2009

SiPM based detector module and digital data acquisition system for PET: Initial results

M Safavi-Naeini
University of Wollongong, mitra@uow.edu.au

D. Franklin
University of Wollongong, danielf@uow.edu.au

M Petasecca
University of Wollongong, marcop@uow.edu.au

M L. F Lerch
University of Wollongong, mlerch@uow.edu.au

R Kirkham
Csiro

See next page for additional authors

Authors

M Safavi-Naeini, D. Franklin, M Petasecca, M L. F Lerch, R Kirkham, G Moorhead, P Dunn, G De Geronimo, and Anatoly B. Rosenfeld

SiPM based Detector Module and Digital Data Acquisition System for PET: Initial Results

M. Safavi-Naeini, *Student Member, IEEE*, D. R. Franklin, *Member, IEEE*, M. Petasecca, *Member, IEEE*, M. L. F. Lerch, *Member, IEEE*, R. Kirkham, *Member, IEEE*, G. Moorhead, *Member, IEEE*, P. Dunn, *Member, IEEE*, G. De Geronimo, *Member, IEEE*, and A. B. Rosenfeld, *Senior Member, IEEE*

***Abstract*—A novel detector module configuration based on silicon photomultiplier (SiPM) detectors with LYSO scintillators, a multichannel mixed analog/digital ASIC and an FPGA-based digital data acquisition system has been developed for preclinical positron emission tomography (PET). The SCEPTER ASIC, originally developed for soft X-ray detection at Brookhaven National Laboratory, has a unique architecture incorporating multiple peak detectors and time-to-amplitude converters, and offers an extremely low dead time, making it ideal for coincidence detection. A representative pair of channels in the prototype system has been characterised. Measured energy and timing resolutions are reported for coincidences between a single-pixel SiPM detector and a conventional PMT. A timing resolution of 11.7 ns was achieved during a 48 hour run. Energy resolution for the SiPM device and data acquisition system was found to be 17%.**

I. Introduction

Positron emission tomography (PET) is one of the fastest growing non-invasive imaging modalities used in both clinical and preclinical studies. For research applications involving small animals, PET systems with a small ring diameter are required. A dedicated small animal PET scanner demands high sensitivity and spatial resolution, at least an order of magnitude better than typical clinical PET scanners. Current state-of-the-art animal PET systems are capable of a spatial resolution of the order of one to two millimetres at the centre of their Field of View (FoV) with a typical coincidence detection sensitivity of between 3 and 5 % [1].

Ms. Safavi-Naeini, Dr. Petasecca, Dr. Lerch and Prof. Rosenfeld are with the Centre for Medical Radiation Physics at the University of Wollongong, Australia.

Dr. Franklin is with the School of Electrical, Computer and Telecommunications Engineering at the University of Wollongong, Australia.

Mr Kirkham, Dr. Moorhead and Dr. Dunn are with CSIRO Materials Science and Engineering.

Dr. De Geronimo is with Brookhaven National Laboratory.

PET spatial resolution deteriorates towards the edge of the FoV. This is a result of parallax error, which increases uncertainty in the placement of the Line of Response (LoR), creating an elongation artifact in the reconstructed image (Figure I). The resulting non-uniformity in image resolution may be reduced by accurately determining the depth at which gamma photons interact with the scintillation crystal (Depth of Interaction or DoI). Techniques for determining DoI include placing photodetectors on opposite sides of a single scintillator slab and using the relative pulse heights to determine DoI (phoswitch detector), using a non-uniform optical coupling between adjacent pairs of scintillator crystals, in which DoI is determined from the relationship between the energies measured in the adjacent crystals, and segmenting (optically isolating) the scintillator and detectors in the radial direction [2], [3], [4]. Simulation based studies have also suggested alternative system geometries such as box-like formation (as apposed to cylendrical) to optimise photon sensitivity [5].

