

Development of a compact DOI–TOF detector module for highperformance PET systems

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Abstract To increase spatial resolution and signal-to-noise ratio in PET imaging, we present in this paper the design and performance evaluation of a PET detector module combining both depth-of-interaction (DOI) and time-offlight (TOF) capabilities. The detector module consists of a staggered dual-layer LYSO block with $2 \text{ mm} \times 2 \text{ mm} \times 7 \text{ mm}$ crystals. MR-compatible SiPM sensors (MicroFJ-30035-TSV, SensL) are assembled into an 8×8 array. SiPM signals from both fast and slow outputs are read out by a 128-channel ASIC chip. To test its performance, a flood histogram is acquired with a 22 Na point source on top of the detector, and the energy resolution and the coincidence resolving time (CRT) value for each individual crystal are measured. The flood histogram shows excellent crystal separation in both layers. The

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average energy resolution at 511 keV is 14.0 and 12.7% at the bottom and top layers, respectively. The average CRT of a single crystal is 635 and 565 ps at the bottom and top layers, respectively. In conclusion, the compact DOI–TOF PET detector module is of excellent crystal identification capability, good energy resolution and reasonable time resolution and has promising application prospective in clinical TOF PET, PET/MRI, and brain PET systems.

Keywords Positron emission tomography \cdot SiPM \cdot Depth of interaction (DOI) \cdot Time-of-flight (TOF)

1 Introduction

Silicon photomultipliers (SiPMs), due to their insensitivity to magnetic field, are promising photo sensors for simultaneous positron emission tomography and magnetic resonance imaging (PET/MRI) [1]. They are advantageous in their compact size, high gain, high quantum efficiency, and low working voltage [2]. In recent years, improvements in the SiPM technique pave the way in developing hybrid PET/MRI [3]. Many research groups are interested in designing magnetic-field-insensitive PET systems which can be inserted into existing MRI systems for small animal imaging [4, 5] and human brain imaging [6, 7].

This work is aimed at developing a high-performance detector module for modern PET systems. In modern PET/ MRI systems, an MR-compatible detector is desired, and the limited space in combined PET/MRI requires a compact detector. However, for a PET detector ring of reduced diameter, the parallax error effect worsens due to the increase in gamma rays in oblique incidence to the PET detectors [8], which introduces serious implications for PET spatial resolution. Therefore, the depth-of-interaction (DOI) encoding capability is critical for the system to achieve a high imaging performance. DOI PET detectors were developed by monolithic technique [9, 10] and pixelated technique [11–16]. The former requires multichannel readout and complex position decoding algorithms [10], hindering its commercial clinical applications, while the pixelated technique, including light sharing [11], dualend readout [12–14], and pulse shape discrimination [15, 16], is more practical.

Another highly desired feature in PET imaging is the high signal-to-noise ratio (SNR), so that patient acquisition time or injected dose can be optimized without compromising image quality. It is also critical for dynamic imaging. Time-of-flight (TOF) is a promising technique that significantly enhances the SNR in reconstructed image [17]. The new generation of SiPM, of low dark noise and high photon detection efficiency (PDE), has demonstrated excellent timing performance [18-22]. One-to-one coupling between the scintillator and SiPM element and individual readout is a straightforward design. However, it requires multi-channel readout, but the intrinsic spatial resolution is limited by the SiPM element size. Thus, light sharing and signal multiplexing with a SiPM array are normally employed, while it reduces the timing performance due to the complex light propagation and noise increase by combining a large number of SiPMs. A tradeoff between performance and the number of readout channels, an array with 4×4 SiPMs is the most common design, which usually has a total size less than 20 mm \times 20 mm [22–25]. Several groups developed large SiPM arrays with good crystal distinction, but the timing performance was not good enough for TOF PET systems [26-28].

In this paper, a dual-layer-offset (DLO) [29, 30] LYSO array coupled to an 8×8 SiPM array is designed. It is read out by an ASIC chip. Excellent crystal identification is achieved, and DOI and TOF information is obtained simultaneously.

2 Materials and methods

2.1 Design of DOI detector

The scintillator block is a home-made DLO LYSO array with a crystal size of $2 \text{ mm} \times 2 \text{ mm} \times 7 \text{ mm}$ (SIPAT, China). All surfaces of the crystals are polished. The top layer is a 15×15 array of $31.6 \text{ mm} \times 31.6 \text{ mm}$, while the bottom layer is a 16×16 array of $33.6 \text{ mm} \times 33.6 \text{ mm}$. The top layer is placed half crystal offset in two dimensions on the bottom layer (Fig. 1a). Due to the staggered design, the detector has better sampling density than common designs and can achieve better image performance for a whole system [31]. Between the crystals are ~ 0.1 -mm-thick enhanced spectral reflector films (3 M).

