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MASTER THESIS WORK

TEST BENCH OF SILICON PHOTOMULTIPLIER FOR POSITRON EMISSION TOMOGRAPHY APPLICATION

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Test bench of Silicon Photomultiplier for Positron Emission Tomography Application

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Abstract. Typically the medical imaging devices have been based on scintillator crystals coupled to photomultipliers tubes. But the problems to combine them with high electromagnetic fields and the relatively high cost per unit surface, opens new opportunities on the field for a different type of detector named silicon photomultiplier. These ones offer an alternative solution to combine the high gain of the photomultipliers tubes, and the insensitiveness to the magnetic field, high quantum efficiency and compact structure of the avalanche photodiodes. That allows an increasing quality on the medical imaging technics, such as positron emission tomography, which implies a better and early detection of different diseases. In this study a very promising results for coincidence time resolution and single photon time resolution coming from the silicon photomultiplier combined with a readout electronics developed by SiUB group are shown, where we are trying to reach the limit of the technology.

Keywords: Silicon photomultiplier, positron emission tomography, time resolution, Geiger mode, scintillation crystal.

1. Introduction

There is a lot of research effort on improving the time resolution of the medical imaging equipment that is relate to the spatial resolution, it means the minimum part of the organ under study that could be resolved. The positron emission tomography (PET) is a molecular diagnosis technique that allows an *in vivo* tracing of different biological processes, whose purpose is to show the cellular or molecular activity, not the anatomy of the patient. The process implies the administration of a radio-tracer, namely a biomolecule labelled with a radioactive atom emitting positrons (β^+). The biomolecule is chosen such that it will preferentially accumulate in the area of interest resulting in high radio-tracer concentration in this region. During PET imaging, the positron emitted by the radio-tracer undergoes annihilation with an electron and this process produces two 511 keV gamma-rays that travel approximately 180 ° from each other. Near-simultaneous detection of these two 511 keV gamma-rays with detectors placed opposite each other constitutes the signal generation process for PET [1]. It is also interesting the combination of the molecular tracing and the structural anatomy, by means of a hybrid system that combines computed tomography (CT) or magnetic resonance imaging (MRI) with PET, all on the same device and performing the two scans simultaneously. With the PET-CT combination it is possible to have the physiological uptakes and also the attenuation factor and the pathological uptakes provided by the CT. On the other hand, the problem to combine MRI with PET is that for the first system an electromagnetic (EM) field with a high intensity is created. This implies that it is complicated to merge with PET on the same room because this second system uses a ring of photomultipliers tubes working on coincidence mode and the electron cascade principle, which the PMTs are based on, does not work under magnetic fields and they should be located on a separated room with the proper EM shield. Moreover, the light coming from the scintillator

crystal needs to travels through an optical fiber to reach the PMT. This system involves a very complex and precision technology that makes it very expensive and limited by many those factors.

Due to this fact, nowadays the research groups are evaluating and implementing a different detector used for PET called with several names such as silicon photomultipliers (SiPM) or multipixel photon counter (MPPC). They are based on solid state technology and similar to the avalanche photodiodes (APDs), they work under Geiger mode (GM), but in contrast to these lasts ones, SiPM gives information on light intensity. The structure is a photodiode with many micro-cells connected in parallel on a common silicon substrate. Each micro-cell is represented by a p/n junction working in GM and connected in series with its integrated passive quenching resistance [2]. Thanks to that, it is insensitiveness to the EM fields and the integration of a PET-MRI could be done on the same device and room, coupling directly the scintillator crystal to the SiPM and processing the signal simultaneously with the MRI scan.

In the following sections of this work we can find the basic concepts in order to understand how the device works, including basic notions for the readout electronics, and then the results obtained performing the corresponding setup that allows extracting the data of interest, such as coincidence time resolution or single photon time resolution. Always aiming for the best values in order to reach the limit of the technology and compete with the current ASICs used for positron emission tomography [3-5].

2. How it works (Theory)

In this section we will see the basic concepts that explain how the SiPM work as well as the electronics that process their signal and also the scintillator crystal.