Current PET system use several types of silicon photodetectors such as PIN detectors and avalanche photodiodes (APDs), with silicon photomultipliers (SiPMs) becoming an increasingly popular alternative in recent years. SiPMs reduce the gain uncertainty of individual APDs with respect to bias and temperature, and offer a much higher gain that the PIN diodes. They offer a very fast rise time, making them ideal detectors for studies requiring a sub-nanosecond timing resolution.

In order to perform high-resolution imaging of organs or sub-organ structures in rodents, the University of Wollongong's Centre for Medical Radia-

978-1-4244-3962-1/09/$25.00 ©2009 Crown

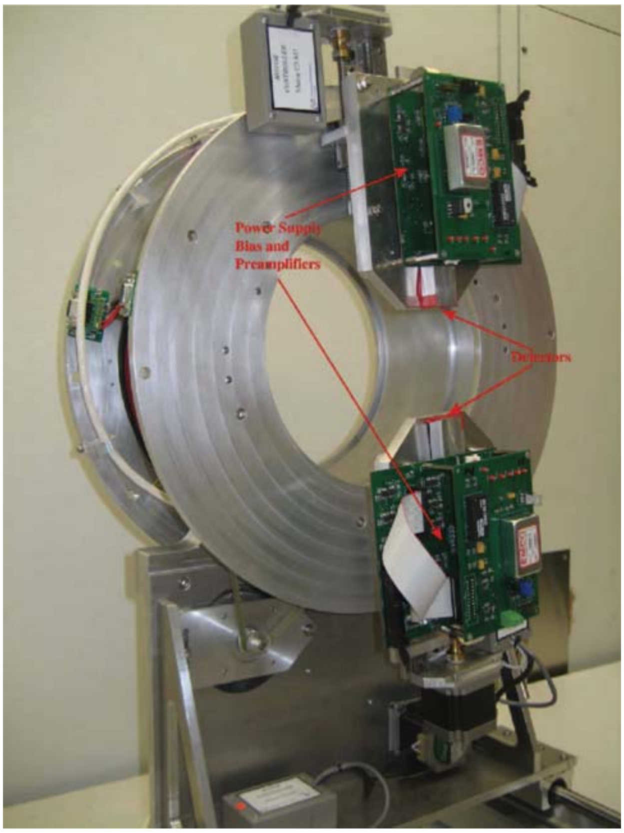
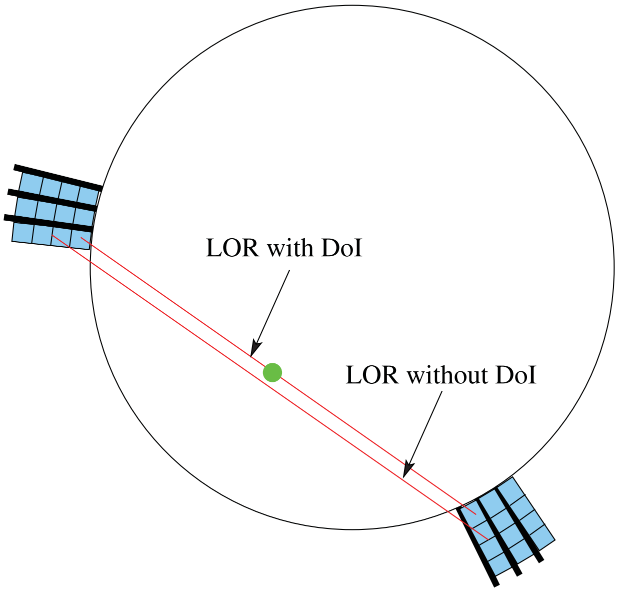


Fig. 1. CMRPet: Two Detector Heads Mounted on Gantry



tion Physics (CMRP) is developing an animal PET system, CMRPET, with a theoretical sub-millimeter resolution and high sensitivity. The CMRPET design utilises a novel detector module with SiPM detectors and LYSO scintillator arrays placed in an edge-on orientation combined with one-to-one scintillator/detector optical coupling. This design will provide depth of interaction capability and minimises dead time for each module.

