Published in final edited form as:

IEEE Nucl Sci Symp Conf Rec (1997). 2012 ; 2012: 2123–2127. doi:10.1109/NSSMIC.2012.6551486.

A Prototype Detector for a Novel High-Resolution PET System: BazookaPET

Ryeojin Park,

Department of Medical Imaging, University of Arizona, Tucson, AZ, 85724 USA. They are also with the College of Optical Sciences, University of Arizona, Tucson, AZ 85721 USA

Brian W. Miller [Member IEEE],

Department of Radiology, University of Arizona, Tucson, AZ 85724 USA and is now appointed in Pacific Northwest National Laboratory, Radiation Detection and Nuclear Sciences Group, National Security Directorate, Richland, WA 99352 USA

Abhinav K. Jha,

Department of Medical Imaging, University of Arizona, Tucson, AZ, 85724 USA. They are also with the College of Optical Sciences, University of Arizona, Tucson, AZ 85721 USA

Lars R. Furenlid [Member IEEE],

Department of Medical Imaging, University of Arizona, Tucson, AZ, 85724 USA. They are also with the College of Optical Sciences, University of Arizona, Tucson, AZ 85721 USA

William C. J. Hunter [Member IEEE], and

Department of Radiology, University of Washington, Seattle, WA 98195 USA

Harrison H. Barrett [Fellow IEEE]

Department of Medical Imaging, University of Arizona, Tucson, AZ, 85724 USA. They are also with the College of Optical Sciences, University of Arizona, Tucson, AZ 85721 USA

Ryeojin Park: rpark@email.arizona.edu

Abstract

We have designed and are developing a novel proof-of-concept PET system called BazookaPET. In order to complete the PET configuration, at least two detector elements are required to detect positron-electron annihilation events. Each detector element of the BazookaPET has two independent data acquisition channels. One side of the scintillation crystal is optically coupled to a 4×4 silicon photomultiplier (SiPM) array and the other side is a CCD-based gamma camera. Using these two separate channels, we can obtain data with high energy, temporal and spatial resolution data by associating the data outputs via several maximum-likelihood estimation (MLE) steps. In this work, we present the concept of the system and the prototype detector element. We focus on characterizing individual detector channels, and initial experimental calibration results are shown along with preliminary performance-evaluation results. We measured energy resolution and the integrated traces of the slit-beam images from both detector channel outputs. A photo-peak energy resolution of ~5.3% FWHM was obtained from the SiPM and ~48% FWHM from the CCD at 662 keV. We assumed SiPM signals follow Gaussian statistics and estimated the 2D interaction position using MLE. Based on our the calibration experiments, we computed the Cramér-Rao bound (CRB) for the SiPM detector channel and found that the CRB resolution is better than 1 mm in the center of the crystal.

Index Terms

silicon photomultiplier (SiPM); image intensifier; monolithic $\text{LaBr}_3\text{:Ce}$; PET imaging; molecular imaging; CCD-based gamma camera

I. Introduction

In the past few decades, a tremendous interest in small-animal PET scanning has arisen since it provides a tool for understanding a wide range of biological processes [1]–[4]. In order to provide greater specificity and accuracy, a high-spatial-resolution PET scanner is required and various designs have been proposed to achieve this high spatial resolution [5]–[10]. Recent advances in CCD technology have allowed for high-spatial-resolution small-animal single-photon-emission-computed-tomography (SPECT) systems using CCD-based gamma cameras. BazookaSPECT has demonstrated its capability as a high-resolution, photon-counting gamma-ray detector [11], [12]. Because the light pulses of individual gamma-ray events are well separated spatially and can be segmented at each frame, a high event-detection rate ($\sim 1.1 \times 10^6$ events/second) can be achieved by the large-area BazookaSPECT [13]. Despite its excellent spatial resolution, a CCD-based gamma camera has not been considered as a PET detector because its temporal resolution is limited by the frame rate, and it provides only minimal energy resolution.

In this work, we extend the high-spatial-resolution concept of BazookaSPECT to PET by making use of a secondary photodetector with better energy and timing information. We are able to correlate events from these two detectors by correlating event positions. We can then retrospectively form a joint estimate of position using information from both detectors. In order to detect positron-electron annihilation events, BazookaPET consists of at least two detector elements (Fig. 1a). A detector element consists of a scintillation crystal with two independent detector channels (Fig. 1b). One detector channel is a silicon photomultiplier (SiPM) array, which is considered as a more conventional PET detector and the other is an image intensifier with CCD, which is the same as the BazookaSPECT configuration. We refer to the SiPM detector channel as the SiPM side and the other detector channel as the Bazooka side. Although there has been a prior effort to combine an EMCCD with SiPM arrays to improve the photon-counting performance and energy resolution of the EMCCD [14], [15], the configuration was intended for a small-animal SPECT application.

