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Silicon photo-multipliers as photon detectors for PET

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ABSTRACT

A novel photon detector, the Silicon Photomultiplier (SiPM), has been tested in view of a photosensitive device for Positron Emission Tomography (PET), which is one of the most important non-invasive methods for in-depth and in vivo imaging of tissue. Such a device offers advantages over existing ones, including application in a magnetic field, better resolution and easier operation. Two PET modules have been constructed by coupling an array of Hamamatsu 3 mm × 3 mm Multi-pixel photon Counters (MPPCs) and Photonique 2.2 mm × 2.2 mm Solid-state Photomultipliers (SSPMs) to LYSO scintillation crystals. The results of studies of the two module prototypes are presented.

1. Introduction

Positron Emission Tomography (PET) is a non-invasive method for in-depth and in vivo imaging of tissue. In contrast to X-ray tomography (CT) and that with Nuclear Magnetic Resonance Imaging (MRI), which give information about tissue structure and pathological variations in the morphology, PET detects physiological, biochemical and pathological processes in the living organism. The positron emitted in a β^+−decay of the nucleus slows down in the tissue and subsequently annihilates with a nearby electron. The annihilation gamma-rays of 511 keV are usually detected indirectly, through scintillation in inorganic crystals. Photon detectors, like Photomultiplier Tubes (PMTs), detect the scintillation light. The majority of PET devices use PMTs, but due to their size, relatively poor ratio of active to total surface and high price, which is a significant fraction of the total cost of the device, it is worthwhile to search for alternative detectors of visible photons. The sensitivity of PMTs to magnetic fields, and the increasing requirement to unify different image modalities in one measurement, provide an additional reason to search for new detectors. One would like to incorporate a PET apparatus inside a MRI magnet for simultaneous imaging of tissue function and of density. Semiconductor sensors, photodiodes and avalanche photodiodes seem to be much closer to an ideal detector. In particular, a new type of semiconductor detector, the Silicon Photomultiplier (SiPM) looks very promising [1–3].

2. The apparatus

In this study two PET modules have been constructed by coupling a 4 × 4 array of LYSO scintillator crystals at a pitch of 4.5 mm to an array of SiPMs of the same pitch using optical grease. The scintillation array consisted of 4.3 × 4.3 mm^2 LYSO scintillation crystals of length of 20 mm encased in $BaSO_4$ reflector. In the first module the 2.2 mm × 2.2 mm Photonique 0607 SSPMs (packaged on 4 mm × 4 mm printed circuit board with two pins) and in the second the 3 mm × 3 mm Hamamatsu S10931-100P surface mount type MPPCs were used (see Table 1 and Fig. 1). The main advantages of the latter are higher efficiency and larger sensitive area. Such a PET module with SiPMs is much more compact than existing ones, offers lighter mechanical design, works in a magnetic field, has a better resolution and is easier to operate.

The printed circuit boards which house the SiPMs contain a single bias voltage connector and the outputs for all the channels (Fig. 1). Due to large variation of the working voltage of the delivered Photonique SSPMs (0.76 V RMS), the voltage divider circuit was integrated on the board to adjust the correct bias voltage (Fig. 2). The delivered Hamamatsu MPPCs have much smaller variation of the working voltages of 0.1 V RMS.

Fig. 3 shows the signals due to interaction of 511 keV photons in the LYSO scintillators. Due to smaller capacitance, the signals in the SiPMs by Photonique are on average 10 times smaller than the signals in Hamamatsu MPPCs. The pulse shapes from SiPMs by Photonique exhibit in addition to the scintillator decay time of about 50 ns also the second much slower one of the order of about 1 μs, which might be due to internal quenching resistance.

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Table 1
Parameters of the SiPMs used in the PET module.