2.1. Silicon Photomultiplier (SiPM)

The Silicon Photomultiplier (SiPM) is a multipixel semiconductor photodiode, where each pixel joint together on common silicon substrate on common load. The typical size of the pixel fluctuates depending on the manufacturer and the final specifications from 20-100μm. Every pixel operates in Geiger mode, under bias voltage (V_{bias}) of 5-15% over the breakdown voltage ($V_{breakdown}$), so each carrier generated by photons or thermally gives rise to a Geiger-type discharge. This Geiger discharge is stopped when the voltage goes below the breakdown value due to an external polysilicon resistor (R_{pixel}) that decouples electrically each pixel; the typical values of these resistors are (100-500)$k\Omega$. The single pixel gain is determined by the charge accumulated in the pixel capacity (C_{pixel}); typical values of this capacity are (90-150)fF [6]. With these two parameters it is possible to obtain the discharge time (τ) as can be seen in equation (2.2.1):

$$\tau < C_{pixel} \cdot R_{pixel} \sim 10^{-8} s \qquad (2.2.1)$$

The charge of the pixel (Q_{pixel}) can be also calculated from the equation (2.2.2), knowing that the V_{bias} is only few volts above the $V_{breakdown}$, and therefore the gain of the single cell (G_{pixel}) can be also calculated:

$$Q_{pixel} = C_{pixel} \cdot (V_{bias} - V_{breakdown}) \sim 100 fC \qquad (2.2.2)$$

$$G_{pixel} = \frac{Q_{pixel}}{e} \sim 10^{6} \qquad (2.2.3)$$

It can be seen that the result for the gain from the equation (2.2.3) is on the same order as the vacuum PMT gain.

The total number of pixels is between 100-4000mm^{-1}. Because all SiPM micro-cells work together in parallel on a common load, the output signal is the sum of the signals from all fired pixels. So such a device, where each pixel operates digitally as a binary element, works as an analogue device and it can measure the light intensity.

The topology of a single pixel is similar to the APD structure. A few micron epitaxy layer on low resistive p substrate forms the drift region with low built-in electric field as can be seen in figure 1. The thin depletion region between the p^+ and n^+ layers with very high electric field, up to 10^6V/cm is created, where the conditions for Geiger mode discharge take place ($V_{bias} > V_{breakdown}$). The electrical decoupling between the adjacent pixels is provided by polysilicon resistive strips and uniformity of the electric field within a pixel by the n^- guard rings around each pixel. All pixels are connected by common aluminium (Al) strips, in order to readout the SiPM signal.

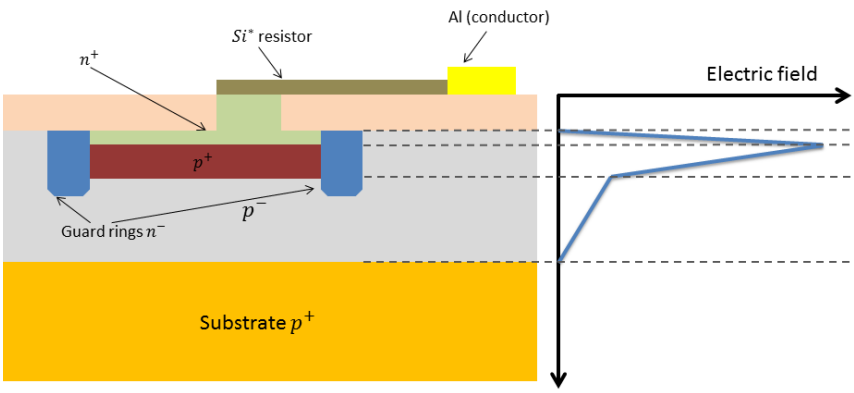


Figure 1: Schematic representation of a single pixel structure and the corresponding electric field shape.

The physical mechanism upon which avalanche gain depends is the impact ionization. It occurs when the electric field in the depletion region is strong enough: an electron colliding with a bound valence electron transfers enough energy to the electron to ionize it. This creates an additional electron-hole pair producing current gain. The additional carriers, in turn, can gain sufficient energy from the electric field to cause further impact ionization, creating an avalanche of carriers [7].

Another important parameter of the SiPM is the photon detection efficiency (PDE) that is defined as the ratio between the number of output pulses of the device exceeding its dark count rate and the number of photons impinging on the detector surface. Therefore, in order to be detected by a SiPM, a photon must be focalized in the active area of the detector, it should generate a primary carrier (more precisely an electron hole pair) and the primary carrier should succeed in triggering an avalanche. Consequently, the PDE of a SiPM depends on three parameters: the quantum efficiency QE, the triggering probability and the geometrical efficiency (GE).

