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Investigation of a Multi-Anode Microchannel Plate PMT for Time-of-Flight PET

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Abstract

We report on an investigation of a multi-anode microchannel plate PMT for time-of-flight PET detector modules. The primary advantages of an MCP lie in its excellent timing properties (fast rise time and low transit time spread), compact size, and reasonably large active area, thus making it a good candidate for TOF applications. In addition, the anode can be segmented into an array of collection electrodes with fine pitch to attain good position sensitivity. In this paper, we investigate using the Photonis Planacon MCP-PMT with a pore size of 10 μm to construct a PET detector module, specifically for time-of-flight applications. We measure the single electron response by exciting the Planacon with pulsed laser diode. We also measure the performance of the Planacon as a PET detector by coupling a 4 mm $\times$ 4 mm $\times$ 10 mm LSO crystal to individual pixel to study its gain uniformity, energy resolution, and timing resolution. The rise time of the Planacon is 440 ps with pulse duration of about 1 ns. A transit time spread of 120 ps FWHM is achieved. The gain is fairly uniform across the central region of the Planacon, but drops off by as much as a factor of 2.5 around the edges. The energy resolution is fairly uniform across the Planacon with an average value of $18.6 \pm 0.7\%$ FWHM. While the average timing resolution of 252 ± 7 ps FWHM is achieved in the central region of the Planacon, it degrades to 280 ± 9 ps FWHM for edge pixels and 316 ± 15 ps FWHM for corner pixels. We compare the results with measurements performed with a fast timing conventional PMT (Hamamatsu R-9800). We find that the R9800, which has significantly higher PDE, has a better timing resolution than the Planacon. Furthermore, we perform detector simulations to calculate the improvement that can be achieved with a higher PDE Planacon. The calculation shows that the Planacon can achieve significantly better timing resolution if it can attain the same PDE as the R-9800, while only a 30% improvement is needed to yield a similar timing resolution as the R-9800.

Index Terms

time-of-flight PET; scintillation detectors; photomultipliers; microchannel; timing resolution

I. INTRODUCTION

In conventional PET, the location of an individual positron is determined after it decays to form a pair of back-to-back 511 keV photons. The three-dimensional location of each positron could be directly determined by accurately measuring the difference in arrival times of the two annihilation photons. In other words, the position of the positron would be constrained to a point rather than a line; so three-dimensional images could be obtained without a reconstruction algorithm. However, the position along the line is localized to $\Delta x = (c/2)\Delta t$ where Δx is the position error, c is the speed of light, and Δt is the error in the timing measurement. Currently achievable timing resolution of a few hundreds picoseconds only constrain the positron position to a few centimeters. While this does not improve the spatial resolution, it has been shown to reduce the statistical noise in the reconstructed image if the

line segment was shorter than the size of the emission source [1–5]. Simple argument estimates that the statistical noise variance will improve by a multiplicative factor $f=D/\Delta x$ where D is the size of the emission source. Equivalently, as signal-to-noise ratio (SNR) is proportional to the square root of counts, the improvement in SNR can therefore be estimated by $SNR_{TOF}=\sqrt{f}\cdot SNR_{conv}$ where SNR_{TOF} is the SNR in the time-of-flight (TOF) image and SNR_{conv} is the SNR in the conventional image. Recently, due to advances in detector technology and readout electronics, there has been a renewed interest in TOF PET [6–8]. Significant improvement in the timing resolution has been realized using recently developed scintillators such as cerium-doped lutetium orthosilicate (LSO) [9] and lanthanum bromide (LaBr₃) [10]. Commercial TOF PET camera with 500–600 ps FWHM coincidence timing resolution has been developed [11].

While advances in the readout electronics have significantly reduce their contribution to the timing resolution, the timing performance of the detector will ultimately limit the achievable timing resolution. It has been shown that the choice of PMT and light sharing introduced by the scintillator array and light guide can degrade the timing resolution of the detector module [12]. Fig. 1a shows the components of a commercial PET block detector module, which includes the scintillator crystal array, light guide, and photomultiplier tube (PMT) as parts of the timing chain. Therefore, a PMT or a photodetector with excellent timing properties is needed to achieve the best timing resolution. The timing properties include fast rise time, low transit time spread (TTS), and high particle detection efficiency (PDE). It is also well known that the optimum detector configuration that will yield the best timing resolution is to couple the scintillator crystal directly onto the PMT as shown in Fig. 1b. However, the timing resolution is very dependent on the crystal geometry, which for PET detector is typically a long narrow crystal, which provides a high efficiency to stop the incoming 511 keV photons. Thus, for a given crystal geometry, the best timing resolution can be achieved by coupling the crystal directly to a fast photodetector with excellent timing properties. In order to construct a practical detector module with position sensitivity without degrading the timing resolution, each pixel in the scintillator crystal array would be coupled one-on-one to a fast photodetector array. Since the scintillator crystals array are closely packed, the pixel or anode of the photodetector array has to be closely packed as well. In addition, the photodetector array has to provide a reasonably large detection area with minimum dead area around the periphery in order to tile detector modules together for TOF PET applications.