Producer SiPM type	Photonique 0607	Hamamatsu S10931-100P
Package type	printed circuit board with pins	surface mount
Size (mm^2)	2.2 $\times$ 2.2	3 $\times$ 3
Operating voltage(V)	25	70
Gain	7.5 10^5	2.4 10^6
Terminal capacitance (pF)	320	320
Number of pixels	1700	900
Pixel size (μm)	40	100
Peak PDE	0.25	0.75

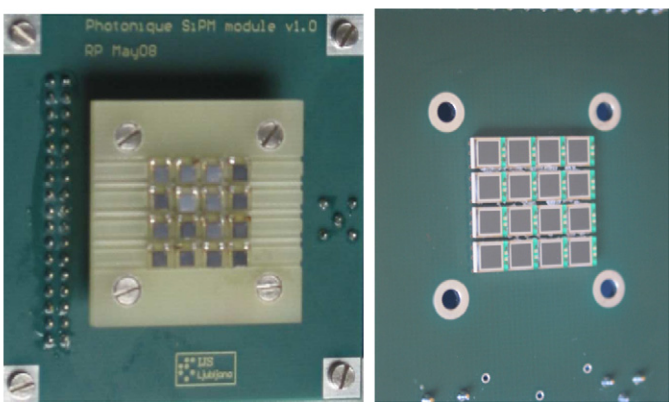

Fig. 1. 4×4 array of the Photonique 0607 SSPMs (left) and Hamamatsu S10931-100P MPPCs (right) arranged in a matrix with a pitch of 4.5 mm.

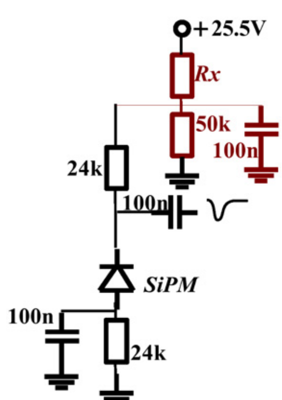

Fig. 2. An electronic scheme of the single SiPM channel in case of the Photonique SSPMs. The resistances R_x are adjusted individually to match the recommended working voltage.

3. Results

Due to large area and consequently higher noise rate, the single photon signals cannot be distinguished from the background at room temperature. We have therefore measured several 1 mm^2 devices obtained by the same producers.

The surface sensitivity is assessed by exposing smaller 1 mm^2 SiPMs of the same pixel size to a pulsed 5 μm wide laser beam of intensity at the single photon level. The SiPM is moved relative to the light source to produce two-dimensional scans of the count rate over the surface of the silicon photomultiplier. The measured sensitivity shows that it is fairly uniform across the SiPM surface. Scans over a smaller area reveal the inner structure of individual pixels (Fig. 4).

