Application of MPPC to Positron Emission Tomography

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The main challenges in the design of next generation Positron Emission Tomography machines are the improvement of the spatial resolution and the combined operation with magnetic resonance. The Micro Pixel Photon Counter by Hamamatsu is a good candidate for this application. Its small size (1×1 mm^2) and high photo-detection efficiency in the blue spectral region allows the direct readout of a highly segmented scintillator matrix with improvements on the detector's spatial resolution. Furthermore, this photodetector is insensitive to magnetic fields up to 5 T, making its application in a magnetic resonance environment possible.

The aim of the study presented in this paper is the characterization of a system of scintillator crystal readout via MPPC. Crystals of $1 \times 1 \times 15$ mm^3 and $3 \times 3 \times 15$ mm^3 are directly coupled to a MPPC of the same size active area, and the energy resolution at 511 keV is measured. The coincidence time resolution of two such detector units is investigated. A first comparison of the performances of LSO (Lutetium Orthosilicate, or Lu_2SiO_5) and LFS (Lutetium Fine Silicate, newly developed by Lebedev Physical Institute, LPI) is given.

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Introduction

Positron Emission Tomography (PET) is a non-invasive medical imaging technique [1, 2]. A β^+ emitter is used to mark a tracer (generally a glucose molecule) which is injected into a living organism. The two 511-keV photons products of e^+e^- annihilation inside the organism are detected in coincidence, and their line of response is reconstructed. The analysis of several lines of response via well established mathematical techniques allows to reconstruct the 3D shape of the organs where the tracer is absorbed. The typical detector block for commercial Positron Emission Tomography consists of a pixelated BGO crystal, readout by few photomultiplier tubes. The signals from the photo-detectors are weighted by a resistive chain and the interaction point in the scintillator is reconstructed.

The energy resolution at the 511 keV photopeak and the time response are two relevant parameters for the optimization of a PET system. The former makes it possible to separate photons which have undergone Compton scattering from those which have lost energy only due to photo-electric effects. Compton-scattered photons change direction and contribute to the reconstruction of fake lines of response. These events are regarded as background to the PET measurements. A standard BGO-based PET detector [1] provides $10-13\%$ energy resolution at 511 keV, an optimal design would require $5-10\%$.

The time resolution of the detector determines the width of the coincidence window for the two photons. It is determined by the decay time of the scintillator and the response of the readout photodetector. In the BGO case, the time resolution is dominated by the decay time of the scintillator, ~ 300 ns. A shorter coincidence window ($O(10)$ ns) would be optimal, as it reduces the random coincident background. The time information can be also used to directly improve on the position resolution as it is done for Time-of-Flight PET (TOF-PET). A time resolution of 500 ps FWHM corresponds to a spacial resolution of 7.5 cm [3, 4].

Recent developments of new technologies for PET are mainly focused on the improvement of energy and time resolution of the detectors, which both influence the spatial resolution of the system. Many new fast crystals have been produced in the last ten years. LSO has a decay time of 40 ns and emits photons mostly at 420 nm wavelength. It shows a typical energy resolution of 10 % at 511 KeV and a time resolution of 475 ps when read out by a photomultiplier tube. Many faster crystals are avalaible nowadays, characterized by a peak light emission in the blue and ultra-violet spectrum. The characteristics of the crystals dictate the requirements of the photodetectors for PET. They need a high photo-detection efficiency in the blue spectral range and an excellent time response. The design of the second generation PET systems points toward a finer segmentation of the scintillator matrix. The ideal case is that crystals with a section of 2x2 mm^2 are read out by photodetectors of the same size. PIN diodes or APD are often used in such designs.