Placing pixellated photodetectors and crystals in an edge on formation is a unique feature of CMRPet, providing a good estimate of the LoR. The high intrinsic gain of the SiPM detector along with the total volume of scintillator material provides the necessary sensitivity [6]. The CMRPet system has a radial depth of four 3×3 mm pixels, offering a theoretical spatial resolution at the centre of the field of view of 0.68 mm.

The significant increase in detectors per module (16 SiPM detectors compared to the four PMTs traditionally used) requires the use of sophisticated analog/digital pulse processing methods. This is achieved via the SCEPTER ASIC from Brookhaven National Laboratory, coupled to a powerful signal processing and data acquisition system (CSIRO's HYMOD) which pre-processes and logs the large volume of resulting data for offline coincidence detection and image reconstruction.

This paper presents initial measurements of timing and energy resolutions performed using a scaled-down version of the CMRPET system, with only one pair of channels in coincidence. To provide a useful comparison with more conventional hardware, detectors, a SiPM detector is used for the first channel and a photomultiplier tube (PMT) is used for the second. This experimental configuration is discussed in Section II. Measurements of timing and energy resolutions were performed with this configuration, and the results are presented in Section III. Section IV summarises the achievements to date and outlines the directions for ongoing research.

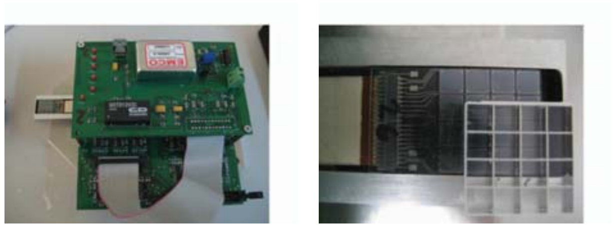
II. Materials and Methods

The experimental apparatus is described in the following sections.

A. Detector and Preamplifier

A 3×3 mm SiPM photodetector (SENSL SPMMicro 3035) was biased at 28.2 V and coupled to a matching $3 \times 3 \times 3$ mm LYSO scintillator via optical grease. The scintillator was chosen due to its high light yield and short decay time.

A photomultiplier tube (Hamamatsu R2300 biased at 850 V) coupled to another $3 \times 3 \times 3$ mm



(a) SiPM detector and its associated preamplifier circuit (b) SiPM detector arrays and scintillator

Fig. 2. Detector arrays, scintillator and preamplifier

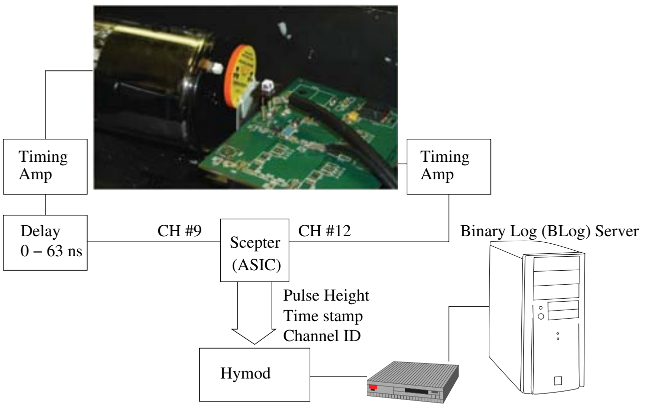


Fig. 3. Single pixel SiPM detector in coincidence with the PMT with digital readout system

LYSO scintillator crystal was placed 3 cm away from the SiPM, with a ^{22}Na point source placed at the midpoint of the two detectors. The output signals were fed to a dual input/output preamplifier based on a fast wideband Darlington amplifier (GALI-5+) with a fixed gain of 17.5 dB.

