MCP-PMT timing at low light intensities with a DRS4 evaluation board

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Abstract
Positron emission tomography (PET) is one of the most important diagnostic tools in medicine, allowing three-dimensional imaging of functional processes in the body. It is based on a detection of two gamma rays with an energy of 511 keV originating from the point of annihilation of the positron emitted by a radio-labeled agent. By measuring the difference of the arrival times of both annihilation photons it is possible to localize the tracer inside the body. Gamma rays are normally detected by a scintillation detector, whose timing accuracy is limited by a photomultiplier and a scintillator. By replacing a photo sensor with a microchannel plate PMT (MCP-PMT) and a scintillator with Cherenkov radiator, it is possible to localize the interaction position to the cm level. In a pioneering experimental study with Cherenkov detectors using PbF_2 crystals and microchannel plate photomultiplier tubes MCP-PMT a time resolution better than 100 ps was achieved. In this work a DRS4 digital ring sampler chip was used to read out single photon output signals from two different MCP-PMTs (Hamamatsu R3809 and Burle 85001) with a sampling rate of 5×10^9 samples/s. The digitized waveforms were analyzed and a comparison between the two detectors timing response was made. The time resolutions achieved were (161 ± 2.21) ps and (220 ± 2.63) ps FWHM for the Hamamatsu and Burle MCP-PMT respectively. No significant variances were observed in the study of the behavior of the FWHM when both MCP-PMT were scanned.

Key words: *positron computed tomography; Cherenkov counters; microchannel electron multipliers; photomultipliers.*

Estudio de la respuesta temporal de dos MCP-PMT a bajas intensidades de la luz utilizando un DRS4

Resumen
La tomografía por emisión de positrones (PET) es una importante herramienta en el diagnóstico médico ya que permite la obtención de imágenes tridimensionales de los procesos funcionales en el cuerpo. La técnica está basada en la detección de los dos cuantos gamma de 511 keV originados en la aniquilación del positrón emitido por el radiofármaco administrado al paciente. Midiendo la diferencia en la llegada de los dos cuantos gamma es posible determinar la posición en la que ocurrió la aniquilación. En los equipos convencionales son utilizados detectores centellantes cuya respuesta temporal está limitada por el fotomultiplicador y el cristal centellante. Remplazando el fotomultiplicador por un PMT (MCP-PMT) y el cristal centellante por un detector Cherenkov, es posible localizar la posición en la que ocurrió la aniquilación con una exactitud a nivel de pocos centímetros. En previos resultados experimentales utilizando detectores Cherenkov con cristales de PbF_2 y MCP-PMT se alcanzó una respuesta temporal de menos de 100 ps. En este trabajo fue utilizado un chip DRS4 con una velocidad de procesamiento de las señales de 5×10^9 samples/s para la lectura de la salida de fotones únicos de los dos MCP-PMT estudiados (Hamamatsu R3809 y Burle 85001). Las señales digitalizadas fueron analizadas y se realizó una comparación entre la respuesta temporal obtenida para ambos MCP-PMT. El tiempo de respuesta obtenido en términos de FWHM fue de (161 ± 2.21) ps y (220 ± 2.63) ps para los MCP-PMT Hamamatsu y Burle respectivamente. No se detectaron variaciones significativas en el FWHM al escanearse la superficie activa de ambos MCP-PMT .

Palabras clave: *medicina nuclear; Cuba; cámaras gamma; fotomultiplicadores; amplificadores.*