2.2. Scintillator Crystal

Scintillation spectrometers consisting of inorganic scintillation crystals coupled to appropriate optical detectors are one of the most widely used classes of detectors for gamma-ray spectroscopy. An incoming gamma-ray when stopped inside a scintillation crystal deposits its energy within the crystal, and an optical pulse (photons on the visible range) is produced by the scintillator, the amplitude or number of photons produced is proportional to the energy deposited in the scintillation crystal [8]. This optical pulse emitted by the scintillator can then be detected by optical detectors such as photomultiplier tubes or silicon photodiodes, in this case a SiPM, to provide an electronic charge pulse that can be processed with a pulse processing electronics, which will be explained on the next chapter.

There are a few requirements than a scintillator crystal must fulfil. It has to have a high energy resolution because it allows rejection of scattered events. High time resolution is required for rejection of random events. The material has to have a high mass density in order to stop the gamma photon on the minimum space length. Moreover, the time response is important to get time-of-flight (TOF) information and obtain a better localization of the depth-of-interaction

(DOI), it means the position where the photon impacts inside the crystal. Such localization in TOF-PET can lead to enhanced signal-to-noise ratio in the reconstructed image [9]. Furthermore, a short decay time is desirable to process each pulse individually at high counting rates, as well as to reduce the number of random coincidence events. On the following table 1 the main scintillator crystals used nowadays and some from the past can be seen.

Table 1. Properties of Inorganic Scintillator Crystals

Material	Light Yeld [Photons/MeV]	Wavelength Emission Peak [nm]	Radiation Length (511keV) [cm]	Luminescence Decay Time [ns]
LSO	24000	420	1.2	40
BGO	8200	505	1.1	300
GSO	7600	430	1.5	60
LFS	38000	425	1.15	35
LuAP	11300	365	1.05	18
LPS	26300	385	1.38	38
$LaBr_3$:Ce	61000	380	2.1	32
NaI(Tl)	38000	415	3.3	230
BaF_2	2000^a, 10000^b	220^a, 310^b	2.3	0.6^a, 620^b

a-fast component, b-slow component

The table 1 shows a summary of the most used scintillator crystals for PET, and the ones that are after the double green line are the ones that were used on the past. BGO, GSO and LSO are commonly used in PET instrumentation due to their high gamma ray stopping efficiency. BGO, however, has low light output and relatively slow response. The light output of GSO is also low, though its response is faster than BGO. LSO is brighter and faster than BGO and GSO, which has made it one of the dominant scintillators for PET. However, LSO also has some drawbacks. Energy resolution of LSO is variable and is limited by its nonproportionality. NaI(Tl) has been used in PET designs in past in view of its brighter response, lower cost and better energy resolution (compared to LSO, GSO and BGO). However, its slow response and low gamma-ray stopping efficiency limit its performance in 3-D PET imaging. BaF2 has been considered for PET in past due to its very fast component (0.6 ns decay time). However, the amount of light covered by the fast component of BaF2 is only about 2000 photons/MeV [10]. The gamma-ray stopping efficiency of BaF2 is also not very high. Nowadays most of the SiPM are mounting LFS because its equilibrium between these five parameters, a high light yeld, a relatively short radiation length and a very good luminescence decay time.

2.3. Signal processing board

The readout of the signal coming from the SiPM is typically done by an Analog to Digital Converter (ADC) implemented in an ASIC design. If the system has a moderated resolution requirement (as in PET systems), the ADC circuit can be replaced by a Time over Threshold (ToT) in combine with a Time to Digital Converter (TDC). With this implementation the power consumption decreases in comparison with the ADC, and it becomes more robust and simple. The SiUB group has developed an ASIC named FlexToT that uses this ToT+TDC combination in order to read the signal coming from the SiPM, and the basic operation concept will be explained as follows.

