Published in final edited form as:

Nucl Instrum Methods Phys Res A. 2014 January 21; 735: . doi:10.1016/j.nima.2013.10.008.

A large area, silicon photomultiplier-based PET detector module

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Abstract

The introduction of silicon photomultipliers (SiPM) has facilitated construction of compact, efficient and magnetic field-hardened positron emission tomography (PET) scanners. To take full advantage of these devices, methods for using them to produce large field-of-view PET scanners are needed. In this investigation, we explored techniques to combine two SiPM arrays to form the building block for a small animal PET scanner. The module consists of a 26×58 array of $1.5 \times 1.5\text{mm}^2$ LYSO elements (spanning $41 \times 91\text{mm}^2$) coupled to two SensL SiPM arrays. The SiPMs were read out with new multiplexing electronics developed for this project. To facilitate calculation of event position with multiple SiPM arrays it was necessary to spread scintillation light amongst a number of elements with a small light guide. This method was successful in permitting identification of all detector elements, even at the seam between two SiPM arrays. Since the performance of SiPMs is enhanced by cooling, the detector module was fitted with a cooling jacket, which allowed the temperature of the device and electronics to be controlled. Testing demonstrated that the peak-to-valley contrast ratio of the light detected from the scintillation array was increased by $\sim 45\%$ when the temperature was reduced from 28°C to 16°C . Energy resolution for 511 keV photons improved slightly from 18.8% at 28°C to 17.8% at 16°C . Finally, the coincidence timing resolution of the module was found to be insufficient for time-of-flight applications (~ 2100 ps at 14°C). The first use of these new modules will be in the construction of a small animal PET scanner to be integrated with a 3T clinical magnetic resonance imaging scanner.

Keywords

PET; Silicon Photomultipliers; Nuclear Medicine Instrumentation

1. Introduction

Development of a combined MRI-PET system has received considerable attention over the last two decades. Indeed, this combination was first proposed in the mid-1990s (Christensen et al 1995, Raylman et al 1996). It wasn't until the early twenty-first century, however, when arrays of avalanche photodiodes (APD) became available, that practical systems were constructed. In one of the first efforts, a UC Davis-University of Tübingen collaboration

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created an MR-compatible PET scanner insert that utilized APDs (Pichler et al 2006). This device was constructed from detector modules consisting of a 10×10 array of $2 \times 2 \times 12$ mm³ Lutetium Orthosilicate (LSO) elements coupled to a 3×3 array of APDs through a 3.5mm thick acrylic light guide. The active area of the detector module is approximately 2×2 cm². The PET scanner was integrated with a small MRI RF coil for use with a 7 T MRI animal scanner. Grazioso *et al.* from Siemens Molecular Imaging in Knoxville, TN have also utilized APDs to produce MRI-compatible PET detector modules designed to be placed inside the imaging region of a 1.5 T clinical MRI scanner (Grazioso *et al* 2006). This device consists of 8×8 arrays of $2 \times 2 \times 20$ mm³ LSO elements coupled to 2×2 arrays of APDs. The modules were successfully tested inside the bore of a Siemens 1.5 T Symphony MRI scanner. This work led to creation of the first commercially available MRI-PET scanner by Siemens. While the scanners produced with these modules had relatively good characteristics, performance was ultimately limited by the relatively low signal-to-noise ratio (SNR) due to the low gain and temperature-dependent noise of the APDs.

