Initial results on Sipm array based on a symmetric resistive voltage division readout

S. David, M. Georgiou, E. Fysikopoulos, N. Efthimiou, T. Paipais, L. Kefalidis and G. Loudos

Abstract—The aim of this study is to design and apply a symmetric resistive voltage division matrix to read out the compact silicon photomultiplier array (SPMArry2 by SensL) photodetector for possible applications as gamma or PET probes for localization of cancer tumors. We applied a symmetric resistive voltage division circuit, reducing the 16 voltage outputs, which are provided from SensL’s evaluation board, to 4 position signals. We acquired raw images and crystal maps from various pixilated scintillators under $^{99m}$Tc and $^{22}$Na excitation, emitting mainly at 140keV and 511keV, respectively. The exact values of the resistors as well as the summing amplifiers used in the summation and division stages are given. Moreover, experimental evaluation in terms of energy and spatial resolution is reported. A clear visualization of the discrete 2mm$^2$ pixilated CsI:Tl scintillator elements under 140keV excitation - with better than 23% energy resolution - was achieved. Raw images of the crystals maps acquired shown visualization of the discrete 1mm$^2$ CsI:Na scintillator elements under 511keV irradiation. Under 511keV excitation we achieve an energy resolution of ~35% for BGO and 18% for CsI:Na pixilated scintillators, respectively. All measurements were carried out at room temperature (~25°C), without additional cooling.

I. INTRODUCTION

Silicon Photomultipliers (SiPMs), also known as multi-pixel photon counters (MPPCs) or solid-state photomultipliers (SSPMs), represent an alternative solution that to a large extent combines the advantages of PMTs and APDs [1]. They have high gain (equal to that of PMTs) and are operated at low bias voltages (<80V) [2]. They are relatively insensitive to magnetic fields and thus are good candidates for MR compatible applications such as PET/MRI [3]. The main drawback of SiPMs is that they require cooling circuits due to their high sensitivity to temperature variations, especially in large SiPM blocks [4-5]. Moreover, non-proportional response at high light flux - for example in applications that detect high energy photons using bright scintillators - has been reported due to the saturation of limited fired cells. Nevertheless, nowadays the SiPM’s can have more than 3000 cells per pixel and can provide almost linear response at most of the nuclear imaging applications. Several studies have been reported, where no cooling circuit is required to achieve adequate signal-to-noise levels in imaging applications, at least for small systems [6]. Excellent timing properties of SiPMs make them also promising devices for time-of-flight PET imaging applications [7-8]. SiPM discrete arrays are very flexible and have the potential to be a suitable photodetectors for gamma or PET probes, used for localization of cancer tumors. Astonishing progress has been made over the past 5 years, with design and performance improvements in arrays produced by SiPMs manufacturers.

Although continuous evolution of SiPMs provides new modules with improved characteristics, at the moment most studies validate single detector modules; few studies have explored the construction of a detector module designed for dedicated applications. In this study, we report results of the SPMArry2 (4x4 element array of 3x3mm$^2$ SiPMs) optical detector coupled to various pixilated scintillators for possible application as Gamma and PET nuclear probe. Evaluation was carried out with $^{99m}$Tc and $^{22}$Na isotopes, emitting mainly at 140 KeV and 511KeV, respectively. A symmetric resistive voltage division matrix was applied which reduces 16 outputs - provided from SensL’s evaluation board - to 4 position signals. Results from the experimental evaluation in terms of energy and spatial resolution are reported. All the measurements were applied at room temperature (25°C), without additional cooling circuit.

II. MATERIALS AND METHODS

SensL’s scalable silicon photomultiplier array (SPMArry2) is a commercially available, solid-state, large array detector based on silicon photomultiplier technology [9]. It consists of 16 pixel elements covering an active area of 13.6 mm$^2$. Each pixel has 3640 number of microcells connected in parallel, with individual cell dimensions equal to 35 μm. The pixel array is over-molded with epoxy to completely encapsulate the
pixels, bondwires and substrate bondpads. This optical detector array offers 3 side tileable packaging to allow the SPMArray2 to be tiler for large area detection systems.