By associating the Bazooka-side readouts with the SiPM side, BazookaPET is intended to achieve fast coincidence-timing resolution, good energy resolution and high spatial resolution. Key to reaching these goals is the use of maximum-likelihood estimation (MLE) for both the Bazooka-side and the SiPM-side readouts. However, this paper focuses on the calibration and characterization of each detector channel before we investigate the association between the two detector channels.

II. Materials and Methods

A. Motivations and Detector Design

A schematic of the detector element of BazookaPET and the flowchart of data acquisition and MLE steps to estimate the 3D interaction position ($\mathbf{r}$) and energy is shown in Fig. 2. When a 511 keV gamma ray traverses through the 4×4 SiPM array and interacts within the scintillation crystal, a portion of the scintillation light is collected by the SiPM. The histograms of all 16 pixel outputs are shown in Fig. 2 when a gamma-ray beam collimated by a slit was incident on the crystal. The light from each scintillation event is also imaged onto the microchannel-plate image intensifier by a pair of F/1.4 lenses, with both lenses focused to infinity in what we refer to as a back-to-back lens configuration. The back-to-back lens configuration reduces aberrations while maintaining high light-collection efficiency. Finally, the image is acquired by a CCD camera which has 640×480 pixels with a pixel pitch of 7.4 μm and a frame time of a few milliseconds. The image intensifier gain mitigates the light loss due to the lens coupling.

In order to produce a data set with high spatial and temporal resolution, we need to associate the SiPM signals with the CCD data. In order to moderate the difficulties of the association, several sophisticated MLE steps are required. First, we estimate the 3D interaction position and the energy of the gamma ray using the SiPM signals. Parallel to this MLE, we estimate the depth of interaction (DOI) using the CCD data. It has been shown previously that by considering the light spread of the CCD signals we can estimate the depth of interaction [16]. The lateral interaction position from the Bazooka side is estimated from the centroid position of light spread over a cluster of pixels. As an example of this lateral centroid position estimation, Fig. 2 shows a resulting integrated centroid image when a gamma-ray slit beam was incident on the crystal. After these two position MLE steps, we perform the last MLE which estimates the association of these two data sets.

To generate the ML estimate of $\mathbf{r}$ for given SiPM signals $\mathbf{g}$, we are currently working on the probability model $\text{pr}(\mathbf{g}|\mathbf{r}, \mathcal{E})$ where $\mathcal{E}$ is energy of a gamma ray. For accurate modeling, we combined a Geant4 simulation with Monte Carlo simulations of SiPM response [17], [18]. In the simulations, the 4×4 SiPM array is optically coupled to a 13×13×5 mm³ LaBr₃:Ce crystal. The Monte-Carlo simulation of SiPM response uses scintillation photons generated by the Geant4 simulation as inputs [19]. We have confirmed using simulation that the sampled probability distribution follows a Gaussian model. However, the experimental validation of the model will be investigated in the future.

B. Maximum-likelihood Estimation for SiPM signals

In this work, we considered only the estimation of the 2D interaction position ($\mathbf{R}$) and we will revisit 3D positioning in future work. The SiPM pixel outputs are assumed to be independent, and each SiPM pixel output is assumed to follow a Gaussian distribution:

$$\text{pr}(\mathbf{g}|\mathbf{R}, \mathcal{E}_0) \prod_{m=1}^M \frac{1}{\sqrt{2\pi K_{mm}(\mathbf{R}, \mathcal{E}_0)}} \times \exp \left[-\frac{[g_m - \bar{g}_m(\mathbf{R}, \mathcal{E}_0)]^2}{2K_{mm}(\mathbf{R}, \mathcal{E}_0)} \right]. \quad (1)$$

The ML estimate of the position is found by searching over $\mathbf{R}$ with a fixed energy ε_0 . Since the logarithm is a monotonic transformation, we can take a logarithm of the probability density function without loss of generality:

$$\hat{\mathbf{R}}_{ML} \equiv \underset{\mathbf{R}}{\operatorname{argmax}} \{ \ln[\operatorname{pr}(\mathbf{g}|\mathbf{R}, \varepsilon_0)] \}. \quad (2)$$

