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A sub-millimeter resolution PET detector module using a multi-pixel photon counter array

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Abstract

A PET block detector module using an array of sub-millimeter lutetium oxyorthosilicate (LSO) crystals read out by an array of surface-mount, semiconductor photosensors has been developed. The detector consists of a LSO array, a custom acrylic light guide, a 3×3 multi-pixel photon counter (MPPC) array (S10362-11-050P, Hamamatsu Photonics, Japan) and a readout board with a charge division resistor network. The LSO array consists of 100 crystals, each measuring $0.8 \times 0.8 \times 3$ mm³ and arranged in 0.86 mm pitches. A Monte Carlo simulation was used to aid the design and fabrication of a custom light guide to control distribution of scintillation light over the surface of the MPPC array. The output signals of the nine MPPC are multiplexed by a charge division resistor network to generate four position-encoded analog outputs. Flood image, energy resolution and timing resolution measurements were performed using standard NIM electronics. The linearity of the detector response was investigated using gamma-ray sources of different energies. The 10×10 array of 0.8 mm LSO crystals was clearly resolved in the flood image. The average energy resolution and standard deviation were 20.0% full-width at half-maximum (FWHM) and $\pm 5.0\%$, respectively, at 511 keV. The timing resolution of a single MPPC coupled to a LSO crystal was found to be 857 ps FWHM, and the value for the central region of detector module was 1182 ps FWHM when $\pm 10\%$ energy window was applied. The nonlinear response of a single MPPC when used to read out a single LSO was observed among the corner crystals of the proposed detector module. However, the central region of the detector module exhibits significantly less nonlinearity (6.5% for 511 keV). These results demonstrate that (1) a charge-sharing resistor network can effectively multiplex MPPC signals and reduce the number of output signals without significantly degrading the performance of a PET detector and (2) a custom light guide to permit light sharing among multiple MPPC and to diffuse and direct scintillation light can reduce the nonlinearity of the detector response within the limited dynamic range of a typical MPPC. As a result, the proposed PET detector module has the potential to be refined for use in high-resolution PET insert applications.

1. Introduction

A previously reported positron emission tomography (PET) insert system has successfully achieved high spatial resolution with a novel, virtual-pinhole PET geometry (Tai *et al* 2008, Wu *et al* 2008a). The PET insert was developed to improve the image resolution of an existing, general-purpose animal PET system. Because the previous design was based on photomultiplier tubes (PMT), a significant number of gamma rays were attenuated and scattered by the photosensor and light guides, reducing the overall system sensitivity (Wu *et al* 2008b). By virtue of their physical makeup, a semiconductor photosensor is significantly

less likely to interact with 511 keV gamma rays, and hence may potentially be more suitable for the development of high-resolution PET insert systems.

Recently, there has been much interest in the use of Geiger mode, avalanche photosensors, often referred to as a silicon photomultiplier (SiPM), for gamma ray detector applications. A SiPM consists of multiple small avalanche photo-diodes (so-called microcells), and each microcell is connected to a common electrode structure. When a reverse bias is applied to the microcells at a voltage higher than the breakdown voltage, each microcell operates in Geiger mode and provides a single photon counting capability. That is, if an electron-hole pair is created by the absorption of a light photon, the electron will be rapidly accelerated to ionize more electrons, inducing a cascade of charge (signal) amplification. Since multiple microcells are connected in parallel, the output signal is the sum of Geiger mode signals (discharge) of microcells triggered by incident light photons. After the Geiger discharge, the single pixel recovery time is needed for suppression of the afterpulse effect (Renker 2006) until the microcell is fully recharged and ready to accept the next event.

These devices provide several attractive features (Kovaltchouk *et al* 2005, Moehrs *et al* 2006, Otte *et al* 2005, Spanoudaki *et al* 2007) as follows.

- High internal gain and low noise
- Fast response time
- Low power consumption
- Discrete gain response
- Insensitivity to magnetic fields
- Large range of physical dimension and cellular density.

The small dimensions and ease of configuration into arrays of the SiPM compared to PMT provide a distinct advantage when building a compact, scintillation-block detector. Therefore, SiPM is a promising photosensor that could replace PMT in medical imaging and is particularly suitable for a magnetic resonance (MR)-compatible PET scanner because it is relatively insensitive to magnetic fields (Delso and Ziegler 2009, Moehrs *et al* 2006, Otte *et al* 2005, Yamamoto *et al* 2010).