While there are a few options for the photodetector array, a photodetector employing microchannel plates (MCPs) can offer the most promising properties for TOF PET detector modules. A single MCP normally has a gain of 10^2 to 10^3 . In order to achieve a good signal-to-noise ratio, two MCPs can be stacked together to yield a gain greater than 10^5 . Furthermore, an MCP with small pore size and small length-over-diameter (L/D) ratio would yield excellent timing performance. However, there are compromises on how small the pore and how thin an MCP can be fabricated. In a photodetector with MCPs, other physical factors that contribute to the timing performance are the distance from the photocathode to the front-surface of the MCP and the distance from the back-surface of the MCP to the collecting anodes because these factors affect the electron transit time. In this paper, we evaluate the performance of the Planacon MCP-PMT from Photonis. The MCP exhibits excellent timing properties and the Planacon multi-anode readout provides good position sensitivity [13–14]. It also has a reasonably large detection area with a small dead area around the periphery and is compact. First, we measure the single electron response (SER) using a pulsed laser diode. Next, we evaluate the Planacon as a PET detector by measuring the gain uniformity, energy resolution and timing resolution across the face of the Planacon by coupling a scintillator on it. For comparison, we also perform the measurements with a fast conventional PMT. Finally, we perform detector simulations using the measured

SER to calculate the improvement in the timing resolution as the PDE of the Planacon is improved.

II. MATERIALS AND METHODS

A photograph of the Planacon XA85015/A1 is shown in Fig. 2. The XA85015 is operated at -2700 V and employs two MCPs arranged in a Chevron configuration. The distance from the photocathode to the MCP is about 6 mm and the distance from the MCP to the anode is about 5 mm. The pore size of the MCP is $10\text{ }\mu\text{m}$ with a L/D ratio of 40. Using the recommended voltage divider from the manufacturer as shown in Fig. 3, the voltage between the photocathode and the MCP is 264 V, across the two MCPs is 2173 V, and between the MCP and anode is 264 V (we measured an effective resistance of $4.12\text{ M}\Omega$ when a $5\text{ M}\Omega$ resistor is put in parallel across the MCPs implying that the resistance across the MCPs is about $23.4\text{ M}\Omega$). The dimension of the XA85015 is $59\text{ mm} \times 59\text{ mm} \times 25\text{ mm}$ with an active detection area of $53\text{ mm} \times 53\text{ mm}$. The output of the MCP is read out by 64 anodes arranged in an 8×8 matrix. The size of each anode is $5.9\text{ mm} \times 5.9\text{ mm}$ with a pitch of 6.5 mm between the anodes. The anode structure can be fabricated with finer pad sizes and smaller pitch to provide better spatial resolution. Alternatively, cross-strips anode structure can also be fabricated to reduce the number of readout channels. For comparison, a fast conventional PMT is also used in the measurements. The conventional PMT is a Hamamatsu R-9800 with a super bialkali photocathode operated at 1300 V.

A. Single Electron Response Measurement

The SER of one anode of the XA85015 and R-9800 are measured by exciting the PMTs with a pulsed laser diode driven by the Hamamatsu Picosecond Light Pulser (PLP-01). The laser diode has an emission wavelength of 650 nm and is attenuated with enough neutral density filters so that the average number of photoelectrons per pulse is less than 1. The single electron waveform is measured with a fast digitizing oscilloscope (Tektronix TDS7404, 4 GHz analog bandwidth) directly from the output of the PMT terminated into 50Ω . The single photoelectron TTS of the PMT is measured using a coincidence timing measurement between the reference trigger output of a pulsed laser diode and the PMT signal.

B. Detector Measurement

To evaluate the Planacon as a PET detector, a $4\text{ mm} \times 4\text{ mm} \times 10\text{ mm}$ LSO crystal with chemically etched surface finish is end-coupled to it on one of the $4\text{ mm} \times 4\text{ mm}$ face with optical grease while the rest of the crystal surface are wrapped with four layers of Teflon tape. In order of minimize systematic effect, the same crystal is used in all the measurements. The crystal is positioned on the individual anode using a mask fabricated from Delrin. The mask has circular holes large enough for the crystal to snugly slide in. The centers of the holes correspond to the center of the anodes on the Planacon. Since it is not possible to have 64 holes machined out, the mask only has 30 holes to uniformly sample the face of the Planacon as shown in Fig. 4.