The SensL SPMArray2-A0 preamplification electronics and a SPMArray2-A1 evaluation board which provides the pixels voltage output were used in this study. The preamplification board mainly consists of 16 differentials fast amplifiers (AD8132). The evaluation board provides 16 individual pixels voltage outputs, plus one more output which is the sum signal. More information about these modules can be found at manufacturer’s website [9]. The SPMArray2 detector works properly at a low bias voltage of -30V. The breakdown voltage that initiates the avalanche is at -29V. We supply the SPMArray’s bias voltage directly through the external power supply bypassing the bias regulation circuit as shown in the diagram, so that we can adjust it if necessary. All measurements presented in this study were contacted with -29.7V bias voltage.

The characteristics of the pixilated scintillator arrays, the dimensions of crystal’s array pitch, as well as the quantum detection efficiency and the spectral matching factors between the scintillation light and the photon detection efficiency of the SPMArray2 used in this study are reported in Table 1. BGO and CsI scintillators (doped with Tl and Na activators) are currently used in nuclear medicine and dedicated imagers and were tested.

**Table I. Scintillator Detector Parameters**

<table>
<thead>
<tr>
<th>Detector characteristics</th>
<th>3x3x5 mm³</th>
<th>2x2x5 mm³</th>
<th>1.5x1.5x5 mm³</th>
<th>1x1x5 mm³</th>
<th>2x2x5 mm³</th>
<th>2x2x5 mm³</th>
</tr>
</thead>
<tbody>
<tr>
<td>CsI:TI</td>
<td>0.22</td>
<td>0.22</td>
<td>0.25</td>
<td>0.2</td>
<td>0.3</td>
<td></td>
</tr>
<tr>
<td>CsI:TI</td>
<td>54000</td>
<td>54000</td>
<td>41000</td>
<td>41000</td>
<td>8300</td>
<td></td>
</tr>
<tr>
<td>CsI:Na</td>
<td>550</td>
<td>550</td>
<td>420</td>
<td>420</td>
<td>480</td>
<td></td>
</tr>
<tr>
<td>CsI:Na</td>
<td>1000</td>
<td>1000</td>
<td>630</td>
<td>630</td>
<td>300</td>
<td></td>
</tr>
<tr>
<td>BGO</td>
<td>85.7</td>
<td>68.9</td>
<td>90.3</td>
<td>90.3</td>
<td>99.8</td>
<td></td>
</tr>
<tr>
<td>Light yield [photons/MeV]</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Wavelength of emission max. [nm]</td>
<td>1000</td>
<td>1000</td>
<td>630</td>
<td>630</td>
<td>300</td>
<td></td>
</tr>
<tr>
<td>Decay time (ns)</td>
<td>12.2</td>
<td>9.6</td>
<td>22.9</td>
<td>22.9</td>
<td>38.2</td>
<td></td>
</tr>
<tr>
<td>QDE @ 140 keV (%)</td>
<td>0.86</td>
<td>0.86</td>
<td>0.56</td>
<td>0.56</td>
<td>0.80</td>
<td></td>
</tr>
<tr>
<td>QDE @ 511 keV (%)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Matching Factor %</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

We evaluated CsI scintillator arrays of various pixel size elements under $^{99m}$Tc excitation for possible applications in gamma probes. The detection efficiency of those scintillators is very high as computed with the quantum detection efficiency [10]. For PET applications we evaluated the BGO and CsI:Na scintillator arrays with 5mm thickness.

The quantum detection efficiency (QDE) reported in Table I was calculated according to the exponential law.

$$QDE = 1 - e^{-\mu_x x}$$

Where $\mu_x$ is the linear attenuation coefficients of the examined scintillator materials and $x$ is the coating thicknesses of the scintillator mass [10]. The spectral matching factor reported in the last row of the table indicates the degree of matching of the produced scintillation emission light by the scintillators and the spectral sensitivity of SPMArray2 optical detector. The spectral matching factor $\alpha_s$ was calculated by the ratio:

$$\alpha_s = \frac{\int S_P(\lambda)S_D(\lambda) d\lambda}{\int S_P(\lambda) d\lambda}$$

Where $S_P$ is the spectrum of the light emitted by the scintillator, $S_D$ is the spectral sensitivity of the SPMArray2 and $\lambda$ denotes the wavelength of the emitted light [11].