A block detector module composed of an array of SiPM for imaging will have a dead space between neighboring sensors, which inevitably reduces the light collection efficiency and degrades positioning accuracy. As a result, control of the scintillation light collection efficiency is critical in order to obtain a good flood image. Another factor affecting block detector performance is the number of microcells. Generally, a larger number of microcells per photosensor yields a wider dynamic range and increases the linearity of the detector response. However, due to the fine gaps between microcells, increasing the number of microcells decreases the fill factor, or the fraction of the device's surface which is photosensitive, thereby reducing the sensor's detection efficiency. The trade-off between dynamic range and photon detection efficiency (PDE) is an important factor in block detector design (Renker 2007). The PDE is given by the following equation:

$$\text{PDE} = \varepsilon_{\text{geometric}} \times \varepsilon_{\text{avalanche}} \times \text{QE}, \quad (1)$$

where $\varepsilon_{\text{geometric}}$ is the geometric efficiency of the photosensor, $\varepsilon_{\text{avalanche}}$ is the probability of an avalanche occurring when a photon interacts with the microcell and QE is the quantum efficiency of the microcells, i.e. the likelihood of a photon being absorbed in the microcell. The geometrical efficiency is directly proportional to the fill factor. Again, a large number of

microcells in a sensor provide a wide dynamic range and linear response to light intensity, but at the expense of the PDE.

This study introduces a block detector design using high PDE photosensors while not seriously sacrificing the dynamic range. One approach to this goal is to share the scintillation light across a multiplicity of SiPM photosensors. In such a scenario, the light guide plays a key role in the detector module's performance, as it must efficiently transfer the scintillation light to the photosensors, with appropriate weighting for accurate positioning while avoiding photosensor saturation.

The aim of this study is to construct a novel PET detector module using an array of sub-millimeter LSO crystals coupled via a light guide to an array of Hamamatsu multi-pixel photon counters (MPPC, a family of SiPM) (Gomi *et al* 2007). The basic suitability of these components for high-resolution PET detector modules was evaluated through measurement of the module's flood image, energy resolution, timing resolution and the linearity of the output signals.

2. Materials and methods

2.1. Block detector module

Nine MPPC (model# S10362-11-050P, Hamamatsu Photonics, Japan) were employed to constitute a block detector module. The dimensions of each MPPC sensor are $2.4 \times 1.9 \times 0.8$ mm³ and the sensitive area is 1×1 mm² as illustrated in figure 1. The sensitive area is a set of 400 microcells. The entire MPPC is covered with a transparent resin film of 0.3 mm thickness. It has two long metal strips on the bottom surface to provide the electrical contacts.

The nine MPPC were arranged in a 3×3 array held by a Teflon base with nine pockets as shown in figure 2(a). The total area occupied by the nine units of MPPC was 8.6×8.6 mm², with a sensing area of 7.8×7.8 mm² (figure 2(b)) for decoding a 10×10 LSO array. The dimensions of each LSO pixel are $0.8 \times 0.8 \times 3$ mm³ and the pitch is 0.86 mm due to the reflector between pixels. Therefore, the cross-sectional dimension of the LSO array is the same as that of the MPPC array (including the packaging materials around the MPPC elements). The LSO crystal used in this study was actually a 24×12 array that was significantly larger than the sensing area. Only the central 10×10 subarray was coupled to the custom light guide. The extra LSO pixels were masked with black tape. To equally space the off-centered sensitive area of the MPPC sensors, the left three sensors were rotated 180° (figure 2). Optical grease was applied to the MPPC entry windows. The white Teflon base consists of two pieces that hold (from top down) an LSO array, a light guide, the nine MPPC and 18 pogo-pins (Interconnect Devices Inc.). The gold-plated pogo-pin, which has 4.57 mm length and 0.69 mm diameter, conducted the MPPC output signal with 50 Ω impedance.

After all the components were assembled together, the base was mounted to a custom-made readout board (figure 3(a)) that consists of current feedback charge-sensitive pre-amplifiers and a resistive charge divider (Siegel *et al* 1996). The output signals of the nine MPPC are multiplexed by a charge division resistor network to generate four position-encoded analog outputs. The signals were fed into nuclear instrumentation module (NIM) spectroscopy amplifiers (N568B, CAEN) with a shaping time of 0.2 μs. The peaks of the shaped pulses were digitized by a data acquisition board (PCI-416, Dattel Inc.) to create a list-mode data file which contains the four digitized output signal. The sum of four output signals corresponded to a total energy deposition of each event and the centroid of deposited energy was determined by Anger logic for event localization (Siegel *et al* 1996). A complete diagram of the detector module designed in this study is shown in figure 3. A common high voltage was supplied to the nine

MPPC without individual regulation since the sensors were pre-sorted devices in consideration of gain and dark currents (table 1).

