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Measurements of entrance-surface vs. conventional single-ended readout of a monolithic scintillator

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Abstract

Availability of compact high-gain, low-noise Silicon Photomultipliers (SiPM) prompts us to examine readout sensors on the entrance surface (SES) as compared to the conventional single-ended readout with sensors on the opposing surface. We measured detector response statistics versus 3D position for these configurations using an 8×8 SiPM array on a 15-mm-thick by 32-mm-wide LYSO block. We calibrate an independently distributed multivariate-normal likelihood model and use it to generate maximum-likelihood estimates of 3D interaction position. Spatial resolution improved 14% and timing resolution improved 10% for the SES device. Bias was unaffected. Photodetection efficiency of our prototype SiPM may have limited further improvement in positioning and timing performance. In future work, we will look to utilize SiPM arrays with enhanced photodetection efficiency.

I. Introduction

Arrays of small crystals have often been used in preclinical positron emission tomographs (PET) to achieve fine-spatial-resolution images [1]–[5]. However, as crystal cross-section decreases, cost quickly increases while photodetection efficiency (fill factor) and signal-to-noise ratio (light collection) decrease. We have previously investigated use of statistical-

based 3D-positioning methods [6–8] in continuous crystal detectors [9] as an alternative for high-resolution PET.

More recently, the availability of compact high-gain, low-noise Silicon Photomultipliers (SiPM) prompted us to conduct simulation studies of readout sensors on the entrance surface (SES) as compared to the conventional single-ended readout with sensors on the opposing surface [10,11]. These simulation studies lead us to believe that there is improved intrinsic 3D-spatial resolution. Presently, we report experimental results comparing SES versus conventional single-ended readout of a monolithic detector.

II. Design and Methods

A. Detector, readout, and acquisition hardware

We measured the responses for conventional and SES readouts of a 32-mm-wide by 15-mm-thick square block of LYSO scintillator (Fig. 1). Light was collected on one end with an 8×8 SiPM array provided by Philips Healthcare (Fig. 2 and Table 1). In either case, the opposing scintillator surface was covered with 6 layers of Teflon and the sides were painted with flat black enamel paint.

We used modified commercial time-of-flight electronics from Philips Healthcare (Gemini TF-scanner hardware) to acquire 60-channels per detector; corner channels of the 8×8 array are omitted due to this channel limit (Fig. 2b). We connected SES to preamplifiers with custom flexible printed cables (FPC) from MicroConnex®, WA (Fig. 2a). FPC mounting failed for 2 channels of the SES readout (Fig. 2b).

B. Detector-response calibration setup

The detector assembly and preamps were kept at 10 °C in a climate-controlled dark box. We established a nominal operating bias at this temperature of −30.25V (−2.85V average excess bias across 60 channels), which optimized energy resolution at 511-keV for the SiPM array used for conventional readout. Likewise, we chose −30.15V bias (−2.85V average excess bias) for the SES device.

We begin with measuring the SiPM relative gains, which we use for event filtration [7,8] of the detector-response calibration data discussed below. We positioned a pulsed (6-kHz, 50-nsec-width) blue (500-nm) LED (Nichia NSPB500AS) over the center of each bare channel and acquired baseline-corrected spectra for 2.5×10^5 pulses.

For each detector configuration, we then use a normal-incidence 511-keV photon beam to calibrate the 3D-function of detector response statistics (mean and variance) and to generate test data to measure the resolution of 3D interaction position (see Figs. 3–4). Beam width in this setup diverges from 0.3-mm full width at half maximum (FWHM) at the entrance surface to 0.5-mm FWHM at the exit surface. We move the calibration beam in 1-mm steps to an array of positions spanning the active detector area. At each position, fifty thousand events were recorded. For the sake of measuring the detector response as a function of known interaction position, we wish to keep only photoelectric events.

C. Compton filtering

We aim to calibrate the detector response as a function of interaction location. To avoid increased energy-deposition variance caused by Compton scatter within and out of the detector, we attempt to filter such events. To do so, we apply an energy window (−25%, +40%) and a spatial filter (20% contour mask on position estimates). In this manner, we remove about 65% of the recorded events from the final calibration data. A detailed discussion of our event filtering method is given in [7]. With this filtered calibration data,

we can compute the mean, $\bar{g}_i(\mathbf{R})$, and standard deviation, $\sigma_i(\mathbf{R})$, of the detector response for each channel, i , versus 3D interaction position, $\mathbf{R}$.

D. Depth separation of calibration events

To decode interaction depth of calibration events, we initially partition the data using signal quartiles of the peak channel (usually the channel closest to the interaction), which tends to be monotonically increasing towards the photo detector; for an exponential interaction distribution, these quartiles correspond to depth bins separated at 3.6 mm, 7.3 mm, and 11.1 mm for 15-mm-thick LYSO. We then compute an initial estimate of detector response statistics $\bar{g}_i(\mathbf{R})$ and $\sigma_i(\mathbf{R})$ and use these to refine depth separation by likelihood sorting. After resorting, we recompute response statistics and iterate the depth-estimation and sorting process. This process is repeated until stable statistics are obtained.