As shown in Fig. 1b, two 8×8 SiPM arrays were fabricated on the same board based on the MicroFJ-30035-TSV SiPM (SensL Technologies Ltd., Ireland), which is an industry-leading low dark noise, MR-compatible, and high PDE sensor through silicon via package [20]. Each individual SiPM has 5676 microcells ($35 \ \mu m \times 35 \ \mu m$) with a package size of 3.16 mm \times 3.16 mm. The 8×8 array has an overall dimension of 33.7 mm \times 33.7 mm with 1-mm gaps between adjacent SiPM pixels. In this study, only one of the two SiPM arrays was used.

2.2 Data acquisition system

The output channels of the 8×8 SiPM array including 64 standard outputs and 64 fast outputs (the derivative of the internal fast switching of the microcell) are directly fed to a 128-channel ASIC called EXYT [32] on back side of the SiPM board. The standard output channels are multiplexed by on-chip resistor networks to generate 3 analog outputs (*E*, *X*, and *Y*) for energy and positions decoding. The fast output channels are used to generate signal *T* for timing information. Each fast output channel has a 5-bit current DAC for local threshold adjustment to reduce non-uniformity of the threshold of timing discriminators. The threshold for each fast channel is scanned in advance.

The pulses of E, X, and Y are continually sampled by a free-running ADC chip with 80 MHz sample rate (12-bit AD9634, Analog Devices) and processed by programmable logic circuits in FPGA (Spartan-6, Xilinx). Energy is calculated in FPGA by a digital integration mode. At last, the data are transmitted to a PC through Gigabit Ethernet. Figure 2 shows a printed circuit board with two ASIC chips and the readout electronics for the detector module.

2.3 Experiments

2.3.1 Flood source measurement

The detector was covered by a light-tight box and worked at room temperature (25 °C) without cooling. Power supply voltage for SiPMs was 27.5 V (over-voltage 3 V). A ²²Na point source (about 3 μ Ci) was placed on top of the LYSO array. Due to the low activity of the ²²Na source, the influence of intrinsic ¹⁷⁶Lu radiation in LYSO must not be disregarded [33]. Thus, the intrinsic ¹⁷⁶Lu radiation was measured under the same condition as the ²²Na study for energy spectrum calibration.



Fig. 1 Scheme of the DLO LYSO–SiPM detector module (a), and a home-made SiPM board with two 8×8 arrays (b)



Fig. 2 Two ASIC chips for two 8×8 SiPM arrays (a) and readout electronics for the detector module (b)



2.3.2 Timing performance evaluation

detector module

Figure 3 illustrates the experimental setup for timing performance evaluation with a single crystal coincidental to a detector module. The ²²Na source was placed close to the single crystal to irradiate all LORS from each crystal in the detector module. The other experimental conditions were the same as the flood source experiment.

2.4 Data processing

A crystal position map (CPM) was generated from the flood histogram of ²²Na measurement by using an automatic segmentation program [34] we developed based on a stratified peak tracking method. In the program, the peaks of the top layer were identified first with the following procedures:

- 1. Creating a square image of the original flood histogram to increase the distinction between the top and bottom layers;
- 2. Generating a template image of the top layer by using a SVD(singular value decomposition)-based method;
- 3. Determining the initial peaks of the top layer using the *X* and *Y* dimensional projections of the template image; and
- 4. Locating the accurate peaks of the top layer by using a mean-shift algorithm.

Then, responses of the top layer generated using the located top peaks and a Gaussian kernel were subtracted from the original flood histogram. The peaks of the bottom layer were tracked using the similar procedures. At last, a CPM was created by classifying each pixel to its nearest peak and assigning it with the crystal index of that peak.

Energy values (*E*) were histogramed using their positions and the CPM to generate the energy spectra for both the ²²Na and ¹⁷⁶Lu intrinsic radiation experiments. The ¹⁷⁶Lu spectra were subtracted from the ²²Na spectra to generate calibrated the energy spectra. The 511 keV peaks in the calibrated energy spectra were identified and fitted by Gaussian functions $N(\mu, \sigma)$. The energy resolution (η) of each crystal was obtained from the fitted result of $\eta = 2.355 \sigma/\mu$.