2.2. Light guide design

A light guide was designed to control the distribution of scintillation light generated by the LSO array on the MPPC sensitive area to improve localization of each gamma-ray event. A Monte Carlo simulation program, DETECT2000 (Cayouette *et al* 2003), was utilized to assist the light guide and detector module design (Dhanasopon *et al* 2005). To reduce the simulation time, light-emitting sources were only generated for five LSO crystals on the left center section array and were located at the center of each crystal (figure 4). A number of parameters, such as light guide thickness, saw-cut depth, the number of cuts and surface treatment, were examined to evaluate their effect on the 2D flood image and improve design. The light guide design was similar to the approach described by Pichler *et al* (2004), providing a few cuts on each side of the light guide to separate the outer crystal rows in the position profile.

The MPPC sensor package has a large dead space on its face because the sensitive area is only roughly 21.9% of the sensor face and there are relatively large gaps between adjacent sensors in the MPPC detector array. The total coverage of the sensitive areas of nine MPPC is 12.2% of the 10×10 LSO array. The dead space described above lowers the light collection efficiency since the bulk of the scintillation light never reaches a photosensor. This negative effect is also a source of energy resolution degradation because some events might register in the Compton scattering region due to light loss, rather than registering in the photopeak region. To prevent the light photon loss from the dead space and to increase light collection efficiency, a VM-2000 (3M, St. Paul, MN) mask which has nine openings for MPPC-sensitive areas was employed and attached on the bottom surface. Thus, the light photon which meets the VM foil would be reflected back to the other reflector surface, giving it a second chance to be detected by the photosensors. This idea is related to a method described by Hamamoto *et al* (2005). The effect of the reflector on flood image was also estimated by using the Monte Carlo simulation.

After evaluating the above-described parameters, a light guide design that demonstrated the best crystal separation in the 2D position histogram was selected for prototyping. A sheet of clear polycarbonate without tint was utilized to fabricate the light guide in recognition of its mechanical strength and adequate optical properties, including greater than 82% transmission of 420 nm scintillation light through 5.08 mm thickness (Lytle *et al* 1979). The VM-2000 reflector was used to enclose the side surfaces and bottom of the light guide. The saw-cut width of the light guide was slightly wider than 0.15 mm, the thinnest width of jewel-saw blade available for this study. When the saw blade was rotating and cutting the light guide body to make a reflective slot, the actual saw-cut width was enlarged by the sidewall grinding. The saw-cut walls (slots) were filled with magnesium oxide (MgO) powder so they would function as reflective walls.

2.3. Basic performance test

2.3.1. Flood image—For the flood image evaluation, a Na-22 (511 keV) point source was used to acquire a 2D position histogram of the 10×10 LSO array to verify the pixel separation. A Na-22 source of 1.15 MBq was located 75 mm away from the detector surface to provide uniform irradiation. Acquisition was continued until about 5 million events were accumulated. The common high voltage was set to 68.9 V, and the energy threshold for the sum of nine MPPC signals was set to ~200 keV. Each gamma-ray event detected on the flood image was binned to a corresponding region on a crystal lookup table.

2.3.2. Energy resolution—The energy resolution was measured with a 1.70 MBq Ge-68 point source. The four output signals stored in the listmode file were summed to create the total

energy estimation, and the Anger logic determined the location on the flood image. The location was used to determine the crystal of interaction (via the lookup table), and the total energy per crystal was histogrammed into energy spectra. For LSO background correction, a simple subtraction method described by Yao *et al* (2008) was used. The LSO contribution to the energy spectrum was estimated based on a pre-measured long LSO background scan and subtracted prior to analyzing the energy resolution. The energy spectrum of individual LSO might be compressed due to the nonlinear response of the MPPC photosensor. The level of compression varies among crystals due to differences in light collection efficiency, and so the compressed energy spectrum must be corrected before the energy resolution of individual crystals can be estimated. Linearity of each individual crystal was measured (see section 2.3.4) and used to correct for distortion of the energy spectra. The correction procedure is as follows.

- A. Energy spectra of individual crystals were measured using five different gamma-ray sources (Co-57, Tc-99m, Ga-67, I-131 and Na-22 sources).
- B. Find photopeak channels that correspond to the seven gamma-ray energies for each crystal.
- C. Assuming that the detector response is linear at low energy level, apply a linear fit to the first three data points (the three lowest energies) and refer the fitted function as a linear line (LL). Subsequently, apply a second-order polynomial fit to all seven data points. This polynomial fit reveals a saturation curve of the MPPC output response at each crystal (SCV). The constant of SCV represents a bias due to the dark noise of the MPPC, the first-order x -term represents linear response from low gamma-ray energy, and the second-order x -term was used to model curvature in the nonlinear response from higher gamma-ray energy. Thus, the correction factor at a given gamma-ray energy e is derived by

$$\text{Correction factor}(e) = \frac{LL(x)}{SCV(x)} \Big|_{x=e} \quad (2)$$

where $LL(x) = a + bx$ and $SCV(x) = i + jx + kx^2$, and the 100 sets of coefficients a , b , i , j and k were determined by linear and polynomial fit.