In Fig.1 the custom circuit that we implemented in order to reduce the 16 anode signals to 4X and 4Y coordinate signals is shown. Each row of the four individual pixels is connected via 1k resistors (for the X axis) to a spectroscopic inverter, summing amplifier (the AD 8032) as illustrated. In the same way the reduction of the signals was carried out on the Y axis resulting to the 8 position signals.

![Fig. 1. A custom resistive voltage divider network reduced the number of 16 pad signals to 8 signals (4X and 4Y). The network applied for X direction is shown.](image)

Each of the 4 by 4 coordination signals are further reduced to 2X and 2Y signals as shown in the Fig.2. Each signal is connected via two weight resistors to 2 inverter summing amplifiers (AD8092). The exact position of the 4 signals
depends on the resistor value of the division voltage chain. This idea was firstly introduced by Popov et al. and was applied in the position sensitive photomultiplier detectors and recently in large SiPM arrays [12-13].

Data acquisition of the 4 position signals arriving at the resistive divider ends \((X_A, X_B, Y_A, Y_B)\) was performed using a free running sampling technique [14]. Four 12-bit high speed ADCs with 10MspS sampling rate were used for the digitization (ADS5282, Texas instruments) [15]. The digital output of each ADC was continuously stored in a 16x12 shift register. FPGA was used for triggering and signal processing of the pulses (Xilinx Spartan-6). A trigger signal is produced when the sum of the four incoming samples exceeds a given digital threshold. The trigger signal was delayed by 13 clock cycles to make sure that complete pulses are stored [14]. Pulse integration was performed in order to minimize data storage to an external memory. 14 samples per pulse for integration were found to be a good compromise between pulse information of interest and FPGA resources. Ethernet was used to send data to a standard laptop every 1000 events captured in memory.

The centroid position \((X, Y)\) of the incident light pulse distribution is finally calculated using Anger’s equations (3).

\[
X = \frac{X_A - X_B}{X_A + X_B}, \quad Y = \frac{Y_A - Y_B}{Y_A + Y_B}
\]

III. RESULTS AND DISCUSSIONS

The scintillation crystal emission responses and the spectral sensitivity of the silicon photomultiplier that have been used in this study are shown in Fig.3. We observe very good matching between the CsI doped with Thallium and BGO scintillators, as we reported in Table I.

Fig. 4 presents the raw images acquired with the discrete 3mm x 3mm x 5mm pixelated CsI:Tl scintillator array under \(^{99m}\text{Tc}\) excitation. The upper image was obtained by coupling the scintillator plate to the SiPM detector entrance window only with optical grease (BC 630), while the second by using 1mm glass and optical grease between all surfaces. The horizontal profiles of the images indicate a clear visualization of the discrete scintillator elements. The diagrams on the right show the histogram of the energy spectra as calculated from one central pixel, drawn on the raw images. Energy resolution of 20%-21% was measured. A drop in sensitivity in the image acquired with the glass coupling is observed and shifts the photopeak centroid for about 200 channels, from 1480 to 1280 channel.

Fig. 5 presents the raw images obtained (under \(^{99m}\text{Tc}\) excitation) with the discrete 2mm x 2mm x 3mm pixelated CsI:Tl scintillator array are illustrated in Fig.5. The raw image shows a clear visualization of the discrete 2mm x 2mm scintillator elements.
The upper raw image was acquired by using only optical grease coupling, while the bottom with an additional 1 mm glass. The energy resolution achieved with this scintillator (calculated on one central scintillator element) was 23%.

The energy resolution across the entire field of view (FoV) is almost stable as shown in Table II. The CsI:Tl with 3mm x3mm pixel elements shows slightly better mean energy resolution (~21.5%) than the discrete 2mm (~23%), mainly due to the its larger thickness (5mm versus 3mm).

