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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 1.5×1.5 mm^2 LYSO elements (spanning 41×91 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 ~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 3 T clinical magnetic resonance imaging scanner.

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 [3,10]. It was not 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 created an MR-compatible PET scanner insert that utilized APDs [8]. This device was constructed from detector modules consisting of a 10×10 array of $2 \times 2 \times 12$ mm^3 Lutetium Orthosilicate (LSO) elements coupled to a 3×3 array of APDs through a 3.5 mm thick acrylic light guide. The active area of the detector module is approximately 2×2 cm^2. 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 [4]. This device consists of 8×8 arrays of $2 \times 2 \times 20$ mm^3 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 [11]. A number of investigators have created MR-compatible PET detector modules from which MRI–PET scanners can be constructed ([2,6,12,13,19–22]). For example, a group from the Seoul National University constructed a 32.4×28.7 mm^2 SiPM-based PET detector module ([21]). 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 $13.2 \times 13.2 \times 10$ mm^3 piece of LYSO mounted on a 4×4 array of SiPMs ([12]). The use of

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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 $1.3 \times 1.3 \times 10\ mm^3$ LYSO elements coupled to an array of SiPMs ([13]). 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 3 T 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 $1.5 \times 1.5 \times 10\ mm^3$ (pitch = 1.57 mm) LYSO elements separated by ESR reflector (Proteus, Inc., Knoxville, TN). Thus, the active area of the detector is $41.2 \times 91.5\ mm^2$, which is larger than other SiPM-based detector modules reported in the literature ([2,12,7,13,19–22]). 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 16 (4×4) $3.05 \times 3.05\ mm^2$ pixels (4774 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 $48 \times 48\ mm^2$. 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. Fig. 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 mm $\times$ 2.38 mm 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 300 W 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 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 Fig. 2. The cooling system is capable of producing uniform cooling over the surface of the ArraySL-4p9s (variation of $< \sim 0.6$ °C). Fig. 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.

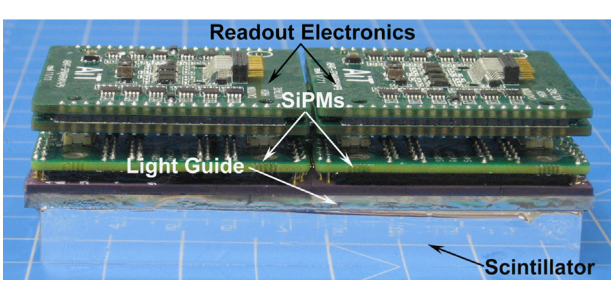

Fig. 1. Picture of the detector unit.

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 four, 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 12 rows and 12 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. Fig. 4 shows a schematic of the readout electronics. The diodes reduce undesirable leakage current amongst the SiPMs. Reduced leakage improves spatial uniformity