- D. For each crystal's energy spectra measured using Ge-68, apply the correction factor to map the original x -axis (energy) to a new variable x' . Then plot the original y -values (counts at a specific energy) against x' to get the corrected energy spectrum.

After the correction, the photopeak of each crystal's energy spectrum was fit to a Gaussian function to measure the energy resolution at 511 keV.

2.3.3. Timing resolution—Timing resolutions of a single MPPC and the MPPC array detector module were measured against a Hamamatsu H5783 PMT coupled to a plastic scintillator (Saint Gobain, Newbury, OH) of dimensions $2 \times 2 \times 8 \text{ mm}^3$. The fast PMT output signal was fed into a constant fraction discriminator (CFD, Ortec 935) and its output was used to generate start input on the time-to-amplitude converter (TAC, Ortec 567) NIM module. The four output signals from the MPPC array detector module were fed into a summing circuit (Fan-In/Out, Phillips Scientific 740), and the output was followed by a fast-filter amplifier (Ortec 579) and a CFD to generate stop input of the TAC. The valid conversion output of TAC was fed into a gate-and-delay generator (Phillips Scientific 794) module to generate a trigger signal. The MPPC array, PMT and TAC outputs were digitized by the PCI-416 data acquisition board, and the list-mode data were post-processed to allow the option of applying energy discrimination to the timing measurement data. The experimental setup is shown in figure 5. Since a timing variation across the crystal array was anticipated, timing resolutions were

obtained at three different points (center, edge and corner of the LSO array) with electronically collimated gamma rays. The distance between the MPPC detector and PMT was 100 mm, and the Na-22 source was located in the middle of the two detectors. For the timing measurement of a single MPPC coupled to a single LSO crystal ($0.9 \times 0.9 \times 4 \text{ mm}^3$) against the PMT detector, a simple T-connector was used instead of the fan-in/out. The distance between the LSO and the plastic scintillator was adjusted to 60 mm.

2.3.4. Linearity test—The MPPC sensors used in this study have 400 microcells each, so there is a risk of the device being optically saturated beyond a certain flux of light photons. A variety of sources (Co-57, Tc-99m, Ga-67, I-131, Ge-68 and Na-22) were placed 75 mm away from the LSO surface ($>5 \times$ detector module width) to investigate the linearity of the detector output at different gamma-ray energies (93, 122, 140, 185, 300, 364, 511 and 1275 keV). The source activities were 50, 18.5, 7, 9.6, 1.7 and 1.15 MBq, respectively. The electronics setup was kept the same, but the energy threshold was decreased to 80 keV for the low energy emitter. Energy spectra of 10×10 LSO pixels were generated from the singles event of each radiation source, and once an LSO background was subtracted from the energy spectra as mentioned above, linearity profiles were created. At least 10 000 counts were stored in the peak channel of the total energy spectrum. A hypothesis of this linearity test was that the output of the MPPC array might be linear compared to the one single MPPC sensor because the scintillation light from a gamma-ray energy would be shared among them and collected in concert with several MPPC. The photopeak channel of a spectrum directly indicates the output amplitude at a particular energy. If the response becomes nonlinear, it implies that the MPPC output signal starts to optically saturate. The same linearity tests with the six different sources were repeated with a single channel MPPC and an H5783 PMT as benchmarks for the performance of the proposed detector. A single LSO crystal measuring $0.9 \times 0.9 \times 4 \text{ mm}^3$ was coupled to a single channel MPPC and PMT to generate the linearity profiles as a function of the gamma-ray energy. For the single MPPC, the same 68.9 V high voltage was applied. The PMT data were used as a reference in comparing the linearity between the single channel MPPC and the MPPC array detector. The single channel MPPC data were used as the baseline to evaluate the improvement in linearity of the detector module at its center and at its corners.

3. Results

3.1. Flood image

The thickness of the fabricated light guide was 1.0 mm. One cut of 0.7 mm depth, which was positioned on the middle of the second LSO pixels, appears to be a possible candidate for edge crystal separation. The Monte Carlo simulation results of five light guides among the various tested geometries are shown in figure 6. The thicker (1.7 mm) light guide allowed excessive light-sharing and resulted in reduced pixel separation in the flood image (figure 6(b)) which made the pixel identification more difficult. The effect of bottom reflective mask on flood image was explicitly disclosed on the simulation result. The 10×10 array of 0.8 mm LSO crystals were successfully resolved in the flood image while using the selected light guide as already expected from the simulation. The finalized 2D position histogram is shown in figure 7(b).