The preamplified signal from both detectors were coupled to timing amplifiers with a 200 ns integration time, and the resulting signals acquired by an Application Specific IC (discussed in detail in Section II-B). The measured pulse heights and times of arrival were recorded by the HYMOD system (discussed in detail in Section II-C) and logged.

A pulse threshold was applied to suppress spurious noise. All measurements were performed at room temperature (21 °C). The experimental apparatus was validated using a precision pulse generator, which showed a timing resolution of approximately 3.5 ns.

B. *SCEPTER Pulse Capture System*

The shaped and gain adjusted signals were coupled to two channels of a 32-channel analog/digital ASIC, SCEPTER. Developed by the Brookhaven National Laboratories, SCEPTER detects shaped and preamplified pulses arriving from the detectors and converts the pulse height and time of arrival to a digital value via a peak detect and Time to Amplitude Converting (TAC) system [7]. It can sequentially record up to eight simultaneous incoming pulses. The time of arrival and pulse height information are read out as analog voltages in successive read request intervals. These can be sampled in the middle of the read request interval at the same time as channel addresses are read out digitally. The readout process, which is independent from the pulse recording method, minimises the dead time of the system. By analysing the timestamps and the pulse energies, it is possible to identify coincidences, allowing LoRs to be constructed for sinogram binning and image reconstruction.

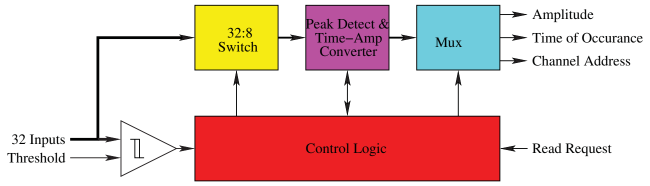


Fig. 4. SCEPTER pulse capture system

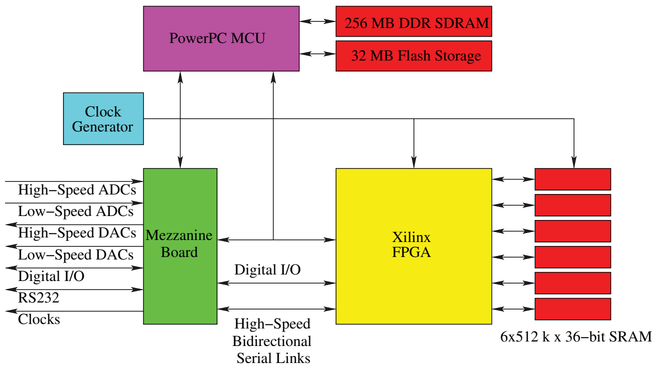
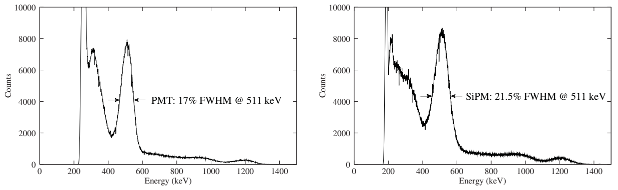


Fig. 5. HYMOD Data Acquisition System

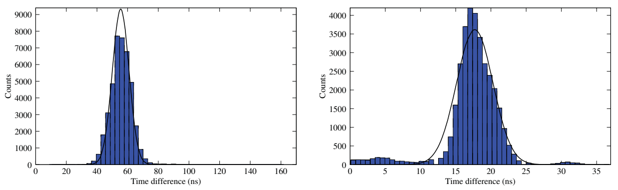
C. *HYMOD data acquisition system*

Data is acquired in real time via the HYMOD PowerPC/FPGA signal processing board developed at CSIRO, which forwards it to a data logger for analysis [8]. HYMOD digitises the analog representations of pulse height and time of arrival, acquires the channel addresses and generates the required control signals for SCEPTER (including the read request clock).