Introduction

Positron emission tomography (PET) is a powerful and sensitive technique in nuclear medicine, allowing three-dimensional imaging of functional processes in the body. It is based on a detection of two gamma rays with an energy of 511 keV originating from the point of annihilation of the positron emitted by a radio-labeled agent. By measuring the difference of the arrival times of both annihilation photons it is possible to localize the tracer inside the body [1]. Each registered coincident 511 keV pair is referred to as an event. The contrast of the image can be further improved by reducing the fraction of background combinations, mostly from random coincidences and from events where gamma rays scatter in the tissue. In a standard (PET), noise contributes uniformly along the line of response since the annihilation point position along this line is not known. This location can be determined by measuring the time difference between the times when two gamma rays were detected, but this requires considerable timing precision. If we want to determine the annihilation point position with and accuracy of 1 cm, we have to measure the time difference with an accuracy of 66 ps. Currently, the best accuracy for the (TOF-PET) is between 300 and 400 ps, which significantly improves the contrast in the large objects imaging [2]. The scintillator crystal requires about 100 ps to reach a maximum light output and then, decreases exponentially with a decay time of 10 ns. With a state of fast light sensors and read-out electronics, the main limitation in time resolution is given by the response time of the scintillator. With such a timing, it is necessary to detect a very large number of photons for a time resolution better than 100 ps.

The search for new materials and mechanisms of light emission is one possible direction for improvement, being followed in particular by several strong research groups in Europe [3]. To significantly improve time resolution of detecting gamma rays in a (PET) device it is necessary to explore new ways of light production, in which as soon as the gamma ray interacts in the detector, the photons are emitted. One possibility is to use Cherenkov light emitted by electrons moving in a material with velocities exceeding the speed of light in the material. Such fast electrons are produced in the interaction of gamma ray through the Photoelectric Effect or Compton Scattering.

In pioneering experimental study with Cherenkov detectors a time resolution better than 100 ps was achieved using PbF_2 crystals and (MCP-PMT). For a Photonis (successor to Burle) model XP85015/A1 (MCP-PMT), the timing resolution was measured with σ = 63 ps at single photon illumination [4]. The crystals of PbF_2 have a further advantage over conventional ones scintillator materials. Due to the large atomic number of lead, the probability of Photoelectric Effect increase and therefore the probability that all gamma ray energy be absorbed in a single interaction with matter. The main problem was the low detection efficiency for annihilation gamma rays as a result of the small number of emitted Cherenkov photons and the relatively low efficiency of photo-sensors. To use this method in a (PET) device the efficiency of gamma ray detection needs to be improved because just few Cherenkov photons are emitted per interaction. It is necessary a photo sensor with a good time resolution and a high efficiency detecting individual photon [5]. In addition to the (MCP-PMT) discussed above, a Silicon Photomultiplier (SiPM) is also suitable as a sensor. It has a considerably better detection efficiency and also a relatively large dark count rate, that needs to be reduced by cooling [6]. The Burle MCP-PMT performed very well as a single photon detector although the number of photons per track is still too small. There are two possible improvements according to the producer, namely the increase of active area up to 85 %, and the increase of the photoelectron collection efficiency from 60 % to 70 % [7].

In this work a DRS4 digital ring sampler chip [8, 9] was used to read out single photon output signals from two different MCP-PMTs (Hamamatsu R3809 and Burle 85001). The digitized waveforms were analyzed and a comparison between the two MCP-PMTs timing response was made. This paper is organized as follows: *Section 2* describes the materials and methods, *Section 3* deals with the discussion of the results and *Section 4* shows the conclusions. The main objective of this work is to study experimentally the time response for two MCP-PMTs in addition to analyzing if there are significant variances for the FWHM over their active zone.

Materials and methods

In our experiment, two MCP-PMTs (Hamamatsu R3809 and Burle 85001), were operated at 3200 V and 2350 V respectively. We are using a setup (figure 1a) in which we have a picosecond PiLas laser with blue (406nm) head. The laser light was first attenuated by neutral density filters and guided into light tight box along the optical fiber, with the far end attached to a computer controlled stage. The repetition rate of light pulses used as a trigger signal was 1kHz. In our experimental setup, the signal is also attenuated before being amplified by an Ortec FTA820. The pulse from the detector should be pre-amplified in order to achieve better performance using digital timing method [7]. The MCP-PMTs are mounted (figure 1b) in such a way that allows moving the laser incidence position from one MCP-PMT to the other one without changing the