3.2. Energy resolution

Mean, standard deviation, minimum (best) and maximum (worst) energy resolution values after the correction were 20.0% full-width at half-maximum (FWHM), $\pm 5.0\%$, 12.0% and 36.5%, respectively. Figure 8 shows the crystal lookup table and the 10×10 energy spectra at each crystal position. The energy spectrum of the corner crystals was observed to be more compressed than that of the center crystal due to the optical saturation described in section 3.4.

The compression effect created by nonlinear response narrows the FWHM of the photopeak even, as it lowers the photopeak channel position (figure 8(b)).

3.3. Timing resolution

Figure 9 shows the timing spectra of the MPPC detectors measured with different energy windows. Timing resolution of the single channel MPPC was 904 ps FWHM when no energy window was applied, and the value was improved to 857 ps and 856 ps, respectively, when $\pm 10\%$ and $\pm 5\%$ energy windows were applied. The 511 keV photopeak of the plastic-PMT detector was unclear on the energy spectrum. However, if the low energy events from the plastic-PMT detector were rejected and a $\pm 5\%$ energy window around 511 keV is applied for the MPPC detector, the timing resolution was improved to 690 ps (figure 9(a)). When the center of the LSO-MPPC array detector module was irradiated with electronically collimated annihilation photons, the timing resolution was 1252 ps, 1182 ps and 1152 ps with a wide open $\pm 10\%$ and $\pm 5\%$ energy window around the 511 keV, respectively (figure 9(b)). When the edge or the corner of the detector module was irradiated with electronically collimated annihilation photons, the timing resolution was slightly worse (1296 ps or 1312 ps with $\pm 10\%$ energy window, respectively).

3.4. Linearity

The photopeak channels of various gamma-ray energies are summarized in table 2. The 1275 keV gamma ray of Na-22 would saturate the MPPC detector. As a result, no photopeak can be seen on either a single channel LSO-MPPC detector or the corner region of the detector module using a MPPC array. For the single MPPC sensor, the saturation effect appears at very low gamma-ray energy (worse than for the corner crystals of the 10×10 LSO array). Figure 10 shows surface plots of the photopeak channels of 10×10 LSO pixels according to four different gamma-ray energies. Due to saturation, the photopeak location at the corner of the LSO array was lower than at the center. Figure 11 shows the linearity curves of different detectors as a function of gamma-ray energy. These curves were then used to predict the peak channel at which the detector response should be linear for a given gamma-ray energy. Nonlinear response of the 400 microcell MPPC is clearly seen when directly coupled to an LSO crystal (figure 11 (a)). Nonlinearity was also observed for the corner crystals in the MPPC array-based detector module, but significantly less for central crystals (figure 11(b)). The saturation was not observed in the center of the crystal in the range of 93 keV to 300 keV. The saturation of the central crystals was about 6.5% for 511 keV and 28.9% for 1275 keV. The corner crystals showed the saturation effect starting from 200 keV, and the photopeak channel was decreased by 29.4% for 511 keV gamma rays.

4. Discussion and conclusion

This block detector using nine MPPC arranged in a 3×3 array could successfully resolve the 10×10 array of 0.8×0.8 mm² LSO crystals. The flood image will likely improve when the light guide is further optimized and precisely fabricated. The non-uniformly distributed MgO powder used to fill in the saw-cut slot was examined under a microscope; it might cause uneven reflectivity that could affect the flood image. The area of the LSO array was about 74 mm² and the total coverage of sensitive area provided by the nine MPPC was 9 mm², so the optical coupling ratio was only about 12% (section 2.2). However, the LSO crystals were clearly separated on the 2D position histogram using the custom-made light guide with a bottom reflection mask. As shown in the DETECT2000 Monte Carlo simulation results, without the mask, the events overlapped and some pixels were indistinguishable. The simulation results also demonstrated that one saw-cut would be enough to separate the rows and columns of the 10×10 crystals and that was confirmed by the real measurement (flood image). Since the saw-cuts were another source of light loss, fewer saw-cuts are preferred. It became clear during the

assembly of the many small components that alignment was critical for good detector module performance. Misalignment of the photosensors with respect to the light guide and/or the LSO array created poor flood images with unresolvable rows and columns.

The resistive charge divider circuit reduced nine MPPC signals to four, and simplified the readout system. However, the impedance matching must be carefully considered between the input of the resistive network and the output of the MPPC signal (Seifert *et al* 2008), so as to minimize the loading effect, especially when the two components are directly coupled. The resistive network optimization is beyond the scope of this work and shall be carefully investigated in a future study.