The FlexToT (Flexible Time Over Threshold) ASIC has been designed using AMS SiGe BiCMOS 0:35mm technology. It has 16 channels and each one is composed by an input current stage where the input current signal coming from an MPPC or SiPM is copied in three stages. For energy and pile-up stages the current is also scaled. The copied signals are transformed for time, energy and pile-up measurements into binary signals compatible with digital electronics such as FPGA. The time stage is composed by a fast comparator which outputs the width of the

input signal for a selected threshold. The energy stage contains the Time over Threshold (ToT) circuit composed by an integrator and a Schmitt trigger. The typical ToT design, composed by a charge amplifier, a shaper and a discriminator, has been replaced to obtain a linear Pulse Width Modulated (PWM) energy signal. An external Time to Digital Converter (TDC) implemented on an FPGA digitizes the energy output signals from the ASICs. The pile-up stage is composed by a differentiator and a fast comparator to discriminate events with more than one pulse [11]. Figure 1 shows the block diagram of the FlexToT ASIC.

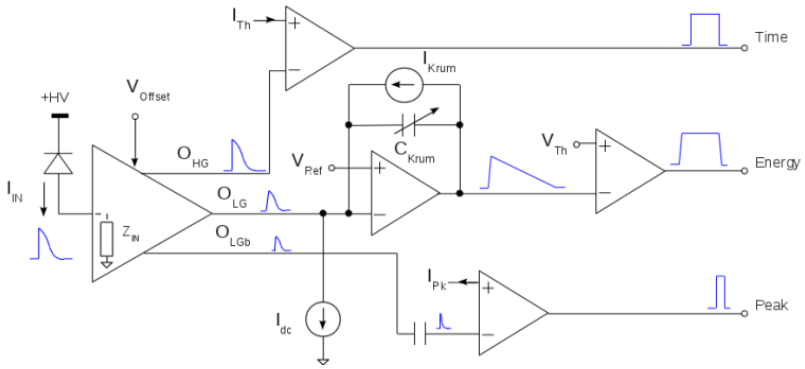
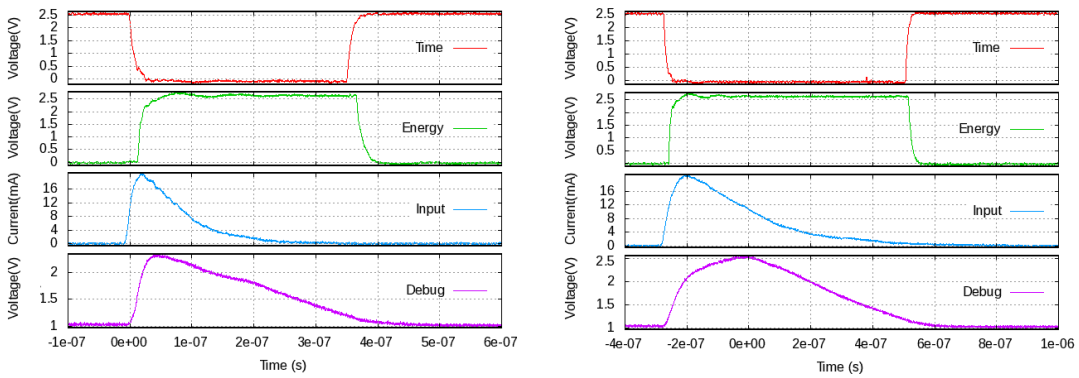


Figure 2: Simplified block diagram of the ASIC FlexToT designed by SiUB.

Flexibility is one of the main characteristics of the design. The ToT circuit implementation presents the possibility of modifying a set of parameters to adapt to different scintillator crystals with different time constants or different sensors with a wide range of overvoltages. Figure 2 illustrates analog signal processing with the FlexToT ASIC, for two different input current signals with different time constant.



(a) **(b)**

Figure 3: Typical signals coming from the SiPM taken by the oscilloscope Agilent MOS-X 3034A

The figure 3a shows the performance for a input signal with a τ = 36ns, and the time constant of the figure 3b is τ = 110ns. The figures in the first and the second row show the output signals in the time and energy stage. The 3rd row shows the input current signal and the 4th the signal obtained with integration in the energy stage. It can be seen that the width for both time and energy signals and the debug shape are quite different as a result of a variation on the decay time of the input signal.

