *IEEE Trans. Nucl. Sci.* 2009; 56(1):189–196. [PubMed: 20191099]
14. Maas MC, et al. Experimental characterization of monolithic-crystal small animal PET detectors read out by APD arrays. *IEEE Trans Nuc Sci.* 2006; 53(3):1071–1077.
15. LeBlanc JW, Thompson RA. A novel PET detector block with three dimensional hit position encoding. *Nuclear Science Symposium Conference Record*, 2003 IEEE. 2003; 3:1718–1722.
16. Ling T, Lee K, Miyaoka RS. Performance comparisons of continuous miniature crystal element (cMiCE) detectors. *IEEE TNS*. 2006; 53(6):2513–2518.
17. Ling T, Lewellen TK, Miyaoka RS. Depth of interaction decoding of a continuous crystal detector module. *Physics in Medicine and Biology*. 2007; 52:2213–2228. [PubMed: 17404465]
18. Milster TD, et al. A full-field modular gamma camera. *J Nucl Med*. 1990; 31(5):632–639. [PubMed: 2341900]
19. Gray RM, Macovski A. Maximum a posteriori estimation of position in scintillation cameras. *IEEE TNS*. 1976; 23(1):849–852.
20. Clinthorne NH, Rogers WL, Shao L, Koral KF. A hybrid maximum likelihood position computer for scintillation cameras. *IEEE TNS*. 1987; 34(1):97–101.
21. Miyaoka RS, Li X, Lockhart C, Lewellen TK. New continuous miniature crystal element (cMiCE) detector geometries. *Nuclear Science Symposium Conference Record*, 2009. NSS '09. IEEE. 2009:3639–3642.
22. Li X, Lockhart C, Lewellen TK, Miyaoka RS. A high resolution, monolithic crystal, PET/MRI detector with DOI positioning capability. *IEEE EMBS*. 2008:2287–2290.
23. Miyaoka RS, Li X, Lockhart C, Lewellen TK. Comparison of detector intrinsic spatial resolution characteristics for sensor on the entrance surface and conventional readout designs. *IEEE TNS*. 2010; 57(3):990–997.
24. Daube-Witherspoon ME, Muehllehner G. Treatment of axial data in three-dimensional PET. *J Nucl Med*. 1987; 28(11):1717–1724. [PubMed: 3499493]
25. Miyaoka RS, Ling T, Lockhart C, Li X, Lewellen TK. Calibration procedure for a continuous miniature crystal element (cMiCE) detector. *IEEE TNS*. 2010; 57(3):1023–1028.
26. Howe MA, Cox GA, Harvey PJ, McGirt F, Rielage K, Wilkerson JF, Wouters JM. Sudbury neutrino observatory neutral current detector acquisition software overview. *IEEE TNS*. 2004; 51(3):878–883.
27. Li X, Hunter WCJ, Lewellen TK, Miyaoka RS. Design of a trapezoidal slat crystal (TSC) PET detector for small animal PET/MR imaging. *Nuclear Science Symposium Conference Record*, 2010. NSS '10. IEEE. 2010 in press.
28. Karp JS, Muehllehner G, Mankoff DA, Ordonez CE, Ollinger JM, Daube-Witherspoon ME, Haigh AT, Beerbohm DJ. Continuous-slice PENN-PET: A positron tomograph with volume imaging capability. *J Nucl Med*. 1990; 31:617–627. [PubMed: 2341898]
29. Pierce L, Miyaoka RS, Lewellen TK, Alessio A, Kinahan PE. Determining block detector positions for PET scanners. *Nuclear Science Symposium Conference Record*, 2009. NSS '09. IEEE. 2009:2976–2980.
30. Lewellen TK, Miyaoka RS, MacDonald LR, Haselman M, DeWitt D, Huack S. Evolution of the design of a second generation firewire based data acquisition system. *Nuclear Science Symposium Conference Record*, 2010. NSS '10. IEEE. 2010 in press.
31. DeWitt D, Johnson-Williams NG, Miyaoka RS, Li X, Lockhart C, Lewellen TK, Hauck S. Design of an FPGA-based algorithm for real-time solutions of statistics-based positioning. *IEEE TNS*. 2010; 57(1):71–77.
32. Johnson-Williams NG, Miyaoka RS, Li X, Lewellen TK, Hauck S. Design of a real time FPGA-based three dimensional positioning algorithm. *IEEE TNS*. accepted for publication.
33. Miyaoka RS, Ling T, Lewellen TK. Effect of number of readout channels on the intrinsic spatial resolution performance of a continuous miniature crystal element (cMiCE) detector. *IEEE Trans Nuc Sci*. Oct; 2007 54(5):1561–7.

34. Lee K, Kinahan PE, Miyaoaka RS, Jae-Seung K, Lewellen TK. Impact of system design parameters on image figures of merit for a mouse PET scanner. *IEEE TNS*. 2004; 51(1):27–33.

Fig. 1. cMiCE design PET detector with $50 \times 50 \times 8$ mm³ LYSO crystal coupled to H8500 8×8 multi-anode PMT. Crystal is coupled to PMT using optical grease and secured in place using black electrical tape as shown in picture.

Figure 2. Photograph of prototype cMiCE scanner showing two cMiCE detector modules and a rotating stage. Point sources were placed on an imaging platform and data were collected at 30 view angles (i.e., 6 degree angular steps).

Figure 3.
FWHM contour plots using four levels of DOI SBP positioning: (a) Detector 1, (b) Detector 2. The test locations are indicated by the dots, which are separated by 2.026 mm.

Figure 4.

Image of reconstructed point sources using DOI information. The point source was positioned at approximately 1, 5, 10 and 15 mm from the center of the field of view. Profile through points at ~1 mm and ~5 mm offset (i.e., dashed line) is plotted in Figure 6.

Figure 5. Profile through points at ~1 mm and ~5 mm offset from the center of the imaging field of view as shown in Figure 5.

Table 1

Reconstructed image resolution using DOI information.

Radial Offset	Radial FWHM (mm)	Tangential FWHM (mm)	Axial FWHM (mm)
1 mm	1.11 ± 0.04	0.96 ± 0.07	0.93 ± 0.05
5 mm	1.08 ± 0.05	0.95 ± 0.01	1.02 ± 0.05
10 mm	0.97 ± 0.04	1.01 ± 0.04	1.32 ± 0.06
15 mm	1.06 ± 0.13	1.06 ± 0.09	1.69 ± 0.05
Average (4 points)	1.05 ± 0.08	0.99 ± 0.07	1.24 ± 0.31

Table 2

Reconstructed image resolution without DOI information.

Radial Offset	Radial FWHM (mm)	Tangential FWHM (mm)	Axial FWHM (mm)
1 mm	1.24 ± 0.11	1.16 ± 0.04	0.91 ± 0.05
5 mm	1.20 ± 0.10	1.17 ± 0.08	0.99 ± 0.06
10 mm	1.16 ± 0.04	1.23 ± 0.12	1.23 ± 0.09
15 mm	1.33 ± 0.19	1.20 ± 0.07	1.68 ± 0.11
Average (4 points)	1.23 ± 0.12	1.19 ± 0.08	1.20 ± 0.32