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Characterization of a module with pixelated CdTe detectors for possible PET, PEM and compton camera applications

G. Ariño-Estrada\textsuperscript{a,1}, M. Chmeissani\textsuperscript{a}, G. de Lorenzo\textsuperscript{a}, C. Puigdengoles\textsuperscript{a}, R. Martínez\textsuperscript{b} and E. Cabruja\textsuperscript{b}

\textsuperscript{a}Institut de Física d’Altes Energies, IFAE, 08193 Bellaterra, Barcelona, Spain

\textsuperscript{b}Centro Nacional de Microelectrónica, IMB-CNM (CSIC), 08193 Bellaterra, Barcelona, Spain

E-mail: garino@ifae.es

ABSTRACT: We present the measurement of the energy resolution and the impact of charge sharing for a pixel CdTe detector. This detector will be used in a novel conceptual design for diagnostic systems in the field of nuclear medicine such as positron emission tomography (PET), positron emission mammography (PEM) and Compton camera. The detector dimensions are 10 mm $\times$ 10 mm $\times$ 2 mm and with a pixel pitch of 1 mm $\times$ 1 mm. The pixel CdTe detector is a Schottky diode and it was tested at a bias of -1000 V. The VATAGP7.1 frontend ASIC was used for the readout of the pixel detector and the corresponding single channel electronic noise was found to be $\sigma < 2$ keV for all the pixels. We have achieved an energy resolution, FWHM/$E_{\text{peak}}$, of 7.1%, 4.5% and 0.98% for 59.5, 122 and 511 keV respectively.

The study of the charge sharing shows that 16% of the events deposit part of their energy in the adjacent pixel.

KEYWORDS: Solid state detectors; Gamma camera, SPECT, PET PET/CT, coronary CT angiography (CTA); Pixelated detectors and associated VLSI electronics

\textsuperscript{1}Corresponding author.
1 Introduction

PET is a nuclear medicine technique capable of providing in-vivo functional imaging results without any invasive process. In such a technique, a radiotracer is injected into the body of the patient. In the case of PET, this must be a positron emitter. The principle operation of PET is the detection of pair of photons created in a positron-electron annihilation. Such an interaction happens when the radiotracer emits a positron that interacts with an electron of the surrounding matter.

In PET or PEM, the energy resolution plays a decisive role in eliminating events that have undergone a Compton scattering and thus have the wrong line of response (LOR). In Compton camera, the energy resolution plays a critical role in defining the systematic error in the reconstruction of the Compton cone angle, and at the same time to insure that the full energy of the photon is contained in the scatterer and the absorber detectors. In [1] the scattered fraction is estimated to be 27% whereas for the [2], with an energy resolution of 5 keV FWHM, one expects scatter fraction of 3%. Such a low scatter fraction increases significantly the signal purity. This allows one to obtain an image with the same signal-to-noise ratio as state-of-the-art clinical devices [2] with an order of magnitude less statistics. CdTe detectors can provide such a good energy resolution. Moreover, CdTe detectors with 2 mm thickness have shown a good timing resolution that fits the timing coincidence of PET events [3] and this allows the user to operate the PET scanner at a high dose of radiotracer without pile-up effect.

Current commercial PET scanners use scintillation crystals coupled to photomultiplier tubes (PMTs) to detect, indirectly, those photons [1, 4, 5]. The crystal detectors offer a good detection efficiency for energies below 1 MeV, fast timing properties and good energy resolution. Scintillator crystals have an intrinsic limitation when it comes to energy resolution, and in the case of NaI, BGO and LSO, it is around 6% FWHM at 662 keV [6]. On the other hand, PMTs are incompatible with strong magnetic fields. Other photosensors like avalanche photodiodes (APD) or silicon photomultipliers (SiPM) are used to overcome this limitation. Finally, the thickness of the scintillating
crystals typically used in clinical PET scanners is around 20 mm, and complex DOI [15] measurement techniques must be employed to reduce the so-called radial astigmatism due to the parallax effect. In practice, it is hard to obtain DOI resolution lower than 3 mm FWHM. Different detector module designs employing these photosensor detectors have been evaluated in [7–9].

Semiconductor detectors are becoming a good alternative to overcome the limitations of scintillation detectors. They offer an excellent energy resolution, <1% FWHM/E at 511 keV [3], and they can be segmented into voxel arrays with pixel pitch down to less than 1 mm to improve the spatial resolution in all directions. In some modality they are also insensitive to strong magnetic fields when the electric and magnetic fields are parallel. PET scanners employing CdTe or CdZnTe have been constructed and tested as detailed in [10, 11].