The Micro Pixel Photon Counter (MPPC) [5] is an excellent candidate for this application. It is a silicon photodetector, with variable size between 1x1 mm^2 and 3x3 mm^2. It consists of an array of p-n junction pixels biased above the breakdown voltage. They are passively quenched with an external resistor. Each pixel response amounts to the same charge and is hence not proportional to the energy of the impinging photon. The signal output is the sum of the charges of all pixels, which is proportional to the incident flux of photons. The gain of the device ranges between 10^5 and 10^6. The MPPC shows a high sensitivity in the 420 nm spectral region. The typical low dark current

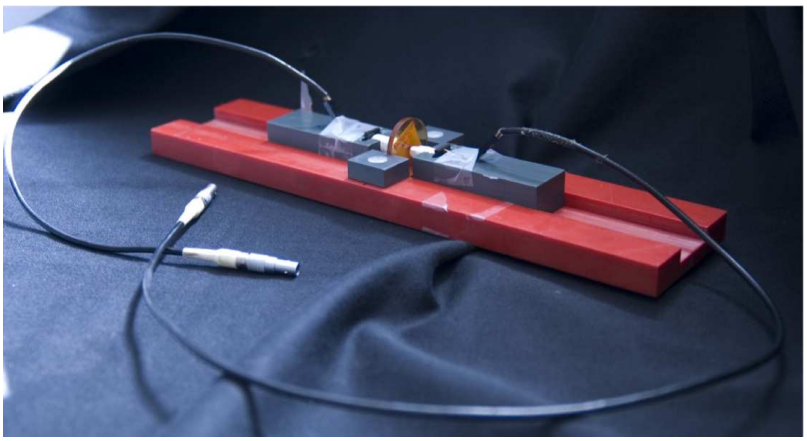

Figure 1: Setup for the energy and time resolution measurements of LSO crystals. The scintillators (white) are directly read out by the MPPC and aligned with a ^{22}Na source positioned in the middle. The setup allows a reproducibility of the measurements of 3% for the 3×3 mm^2 detectors and 10% for the 1×1 mm^2 ones

($< 1\ \mu$A), the low bias voltage ($\sim$ 70V) and the high gain largely simplify the readout electronics.

The test of a possible application of MPPC in PET readout is the topic of this work. A first characterization of a basic detector unit is presented. A LSO crystal ($1 \times 1 \times 15$ mm^3 or $3 \times 3 \times 15$ mm^3) is read out by a MPPC of the same size active area. The energy resolution of one detector unit is measured at 511 keV. The time resolution of two detector units in coincidence is extracted. A first comparison between LSO and LFS (Lutetium Fine Silicate) is presented.

1. Experimental Setup

This study is based on five samples of 1x1 mm^2 MPPCs (400 pixels) and five samples of 3x3 mm^2 MPPCs (3600 pixels). The active silicon is protected by a special plastic package. The suggested operation voltage is 76 V and 69.9 V respectively, with a spread of 0.1 V between the five pieces in each sample. The dark rate at 0.5 pixels is estimated to be 220 kHz and 3 MHz respectively. The gain of the devices is 7×10^5. The pairs of $1 \times 1 \times 15$ mm^3 and $3 \times 3 \times 15$ mm^3 LSO crystals (Hilger) and of $3 \times 3 \times 15$ mm^3 LFS crystals (Lebedev Physical Institute, LPI) are wrapped in a 2-mm thick Teflon layer. One end of the crystal is left free and coupled with optical grease to a MPPC of equal active area.

The setup is shown in Figure 1. Two special holders (gray) are machined to fix the photodetector to the crystal front face. The precision in the relative allignmet between the MPPC and the crystal is of order of 0.3 mm. The holders are positioned face to face on a rail (red) on either side of the ^{22}Na source (orange disk). The setup has one degree of freedom in the distance of the two detectors from each other and from the source.

Without using any amplification, the signals from the MPPC are integrated by the VME QDC Lecroy 1182, in a gate generated when the signals are in coincidence using NIM logic modules. The time resolution measurements are performed using the analysis software tools of a 4-GHz True-Analog Bandwidth oscilloscope (TDS7404B by Tektronix). The two signals from the detector elements are directly sent to the inputs of the oscilloscope, where they are discriminated using

the internal discrimination logic of the oscilloscope. After discrimination a logic coincidence is formed. The time difference between the two signals in then computed on line, with a deidcated software. In acquisition mode the oscilloscope provides a sampling rate of 20 GS/s, resulting in a time resolution of 50 ps.