E. Interaction-position estimation

We determine 3D position by maximum likelihood estimation using an independently distributed multivariate-normal likelihood model (Eq. 1). We then compute the FWHM and bias for an ensemble of 20,000 test events gathered in a similar manner as the calibration data, using a normal 511-keV gamma-ray beam.

$$\hat{\mathbf{R}} = \underset{\mathbf{R}}{\operatorname{argmax}} \prod_{i=1}^N \operatorname{Norm}\left(g_i | \bar{g}_i(\mathbf{R}), \sigma_i(\mathbf{R})\right). \quad (1)$$

III. Results

We first look at the pulse shapes for conventional and SES readout configurations to determine if the added FPC cable affects the signal transfer, which it does not (Fig. 5). Next, a map of calibrated gain values for two SiPM arrays is given in Fig. 6. Peak-channel signal distributions for depth-separated data are shown at one calibration-beam position for the conventional and SES readout configurations in Fig. 7. An example of the detector response statistics as a function of beam position and depth is shown for one SiPM channel in Fig. 8. Estimate resolution (FWHM) and bias for the conventional and SES readout configurations are reported in Fig. 9 and Table 2.

IV. Discussion

We observe improved depth sensitivity in the SES readout configuration over the conventional single-ended readout; this result is because more events occur closer to the SES photodetectors due to exponential attenuation. With better depth positioning, we can reduce signal variance, which then impacts lateral positioning performance. Lateral detector spatial resolution improved by 14% using SES rather than conventional readout; this improvement is in spite of two problematic channels in the SES device due to FPC failures. Resolution in both devices was limited by an effective photon detection efficiency of ~10% in the prototype SiPM devices that we used. The average estimate bias is comparable for the two devices. Coincidence timing resolution measured to be 10% better for SES versus conventional readout.

The addition of FPC cables to reroute photodetector signals from the entrance surface to preamps on the back surface do not appear to affect pulse shape. Furthermore, although we expect 2–5% attenuation of 511 keV photons given the material profile of SiPM tile, no significant scatter was observed experimentally. However, we are considering dislocating or

decreasing the profile of passive components on the SiPM tile (i.e. capacitors). Finally, the FPC connection failures on the SES device are a result of having to remove the existing connectors to attach the FPC. Future tiles will not have this problem since we will start with a bare tile (no existing connectors).

Acknowledgments

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Fig. 1. Conventional vs. SES readout of a monolithic scintillator. Singles for SES are routed with flexible printed cables (FPC) to readout electronics on the opposite side.

Fig. 2.

(a) Front & back views of 8x8 SiPM arrays provided by Philips Healthcare. Flexible printed cables (FPC) have been added to one of the arrays for the SES readout configuration. (b) Corner channels were not used due to 60-channel acquisition setup and FPC connections to channels 43 and 56 failed for the SES readout array.

Fig. 3. SES-readout setup. Oversized FPC cables are being used for this initial characterization to allow flexibility of detector size and placement.

Fig. 4.

Coincidence-beam setup for detector response calibration. In addition to coincidence collimation, we also used geometric collimation (5-mm-thick tungsten with 2-mm bore) to allow for a high photon flux within the calibration beam while keeping the overall count rate low to minimize random coincidences. Furthermore, we masked the coincidence detector (25-mm-thick lead with 2-mm bore) to disallow small-angle scatter that would otherwise blur the coincidence beam.

Fig. 5. Pulse shape of preamplifier output for 511-keV-windowed events. Conventional and SES pulse peaks were scaled to match. Repeated measurements of these pulse shapes indicate very little difference.

Fig. 6. Relative-gain maps of SiPM arrays used for (left) the conventional setup, and (right) for the SES setup. Corner channels of these 8×8 array are missing due to 60-channel limit of acquisition hardware.

Fig. 7.

Spectral distribution of depth-separated event data for photodetector channel (channel 4 in each case above) nearest the calibration beam position. The top two plots are of the Conventional detector with beam centered on the channel (left) and near its edge (right). The bottom plots are at similar positions for the SES detector.

Fig. 8. Calibrated detector response statistics versus 3D position shown for one of 60 channels. Columns (A): SES mean, (B) SES standard deviation, (C) Conventional mean, and (D) Conventional standard deviation. Each row corresponds to one of four depth bins. A/D units are used throughout.

Fig. 9. SBP-positioning results for Conventional readout (right) and SES readout (left). Shown are: (top) half-max contours of the SBP-estimate distributions of events for an array of beam positions (2-mm spacing); (middle) geometric average half-max contour widths at each test position on a 1-mm grid (FWHM) over array; (bottom) ensemble bias of SBP positioning.

Table 1

Specifications of prototype SiPM array provided by Philips Healthcare for the conventional-SES comparison. Sizes below are square.

Property:	Value	Property:	Value
8×8 package size:	32.55 mm	Dark count rate	5 Mcps
SiPM size (pitch):	3.950 mm	Diode capacitance	170 pF
Active size:	3.738 mm	Quench capacitance	92 pF
Cell size (pitch):	58.4 μ m	Quench resistor	80 Ω /cell
Cell fill factor:	45%	Breakdown bias	27–31 V
Cells/SiPM:	4096	Typical excess bias	3–5 V
PD Efficiency:	10–15%		

Table 2

Performance summary of Conventional vs. SES readout computed (average) from results in Figure 8.

	Avg. Position FWHM [mm]	Avg. Position Bias [mm]	Avg. Coinc. Timing FWHM (ns)
Conventional	1.54	0.251	1.13
SES	1.33	0.248	1.02