Perhaps the most important development in the creation of practical and high performing MR-compatible PET detectors was the development of arrays of silicon photomultipliers (SiPM). These devices have higher gain than APDs, comparable to photomultiplier tubes (on the order of 1×10^6), and have the same insensitivity to magnetic fields as APDs (Roncali and Cherry 2011). A number of investigators have created MR-compatible PET detector modules from which MRI-PET scanners can be constructed (Chagani 2009, Schaart 2009, Yamamoto 2010, Llosa 2011, Schulz 2011, Zorzi 2011, Wang 2012, Yoon 2012). For example, a group from the Seoul National University constructed a 32.4×28.7 mm² SiPM-based PET detector module (Yoon 2012). The energy resolution of the detector was reported to be 13.9% for 511 keV photons. A group from the Netherlands developed a PET detector module utilizing a single $13.2 \times 13.2 \times 10$ mm³ piece of LYSO mounted on a 4×4 array of SiPMs (Schaart 2009). The use of a monolithic piece of scintillator permits the assessment of each photon's depth-of-interaction (DOI). Finally, Schulz *et al.* developed a MRI-compatible PET detector module consisting of a 22×22 array of $1.3 \times 1.3 \times 10$ mm³ LYSO elements coupled to an array of SiPMs (Schulz 2011). While each of these efforts produced good PET detector modules, their active areas were relatively small and did not take full advantage of the potential performance SiPMs by not cooling them to low temperatures (below 22°C).

2. Material and Methods

The goal of this project was to create a SiPM-based PET detector module that will be used as a building block of a large field-of-view (FOV) PET small animal scanner for use with a 3T clinical MRI scanner. A cooling system for the module was constructed to aid in stabilizing and enhancing the performance of the SiPMs. Finally, a multiplexing scheme was used to reduce the total number of data acquisition channels, facilitating construction of practical and cost effective PET scanners.

2.1 Detector Design

The new detector module utilizes a 26×58 array of $1.5 \times 1.5 \times 10$ mm³ (pitch=1.57 mm) LYSO elements separated by ESR reflector (Proteus, Inc., Knoxville, TN). Thus, the active area of the detector is 41.2×91.5 mm², which is larger than other SiPM-based detector modules reported in the literature (Chagani 2009, Schaart 2009, Yamamoto 2010, Llosa 2011, Zorzi 2011, Schulz 2011, Wang 2012, Yoon 2012). The LYSO array was coupled to two SensL ArraySL-4p9s (SensL Technologies, LTD., Cork Ireland). These devices are made-up of a matrix of 3×3 ArraySL-4 SiPMs. Each ArraySL-4 has sixteen (4×4) 3.05×3.05 mm² pixels (4,774 microcells). Their typical gain is 2.4×10^6 , at room temperature and recommended bias. Thus, each of the ArraySL-4p9s produces $3 \times 3 \times 4 \times 4 = 144$ channels

of data; the light-sensitive area is $48 \times 48 \text{ mm}^2$. A 2 mm-thick piece of acrylic was used to optically couple the two SiPM arrays to the scintillator array. The light guide is necessary to spread light from the LYSO elements to bridge the optical seam between the two SiPM arrays. Hence, the scintillation light produced by elements located at the seam where the two SiPM arrays meet was transmitted to the active areas of the arrays. The light guide also spreads scintillation light amongst the pixels of the SiPM arrays to facilitate calculation of event position in the scintillator array. Figure 1 shows the complete detector unit. Note that the SiPM arrays are larger than the scintillator array. This geometry was chosen so that the scintillator elements at the edges of the array were positioned inside the light-sensitive area of the SiPM array, facilitating identification of all of the elements of the scintillator array. This capability will reduce gaps in detector sensitivity when the modules are combined in a ring to form a scanner.