Here, $\mathbf{g} = \{g_1, \dots, g_M\}$ is a SiPM-data output vector and $\varepsilon_0 = 662$ keV. An exhaustive search algorithm was implemented to find the ML estimate. We used the *likelihood windowing* method to filter events [20]–[22]. After we found $\hat{\mathbf{R}}_{ML}$, we accepted the event only if the resulting maximized log likelihood, $\ln[\operatorname{pr}(\mathbf{g}|\hat{\mathbf{R}}_{ML}, \varepsilon_0)]$ exceeded some threshold. The mean, $\bar{g}_m(\mathbf{R}, \varepsilon_0)$ and variance, $K_{mm}(\mathbf{R}, \varepsilon_0)$ of the SiPM signals were obtained by calibration experiments.

The Cramér-Rao bound (CRB) resolution indicates the best possible detector resolution in terms of the variance of the position estimate [22], [23]. Therefore, we also computed CRBs in order to assess the detector resolution (Fig. 3) of the SiPM side from the calibration results. The result shows that the CRB resolution is better than 1 mm in the center of the crystal. The CRB resolution is low at the lower bottom corner of the crystal due to defects in the crystal packaging.

III. Results and Discussion

A. Integrated Traces of a Slit-Beam Images and Energy Resolution Comparison

As an initial attempt to compare the spatial resolution of estimates using the outputs from each detector channel, we illuminated the scintillation crystal with a collimated gamma-ray slit beam of width 0.6 mm at 662 keV. The slit-beam images were acquired simultaneously from both detector channels. The integrated traces of the slit-beam images are shown in Fig. 4. It is evident that the Bazooka side has advantageous spatial resolution even though the optical coupling between the optical window and the crystal was not ideal. We also compared the energy resolution between two detector channel outputs (Fig. 5) using a collimated cross-slit beam at 662 keV. As we expect, the energy resolution from the SiPM side shows a good result of ~5.3% FWHM while the energy resolution from the Bazooka side is ~48% FWHM. In addition, we investigated how the over-voltage degrades energy resolution for the SiPM side (Fig. 6).

B. Light emission from Silicon Photomultiplier

The emission of visible light as a consequence of avalanche processes is a well-known phenomenon [24], [25]. As a main drawback for SiPM-based imaging cameras, this optical cross-talk effect has been actively studied [26]–[28]. Because of the unique design of the BazookaPET detector element, one of the challenges in association between two detector channels is that the Bazooka side directly detects the light emission from the SiPM avalanches (Fig. 7). The images in Fig. 7 are the integrated centroid images of the Bazooka side when a collimated gamma-ray beam was incident on the crystal. As shown in Fig. 7a, when the bias voltage was zero, we were able to identify the gamma-ray events, but when

the bias voltage was applied, the Bazooka side started to detect light emission from the SiPM avalanches (Fig. 7b). It was found that the wavelength of the radiation emitted by the SiPM is predominantly greater than 550 nm [29]. Therefore, we inserted a short-pass filter along with IR-absorbing glass to block the light emission. Since the peak emission wavelength of the LaBr₃:Ce is ~380 nm, the scintillation light was not blocked. After inserting filters, most of the unwanted visible light from the SiPM was suppressed (Fig. 7d). This allows us to distinguish the gamma-ray events from the random light emission from the SiPM avalanches.

IV. Conclusion and Future Work

In conclusion, we have completed the design and construction of a prototype detector element for the BazookaPET and presented the initial calibration results of the SiPM side. We have also shown preliminary performance-evaluation results between two detector channels. Despite the defects in the crystal packaging, we showed that the spatial-resolution advantage of the Bazooka side is evident. Higher spatial resolution on the Bazooka side can be achieved with better optical coupling between the crystal and the optical window. The timing resolution will be determined by the SiPM-side readout performance (~100 MHz). We measured ~5.3% FWHM energy resolution on the SiPM side without temperature regulation. Further improvements in the stability of the SiPM with temperature control can lead to improved energy resolution [30]. On the Bazooka side, ~48% FWHM energy resolution indicates that we can use only the position estimate to associate between two detector channel outputs. We have also addressed the challenge of the Bazooka side detecting the photon emission from SiPM avalanches and established a method to suppress the unwanted light emission from the SiPM. Our next task is to develop the association algorithm between the two detector channels and investigate the light spread to achieve good estimates for depth of interaction.