MCP-PMT position. The DRS4 chip used it is basically equivalent to a four channel 5×10^9 samples/s digital oscilloscope. For each channel the analog waveforms are stored with a sampling speed up to 5×10^9 samples/s in a ring buffer composed by 1024 sampling cells (capacitors). After all the waveforms signal processing steps were completed, the Constant Fraction time is calculated using the following expressions:

$$\Delta Cf_t(Channel) = Cf_t(Channel) - Cf_t(TriggerSignal) \quad (1)$$

$$Cf_t(Channel) = t_n + \frac{(t_0 - t_n) * (Y_{cf} - Y_{i-1})}{(Y_i - Y_{i-1})} \quad (2)$$

where t_n: is the time sample after the Threshold and before a certain definite value (Cfrac) fixed at 40 % of the maximum voltage of the signal in each sample. t_0 is the time sample after a (Cfrac), and Y_i, Y_{Cf}, Y_{i-1} are the digitized voltage sample before, during and after the (Cfrac). The algorithm used for calculating the Constant Fraction time is shown in figure 2. It also shown how the digitized signal looks like.

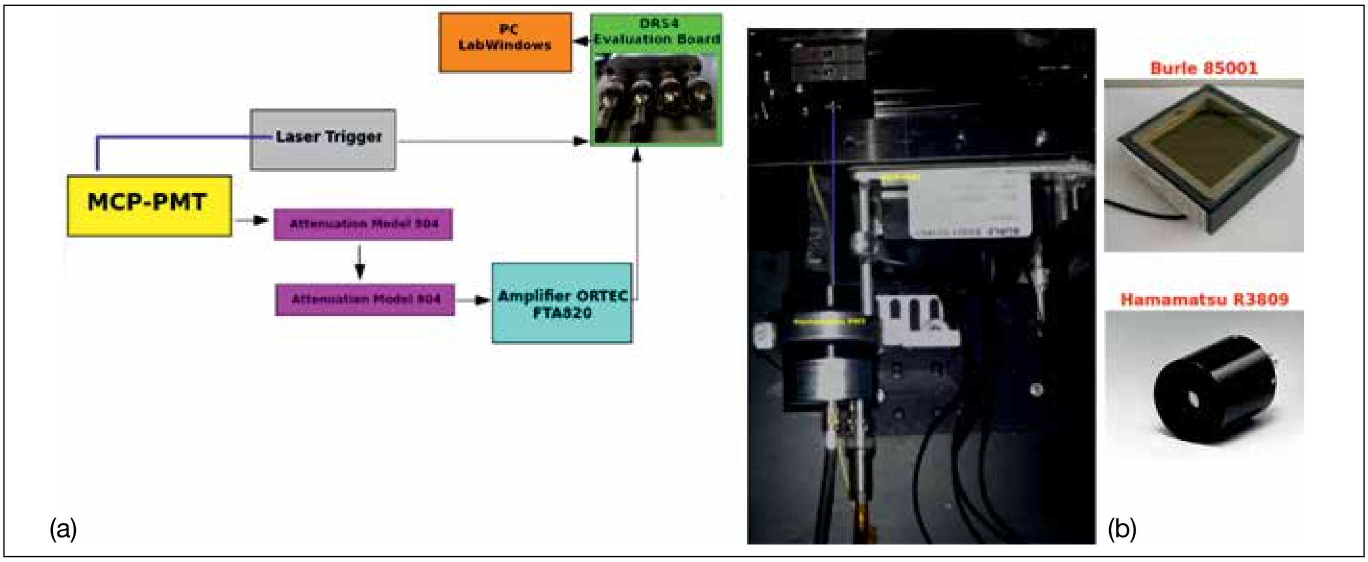

Figure 1. (a) Setup diagram of the Experimental setup. (b) Parallel Mount of the Hamamatsu and Burle MCP-PMT.