(a) $^{22}\mathrm{Na}$ spectra, PMT (b) $^{22}\mathrm{Na}$ spectra, SiPM

Fig. 6. Energy resolution, detectors coupled to identical to 3 x 3 x 3 mm^3 LYSO scintillators



(a) SiPM/LYSO - PMT/LYSO (b) Pulser

Fig. 7. Coincidence timing spectra with optimal read request frequency of 9.1429 MHz. A timing resolution of 11.7 ns FWHM was achieved for coincidences between SiPM/LYSO and PMT/LYSO, while the measured resolution for the pulser was 6.1 ns.

III. Results: Timing & Energy Resolution

A timing resolution of 11.7 ns FWHM was obtained from the HYMOD-based DAQ system. The measured energy resolutions were 17% and 21.5% FWHM (at 511 keV) for the PMT and SiPM detector respectively.

Timing resolution may be further improved by additional power supply noise filtering and using a shorter shaping time. Energy resolution can be enhanced by providing better regulation of detector temperature, in order to eliminate thermally-induced SiPM gain variation over the acquisition period.

IV. Conclusion

A novel hybrid analog/digital prototype PET data acquisition system with SiPM detectors has been designed and partially tested. The measured timing and (SiPM) energy resolutions of 11.7 ns and 21.5% respectively clearly demonstrate that the system has great potential for use in preclinical PET instrumentation.

F_{clock} (MHz)	SiPM-PMT t-res (ns)	Pulser t-res (ns)
3.0476	13.9	6.5
5.8182	15.6	**5.5**
9.1429	**11.7**	6.1

Integration of all system components, including multichannel analog signal processing electronics, automatic gantry control (using a CMRP-developed high-resolution gantry system) and image reconstruction and processing software, is being finalised now.

References

[1] T. K. Lewellen, Recent developments in pet detector technology, Journal of Phys. Med. Biol. 53 (2008) 287–317.

[2] Y. H. Chung, Y. Choi, G. Cho, Y. S. Choe, K. H. Lee, B. T. Kim, Characterisation of dual layer phoswich detector performance for small animal pet using monte carlo simulation, Phys. Med. Biol 49 (2004) 2881–2890.

[3] Y. Shao, K. Meadors, R. W. Silverman, R. Farrel, L. Ciriginano, R. Grazioso, K. S. Shah, S. R. Cherry, Dual apd array readout of lso crystals: Optimiziation of cyrstal surface treatment, IEEE Trans. Nucl. Sci 49 (3) (2002) 649–654.

[4] A. Blanco, N. Carolino, C. M. B. A. Correia, R. F. Marquies, P. Fonte, D. G. Diaz, A. Lindote, M. I. Lpes, M. P. Macedo, A. Policarpo, An rpc prototype with high spatial reolution, Nucl. Inst. And Meth. A 533 (2004) 139–143.

[5] F. Habte, A. M. K. Foudray, P. D. Olcott, C. S. Levin, Effects of system geometry and other physical factors on photon sensitivity of high-resolution positronemission tomography, Phys. Med. Bio. 52 (2007) 3753–3772.

[6] D. R. Schaart, H. T. V. Dam, S. Seifert, R. Vinke, P. Dendooven, H. L oner, F. J. Freek, J. Beekman, A novel, sipm-array-based, monolithic scintillator detector for pet, Journal of Phys. Med. Biol. 54 (2009) 3501–3512.

[7] P. O'Connor, G. D. Geronimo, A. Kandasamy, Amplitude and time measurement asic with analog derandomization: First results, IEEE Transactions on Nuclear Science 30 (4) (2003) 892–897.

[8] D. P. Siddons, A. Dragone, G. D. Geronimo, A. Kuczewski, J. Kuczewski, P. O'Connor, Z. Li, C. G. Ryan, G. Moorhead, R. Kirkham, P. Dunn, A high speed detector system for x-ray fluorescence microscopy, in: IEEE NSS Conf. Rec., 2006, pp. 725–728.