3. Experimental Setups and Results

In order to evaluate the FlexToT ASIC and obtain the results to compare with the state of the art, three measurements have been done on different setup configurations. The following sections explain the results obtained for these three main important parameters, where the first one is related to an intrinsic parameter of the SiPM and the next two parameters are partially limited by the quality of the ASIC; it means that they will show the position of our ASIC with respect to the existing ones.

3.1. Dark Count Rate

The dark count rate (DCR) of a SiPM represents the number of output pulses per second while the device is under dark coinditions. This parameter is the intrinsic noise limit of the device, in particular when it has to be used for single photon detection. As for the case of GM-APD, the main phenomena contributing to the dark pulses are:

- Thermal generated carriers, enhanced by trap-assisted tunneling (TAT) is explained by Shockley-Read-Hall (SRH) generation-recombination theory [12], it shows that in a semiconductor at thermal equilibrium, the lattice vibrations can break some bounds between neighbouring atoms. As a consequence of breaking these bounds, electron-hole pairs can be generated. In terms of energy bands, this process can be described by the transition of a valence electron to the conduction band leaving a hole in the valence band and it is called carrier generation. The dominant generation process in silicon is an indirect transition via localized energy states in the forbidden gap (i.e. the silicon is an indirect semiconductor). These intermediate-level states called generation centers are due to the presence of impurities and crystal defects. Since these generation centers can also absorb carriers, they act as recombination centers. When the SiPM is operating under GM, these electron-hole pair will be collected and therefore it will generate a signal that in this case corresponds to a thermal generated effect, it means noise contribution.

- Band-to-band tunnelling tells that when a high electric field is applied on a p-n junction, it stimulates the ionization of the deep-level defects and determines the emission and capture processes. The effects of the tunnelling phenomena are to increase the emissivity of the generation-recombination centers and they have a strong dependence on the electric field profile (usually they are active at electric fields exceeding $10^6 V/cm$). When the bias voltage over the diode exceeds the breakdown voltage and the diode reaches the Geiger mode condition (i.e. the electrical field in the depletion region is high enough that electron and holes are created by impact ionization), the ionization rates of electrons and holes are so high that the electrons and holes are multiplying very fast, with a multiplication factor approaching infinity. Using quenching mechanisms, the multiplication factor is usually limited to a value of the order of 10^6 charges in one avalanch. The number of avalanches per second generated by carriers traversing the depleted region when the device is kept in dark conditions is called dark count rate (DCR) and it represents an important characteristic of a GM-APD device since it determines the intrinsic noise of the device.

For the DCR measurement the SiPM is coupled to one of the electronic FlexToT boards. The system is placed in a light tight box (black box) to prevent any background contamination coming from the ambient light. The board output is connected to an oscilloscope Agilent MSO 9404A that records the number of pulses in a time gate of $100\mu s$. The acquisition is triggered by an external square signal and the measure is repeated over a discriminator time threshold sweep as can be seen in figure 4.

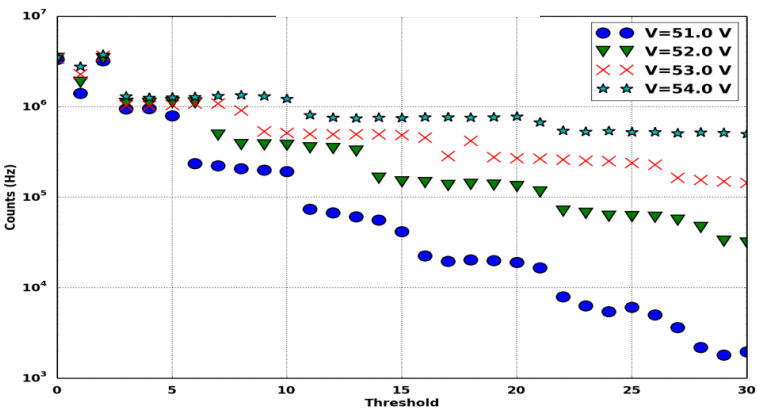


Figure 4: Dark count rate of a Hamamatsu S12572-50C model operated at different overvoltages as a function of the discriminator time threshold, threshold 0 correspond to the lowest value (63 in absolute terms).