2.2 Detector Cooling System

As with most solid-state devices, the performance of SiPMs is affected by temperature. Lowering their temperature reduces thermal (dark current) noise and increases gain. The gain increase is due to the reduction of the breakdown voltage at lower temperatures, while keeping the bias voltage constant. The detector temperature was reduced by enclosing it in a cooling jacket. Specifically, a piece of 4.76 mm thick copper plate was formed into a rectangle to enclose the ArraySL-4p9s and the readout electronics. A channel consisting of $2.38 \text{ mm} \times 2.38 \text{ mm}$ square brass tubing was soldered to the exterior of the enclosure. Barbed tubing connectors were soldered to each end of the channel. A mixture of cooled 50% distilled water and 50% ethylene glycol was circulated through the brass channels, which in turn cooled the copper enclosure. The copper is in contact with the edges of the SiPM array, providing conductive cooling. The liquid is cooled with a mini-chiller (Peter Huber Kältemaschinenbau GmbH, Offenburg Germany). This device can cool the circulating liquid to -10°C and has a cooling power of 300W at 14°C , which is sufficient to cool a maximum of 250 of our modules simultaneously to 14°C (each module dissipates $\sim 1.2 \text{ W}$ of power). To create convective cooling of the module, air is circulated inside the unit via of a 3.175 mm-inner diameter tube integrated with the cooling jacket. This tubing contains seven small holes that permit air to flow across the SiPM readout electronics. Air is supplied by a 4.76 mm-inner diameter tube connected to a medical air receptacle in the research area. The air is pre-cooled by passing cooling fluid through a heat exchanger in contact with the air supply tube. All of the materials used for the cooler were chosen for their high heat conductivity and non-magnetic characteristics. The complete PET detector module is shown in figure 2. The cooling system is capable of producing uniform cooling over the surface of the ArraySL-4p9s (variation of $< \sim 0.6^\circ\text{C}$). Figure 3 shows an infrared image of the front face of the cooled module without the scintillator. Note that the apparent hot areas in the cooling jacket are caused by the reflection of ambient light from the copper and mismatch in the emissivity of the copper and the emissivity setting of the IR camera, which was set to the emissivity of Silicon.

2.3 SiPM Readout and Data Acquisition System

Each SiPM array contains 144 individual elements, so our dual array detector module (two ArraySL-4p9s) produces 288 individual analog outputs. This amount of data makes it challenging to readout one of the devices, let alone a large number of modules that would make up a full scanner. Thus, to make the unit more appropriate for use in larger systems, the number of output channels for each ArraySL-4p9 was reduced from 144 to 4, resulting in eight analog outputs per module. This task was accomplished with a multiplexed readout system developed in collaboration with AiT Instruments (Newport News, VA).

Output signal multiplexing is accomplished in two stages. First, symmetric charge division circuitry decomposes the 144 channels of the 12×12 ArraySL-4p9 to twelve rows and twelve columns. These signals are reduced to four channels by applying a weighted gain to each row and column proportional to its location along each axis. This amplitude encoding produces two signals ($X+$, $X-$) representing an event column location, and two signals ($Y+$, $Y-$) representing an event row location. A new charge division technique developed by AiT Instruments uses diodes instead of resistors to process the SiPM signals prior to amplification. Specifically, a diode is placed between the photodetector and a fast transimpedance amplifier. When the photodetector signal polarity is positive then the diode's low forward resistance allows current flow from the photodetector to the amplifier, while the diode's high reverse resistance blocks current flow from the amplifier to the photodetector. Figure 4 shows a schematic of the readout electronics. The diodes reduce undesirable leakage current amongst the SiPMs. Reduced leakage improves spatial uniformity and event positioning. The nonlinear forward resistance of the diodes attenuates low-amplitude signals, similar to amplitude thresholding, resulting in reduced noise. The diodes isolate individual SiPM capacitance in a group of SiPMs when each photodetector is connected to a transimpedance amplifier, which reduces the effects of cumulative capacitance and charge loss in large arrays of SiPMs. Additionally, the output rise time of the transimpedance amplifier is faster than most directly coupled readout circuits.

The outputs from the readout boards are connected to an interface module (AiT Model# SiPMIM4x4) that accepts signals from up to four ArraySL-4p9s (the equivalent of two of our modules). The module routes the individual analog signals from the readout electronics to a 34-pin connector located on the front of the device. In addition, the module sums the outputs from each ArraySL-4p9. These signals are used to produce the trigger for initiation of an analog-to-digital conversion of the X-Y outputs. Additionally, the interface module supplies bias voltage and power to the readout electronics. Finally, bias voltage and current drawn by each SiPM array, as well as the temperature of the SiPMs (via temperature sensors mounted on the boards) can be monitored using a 24-pin output on the interface module.