Fig. 5 shows a schematic of the experimental setup, which is used to measure the energy and timing spectra of scintillation detectors. In the measurement, a Ge-68 point source is placed between a reference detector and the detector under test (a LSO crystal coupled to the Planacon) to excite them with annihilation 511 keV photons. The reference detector is a 10 mm cube BaF_2 scintillator coupled to a Hamamatsu H-5321 PMT assembly operated at -2300 V. It is mounted on an X-Y translation stage to scan across the face of the Planacon. The timing resolution of the reference detector is measured to be 150 ps FWHM. The anode signal from the Planacon is read out with a fast amplifier with a gain of 5 (Phillips Scientific

775). Both the anode signal from the reference detector and the amplified Planacon signal are read out with constant fraction discriminator modules (Canberra 454 NIM CFD) triggered at 20% of the pulse amplitude. The delay on the CFD is set at 0.6 ns, which yields the best timing resolution for the reference detector. The CFD for reading out the Planacon signal is modified for microchannel plate, which includes changing the internal circuit wiring to optimize the internal delay and minimize the reflection for high bandwidth signal (these modifications was done according to the manufacturer guidelines). The outputs of the CFDs are sent to a time-to-digital converter (Ortec 566 NIM TAC) to obtain a coincidence timing spectrum. The Planacon signal is also amplified by a shaping amplifier and its amplitude is digitized with an analog-to-digital converter (ADC) to obtain a pulse height spectrum. The ADC is part of a multifunction data acquisition module from National Instrument (NI PCI-7833R), which is controlled and read out by a personal computer. Only photopeak events defined to be two FWHM wide centered on the photopeak are considered in the final coincidence timing spectrum. The timing resolution of the detector under test is computed by taking the measured coincidence timing resolution and subtracting (in quadrature) the 150 ps FWHM contribution from the reference detector. For comparison, we also make measurements by coupling the same $4\text{ mm} \times 4\text{ mm} \times 10\text{ mm}$ LSO crystal to the R-9800.

C. Detector Simulation

As shown in Fig. 6, a Monte Carlo analysis is used to model the timing response of the detector, which is described in detail in [15]. A normalized photoelectron emission rate for a $4\text{ mm} \times 4\text{ mm} \times 10\text{ mm}$ LSO crystal at the photocathode of the PMT is obtained by modeling the scintillator response, which includes the generation of the scintillation photons inside the crystal and the subsequent tracking of the individual photons until they exit the crystal to be detected by the PMT. The intrinsic rise time and decay time of LSO is taken to be 0.03 ns and 40 ns respectively [16]. The depth of the generated photons from the entrance follows an exponential distribution given by the interaction length of a 511 keV gamma ray in LSO, which is 12 mm. The tracking of the individual photons inside the crystal uses the optical processes in Geant4 [17]. In addition, the UNIFIED model implemented in Geant4 is employed to model the etched surface finish and to simulate the boundary process of scintillation photons between two dielectric media. Next, a Poisson process is used to generate the individual photoelectron time points from the photoelectron emission rate normalized to a specific mean photoelectron yield. The output signal of the PMT is a convolution of the generated photoelectrons with the SER of the PMT. The SER of the Planacon and R-9800 is taken from the measurements. Detector noise is modeled as a Gaussian distribution and is added to the output signal. The calculated timing spectrum of the detector is determined by applying a constant fraction discriminator to 100,000 generated output signals.

III. RESULTS

A. Single Electron Response

Fig. 7a and 7b show the single photoelectron waveforms averaged over 1024 pulses for the XA85015 and R-9800 respectively. Both the waveforms exhibit fairly symmetrical shapes. The XA85015 pulse has the fastest response with a rise time of 440 ps and a pulse duration of ~ 1 ns. On the other hand, the R-9800 pulse has a rise time of 590 ps and a pulse duration of ~ 2 ns. The detector noise for both the PMTs is estimated to be 1–2 mV rms by observing the amplitude fluctuation on the baseline.

Fig. 8a and 8b show the coincidence timing spectra of the XA85015 and R-9800 respectively when excited with the pulsed laser diode. While the Planacon timing spectrum

has a long tail due to recoil photoelectrons from the front surface of the MCP [18–19], it has the fastest timing response with a TTS of 120 ps FWHM. On the other hand, the TTS of the R-9800 is 260 ps FWHM. By measuring the timing resolution with just the trigger output from the pulse laser diode, the timing resolution of the combined pulsed laser and readout electronics is estimated to contribute about 77 ps FWHM to the TTS measurements.