The characteristic step function is observed and the plateau indicates the number of photo-electrons (pe) that can be triggered. The first plateau corresponds to 1pe, the second one to 2pe and so on. Depending on the overvoltage applied on the SiPM, the length (number of threshold) of the plateau varies. More overvoltage, the bigger the plateau is, but also the DCR that means an increment of the noise. The DCR that is obtained on these measures is about 1 MHz, and it drops to 0.5MHz when the second plateau is reached. This value is very dependence on the temperature and the DCR can take very low values for temperatures below 100K, where a liquid nitrogen environment is needed. With this conditions the DCR establish the value between 0.1-1Hz [13].

Currently the technological effort is focused on reduce the noise contribution related to the optical-crosstalk phenomenon, where a photon that impacts on a certain pixel of the SiPM and trigger the first avalanche, and photon emitted by this first avalanche can trigger a second one in a neighbouring pixel. The photon travels at speed of light so this second avalanche can be considered simultaneously with the first one. There are some methods that deals with the optical-crosstalk like the optical trenches coated with metal between each pixel [14], the trench tries to absorb or reflect the incoming photon in order to avoid the triggerization of an adjacent pixel avalanche.

3.2. Single Photon Time Resolution

For the SiPM the timing resolution is defined as the time jitter between the true arrival time of the photon to the sensor and the instant when the output signal is recorded. The time resolution of photon detection systems is important for a wide range of applications in physics and chemistry. It impacts the quality of time-resolved spectroscopy of ultrafast processes and has a direct influence on the best achievable time resolution of time-of-flight detectors in high-energy and medical physics.

The timing resolution to single photon was studied by means of the setup that can be seen in figure 5. The light source used is a picosecond pulsed diode laser from PicoQuant that has a FWHM of about 50ps set at repetition rate of 2kHz, it works at 641nm (red). The optimum wavelength for the SiPM is around 400nm but the diode laser that works at this wavelength has some fiber coupling problems that we are not able to fix on time [15]. The light pulses travels through the single mode fiber from OZ Optics and then the collimator produces an spot of about 3cm of diameter that pass through an Optical Density (OD) filter from Edmun Optics in order to attenuate the light to the minimum level where the single photon regime can impinge on the SiPM. The entire setup is housed in a thermally insulated black box.

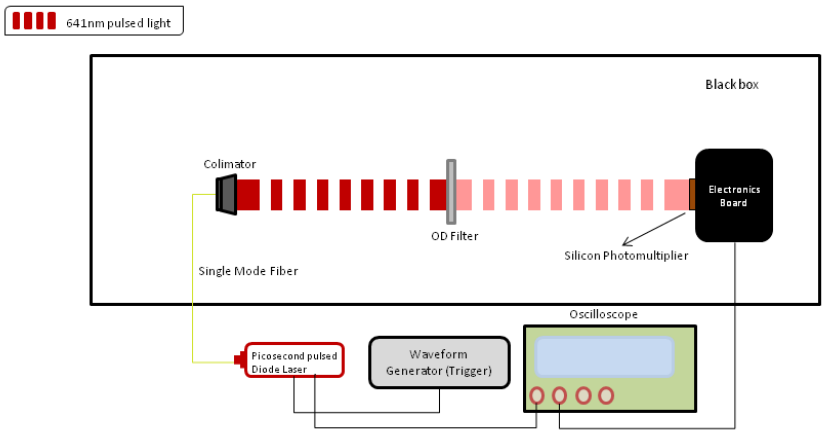
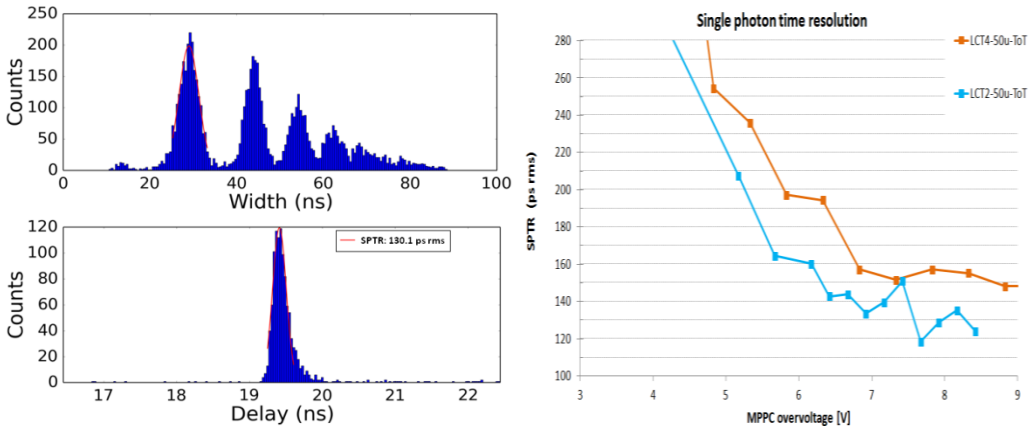