Acknowledgments

The Center for Gamma-Ray Imaging is supported by NIH Grant P41 EB002035. Aspects of this work were supported by NIH grant R01 EB 000803. R. Park, is partially supported by the Technology and Research Initiative Fund fellowship. The authors would like to thank Ronan J. Havelin and H. B. Barber for assistance in image acquisition and insightful discussion. The authors also would like to thank James Proffitt of AiT Instruments for assistance in the SiPM data-acquisition electronics.

References

1. Phelps ME. Positron emission tomography provides molecular imaging of biological processes. *Proc Natl Acad Sci USA*. 2000; 97(16):9226–9233. [PubMed: 10922074]
2. Cherry SR, Gambhir SS. Use of positron emission tomography in animal research. *ILAR J*. 2001; 42(3):219–232. [PubMed: 11406721]
3. Hume S, Gunn R, Jones T. Pharmacological constraints associated with positron emission tomographic scanning of small laboratory animals. *Eur J Nucl Med Mol I*. 1998; 25(2):173–176.
4. Peng H, Levin CS. Recent developments in PET instrumentation. *Curr Pharm Biotechnol*. 2010; 11(6):555–571. [PubMed: 20497121]
5. James SS, Yang Y, Bowen SL, Qi J, Cherry SR. Simulation study of spatial resolution and sensitivity for the tapered depth of interaction PET detectors for small animal imaging. *Phys Med Biol*. 2010; 55(2):N63. [PubMed: 20023331]

6. Chatziioannou AF, Cherry SR, Shao Y, Silverman RW, Meadors K, Farquhar TH, Pedarsani M, Phelps ME. Performance evaluation of microPET: a high-resolution lutetium oxyorthosilicate PET scanner for animal imaging. *J Nucl Med.* 1999; 40(7):1164. [PubMed: 10405138]
7. Schaart DR, van Dam HT, Seifert S, Vinke R, Dendooven P, Lhner H, Beekman FJ. A novel, SiPM-array-based, monolithic scintillator detector for PET. *Phys Med Biol.* 2009; 54(11):3501. [PubMed: 19443953]
8. Moehrs S, Del Guerra A, Herbert D, Mandelkern M. A detector head design for small-animal PET with silicon photomultipliers (SiPM). *Phys Med Biol.* 2006; 51(5):1113. [PubMed: 16481681]
9. Pratz G, Levin CS. Accurately positioning events in a high-resolution PET system that uses 3D CZT detectors. *IEEE Nucl Sci Symp Conf Rec, 2007 IEEE.* 2007; 4:2660–2664.
10. Goorden M, van der Have F, Kreuger R, Ramakers R, Vastenhouw B, Burbach J, Booij J, Molthoff C, Beekman F. VECTor: A preclinical imaging system for simultaneous submillimeter SPECT and PET. *J Nucl Med.* 2013 to be published.
11. Miller BW, Barber HB, Barrett HH, Wilson DW, Chen L. A low-cost approach to high-resolution, single-photon imaging using columnar scintillators and image intensifiers. *IEEE Nucl Sci Symp Conf Rec, 2006 IEEE.* Nov.2006 6:3540–3545.
12. Miller BW, Barrett HH, Furenlid LR, Barber HB, Hunter RJ. Recent advances in BazookaSPECT: real-time data processing and the development of a gamma-ray microscope. *Nucl Instrum Meth A.* 2008; 591(1):272–275.
13. Miller BW, Barber HB, Barrett HH, Liu Z, Nagarkar VV, Furenlid LR. Progress in BazookaSPECT: high-resolution dynamic scintigraphy with large-area imagers. *Proc SPIE.* 2012; 8508:85080F.
14. Heemskerk J, Korevaar M, Huizenga J, Kreuger R, Schaart D, Goorden M, Beekman F. An enhanced high-resolution EMCCD-based gamma camera using SiPM side detection. *Phys Med Biol.* 2010; 55(22):6773. [PubMed: 21030743]
15. Salvador S, Korevaar MAN, Heemskerk JWT, Kreuger R, Huizenga J, Seifert S, Schaart DR, Beekman FJ. Improved EMCCD gamma camera performance by SiPM pre-localization. *Phys Med Biol.* 2012; 57(22):7709. [PubMed: 23123792]
16. Miller BW, Barber HB, Barrett HH, Shestakova I, Singh B, Nagarkar VV. Single-photon spatial and energy resolution enhancement of a columnar CsI (Tl)/EMCCD gamma-camera using maximum-likelihood estimation. *Proc SPIE.* 2006; 6142:61421T.
17. Jha, AK.; Kupinski, MA.; van Dam, HT. Monte Carlo simulation of silicon photomultiplier output in response to scintillation induced light. *IEEE Nucl. Sci. Symp. & Med. Imag. Conf Rec., 2011.* IEEE; Oct. 2011; p. 1693-1696.
18. Jha AK, van Dam HT, Kupinski MA, Clarkson E. Simulating silicon photomultiplier response to scintillation light. *IEEE Trans Nucl Sci.* accepted.
19. Agostinelli S, et al. Geant4—a simulation toolkit. *Nucl Instrum Meth A.* 2003; 506(3):250–303.
20. Barrett, HH.; Myers, KJ. *Foundations of Image Science.* Hoboken, NJ: John Wiley and Sons; 2004.
21. Milster T, Aarsvold J, Barrett H, Landesman A, Mar L, Patton D, Roney T, Rowe R, Seacat R 3rd, et al. A full-field modular gamma camera. *J Nucl Med.* 1990; 31(5):632. [PubMed: 2341900]
22. Barrett HH, JHunter WC, Miller BW, Moore SK, Chen Y, Furenlid LR. Methods for Processing Signals From Gamma-Ray Detectors,” Maximum-Likelihood. *IEEE Trans Nucl Sci.* Jun.2009 56:725–735. [PubMed: 20107527]
23. Hunter WCJ, Barrett HH, Furenlid LR. Calibration method for ML estimation of 3D interaction position in a thick gamma-ray detector. *IEEE Trans Nucl Sci.* Feb; 2009 56(1):189–196. [PubMed: 20191099]
24. Newman R. Visible light from a silicon $p - n$ junction. *Phys Rev.* Oct.1955 100:700–703.
25. Chynoweth AG, McKay KG. Photon emission from avalanche breakdown in silicon. *Phys Rev.* Apr.1956 102:369–376.
26. Vinogradov S. Analytical models of probability distribution and excess noise factor of solid state photomultiplier signals with crosstalk. *Nucl Instrum Meth A.* 2012; 695(0):247–251.
27. Dolgoshein B, et al. Status report on silicon photomultiplier development and its applications. *Nucl Instrum Meth A.* 2006; 563(2):368–376.