Recently, Schaart *et al* (2009) obtained ~14% energy resolution at 511 keV with a 13.2 mm × 13.2 mm × 10 mm monolithic LYSO:Ce³⁺ scintillator coupled to a 4 × 4 array of SiPM (SensL, Inc.). Grazioso *et al* (2005) reported ~21% of average energy resolution at 511 keV by using 2 mm × 2 mm × 20 mm LSO crystals arranged in a 9 × 9 array coupled to a Hamamatsu S8664-55 avalanche photo-diode (APD) consisting of 2 × 2 pixels of the 5 mm × 5 mm sensitive area. The energy resolution results presented by these groups are similar to that of the block detector module proposed here (20.0% FWHM after the linearity correction). Even though these detectors are relatively insensitive to magnetic field when compared to PMT-based detectors, the energy resolution is expected to degrade slightly in a high magnetic field; as described by Pichler *et al* (2006) that 14.6% FWHM energy resolution became 18.7% when a spin echo sequence was applied in a 7 T MRI. This implies that only little degradations are expected on MPPC performance but further investigations are necessary.

The timing resolution for the single MPPC was 904 ps without energy windowing, but it was improved to 690 ps when energy discrimination was applied to both MPPC and PMT detectors. The timing resolution result of a 3 mm × 3 mm sensitive area single MPPC (3600 microcells) was reported to be 240 ps when it was coupled to a 3 × 3 × 10 mm³ LYSO crystal and measured against a LaBr₃-PMT detector under an optimal temperature and bias voltage condition (Kim *et al* 2009). This difference in timing performance may be attributed to the differences in operation conditions (bias and temperature) and the fact that we are using a sub-millimeter LSO crystal for high-resolution imaging applications. The timing resolution may be further improved by optimizing the readout and summing circuitry, as well as the bias voltage and temperature. The current timing performance is comparable to the system introduced by Grazioso *et al* (2005) (870 ps with a 4 mm × 4 mm single LSO crystal coupled to an APD). The timing resolution of our array detector module at three different regions (center-edge-corner) was measured to be over 1 ns FWHM even with energy window applied. This timing result of the sub-millimeter LSO array read out by a MPPC array is comparable to the high-resolution PET detector module introduced by Stickel *et al* (2007) (1.42 ns with a 0.5 mm LSO array coupled to a Hamamatsu H7546 PMT).

Rough photopeaks and multiple peaks can be seen in the energy spectra of some edge crystals. This suggests that a fair number of events are misidentified near the edge of an array. The most likely explanation is inadequate surface treatment of the edge of the light guide, which would cause inconsistent collection of the scintillation light emitted by the edge LSO pixels.

Individual voltage regulation and temperature control for the MPPC photosensor are needed for optimal and consistent results. However, the pre-sorted devices employed in this project worked well under a single bias voltage. Temperature control was not implemented in this study, since the focus was to validate the design of a prototype block detector for PET imaging using an array of MPPC devices. In future implementation, an individual high voltage regulation circuit and a temperature control system will be installed on the readout board.

In the PET detector module described here, the requirement of light-sharing for crystal identification reduces the linearity burden on any given photosensor, thereby permitting the use of MPPC sensors with a small number of microcells for good PDE and little risk of saturation. The detector module's Tc-99m energy spectrum showed no sign of saturation, and the signal response at 511 keV showed slight (6.5%) saturation in the central region of the LSO array. The detector response of the edge crystals is less linear (~15% saturation at 511 keV). For corner crystals, the saturation becomes even worse (~30% saturation at 511 keV). It should be noted that there is maximal light sharing for the central crystals in our detector module and minimal light sharing for the corner crystals. Nevertheless, the 400 microcells per photosensor is likely to be insufficient for the light yield from this detector geometry. The basic performance results demonstrate that the proposed PET detector module can offer sub-millimeter spatial resolution and adequate timing and energy resolution, which make it a suitable candidate for PET insert applications. In addition, this detector block could potentially be used for MR compatible PET applications due to its compactness, electrically advantageous properties and insensitivity to magnetic field (Espana *et al* 2010, Judenhofer *et al* 2007, 2008, Spanoudaki *et al* 2007).