Figure 5: Scheme of the experimental setup used developed for the SiPM single photon time resolution measurement

Then the timing signal output from the FlexToT board is recorded by the oscilloscope Agilent MSO 9404A, as well as the trigger signal coming from the laser (with a repetition ration given by a square signal coming from the waveform generator). The timing resolution to single photon

is studied by measuring the fluctuations of the difference in time between the SiPM timing signal and the laser driver synchronization output, it means the trigger signal from the laser.
The analysis of each set of data consists first in measuring the width of the timing signal from the SiPM and makes a histogram of this data. Then the single photoelectron signal from the histogram is fitted, it corresponds to the first peak, the second one corresponds to 2 photoelectrons and so on. Then the corresponding delay time between the trigger signal and the timing signal is plotted. It is showed in figure 6a.



(a) **(b)**

Figure 6: Results obtained for SiPM timing resolution: a)This data correspond to the SiPM Hamamatsu prototype model LCT2-50μm and overvoltage of 8V, b) Set of data varying the voltage for different SiPM Hamamatsu models

The plot and fitting process is done by means a different Python scripts that allows making histograms and fit different functions, in this case the data follows the Gaussian model. The SPTR values that is shown above is the sigma of the corresponding Gaussian fit. Clearly the trend of the SiPM is to improve the timing resolution towards an increase of the overvoltage until it saturates. The FlexToT board with this SiPM models can achieve good single photon time resolution values around 120ps, comparable to the NINO ASIC from CERN [16].

3.3. Coincidence Time Resolution

The Coincidence Time Resolution (CTR) is related to the spatial resolution where the annihilation of the proton (in PET) has occurred. The smaller the CTR, the bigger the spatial resolution (smaller spot can be observed). Commercial full-body PET currently achieves a coincidence time resolution (CTR) of around 500ps FWHM [17]. Although there is a lot of effort in decreases this number a nowadays the values are around 200ps FWHM with scintillator crystal length of about 10-15mm, that CTR corresponds to a zone of about 3cm of diameter around the annihilation point, sufficient to remove events outside the organ of interest [18]. This value depends a lot on the scintillator crystal coupled to the SiPM, its size, composition... The model of the SiPM used, it is improving constantly. And also the electronics used to process the signal coming from the SiPM [19].

The setup used to perform CTR measurements it is shown in figure 7. As it has been explained in section 1, the radioactive β^+ decay of the Na^{22} of 100kBq with 2mm diamater radioactive source produces a positron, it is been annihilated with an electron that emits two photons of 511keV on the same path but with opposite directions (back-to-back gamma photons). These two photons impinge on the crystals of 20x3x3mm and the photon starts to transfer the energy to the crystal, it means that the scintillator crystal reemits the photon energy as an avalanche of optical photons of about 400nm. The photons arrive to the photodetector (SiPM). Then the SiPM are fed by an external power supply (Keithley Sourcementer 2611A) at a certain overvoltage and the signal that produces is processed by the FlexToT board as it has been explained in section 2.3. The entire setup is housed in a thermally insulated black box. The enviorenemnt temperature is about 20 ± 0.1℃. As it has been explained the temperature

fluctuation can cause a change on the SiPM gain and breakdown, so it needs to stay constant during an entire run of CTR measurement.