The Voxel Imaging PET (VIP) Pathfinder project [12] proposes a novel PET scanner design employing semiconductor detectors. The scanner is based on the stack of modular units (the VIP module) made of voxelized CdTe. Each voxel is 1 mm × 1 mm × 2 mm [13]. The stack has a density of ~470 voxels/cm³. The readout process is completely independent for each voxel. A dedicated ASIC, the VIP ASIC, is being designed for this purpose [14]. It includes a charge integrator amplifier (preamp), a shaper filter, a peak-and-hold stage, an analog-to-digital converter (ADC), a trigger on the preamp output and a time-to-digital converter (TDC).

The geometry of the VIP module can adopt either trapezoidal or rectangular shape for the CdTe detector. The trapezoidal shape is thought to be employed to build a full seamless PET ring. Simulations test of sensitivity and spatial resolution of the VIP-PET scanner performance are shown in [2].

In the VIP design, and based on the current noise measurement, the trigger threshold is set to 8 keV. The preamp output is fed into the discriminator. Under this condition the rise time will depend on the electron-hole mobility. Lab measurements show a coincidence time resolution of 12.5 ns FWHM [3].

One main feature of the VIP PET design is the spatial resolution in the radial direction. With VIP PET, the radial direction is pixelized with 1 mm pitch pixels thus reducing the degradation due to the penetration of photons in thick crystals.

The CdTe diodes have less detection efficiency than scintillation detectors for typical PET gammas. In order to compensate for it, as many detectors as needed can be placed in the radial direction of the VIP module. The passive layer between detectors shows a negligible interaction with the gammas and doesn’t create significant extra Compton photons. In the VIP design 4 cm of CdTe are placed in the radial direction, this amount of CdTe stops 70% of incoming 511 keV photons resulting in a 50% efficiency for the detection of the annihilation photon pairs reaching the detector.

The VIP project proposes a solution to the limitations of state-of-the-art PET scanners based on scintillators detectors.

In this work, the characterization of such module is presented. The noise of all channels has been measured and kept under control for optimum energy resolution measurements. The ²⁴¹Am, ⁵⁷Co and ²²Na radioactive sources have been used to evaluate the energy resolution at different energies.
2 Experimental setup

The module consists of 4 CdTe diodes of size 10 mm × 10 mm × 2 mm fabricated by ACRO-RAD LTD. They are made with Schottky contacts with the pixel electrode (anode) made of Al/Au/Ni/Au/ALNl and the cathode made of Pt. The anode is segmented into 100 electrodes of 1 mm × 1 mm pixel pitch. The cathode is single-electrode connected to the high voltage power supply (HVPS).

Each diode is mounted on a glass substrate of 10 mm × 10 mm surface. The anode electrodes are bump bonded to the glass substrate using bump depositions of BiSn with diameter of 250 µm. The VATAGP7.1 ASIC [16] has been used for the signal processing. The ASIC can process up to 128 channels.

To save resources, 98 electrodes of each diode have been paired and connected to one readout channel. One electrode on each diode has been directly connected to a readout channel. Another one has been connected to the guard ring. The guard ring is 50 µm wide and it is separated 5 µm from the electrodes on the edge. Each diode finally consists of 49 channels with 1 mm × 2 mm pixel pitch plus one channel with 1 mm × 1 mm. Two diodes (100 channels in total) have been connected to one VATAGP7.1 ASIC. Each VATAGP7.1 channel has its independent readout circuit. It consists of a charge integrator amplifier coupled in parallel to a fast and a slow CR-RC filter stages. The fast CR-RC filter has a peak time of 50 ns and feeds a discriminator that triggers the readout process. The slow CR-RC filter has a peak time of 500 ns and is coupled to a track-and-hold stage to measure the pulse amplitude.

The analogue outputs of the channels are digitized using an analogue-to-digital converter (ADC). An FPGA is used to buffer the digitized data and to set the control register of the ASIC.

A Labview application has been designed for the data acquisition (DAQ). The application sets the ASIC configuration and classifies and stores the output data. This software also controls the HVPS that is used to bias the diodes. The bias voltage is ramped down every 5 min for 30 s and ramped up again to avoid detector polarization effects.

A picture of the setup including all the devices aforementioned is shown in figure 1. A scheme depicting the data flow in the acquisition process is shown in figure 2.