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References

- Cayouette F, Laurendeau D, Moisan C. DETECT2000: an improved Monte-Carlo simulator for the computer aided design of photon sensing devices. *Proc SPIE* 2003;4833:69–76.
- Delso G, Ziegler S. PET/MRI system design. *Eur J Nucl Med Mol Imaging* 2009;36:S86–92. [PubMed: 19104809]
- Dhanasopon AP, Levin CS, Foudray AMK, Olcott PD, Habte F. Scintillation crystal design features for a miniature gamma ray camera. *IEEE Trans Nucl Sci* 2005;52:1439–46.
- Espana S, Fraile LM, Herraiz JL, Udias JM, Desco M, Vaquero JJ. Performance evaluation of SiPM photodetectors for PET imaging in the presence of magnetic fields. *Nucl Instrum Methods Phys Res A* 2010;613:308–16.
- Gomi S, et al. Development and study of the multi pixel photon counter. *Nucl Instrum Methods Phys Res A* 2007;581:427–32.
- Grazioso R, Aykac M, Casey ME, Givens G, Schmand M. APD performance in light sharing PET applications. *IEEE Trans Nucl Sci* 2005;52:1413–16.
- Hamamoto M, Inadama N, Murayama H, Yamaya T, Tsuda T, Ono Y. DOI detection capability of 3D crystal array standing over two PMTs. *IEEE Nucl Sci Symp and Med Imaging Conf Rec* 2005:M3–115.
- Judenhofer MS, Catana C, Swann BK, Siegel SB, Jung W-I, Nutt RE, Cherry SR, Claussen CD, Pichler BJ. PET/MR images acquired with a compact MR-compatible PET detector in a 7-T magnet. *Radiology* 2007;244:807–14. [PubMed: 17709830]
- Judenhofer MS, et al. Simultaneous PET-MRI: a new approach for functional and morphological imaging. *Nat Med* 2008;14:459–65. [PubMed: 18376410]
- Kim CL, Wang G-C, Dolinsky S. Multi-pixel photon counters for TOF PET detector and its challenges. *IEEE Trans Nucl Sci* 2009;56:2580–5.
- Kovaltchouk VD, Lolos GJ, Papandreou Z, Wolbaum K. Comparison of a silicon photomultiplier to a traditional vacuum photomultiplier. *Nucl Instrum Methods Phys Res A* 2005;538:408–15.
- Lytle JD, Wilkerson GW, Jaramillo JG. Wideband optical transmission properties of seven thermoplastics. *Appl Opt* 1979;18:1842–6. [PubMed: 20212560]

- Moehrs S, Del Guerra A, Herbert DJ, Mandelkern MA. A detector head design for small-animal PET with silicon photomultipliers (SiPM). *Phys Med Biol* 2006;51:1113–27. [PubMed: 16481681]
- Otte AN, Barral J, Dolgoshein B, Hose J, Klemin S, Lorenz E, Mirzoyan R, Popova E, Teshima M. A test of silicon photomultipliers as readout for PET. *Nucl Instrum Methods Phys Res A* 2005;545:705–15.
- Pichler BJ, Judenhofer MS, Catana C, Walton JH, Kneilling M, Nutt RE, Siegel SB, Claussen CD, Cherry SR. Performance test of an LSO-APD detector in a 7-T MRI scanner for simultaneous PET/MRI. *J Nucl Med* 2006;47:639–47. [PubMed: 16595498]
- Pichler BJ, Swann BK, Rochelle J, Nutt RE, Cherry SR, Siegel SB. Lutetium oxyorthosilicate block detector readout by avalanche photodiode arrays for high resolution animal PET. *Phys Med Biol* 2004;49:4305–19. [PubMed: 15509067]
- Renker D. Geiger-mode avalanche photodiodes, history, properties and problems. *Nucl Instrum Methods Phys Res A* 2006;567:48–56.
- Renker D. New trends on photodetectors. *Nucl Instrum Methods Phys Res A* 2007;571:1–7.
- Schaart DR, van Dam HT, Siefert S, Vinke R, Dendooven P, Lohner H, Beekman FJ. A novel, SiPM-array-based, monolithic scintillator detector for PET. *Phys Med Biol* 2009;54:3501–12. [PubMed: 19443953]
- Siefert S, Schaart DR, van Dam HT, Huizenga J, Vinke R, Dendooven P, Lohner H, Beekman FJ. A high bandwidth preamplifier for SiPM-based TOF PET scintillation detectors. *IEEE Nucl Sci Symp and Med Imaging Conf Rec* 2008:NM1–2.
- Siegel S, Silverman RW, Shao Y, Cherry SR. Simple charge division readouts for imaging scintillator arrays using a multi-channel PMT. *IEEE Trans Nucl Sci* 1996;43:1634–41.
- Spanoudaki VC, Mann AB, Otte AN, Konorov I, Torres-Espallardo I, Paul S, Ziegler SI. Use of single photon counting detector arrays in combined PET/MR: characterization of LYSO-SiPM detector modules and comparison with a LSO-APD detector. *J Instrum* 2007;2:12002.
- Stickel JR, Qi J, Cherry SR. Fabrication and characterization of a 0.5-mm lutetium oxyorthosilicate detector array for high-resolution PET applications. *J Nucl Med* 2007;48:115–21. [PubMed: 17204707]
- Tai Y-C, Wu H, Pal D, O'Sullivan JA. Virtual-pinhole PET. *J Nucl Med* 2008;49:471–9. [PubMed: 18287272]
- Wu H, Pal D, O'Sullivan JA, Tai Y-C. A feasibility study of a prototype PET insert device to convert a general-purpose animal PET scanner to higher resolution. *J Nucl Med* 2008a;49:79–87. [PubMed: 18077526]
- Wu H, Pal D, Song TY, O'Sullivan JA, Tai Y-C. Micro insert: a prototype full-ring PET device for improving the image resolution of a small-animal PET scanner. *J Nucl Med* 2008b;49:1668–76. [PubMed: 18794253]
- Yamamoto S, Imaizumi M, Kanai Y, Tatsumi M, Aoki M, Sugiyama E, Kawakami M, Shimosegawa E, Hatazawa J. Design and performance from an integrated PET/MRI system for small animals. *Ann Nucl Med* 2010;24:89–98.
- Yao R, Ma T, Shao Y. Lutetium oxyorthosilicate (LSO) intrinsic activity correction and minimal detectable target activity study for SPECT imaging with a LSO-based animal PET scanner. *Phys Med Biol* 2008;53:4399–415. [PubMed: 18670052]

