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First tests in the application of silicon photomultiplier arrays to dose monitoring in hadron therapy

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ABSTRACT

A detector head composed of a continuous $LaBr_3$ crystal coupled to a silicon photomultiplier array has been mounted and tested, for its use in a Compton telescope for dose monitoring in hadron therapy. The $LaBr_3$ crystal has 16 mm $\times$ 18 mm $\times$ 5 mm size, and it is surrounded with reflecting material in five faces. The SiPM array has 16 (4 $\times$ 4) elements of 3 mm $\times$ 3 mm size. The SPIROC1 ASIC has been employed as readout electronics. The detector shows a linear behavior up to 1275 keV. The energy resolution obtained at 511 keV is 7% FWHM, and it varies as one over the square root of the energy up to the energies tested. The variations among the detector channels are within 12%. A preliminary measurement of the timing resolution gives 7 ns FWHM. The spatial resolution obtained with the center of gravity method is 1.2 mm FWHM. The tests performed confirm the correct functioning of the detector.

1. Introduction

Hadron therapy is a technique that allows administration of radiation dose very precisely in the location area of the tumor reducing the damage to the healthy tissue. However, different factors such as patient misposition, tumor size variations, weight changes and organ motion can result in an uncertainty in the tumor location. Therefore it is crucial to have methods that ensure that the dose is accurately delivered to the tumor during the treatment [1]. Positron emission tomography (PET) combined with Monte-Carlo simulations is currently employed for dose monitoring purposes. A PET detector near the therapy room, or in combination with the treatment (known as in-beam PET), images the distribution of positrons emitted by the target nuclei activated during proton therapy, or also by the projectile nuclei in the case of carbon ion beams. However, this method has some limitations. An acquisition time of more than 10 min is necessary due to the relatively low induced activities, interfering with the treatment routine and resulting in a progressive activity loss due to the metabolic wash-out of the radioactive nuclei in the patient. In-beam PET scanners must be segmented for their integration with the treatment equipment, and the limited-angle acquisition complicates the correlation of the activity imaged and the dose.

The fragmentation of the nuclei during therapy is also followed by the emission of other particles such as gammas and neutrons. Prompt gammas from the excited nuclei are emitted with energies up to about 15 MeV, and can also be used for dose monitoring. The emission of these gammas can be correlated with the Bragg peak [2].

The use of a Compton telescope is an interesting option to locate the origin of these gammas and to estimate the delivered dose [3]. The most common Compton detector configuration employed in medical imaging, which is based on a scatterer and an absorber [4–6], is not optimal for this application, given the high energies of the gamma rays, and the fact that their energy is unknown. The three-Compton technique can be employed instead. In this approach, the gamma-ray origin is determined by Compton kinematics, with the position and energy information provided by three interactions, two Compton plus a third interaction of any type in three different detector heads [7,8]. With this method, however, the efficiency is reduced with respect to the two-interaction determination.

A Compton telescope design is under study for this application, composed of several detector layers made of continuous $LaBr_3$ crystals coupled to silicon photomultiplier (SiPM) 2D arrays. $LaBr_3$-based detectors can provide a high energy resolution, a fast response, and a high efficiency as compared to silicon detectors [9].

Simulations that include the production of gammas in the tissue and the detector response are being carried out with GEANT4 to fully understand the physical processes, and to determine the optimum geometry and estimate the performance of the telescope. A method based on maximum likelihood (ML) estimation, that will allow us to enhance the efficiency by reconstructing the gamma-ray origin when at least two interactions occur, is also under development.

Initial experimental tests have been performed with a $LaBr_3$ crystal coupled to a SiPM array. In the future, the detector is expected to have a high number of readout channels, and thus, an

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ASIC is essential for its readout. The SPIROC1 ASIC [10], developed at the Linear Accelerator Laboratory (LAL, Orsay, France), that is specifically designed for SiPMs for the International Linear Collider (ILC) hadronic calorimeter, has been employed. The goals of the initial measurements are to assess the feasibility of operating the detector, and to test the system performance.

2. Experimental setup

2.1. Detector

The detector is composed of a continuous 16 mm × 18 mm × 5 mm $LaBr_3$(Ce) crystal (BrilLanCe® 380 from Saint-Gobain Crystals[1]) coupled to a SiPM 2D array. The crystal (Fig. 1) is surrounded by highly reflective material in five faces, and encapsulated in an aluminum housing. The open face is covered by a 1 mm thick optical guide. The crystal is coupled directly to the photodetector without optical grease.