28. Renker D. Geiger-mode avalanche photodiodes, history, properties and problems. Nucl Instrum Meth A. 2006; 567(1):48–56.
29. Mirzoyan R, Kosyra R, Moser H. Light emission in Si avalanches. Nucl Instrum Meth A. 2009; 610(1):98–100.
30. Roncali E, Cherry SR. Application of silicon photomultipliers to positron emission tomography. Ann Biomed Eng. 2011; 39:1358–1377. [PubMed: 21321792]

Fig. 1. (a) Concept of a two-detector-element BazookaPET system. (b) An exploded view of a single BazookaPET detector element: a scintillation crystal coupled to two independent detector channels, an SiPM array and lens coupling to an image intensifier/CCD.

Fig. 2.

Event detection concept and the data processing flow chart, including MLE steps, for a single BazookaPET detector element. Note that this schematic is for one detector element and that a pair of detector elements is required for PET imaging. The histograms of the SiPM pixel outputs and the integrated centroid image from the Bazooka side were obtained when a gamma-ray slit beam was incident on the crystal. The slit-beam width was 0.6 mm at 662 keV.

Fig. 3.

The Cramér-Rao bound on the SiPM-side resolution of the lateral position x (a) and y (b) for 662 keV gamma-ray interactions. Our measure of resolution is the full-width at half-max (FWHM) computed as $\sqrt{(2\ln(2))[\mathbf{F}^{-1}_{jj}]}$. The lower bottom corner of the crystal shows poor optical coupling between the optical window and the crystal.

(a)

(b)

Fig. 4.
 The integrated traces of a slit-beam images on (a) the SiPM side and (b) the Bazooka side.
 The slit-beam width is 0.6 mm at 662 keV.

Fig. 5. The Cs^{137} (662 keV) energy spectra for (a) the SiPM side ($\sim 5.3\%$ FWHM) and (b) the Bazooka side ($\sim 48\%$ FWHM).

Fig. 6. The Cs^{137} (662 keV) gamma-ray energy spectra of the SiPM side at various bias voltage. Energy resolution degrades rapidly as over-voltage increases.

Fig. 7.

The integrated centroid images from the Bazooka side show the detection of light emission from SiPM avalanches. The 662 keV collimated gamma-ray beam was incident on the crystal. (a) and (b) images were measured with no filters and (c) and (d) images were measured with filters. The filters suppressed most of the random photon emission from the SiPM avalanches.