Figure 1.

A schematic diagram of a multi-pixel photon counter (MPPC) viewed from the top illustrates the size of the sensor body and slightly off-centered $1 \times 1 \text{ mm}^2$ sensitive area. The letters A and C denote anode and cathode, respectively, and an equivalent symbol is shown on the right.

Figure 2.

A photographic image of a white Teflon base with nine pockets holding the MPPC array in place (a) and an illustration of the 3×3 MPPC array (b). The sensitive areas are equally spaced with a 3.4 mm pitch that provides a $8.6 \times 8.6 \text{ mm}^2$ sensing area for 10×10 LSO. The outer dimension of the sensitive area is only $7.8 \times 7.8 \text{ mm}^2$ and that is about 18% less than the LSO array size ($8.6 \times 8.6 \text{ mm}^2$).

Figure 3.

(a) A photograph of the MPPC block detector and the readout board. The perspective drawing on the right (b) shows the detector assembly part consisting of a 10×10 LSO array on top, light guide, reflective mask and MPPC array (from top to bottom). A schematic diagram of the resistive charge division network is shown in (c). The unit of each value is in ohms (Ω).

Figure 4.

A cross-sectional drawing of a light guide tested on the Monte Carlo simulation. The light guide shown in the geometry has three saw-cuts with a different cut depth on each side.

Figure 5.

Experimental setup for timing resolution measurement. The position of the MPPC detector was shifted vertically and horizontally in order to electronically collimate the gamma rays onto a targeted region (center, edge and corner) of the LSO array.

Figure 6. Comparing the five representative light guides among the various Monte Carlo simulation geometries (a) and the corresponding flood images of five LSO crystals (b). N/C indicates no cut available and N/R indicates the no bottom reflective mask adopted.

Figure 7. A sketch of the final design of the block detector (a) and 2D position histogram of the 10×10 LSO pixels of 0.8 mm size crystals (b). This flood image was acquired with a Na-22 point source (511 keV). The image contains about 5 million events.

Figure 8.

Energy spectrum of each LSO crystal in the 10×10 array. (a) The lookup table shows the boundary of each pixel where the energy data were taken to calculate the energy resolution. (b) Energy spectra comparison at the corner, edge and center pixels before and after linearity correction.

Figure 9.

(a) Timing resolution of a single channel MPPC was 857 ps full-width at half-maximum against with H5783 PMT when $\pm 10\%$ energy window was applied. (b) The timing resolution value of the center of the MPPC array versus PMT was 1182 ps with $\pm 10\%$ energy window.

Figure 10.

Surface plots of the photopeak channels of 10×10 LSO pixels at four different gamma-ray energies: 140 keV, 185 keV, 364 keV and 511 keV.

Figure 11.

Photopeak channel versus gamma-ray energy. The linear extrapolation as a function of the gamma-ray energy between 93 keV and 511 keV was calculated by a polynomial fit of the first few data points. (a) A single LSO crystal coupled to a PMT versus an MPPC and (b) LSO crystals in the central region of the detector module versus corner crystals.

Table 1

Statistical data set of the recommended operational voltage, gain and dark current of the nine MPPC from Hamamatsu.

	Minimum	Maximum	Mean	SD
Operational voltage (V)	68.91	68.97	68.95	±0.02
Current gain ^a ($\times 10^5$)	7.48	7.52	7.50	±0.01
Dark counts ^a (k cps)	346	369	359	±8

^aTest condition: 25 °C, $\lambda = 655$ nm.

Table 2

Photopeak channels of various gamma-ray energies at different crystal locations.

Gamma energy (keV)	Peak channels		
	The corner ^a	The center ^b	Single MPPC
93	20	19	15
122	24	23	21
140	25	26	21
185	34	36	27
300	51	57	40
364	50	64	44
511	68	90	55
1275	N/A	170	N/A

^a Mean value of the four corner crystals in the 10×10 LSO array.

^b Mean value of the central four crystals.