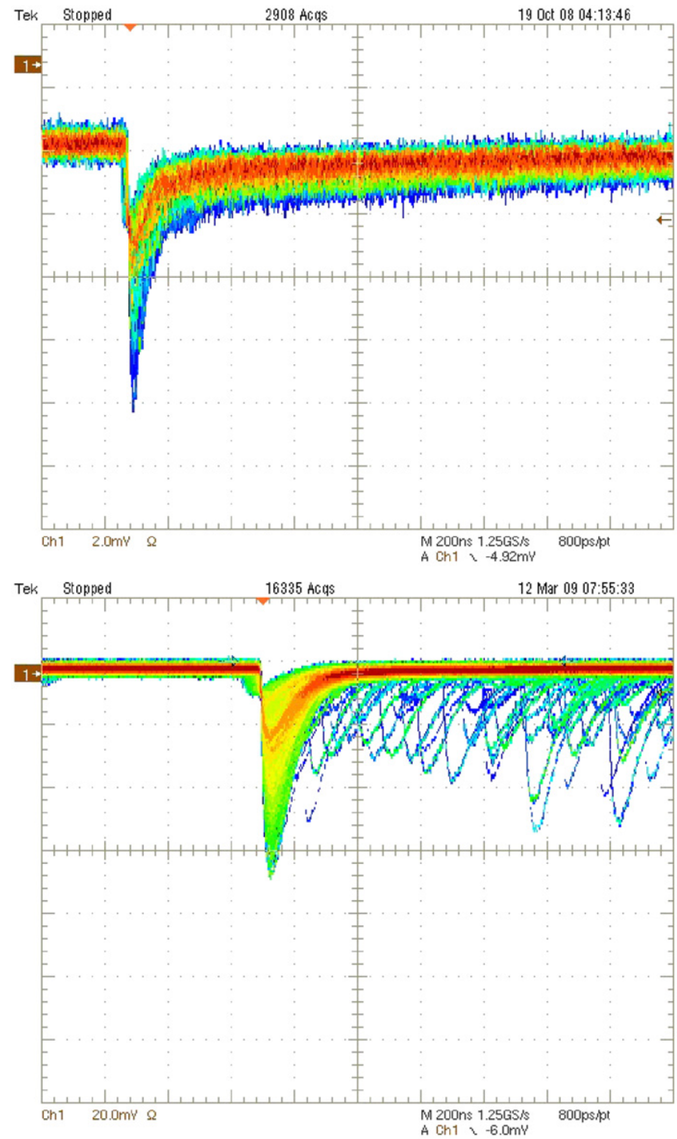

Fig. 3. The SiPM signals due to 511 keV photons: Photonique SSPM (top) and Hamamatsu MPPC (bottom).

In PET scanners, the time difference between two back-to-back photons yields information about the position of positron-electron annihilation in the patient's tissue. This information improves the image quality due to better background rejection. Therefore, the accurate determination of the position of an event depends on the time resolution.

Although not as fast as for example a micro channel plate PMT, the single photon time resolution of the measured SiPM's amounts to 100–200 ns. The measurements of the time resolution at different wavelengths (405 and 635 nm) show that all the SiPMs give a better time resolution in the blue light region (Table 2).

Unamplified signals were fed to the CAEN V965 Dual range QDC. The charge distribution of one Hamamatsu SiPM channel is shown in Fig. 5. Due to the finite number of pixels the energy response of this module is not linear. By using different radioactive sources, we obtained the energy calibration curve in Fig. 6. The measured dependence is well approximated by an exponential approach to a saturation value. Closer inspection reveals a 4% RMS variation of the position of the 511 keV photopeak among the 16 channels of the module. This is much

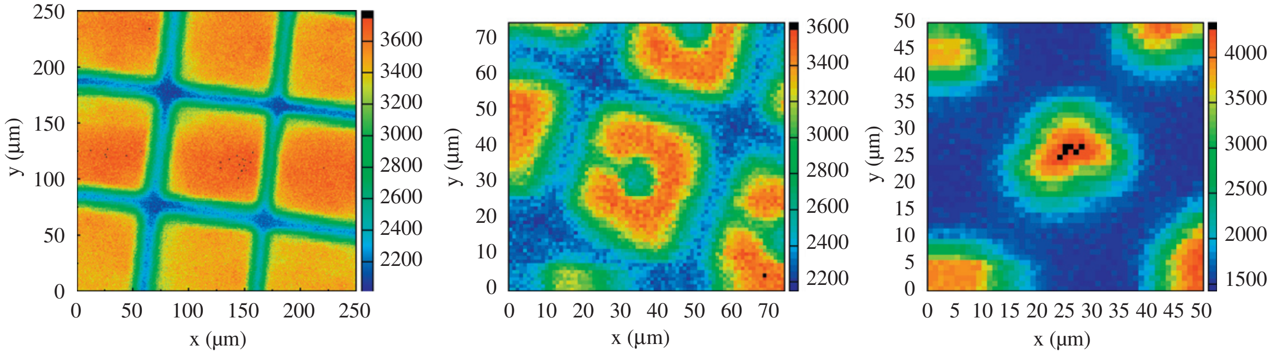

Fig. 4. Two-dimensional scans of the surface sensitivity: HC100 from Hamamatsu (left), S137 by Photonique/CPTA (middle) and E407 by MepHi (right).

Table 2
The time resolution after time-walk correction for different SiPMs.