2. Energy resolution at 511 keV

The energy spectum of a 511 keV photons as measured by one detector is presented in Fig. 2. The photoelectron peak is clearly separated from the energy of Compton-scattered photons. The energy resolution of the detector is extracted using a Gaussian fit to the photoelectron peak. The $\sigma/mean$ of the fit are quoted. Using the $3\times3\times15$ mm^3 LSO crystals, an energy resolution of (10 ± 0.3)% is obtained for 511 keV photons (Fig. 2.a), while the $1\times1\times15$ mm^3 crystals show (14 ± 1.4)% (Fig. 2.b). The errors quoted are due to the systematics of the system. The dominant uncertainty is in the relative allignment between MPPC and LSO. The higher systematic uncertainty of the latter measurement, due to the technical reproducibility and optimization of the coupling, leaves room for improvement.

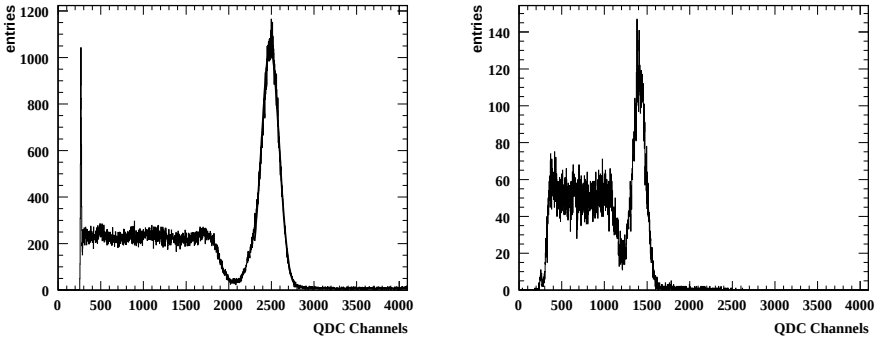

Figure 2: Energy response to a ^{22}Na source of (a) $3\times3\times15$ mm^3 LSO crystal coupled with a 3×3 mm^2 MPPC (3600 pixels) and (b) $1\times1\times15$ mm^3 LSO crystal coupled with a 1×1 mm^2 MPPC (400 pixels).

The energy resolution of the photoelectric peak is the quadratic sum of different contributions, namely the number of corresponding photons (poisson statistics), the intrinsic energy resolution of the scintillator (σ_{intr}), determined mostly by non-linearity of the crystal response, and by the electronic noise and the photodetector noise (σ_{noise}). The total energy resolution is expressed as:

$$\left(\frac{\sigma(E)}{E}\right)^2 \approx \left(\frac{1}{\sqrt{N}}\right)^2 + (\delta_{intr}(E))^2 + \left(\frac{\sigma_{noise}(E)}{E}\right)^2 \tag{2.1}$$

The measurement is performed without any electronic amplification and the intrinsic noise of the MPPC is very low, due to low dark rate and cross talk. Hence σ_{noise} can be neglected. The main contribution is given by the intrinsic resolution of LSO (δ_{intr}(511 keV) $\sim$ 9%, from [6]). The statistical term ($1/\sqrt{N}$) depends on the photodetection efficiency of the photodetector in the spectral range of the scintillation light emission. The poor efficiency in the blue spectral region was the limiting factor in a similar experiment [7] where SiPM (from CPTA) coupled to a $3\times3\times15$ mm^3

crystal provide a resolution of $\sim 35\%$. Furthermore, LSO usually shows $\sim 10\%$ energy resolution at 511 keV if read out by a traditional photomultiplier tube [3]. The energy resolution of MPPC is found to be competitive for PET applications with both the older photodetectors of the same family and the traditional photodetectors.

The $3 \times 3 \times 15$ mm^3 LFS crystal shows an energy resolution of 11% (Fig. 3), comparable to the same-size LSO within the systematic uncertainty of the measurement.