B. Gain Uniformity

Fig. 9 shows the measured center of the 511 keV photopeak, which is used as an estimate of the relative gain response of the XA85015. The gain is relatively uniform in the central anodes with an average value of 113 ± 10 . The gain drops off to an average value of 78 ± 4 along the edges and 53 ± 5 at the corners. The drop in gain along the edges and corners are likely due to the electric field distortion along the boundary, gain variation, and photocathode quantum efficiency variation. For example, along the boundary, all the electric field lines do not terminate at the anode, thus reducing the charge collection efficiency. We have not verified the dominant cause of the variation, which is beyond the scope of this paper.

The gain of one of the central anode of XA85015 is measured as a function of the counting rate. It is well known that LSO exhibits an “after-glow”, which increases the scintillation rate, after the LSO crystal is exposed to room light. This after-glow is caused by charge trapping inside the crystal and can persist for several hours [20]. In order to increase the counting rate, the LSO crystal is first exposed to room light before coupling it to the XA85015 and exciting it with 511 keV photons. The center of the photopeak position and the counting rate is measured as a function of time as the afterglow decays away. The counting rate, which is slightly underestimated because of an applied threshold on the amplitude of the signal, is normalized to the cross-section area of the crystal that is coupled to the XA85015 to obtain the flux. Fig. 10 shows the center of the photopeak position as a function of the flux. While the gain is observed to drop with increasing flux, a reasonably stable gain is obtained below 2×10^4 cps/cm².

C. Energy Resolution

Fig. 11 shows the energy resolution measured with the XA85015. In contrast to the gain uniformity, the energy resolution is relatively uniform across the face of the XA85015 with an average value of $18.6 \pm 0.7\%$ FWHM. While the gain drops along the edges as discussed previously, the signal-to-noise ratio remains approximately the same leading to better uniformity in the energy resolution than in the gain. For comparison, the R-9800 gives a better energy resolution with a value of 15% FWHM. This is due to the fact that the photocathode quantum efficiency (Blue Sensitivity Index of 13 for R-9800 as opposed to 8.5 for the Planacon) on the R-9800 is significantly higher than the Planacon. In addition, the MCP in XA85015 has an open area ratio of 70%, which reduces the charge collection efficiency. As a result, the PDE of the R-9800 is significantly higher than the Planacon, resulting in higher signal-to-noise ratio.

D. Timing Resolution

Fig. 12 shows the timing resolution measured with the XA85015. Similar to the gain uniformity, the timing resolution is relatively uniform in the central anodes with an average value of 252 ± 7 ps FWHM. The timing resolution degrades to an average value of 280 ± 9 ps FWHM along the edges and 316 ± 15 ps FWHM at the corner. For similar reasons as to the gain non-uniformity, the non-uniformity of the timing resolution, especially between the central region and along edges, is due to the lower PDE along the edges of the XA85015. The timing resolution is very dependent on the initial photoelectron rate, which determines the timing of the detector. A higher PDE would yield a higher initial photoelectron rate,

hence a better timing resolution. For comparison, the R-9800 gives a better timing resolution with a value of 221 ps FWHM because it has a significantly higher PDE than the XA85015. Even though the XA85015 has faster SER and better TTS, its timing resolution is limited by the photoelectron statistics.

E. Calculated Timing Resolution

In the detector simulation, we calculated the timing resolution as a function of the mean photoelectron yield ranging from 1000 to 8000 photoelectrons. Based on analytical models as well Monte Carlo simulations and experimental data [15,21–24], it has been shown that the timing resolution scales inversely to the square root of the initial photoelectron rate defined as the mean photoelectron yield divided by the decay lifetime of the scintillator. Fig. 13 shows the calculated timing resolution as a function of the initial photoelectron rate plotted on logarithmic scales. Two lines are plotted in the figure, one corresponding to the XA85015 and the other to R-9800. Because of the faster SER and lower TTS of the XA85015, the calculated timing resolution is better than the R-9800 for a given initial photoelectron rate. However, since the PDE of the R-9800 is significantly higher than the XA85015, the timing resolution of the R-9800 can be better than the XA85015, which validates the measured results.

IV. DISCUSSION AND CONCLUSIONS

We investigated the performances of a multi-anode MCP-PMT as a PET detector for TOF applications and compare the results with a fast conventional PMT. Detector simulation was also performed using the measured SER to calculate the timing resolution as function of the initial photoelectron rate.