For the (Cfra) different cut-off values between 10 and 80 % were tested before selecting the fixed value of 40 % due to the shape of the distribution obtained, the FWHM and the maximum number of counts in the main peak. All the results that will be presented in the next section use the method shown in figure 2 taking into account two consecutive points for the calculation of the Constant Fraction time and fixing during the analysis the (Cfrac) at 40 % of the maximum voltage of the signal in each sample.

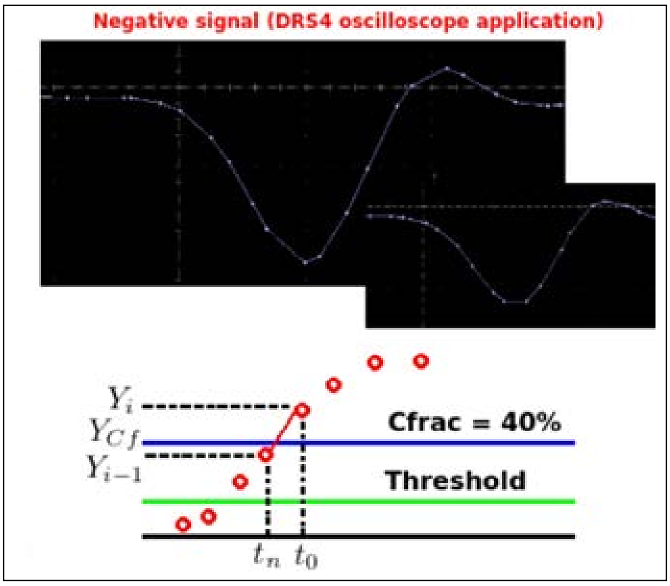

Figure 2. Constant Fraction time calculation method.

Results

With the aim of describing in the best way the shape obtained for the Constant Fraction time distribution, different fitting functions were tested and it was decided to use three Gaussian owing to the fact that it offers a reliable value for the full-width-half-maximum (FWHM) and a better description of the entire distribution not only of the main peak. Different ways to calculate the FWHM, were tested being considered the dependent one of the fitting functions the most appropriate. Figure 3a and figure 3b show for each MCP-PMTs the obtained shape of the Constant Fraction time distribution and the FWHM of (161 ± 2.21) ps and (220 ± 2.63) ps for the Hamamatsu and Burle MCP-PMT respectively.

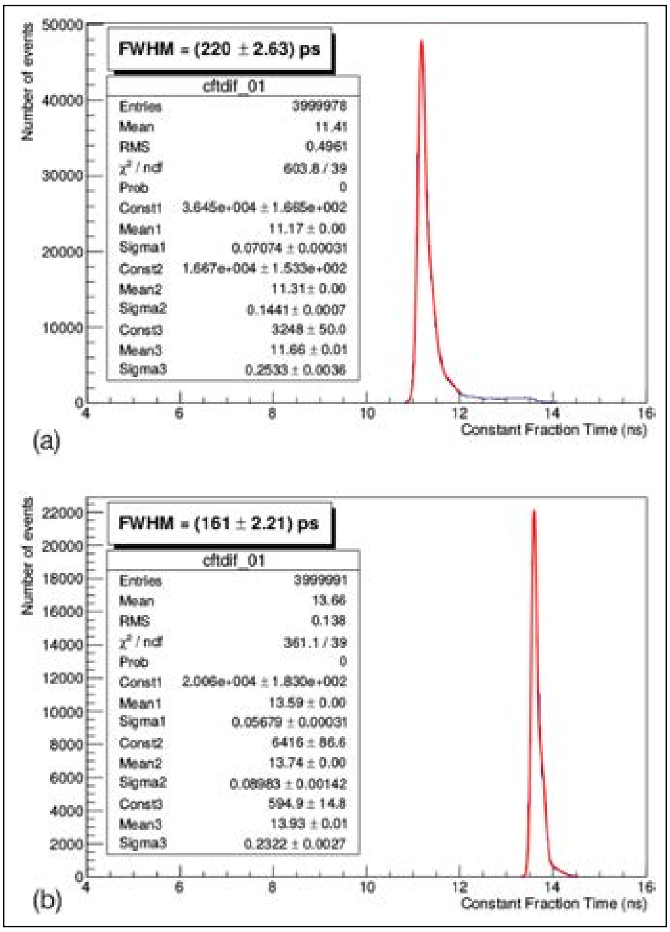

Figure 3. (a) Constant Fraction time Burle 85001. (b) Constant Fraction time Hamamatsu R3809.