The SiPM array (Fig. 2) is made of 16 (4 × 4) multipixel photon counter (MPPC) elements from Hamamatsu Photonics.[2] Each pixel is 3 mm × 3 mm size, and it is composed of 3600 microcells of 50 μm × 50 μm size. The pitch is 4.05 mm in the *X* direction, and 4.5 mm in *Y*. The operating voltages of the pixels range from 71.08 to 71.12 V. A common bias voltage is applied to all elements of 0.5 V over the average operating voltage of the 16 pixels (71.1 V).

2.2. Electronics

The SPIROC1 ASIC is a 36-channel, low noise, high dynamic range front-end circuit. Two variable gain preamplifiers make it possible to achieve the required high dynamic range (up to 2000 photoelectrons). An 8-bit DAC at the preamplifier input allows fine-tuning of the input voltage and the SiPM bias of each channel. Each preamplifier is followed by a shaper with variable shaping time (25–175 ns). For each event recorded, both the pulse amplitude and the time are stored and the time are stored. A fast shaper provides the trigger signal for each channel in parallel. A common threshold is applied to all discriminators that can be adjusted for each channel by means of a 4-bit DAC. The intrinsic timing resolution, measured with a pulse generator as the uncertainty in the time difference between the input pulse and the trigger signal generated by the ASIC, is around 200 ps FWHM.

3. Results

3.1. Detector uniformity

The uniformity of the detector has been estimated by coupling a LYSO crystal of 3 mm × 3 mm × 10 mm size to each of the SiPM pixels in the matrix, and acquiring a ^{57}Co spectrum. The photopeak of the spectrum obtained with each detector element has been fitted. The peak position obtained for each channel, in ADC counts, is shown in Fig. 3 as a function of the pixel position. The variations are within 12%.

3.2. Linearity

The linearity of the detector has been tested by acquiring energy spectra with different sources, with energies ranging from 122 to 1275 keV, and plotting the photopeak position as a function of the source energy. The sources employed with the corresponding energy are listed in Table 1. In order to obtain the energy spectra, the signals acquired in all the photodetector elements in each event are summed to calculate the total energy deposited in the crystal.

Fig. 1. Continuous $LaBr_3$ crystal surrounded by reflective material and encapsulated in an aluminum housing.

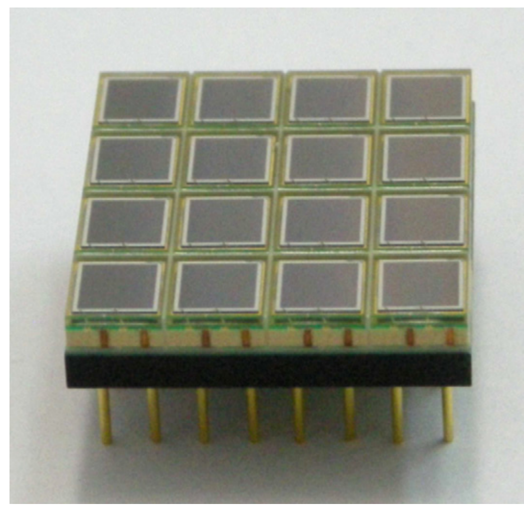

Fig. 2. SiPM array composed of 16 elements of 3 mm × 3 mm size.

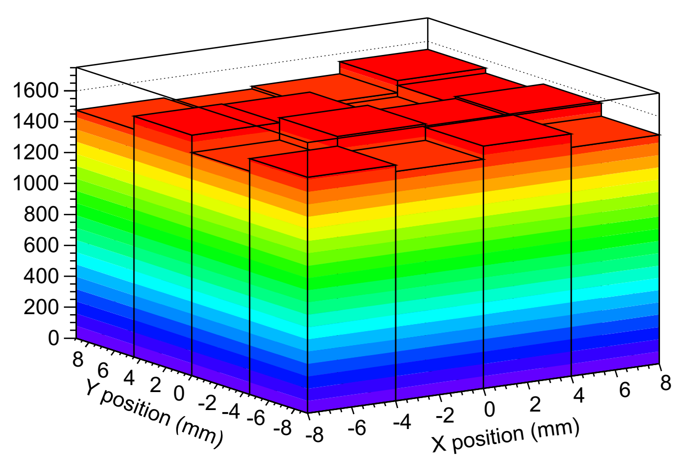

Fig. 3. Position of the ^{57}Co 122 keV photopeak for each channel.

[1] www.crystals.saint-gobain.com
[2] www.hamamatsu.com