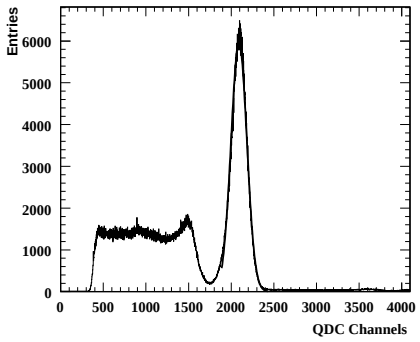

Figure 3: Energy response to a ^{22}Na source of a $3 \times 3 \times 15$ mm^3 LFS crystal coupled with a 3×3 mm^2 MPPC (3600 pixels). The obtained resolution is 11%, comparable with the performances of LSO.

3. Time resolution

The time resolution is defined as the FWHM of the distribution of the time difference between the two signals generated by the detection of two back to back scattered photons. The systematics of this measurement depend mostly on the two thresholds. The first, V_1, is the self-triggering amplitude. When the signals exceed V_1, a trigger pulse is generated. If two signals exceed the threshold at the same time a trigger coincidence is formed. V_1 has to be chosen higher than the typical baseline noise in order to avoid self triggering on background signals. The dark rate of the MPPC sets a lower limit for V_1, as the dark signal coincidence between the two detectors has to be minimal. The requirement on V_1 can be expressed as:

$$(N_{V>V_1}^{fake})^2 << N_{V>V_1}^{signal} \tag{3.1}$$

where $N_{V>V_1}^{fake}$ and $N_{V>V_1}^{signal}$ are the noise and signal rates above V_1. If a coincidence is generated the signals are sampled at another threshold V_2 to determine their time difference. In general V_2 can be higher or lower than V_1. The choice of V_2, as well as the fluctuation of the number of events above threshold, depends on the light yield of the detector. The time resolution of the detector has two components, an intrinsic one proportional to the decay time of the scintillator and a component related to the fake or background events σ_{fake}. The choice of V_1 and V_2 has a strong impact on the later component. The dependences of $\sigma_{fake}(V_1, V_2)$ need to be investigated in more details.

Figure 4.a shows the typical time resolution spectrum with V_1 set to 10 mV ($\sim$ 10 photoelectrons) and V_2 set to 2 mV ($\sim$ 2 photoelectrons)[1]. The FWHM is 1.4 ns. Performing a double

[1]Here and in the following, the correspondence between signal amplitude in mV and number of photoelectrons is