The analog voltage signals from the interface module are routed through a 34-pin cable to an FPGA-based multichannel, analog-to-digital converter unit (ADC) (AiT Instruments, Newport News, VA). Each channel uses an analog delay to compensate for trigger latency, followed by a gated integrator and an ADC to digitize the integrated signal. These units can process a maximum of sixty-four channels (each ADC box can thus accommodate up to eight of our modules). Digitization of the analog signals is triggered by a logic pulse. The logic signals are produced by passing the sum outputs from the interface module (one from each ArraySL-4p9) through a constant fraction discriminator (Philips Scientific Model 715) and then combining the two pulse height-discriminated outputs with a logic OR operation in a Philips Scientific Model 755 logic unit. The digitized outputs are transmitted to the controlling computer via a USB2 connection. Figure 5 shows a schematic of the data acquisition system used to test the module. Determination of the position of a photon interaction point in the scintillator array was performed by calculation of the center-of-mass of the digitized signals from the readout electronics in conjunction with a pre-calculated look-up table (often referred to as a crystal map). This map correlates event position with scintillator pixel element number. Additionally, a pre-measured energy calibration table was applied to the data to calculate the energy deposited in the scintillator. This table contains a factor for each scintillator pixel element that converts ADC channel number to energy. Data acquisition was controlled using the Kmax programming environment (Sparrow Corp., Port Orange, FL).

2.4 Detector Module Testing

Testing was performed to assess the capabilities and limitations of the detector module. First, the dark current as a function of temperature was determined by measuring the current drawn by the SiPMs (without the scintillator array attached) as the temperature was reduced. To gauge the ability of the system to identify scintillator elements as the temperature was reduced, the module was irradiated with six, 4.5 μCi ^{22}Na disk sources (Eckert & Ziegler, Valencia CA) evenly distributed across the face of the module. Data were acquired for 300 s. The applied bias voltage was set to 30.4V, which is the manufacturer-recommended voltage at room temperature. Thus, we were able to assess the effects that changing temperature has on module performance at a constant bias voltage. Note that the gain of an SiPM is directly related to the difference between the bias voltage and breakdown voltage, also known as the overvoltage. As the device is cooled, the breakdown voltage is reduced. Since the bias voltage is constant, the result is an increased overvoltage and consequently increased gain as temperature decreases. It should be noted that SiPM gain could be increased by increasing the bias voltage, but this procedure has the effect of increasing dark current. The resulting data was used to produce a two-dimensional plot of detected event positions. The pixel map is created from single rather than coincidence events. These data were used to calculate the peak-to-valley contrast ratios (PVCr) for each detector element, which is calculated by subtracting the counts present in the region between the positions of adjacent detector elements (valley) from the counts present in the region of a detector element (peak) divided by the counts in the peak. This parameter is a measure of the precision of the center-of-mass calculation used to identify the position of a photon interaction in the detector. To avoid the inherent variability of using intensity profiles to determine the position of peaks and valleys, PVCrs were calculated using software specially created to search for a peak and adjacent valley in the pixel map. The PVCrs for each scintillator element were combined to produce a mean value for the module at a given temperature. Measurements were performed with no cooling of the unit (unit temperature = 28 °C) and with the cooling system turned on (temperature ranged from 22 °C to 16 °C in increments of 2 °C). This range was judged reasonable for standard operation.