SiPM Producer	E407 MepHi	S137 Photonique	H100C Hamamatsu
σ_{red}(ps)	127	182	145
σ_{blue}(ps)	97	151	136

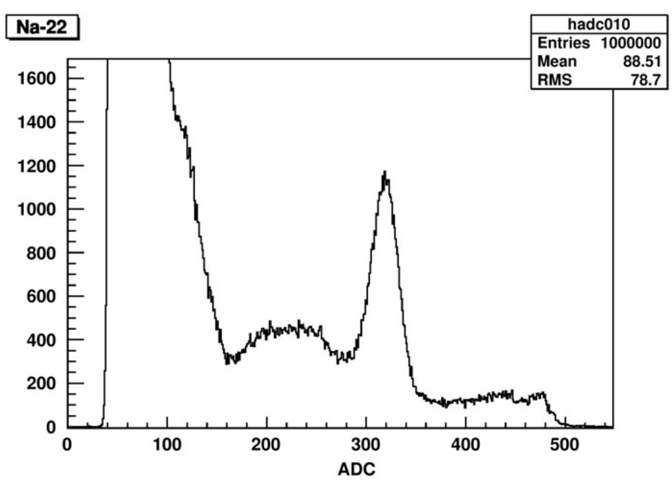

Fig. 5. The single channel response of the PET module from Hamamatsu MPPCs to a ^{22}Na source.

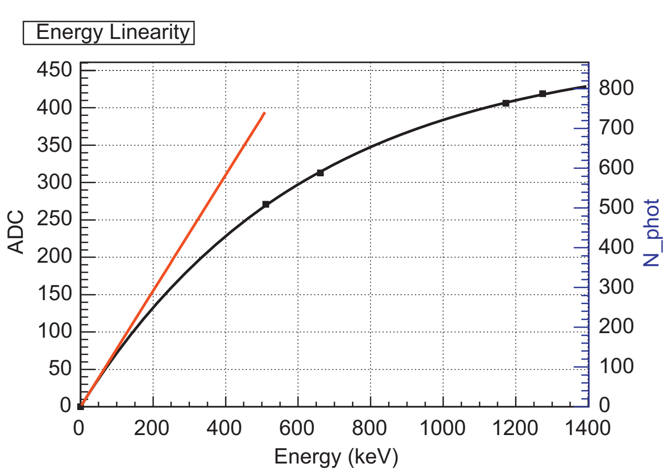

Fig. 6. Energy response of a particular channel of the PET module with Hamamatsu MPPCs.

less than the 16% RMS spread for the PET module of Photonique SSPMs.

Both PET modules were then tested by using the coincidence anihillation γ from an ^{22}Na source. The data acquisition was triggered by a sufficiently large signal from a single SiPM from the other module. The energy resolution of 10% FWHM was measured with Hamamatsu MPPCs. This was better than the best energy resolution of 21% FWHM measured with Photonique SSPMs. We mostly attribute this to the smaller photon detection efficiency, smaller area and also to the not very well controlled coupling of the SiPMs to the crystal matrix.

4. Conclusions

Silicon photomultipliers seem to be a very promising detector for PET. The main advantages are their insensitivity for magnetic fields and their compactness. We have constructed two PET modules using arrays of 4×4 SiPMs in two different packages (printed cicuit board with two pins—Photonique andsurface mount type—Hamamatsu) coupled to a LYSO matrix with pitch of 4.5 mm. The energy resolution of such modules has been investigated. The PET module from Hamamatsu MPPCs performed better than the Photonique one. This might be due to the coupling of the crystals to the SiPMs. We obtained good 10% FWHM energy resolution with little variation between the channels of the Hamamatsu MPPCs. However, due to the limited number of pixels, the response to gamma rays of different energy is not linear. We expect to obtain a better timing resolution for SiPMs with larger pixels. The good energy resolution and fast response might enable a reduction of back-ground in PET imaging and thus improve the image quality. Systematic studies of the module will be performed in the future and the possibility to use the module in the NMR-PET will be explored.

Acknowledgements

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References

[1] P. Buzhan, et al., An advanced study of Silicon Photomultiplier, ICFA Instrumentation Bulletin, 2001.
[2] B. Dolgoshein, et al., Nucl. Instr. and Meth. A 563 (2006) 368.
[3] D. Renker, Nucl. Instr. and Meth. A 567 (2006) 48.