<table>
<thead>
<tr>
<th>Energy resolution at 140KeV</th>
<th>Center of FoV</th>
<th>Edges</th>
<th>Corners</th>
</tr>
</thead>
<tbody>
<tr>
<td>3x3x5mm³ CsI:Tl coupled with grease</td>
<td>21%</td>
<td>22%</td>
<td>23%</td>
</tr>
<tr>
<td>3x3x5mm³ CsI:Tl coupled with 1mm glass</td>
<td>22%</td>
<td>24%</td>
<td>23.5%</td>
</tr>
<tr>
<td>2x2x3mm³ CsI:Tl coupled with grease</td>
<td>22%</td>
<td>24%</td>
<td>23%</td>
</tr>
<tr>
<td>2x2x3mm³ CsI:Tl coupled with 1mm glass</td>
<td>22.5%</td>
<td>23%</td>
<td>23.3%</td>
</tr>
</tbody>
</table>

Following, we imaged a scintillator with smaller pixel elements, in order to find the limits of our resistive circuit under ⁹⁹ᵐTc excitation. The pixilated scintillator that we used was a CsI:Na with pixel size of 1.5mm x 1.5mm x 5mm. Under ⁹⁹ᵐTc excitation no visualization of the 7x7 crystal elements was achieved as shown in Fig.6.

The CsI:Na scintillator has lower light yield than CsI:Tl and its scintillation light does not match well with spectral sensitivity of SPMArray2. In order to have an index of the spatial resolution with this scintillator we constructed a lead mask with a hole of 1mm in diameter and we placed inside a thin plastic tube filled with ⁹⁹ᵐTc as shown in Fig. 7. We irradiated a single crystal, as shown and we took a vertical and a horizontal profile on the raw image.

The FWHM produced from the horizontal profile of the raw image - shown in Fig.8, was found equal to 1.7mm while the vertical 1.6mm.

The FWHM produced from the horizontal profile of the raw image produced by⁹⁹ᵐTc irradiating one discrete scintillator element of the CsI:Na array. The FWHM produced from the horizontal profile was measured equal to 18pixels and the pixel size was equal to 95µm.
A row image produced under $^{22}$Na irradiation, for the 1.5mm x 1.5mm pixelated CsI:Na scintillator shows a clear visualization of all crystal elements as shown in Fig.9. The mean energy resolution achieved by CsI:Na scintillator was equal to 18% at 511keV.

The raw image produced for the BGO scintillator plate with dimensions of 2mm x 2mm x 5mm under 511keV excitation is illustrated in Fig.10.

The energy resolution achieved with BGO scintillator was found equal to 35% for $^{22}$Na excitation, while for $^{137}$Cs (emitting at 662keV) it was 28%. SensL has reported an energy resolution of ~20% for BGO scintillator. In this set up a BGO with crystal thickness of 15mm was used; moreover, energy resolution was measured directly from one pixel element of the array and not from the final image.

Finally, Fig. 11 shows a raw image of the 1mm x 1mm x 5mm CsI:Na pixilated array acquired under 511keV excitation. The scintillator array was optically coupled using optical grease to the SPMArray2 entrance window. The mean energy resolution achieved by the discrete 1mm x 1mm CsI:Na pixilated scintillator was found to be 18%.

IV. CONCLUSIONS

In the present study we have investigated the application of a symmetric resistive voltage division readout for a flexible SensL’s silicon photomultiplier array (SPMArray2) for possible applications in gamma and PET probes. The analog division resistive circuit that we applied allows the visualization of 2mm$^2$ pixellated CsI:TI scintillators under 140keV excitation, achieving better than 23% energy resolution. Moreover, a clear visualization of individual crystals even for 1x1mm$^2$ scintillator elements of CsI:Na scintillators under 511keV excitation was achieved. At this energy we achieved an energy resolution of ~35% for BGO and 18% for CsI:Na scintillators respectively.

At 511keV scintillators brighter than BGO, such as LSO:Ce or LYSO:Ce, are expected to provide improved performance. Some key ideas about the future work are firstly the assessment of the performance with 16 and 8 channel readout outputs, in order to have comparative results. Secondly, the application of the symmetric resistive division matrix directly on the current outputs of the SiPMs, without the evaluation board, and the application of the readout circuit to other SiPMs already available in lab. Finally, experimental evaluation of that circuit with brighter scintillator arrays such as LYSO:Ce or LSO:Ce for possible PET applications will be carried out.

REFERENCES