2.5 Coincidence Timing Resolution

The coincidence timing resolution of the detector module was measured by placing it in coincidence with a fast detector comprised of a 1 cm^3 piece of LSO coupled to a Hamamatsu R2496 PMT (Hamamatsu Photonics, Hamamatsu City, Japan). The spread of electron transit time of this device is 600 ps full-width-at-half-maximum (FWHM). Three 4.5 μCi ^{22}Na disk sources were placed between the detector module and the fast detector (separation = 3 cm). The input signals from the SiPM and PMT were connected to a constant fraction discriminator (Philips Scientific 6915). The resulting pulses were input to a time-to-digital converter (TDC) (ATMD-GPX, ACAM Electronics, Stutensee-Blankenloch, Germany). The fast detector signal supplied the start signal and the signal from the detector module was the stop signal. Coincidence timing resolution was calculated by fitting the resulting timing distribution to a Gaussian function. The FWHM of the fit (corrected for the transit time spread of the PMT used to produce the start signal) is the coincidence timing resolution of the module.

3. Results

The plot shown in figure 6(a) demonstrates the reduction in dark current with reduced temperature. The relationship between dark current and temperature is linear over this temperature range, as demonstrated by the fit of the curve to a line. Figure 6(b) shows a plot of ArraySL-4 gain calculated from SensL data (SensL 2011) as a function of temperature. The gain increases as the temperature is reduced because it is directly related to overvoltage,

which increases with decreasing temperature (assuming a constant bias voltage). Figure 7(a) shows a representative pixel map. Importantly, all of the 26×58 LYSO detector elements of the scintillator array were discernable, even at the edges and seam where the two arrays meet. This finding is important since identification of the elements in these areas can be challenging. Figure 7(b) shows an intensity profile measured from one of the rows of the pixel map shown in figure 7(a). PVCr and energy resolution were measured from maps similar to the one shown in Figure 7. Figure 8(a) shows a plot of PVCr versus temperature. Energy resolution versus temperature is shown in figure 8(b). Finally, figure 9(a) shows representative coincidence timing resolution curves. Figure 9(b) displays the results of the coincidence timing resolution measurements. The curve was fit to the function $T_{\text{Res}} = M1 + M2e^{(M3 \cdot T)}$, which was chosen because it best fit the data. M1 is the temperature-independent component of coincidence timing resolution for the module (T_{Res}), M2 is the amplitude of the exponential component of the timing resolution, M3 is the temperature coefficient of the timing response and T is the temperature of the SiPMs. The best timing resolution achievable with this detector is ~ 2112 ps (the value of the temperature independent component of the curve), which is just slightly below our measured value of 2115 ps at 16°C. It is challenging to put these results in context with the results from other groups because of the large variation in devices and testing methods. Groups measuring coincidence timing resolution for LYSO-based SiPM detectors report results ranging from 120 ps to 1500 ps (Piemonte 2010, Llosa 2009, Seifert 2009, Ahmed 2012, Lee 2011, Shimizu 2013, van Dam 2013). It should be noted that none of these results are for the SensL ArraySL-4p9.

4. Discussion

Construction of PET detectors that are not susceptible to the effect of strong magnetic fields has been significantly advanced by the introduction of SiPMs. Commercially available SiPMs arrays are, however, not of sufficient size to be used to create large-FOV PET scanners. Therefore, they must be tiled to produce larger units. In this investigation, we developed and tested a SiPM-based detector module that will be used as a building block for a planned small animal PET scanner.

Figure 1 shows the detector/electronics components of the unit. The complete PET detector module, detector components and cooling system, is shown in figure 2. This system was capable of producing relatively uniform cooling across the SiPM arrays of the module, as demonstrated in the infrared image shown in figure 3, which is important to maintain uniform performance of the SiPM's across the active area of the detector. As with the detector components, including the electronics shown in figures 4 and 5, the cooling jacket concept can be scaled up for construction of larger detector modules.