The gain is relatively uniform across the central region of the XA85015, but drop off to below 70% of the average gain in the central region along the edges. The gain uniformity across the face of the XA85015 is $\sim 1:2.5$. Improvement in the field distortion along the boundary of the XA85015 would be needed to improve the gain non-uniformity.

The gain of the XA85015 is also observed to depend on the counting rate. This is due to the limit of the MCP strip current, which is $\sim 3 \mu\text{A}$. According to the manufacturer, the gain is stable up to a few percent of the strip current. The gain is observed to be relatively stable below $2 \times 10^4 \text{ cps/cm}^2$. A typical whole-body PET camera with an 80 cm diameter detector ring and a 10 mCi (current maximum clinical protocol level) source at the center is estimated to yield an activity flux of about $1.8 \times 10^4 \text{ cps/cm}^2$ at the detector surface. In addition, the natural radioactivity background count rate of a 20 mm deep LSO crystal is only about 500 cps/cm². Thus, the count rate capability of the XA85015 seems to meet the requirement of a clinical PET detector. A factor of 5 to 10 improvement in the count rate capability of the XA85015 is expected with higher MCP strip current using lower resistance MCP.

The energy resolution and timing resolution of the XA85015 are observed to be poorer than the R-9800. While the energy resolution is relatively uniform across the face of the XA85015, the timing resolution degrades along the edges similar to the gain. These results confirm that the PDE of the XA85015 is poorer than the R-9800 as indicated by the their photocathode quantum efficiency (Blue Sensitivity Index of 13 for R-9800 as opposed to 8.5 for the Planacon).

The initial photoelectron rate can be estimated by overlaying the measured timing resolutions with the XA85015 and R-9800 on the corresponding lines in Fig. 13. The average timing resolution of the central anodes of the XA85015 is used. As expected, the

initial photoelectron rate for the XA85015 (18 photoelectrons / ns) is significantly lower than for the R-9800 (34 photoelectrons / ns) because of the lower PDE of the XA85015. As a validation, we estimate the initial photoelectron rate based on the light output of LSO, the PDE of the PMT and the light collection efficiency of the crystal. The light yield of LSO is about 25,000 photon / MeV [25]. For a 511 keV photon and an effective PDE of 20% (even though the peak quantum efficiency of the super bialkali photocathode is about 30%, the effective PDE is lower when averaged over the emission wavelengths of the scintillation light), approximately 2500 photoelectrons would be generated. This has to be scaled by the light collection efficiency of the crystal, which is calculated to be 0.6 for 4 mm × 4 mm × 10 mm LSO crystals using Monte Carlo simulation. Using a decay lifetime of 40 ns, the calculated initial photoelectron rate is approximately 38 photoelectrons / ns, which agrees reasonably well with the initial photoelectron rate as estimated for the R-9800 in Fig 13. The calculated timing resolution also suggests that significantly better timing resolution can be achieved with the XA85015 than the R-9800 if the PDEs are similar. On the other hand, only a 30% improvement in the PDE of the XA85015 is needed to achieve similar timing resolution as the R-9800.

The performance results of the Planacon, a multi-anode MCP PMT, are very encouraging. The Planacon can potentially provide a viable solution to realize a practical PET detector with a very good timing resolution for TOF PET imaging. However, current version of the Planacon has problems with gain uniformity along the edges and low PDE. We believe these problems are not insurmountable. In particular, the PDE can be improved by applying higher quantum efficiency photocathode as well as higher MCP open area ratio to improve the collection efficiency. We expect improvement in future version of the Planacon.

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Fig. 1.
Components of (a) PET block detector and (b) a scintillation detector with the crystal directly coupling to it.

Fig. 2.
Photograph of the Planacon MCP-PMT (XA85015)

Fig. 3. Recommended voltage divider from the manufacturer (diagram taken from the Planacon specification sheets).

Fig. 4.
Front view of the mask for the Planacon. The mask has 30 holes to place the crystal at the center of the anode.

Fig. 5.
Schematic diagram of the experimental setup for the detector measurement.

Fig. 6.
Block diagram of Monte Carlo timing analysis.

Fig. 7.
Measured SER waveforms of (a) XA85015 and (b) R-9800 averaged over 1024 pulses.

Fig. 8.
Measured transit time spectra of (a) XA85015 and (b) R-9800.

Fig. 9.
Measured photopeak center across the face of XA85015.

Fig. 10.
Measured photopeak center as function of flux for one of the central anode on XA85015.

Fig. 11.
Measured energy resolution across the face of XA85015.

Fig. 12.
Measured timing resolution across the face of XA85015.

Fig. 13. Calculated timing resolution as a function of initial photoelectron rate for XA85015 (solid line) and R-9800 (dashed line).