For the timing performance study, a 450–650 keV energy window was applied to the single crystal and each crystal of the detector module. The coincidence times were histogramed for each crystal and then the coincidence resolving times (CRTs), i.e., full width half maximum (FWHM), were derived by fitting Gaussian functions to the histograms.

3 Results

3.1 Crystal identification and segmentation

Figure 4a shows the flood histogram of the ²²Na source. Almost all crystal responses are clearly distinguished except some at the corner. Figure 4b shows the summed profile of the red box area in Fig. 4a. The average peak-tovalley ratio is 8, indicating that very small decoding error is expected for this detector. The valley counts in the summed profile are mostly contributed by the events with multiple Compton scatters in different crystals. These events cannot be distinguished in a flood histogram of a traditional PMTbased detector [35, 36]; however, due to the excellent decoding ability, the events with multiple Compton scatters in different crystals are separated from the events deposited full energy in a same crystal including photoelectric interactions and multiple Compton scatters. The reconstruction algorithm will be improved in the future by applying different response models for different types of events.

As shown in Fig. 4c, d, peaks of the top and bottom layers were exactly located. As shown in Fig. 4e, the flood histogram was segmented to 481 ($16 \times 16 + 15 \times 15$) zones. Each zone has indices of layer, row, and column. These indices were automatically generated in the procedures. Based on the segmentation results, a crystal position map was generated.

3.2 Energy resolution

Figure 5a, b shows energy spectra of the ²²Na source and the intrinsic¹⁷⁶Lu in a typical crystal with, respectively. The calibrated energy spectrum is shown in Fig. 5c. The energy resolutions and energy channel numbers of 511 keV peaks of the bottom layer are shown in Fig. 5d, e. The average energy resolution (mean value of all crystals) and the average channel number of the 16×16 crystals are $(14.0 \pm 1.6)\%$ and 419 ± 34 , respectively. The energy resolutions and energy channel numbers of the 511 keV peaks of the top layer are shown in Fig. 5f, g. The average energy resolution and the average channel number of the 15×15 crystals are $(12.7 \pm 0.6)\%$ and 417 ± 26 , respectively. The slightly better energy resolution of the top crystals is counterintuitive, as they are expected to have a higher light loss due to longer travel path to the SiPM. One possible reason is the depth variation of the scintillation emission sites in the top crystals is smaller.

3.3 Coincidence resolving time

Figure 6 shows the coincidence resolving times for the single crystal with the top/bottom layer crystals of the detector module. The average CRTs at the top and bottom layer are 565 ± 66 and 635 ± 114 ps, respectively. Similar to the energy resolution performance, the top layer has better timing performance than the bottom layer. Meanwhile, the crystals at the edge of both layers have better timing performance, though their light collection is lower. It may be because that the optical photon collections in various depths of a crystal are different due to the divergent light propagation, which introduces jitter to time determination, and the crystals at the edge.

4 Discussion and conclusion

The results above show that the detector module has excellent crystal identification capability, good energy resolution, and available CRT in both 7-mm-thick layers at



Fig. 4 CPM with the 22 Na source placed on top of the detector module (**a**), the summed profile (**b**) of the *red box area* in the flood histogram, *peaks of the top* (**c**) and *bottom* (**d**) layers, and the segmentation result (**e**) of the crystal position map



Fig. 5 Energy spectra of the 22 Na source (a) and intrinsic 176 Lu (b) the calibrated energy spectra (c), energy resolutions (d) and channel numbers (e) of the *bottom layer*, and energy resolutions (f) and channel numbers (g) of the *top layer*





27.5 V bias and room temperature. The timing performance has the potential of improvement by increasing the bias voltage of SiPMs [21]. However, the standard output signals of the ASIC chip would become saturated. It is because that this version was optimized for MicroFB-30035 SiPM (sensitive to magnetic field), while the B-series have lower signal output than the J-series. A new ASIC chip optimal for MicroFJ-30035-TSV SiPM is under development, and better timing performance shall be expected.

The crystals at the bottom layer with a little poor energy resolution while they have higher light collection than the crystal at the top layer. The crystals at the edge of the array have better coincidence timing resolution while they have lower light collections. These phenomena may be caused by the light collection inconsistent at different crystals and different depths.

In conclusion, the compact DOI–TOF PET detector module we developed is of excellent crystal identification capability, good energy resolution, and reasonable time resolution and has promising application prospective in clinical TOF PET, PET/MRI, and brain PET systems.

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