The data shown in the plots of figure 6 demonstrate the effect that lowering the temperature of the module has on gain and dark current. Dark current has a linear relationship with temperature over the range tested (as demonstrated by the plot in figure 6a)), with a slope of $\sim 2 \mu\text{A}/^\circ\text{C}$. Cooling the SiPMs removes energy from the devices, so the probability of spontaneous electron-hole pairs is lowered, reducing the dark current. Gain increase as a function of temperature (figure 6(b)) is caused by the decrease of breakdown voltage, which is related to the reduced probability of phonon-electron interactions at lower temperatures. Since the bias voltage was fixed, the overvoltage, and hence gain of the device was increased. It should be noted that the reduction of dark current is slightly offset by the increase in gain (the dark current is amplified by same gain factor as the signal). Instead of cooling the SiPMs, it is possible to compensate for the loss in gain caused by increase in device temperature (produced by the power dissipated by the SiPMs and readout electronics) by increasing the bias voltage. This method, however, does not address the increase in dark

current due higher temperatures. Large arrays of SiPMs (288 in the case of our module) read out with multiplexing electronics are especially susceptible to the effects of increased dark current since the signal from each SiPM contributes to detector noise. Thus, cooling was deemed the best method to improve and stabilize our detector's performance.

The pixel map shown in figure 7 illustrates that all of the detector elements are identifiable, even at the edges of the scintillator array. The inability to detect all of the elements of a scintillator array is often caused by reduction in detection efficiency at the edges of the light detection device. In our detector unit, the two ArraySL-4p9s are approximately ~ 2.5 mm wider and longer than the scintillator arrays. Thus, it is possible to efficiently detect light from the edges of the scintillator array since all of the scintillator elements are located within the light sensitive area of the SiPM arrays. This lack of dead areas around the detector reduces degradation in detection sensitivity at the seams of two adjacent modules when they are arranged in a ring scanner, reducing artifacts caused by incomplete angular sampling. Another important finding from the pixel map is the fact that the scintillator elements located at the seam where the two ArraySL-4p9s meet are identifiable. Hence, there are no gaps in the active area of the module. This capability is achieved by the use of a 2 mm-thick acrylic light guide, which allows the light from scintillator elements to be spread from an area where there are no SiPM elements to active areas of the device. The increase in gain and reduction in noise achieved by cooling also contributes to the ability to identify events that occur in detector elements located at the seam between the two SiPMs. The success of this concept means that it is theoretically possible to tile a number of SiPM arrays to create larger detector modules.

Perhaps the most relevant way to gauge how increased signal and reduced noise affects detector performance is by assessing the precision with which events are localized in the scintillator array. To characterize this quantity, we calculated the mean peak-to-valley contrast ratio (PVCr) for the module. The results shown in figure 8(a) demonstrate that the PVCr for the detector module increases by $\sim 45\%$ as the temperature is reduced from 28°C to 16°C . As described above, cooling the detector increases the gain and reduces dark current, which results in increased signal amplitude and reduced noise (mitigated somewhat by the increased amplification of noise). Since these signals are used to calculate event position in the scintillator array with a center-of-mass algorithm, higher amplitude signals in the presence of lower noise makes these calculations more precise. While an increase in PVCr should result in improved spatial resolution, it is not a direct measure of this parameter. Ultimately, improvement in spatial resolution resulting from cooling the SiPMs will have to be assessed once a scanner is constructed.

Increased signal amplitude and reduced noise produced by cooling also improves the energy resolution of the detector by increasing gain. The results in figure 8(b) show that energy resolution improved from 18.8% at 28°C to 17.8% at 16°C for 511 keV photons. This small changes over the temperature range tested is likely due to the fact that the energy resolution for the module is dominated by differences in light yield from the 1508 individual scintillator elements and slight differences in gain amongst the 288 individual SiPMs. Neither of these effects has a strong temperature dependence. The results shown in figure 9 demonstrate that cooling of the SiPMs improves the coincidence timing resolution of our detector module. The coincidence timing resolution is poor compared to that reported by a number of other groups for SiPMs and dSiPMs (Piemonte 2010, Llosa 2009, Seifert 2009, Ahmed 2012, Lee 2011, van Dam 2013). For example, Seifert *et al.* (Seifert 2009) reported an extremely good coincidence timing resolution of 171 ps. Unlike our measurements, this group used a single $3 \times 3\text{mm}^2$ piece of LYSO coupled to a single SiPM (Hamamatsu MPPC) (instead of an array of SiPMs and scintillator elements). Indeed, almost all of the reported coincidence timing resolution results reported in the literature are for single or