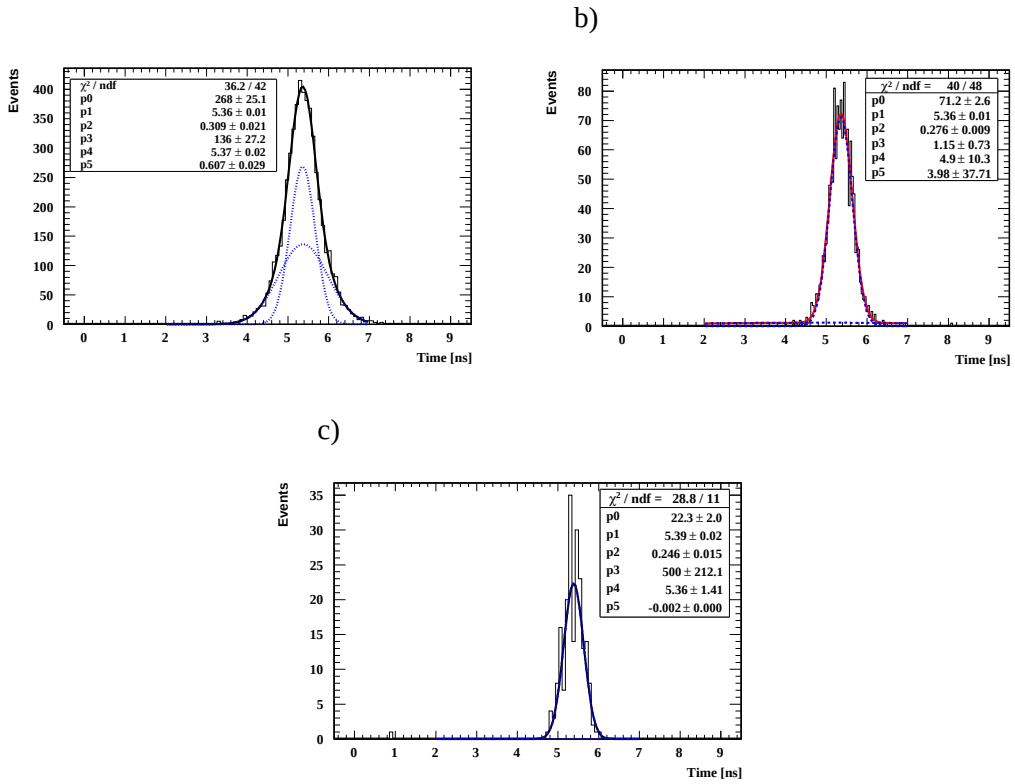

Figure 4: Timing resolution of two LSO crystals detecting in coincidence the 511 keV emission of a ^{22}Na source. A background is superimposed to the clean signal, worsening the FWHM from 700 ps to 1.4 ns (a). Rising the coincidence threshold to 50 photoelectrons (b) and 70 photoelectrons (c), the background disappears and a clean signal with FWHM 650 ps and 578 ps is observed respectively.

Gaussian fit to the spectrum, it is possible to separate two contributions: the signal with a FWHM of 700 ± 60 ps and the background with a FWHM of 1.41 ± 0.07 ns. The background can be identified as originated from events for which one signal is generated via photoelectric effect and the other by Compton scattering. The leading edge of the two signals has a different slope, which worsens the time resolution. The probability of getting these background events is high in this measurement as the coincidence threshold is set to 10 photoelectrons, much lower than the photoelectric peak amplitude, estimated to be $\sim$300 photoelectrons. Raising V_1 to 50 and 70 photoelectrons (Figure 4.b,c), the background disappears, and a time resolution of 650 ± 20 ps and of 578 ± 35 ps FWHM respectively is measured.

It is quoted that a time resolution of 500 ps (FWHM) would be enough to enhance the signal to background ratio by a factor of 2 of the reconstructed image [4]. A similar value is shown for LSO crystal read out by a photomultiplier tube [3]. The measurements with MPPC are not far from what is needed and indicate a promising optimization, achieving results that can compete with the traditional design.

only intended as an indication. The number of firing pixels is strongly related only to the total charge of the signal.

Conclusion

Measurements have been presented which show the possibility to use the Micro Pixel Photon Counter as a good alternative to photomultiplier tubes in Positron Emission Tomography. An energy resolution of 10% at 511 keV is obtained reading a LSO crystal via an MPPC photodetector. A new scintillator crystal from LPI, the LFS, has been compared to LSO, which yields comparable results.

A time resolution better than 600 ps is achieved for the detection in coincidence of two back to back scattered photons from a ^{22}Na source.The background from Compton-scattered photons is either suppressed using a high threshold on the signal amplitude, or accounted for with a second Gaussian function in the fit of the time distribution. This method will have to be further investigated at lower thresholds, but it indicates potential improvement for the time resolution determination. Further systematic studies on the background treatment will follow.

It can already be seen that the application of MPPC to PET detectors open the possibility for much higher segmentation and thereby higher spacial resolution. The flexibility in sizes and shapes offered by MPPC perfectly matches the requirements of PET applications.

As a next step the homogeneity of the crystals and the stability and reproducibility of the system have to be tested on a larger scale. A small prototype detector is being designed made of two modules. One module is assembled as a matrix of 6x6 crystals read out individually via 36 MPPCs. With a 72-channel readout system enabling good energy and time resolution, this prototype should establish MPPC as a possible photodetector for the next generation of PET.

Acknowledgments

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