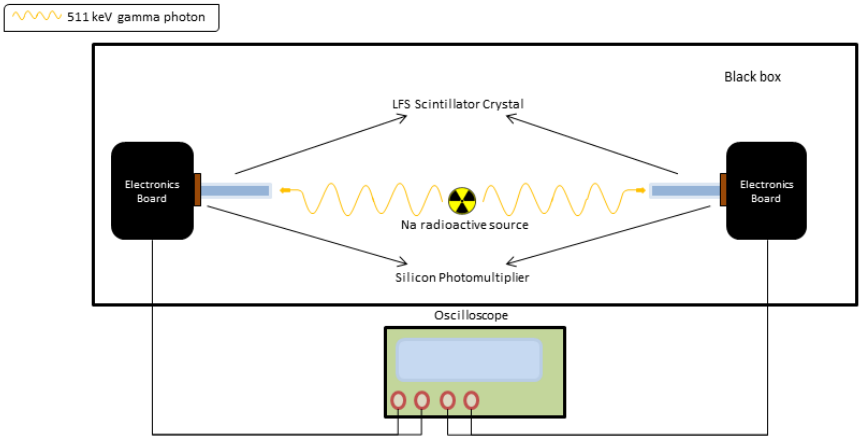
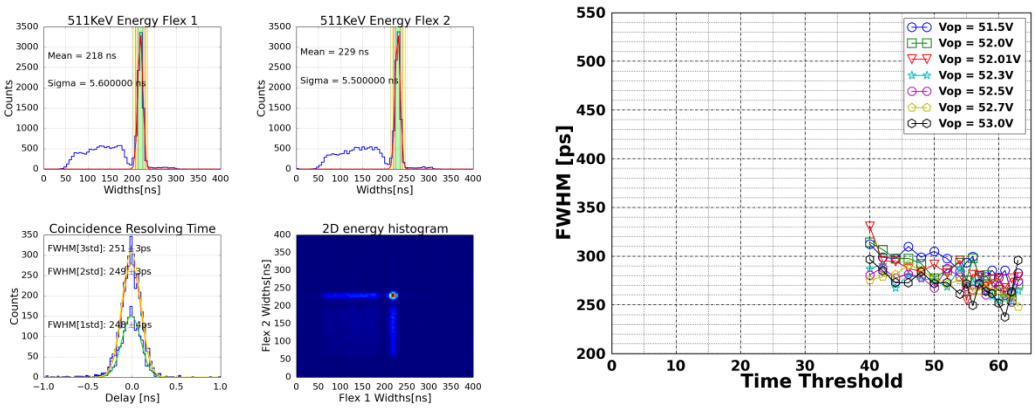


Figure 7: Scheme of the experimental setup used developed for the SiPM coincidence time resolution measurement

The data acquisition is performed by the high sampling rate oscilloscope Agilent MSO 9404A. The energy pulse width from both FlexToT boards and also the time delay between each board is recorded and post processed by another Python script. The results can be seen in figure 8, where it has been histogram the time delay accepting only events from the two photopeaks as can be seen from the Na^{22} spectrum from both boards at the top of the figure 8a. This figure corresponds to the data collected for the prototype LCT2-50μm from Hamamatsu with an operation voltage of 53.0V and a time threshold of 60.



(a) **(b)**

Figure 8: Plot results for the coincidence time resolution measurements, a) CTR measure for a single run with 20000 measures, threshold 60 and 53.0V, b) sweep run over the time threshold and with different voltages.

It is easy to see from figure 8b that the CTR improves with the voltage and also decreasing (the lowest value of time threshold is 63, which correspond to the 0 seen in figure 4) the time threshold. The problem with decreasing the time threshold is that the noise increases because the short noise pulses are allowed. The increment on the voltage increase the gain of the photodetector as well as the photo detection efficiency (PDE), but again the noise (DCR) also increases.

5. Conclusions

In this work we have seen that the ASIC FlexToT working together with the novel technology of the silicon photomultiplier obtains promising results, in the sense of the coincidence and single photon time resolution that is related to the spatial resolution when we talk on image reconstruction for time-of-flight positron emission tomography. With an LFS crystal of 20x3x3mm attached to the SiPM model LCT2 from Hammamatsu, consisting of a 3x3mm

single channel with 3600 micro-cells, we have obtained a SPTR in sigma RMS of 120ps and a CTR in 1 and 2 sigmas of about 250ns. We are on the same numbers of NINO developed by CERN, that is the best ASIC used for the PET purpose but with the advantages that the board that mounts our ASIC is more compact and consume less power.

The next step is to improve the SiPM technology, in the sense of a reducing dark count, improving the photon detection efficiency and the quantum efficiency. On the same path, the FlexToT ASIC will be improved in order to has a better dynamic range and energy resolution.

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