small SiPM-scintillator arrays, in contrast to our results for a 26×58 array of $1.5 \times 1.5\text{mm}^2$ LYSO elements (spanning $41 \times 91\text{mm}^2$) coupled to 288 SiPMs. This effect is demonstrated by the work of Shimizu, *et al.* who compared coincidence timing resolution measured from one SiPM to that for an array of SiPMs (Shimizu 2013). Specifically, the coincidence timing resolution of the single SiPM (MPPC)-scintillator ($5 \times 5 \times 20\text{mm}^3$ LYSO crystal) combination was reported to be 354 ps, for an 8×8 array of SiPMs the coincidence timing resolution increased to 1208 ps. This increase in coincidence timing resolution for the larger array is caused, at least, in part, by choices of readout scheme and the lack of module wide timing calibration.

Our results indicate that coincidence timing resolution has a temperature independent component, in addition to an exponential, temperature dependent component. The most likely explanation for the temperature dependence is linked to the fact that as temperature is decreased, device gain increases, which increases the photon detection efficiency (PDE). PDE is major factor influencing coincidence timing resolution. Another effect of reduced temperature is reduced dark current, which will also affects timing resolution. The temperature independent component is the major contributor to the timing resolution measured for the ArraySL-4p9. While the coincidence timing resolution is affected by the length of the scintillator elements, their surface treatment and the presence of a 2mm-thick light guide, much of the temperature independent component is due to the interaction of the SiPM's intrinsic capacitance with the multiplexed readout. Specifically, this interaction affects the rise time of the output pulses, which degrades timing resolution. Thus, as we have shown, the coincidence timing resolution capabilities of our detector module can be improved modestly by reducing temperature. It will require changes in the structure of the ArraySL-4p9, readout electronics (possibly utilizing specialized ASICs (Solf, 2009)) and timing calibration of the module to significantly improve coincidence-timing resolution.

In summary, we have constructed a large area SiPM-based PET detector module using the SensL ArraySL-4p9 devices. The module has the capability to cool the SiPMs, as well as special electronics for acquiring and processing signals from the unit. Performance as assessed by the measurement of some relevant metrics was promising. The next step is to incorporate twelve of the new modules into a ring to form a small animal PET scanner to be used in conjunction with a 3T MRI scanner.

Acknowledgments

The authors thank Carl Jackson and Wade Appelman from SensL Technologies Ltd. for their assistance with the silicon photomultipliers. This work was supported by funds from the NIBIB (R01 EB007349).

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Figure 1.
Picture of the detector unit.

Figure 2.
Picture showing the complete PET module.

Figure 3.
Infrared image of the front face of the cooled PET detector module.

Figure 4.
Schematic of the SiPM readout electronics.

Figure 5.
Schematic drawing of the detector module data acquisition system.

Figure 6. Results from measurement of dark current (a) and calculation of gain (b) as a function of temperature.

Figure 7.
(a) Pixel map from the detector at a temperature of 16°C; (b) representative intensity profile from a central row (denoted by the white line) of the map shown in (a).

Figure 8.

a) Mean peak-to-valley contrast ratio (PVCR) versus temperature (the error bars show the standard deviation of the mean PVCR value) and (b) energy resolution versus temperature.

Figure 9.

a) Representative coincidence timing resolution curves and b) plot of the module coincidence timing resolution as a function of temperature.