3 Results

3.1 Noise

The diodes have been biased at -1000 V and operated at room temperature. A random trigger has been used to sample the data. A gaussian fit has been applied to the energy distribution of each channel and the standard deviation (σ) has been used to measure the noise spread. The measured σ values are shown in figure 3 left. A channel map of the two diodes is shown in figure 3 right. All channels depict the same noise level.

Channels from 60 to 99 show an increasing noise trend. This is due to the fact that the lines of these channels in the pcb are longer and have larger capacitance value.
3.2 Calibration

The response of the setup has been studied for different input energies. The pixel CdTe module has been exposed to three radiative sources: $^{241}$Am, $^{57}$Co and $^{22}$Na. The most probable emissions of these radionuclides are photons with energies of 59.5, 122 and 511 keV respectively. The pedestal noise and the photopeaks of the three radiative sources have been used to calibrate the ADC of each pixel channel by assuming a linear response. The calibration of channel 22 is shown in figure 4 as an example. Negligible deviations from the linearity have been observed and the angular coefficient of the linear fitting has been used as the total gain.
Figure 3. Left: $\sigma$ of the noise distribution for all channels. Points colored consistently with figure 3 right. Right: Diode 1/2 channel map.

Figure 4. Detector response (ADC counts) for input energies of 0, 59.5, 122 and 511 keV in channel 22. Fit function $f(x) = p[0] + p[1] \times x$.

3.3 Energy resolution

The energy resolution has been measured channel by channel, using the main photopeak for each of the 3 radioactive sources. For each trigger in a given pixel, also the energy of the neighbour pixels is also digitized. To reduce the charge sharing effect [17] on the energy resolution, events that contain any neighbour pixel with energy bigger than 2 sigma-noise from the baseline, are rejected. The FWHM of the photopeaks have been measured by linear interpolation. The FWHM at 59.5, 122 and 511 keV for all the channels are shown in figure 5.

The spectroscopies of the three radionuclides obtained with pixel 22 are shown as an example in figure 6.

In figure 6 left, the $^{241}$Am photopeak shows a FWHM of 4.2 keV (7.1%). In figure 6 center, one can see the FWHM of the 122 keV peak is 5.5 keV (4.5%). The 136 keV peak can be easily
Figure 5. FWHM at 59.5, 122 and 511 keV photopeaks for every pixel-channel.

Figure 6. Left: $^{241}$Am spectroscopy. Center: $^{57}$Co spectroscopy. Right: $^{22}$Na spectroscopy.

tagged. In 6 right, the 511 keV has a FWHM of 5.0 keV (0.98%). A small peak on the left of the main one can be observed. It is due to the K-edge escape peaks of Cd and Te ($\sim 26$–31 keV).

3.4 Charge sharing and its impact of PET event detection efficiency

Pixels with a size of 1 mm × 2 mm, that are not adjacent to the guard ring show that, on average, 16% of the photons have deposited part of their energy, >2 sigma noise and at bias of 500 V/mm, in the adjacent pixel. For a pixel size of 1 mm × 1 mm, and under the same working conditions, one can scale the charge sharing effect down to 21.3% of the photons. Given that the VIP PET event selection requires the energy of the photon to be 503 keV < E-photon < 519 keV, by summing up the charges in the adjacent pixel, one can recover 88% of the photons that undergone charge sharing process. This implies that the detection efficiency for PET event of a pixel of 1 mm × 1 mm is reduced to 97%.

4 Conclusions

A module with 4 pixelated CdTe diodes, with pixel pitch of 1 mm × 2 mm, and 2 mm thick has been assembled and characterized at a bias of 500 V/mm and at room temperature. Two V ATAGP7.1 ASIC have been used for the readout of 200 pixel channel from the 4 CdTe detectors. The electron-
ics noise, including the detector leakage, was found to be less than 2 keV per channel. The pixel energy resolution at 59.5, 122 and 511 keV is 7.1%, 4.5% and 0.98% respectively after eliminating events with charge deposited in more than one pixel. These results are in agreement with other published results with similar detector designs [18–20] and with [3]. These measurements confirm that the assumptions made for the simulation and the evaluation of the VIP PET detector in [13] are realistic, given that the energy resolution at 511 keV was expected to be <1%.

The future work is to include the time stamp for every event, to allow us to operate the detectors in coincidence mode in order to reconstruct image of small size phantoms with PET tomographic techniques.

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