The FWHM is dependent in each case from the fitting functions. In order to evaluate how the time response changes over the surface, a 1-D scan (359.5 µm each step) was measured of the MCP-PMT. Figures 4-5 show the results obtained for each case. The scans for the Burle MCP-PMT (100 steps, approximately 3.6 cm) were made on the (X-axis) and for the Hamamatsu MCP-PMT (45 steps, 1.6 cm) on the (Y-axis). A Constant Fraction time distribution with a million of events for each step reduces the statistical error. For the case of Burle MCP-PMT the FWHM stays above 200 ps and for the Hamamatsu MCP-PMT the FWHM is above 150 ps and less than 200 ps.

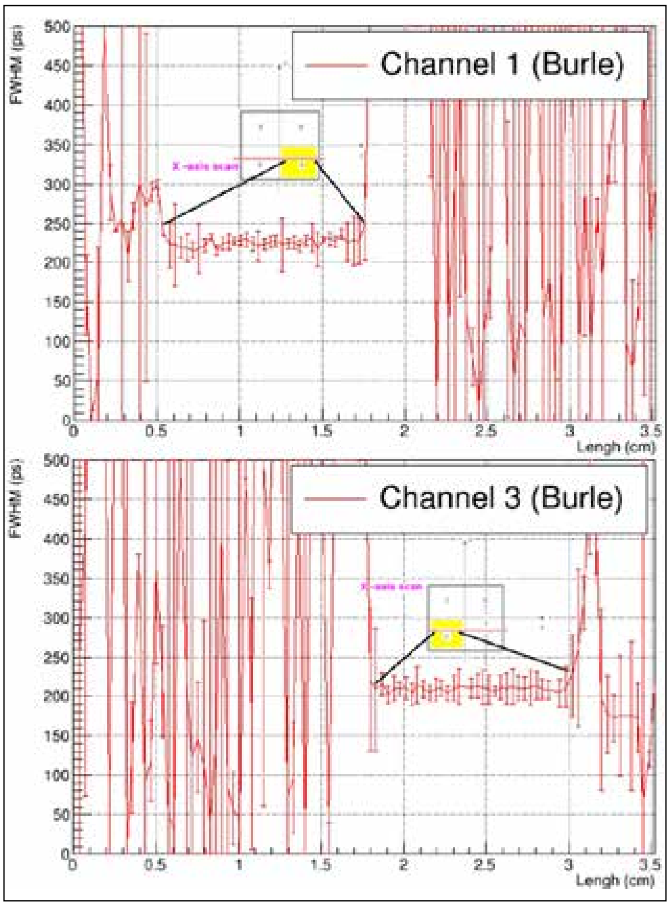

Figure 4. Channel1-Channel3 (two neighboring channels in the direction of the scan) scan for MCP-PMT.

The measured FWHM is mostly constant over the active surface (from approximately 0.6 - 1.7 cm and from 1.8 - 3 cm in figure 4; from approximately 0 - 1.2 cm in figure 5). When the laser illumination falls outside of the active surface of the MCP-PMT, the fit does not converge. The scan across the upper two channels for the Burle MCP-PMT reports the same results of the scan across the lower two ones. The standard deviation for the FWHM over the active surface was 2.92 ps for Hamamatsu and 4.82 ps for Burle Channel 1, 4.62 ps for Burle Channel 3. It can be said that the FWHM does not vary substantially over the MCP-PMT active surface.

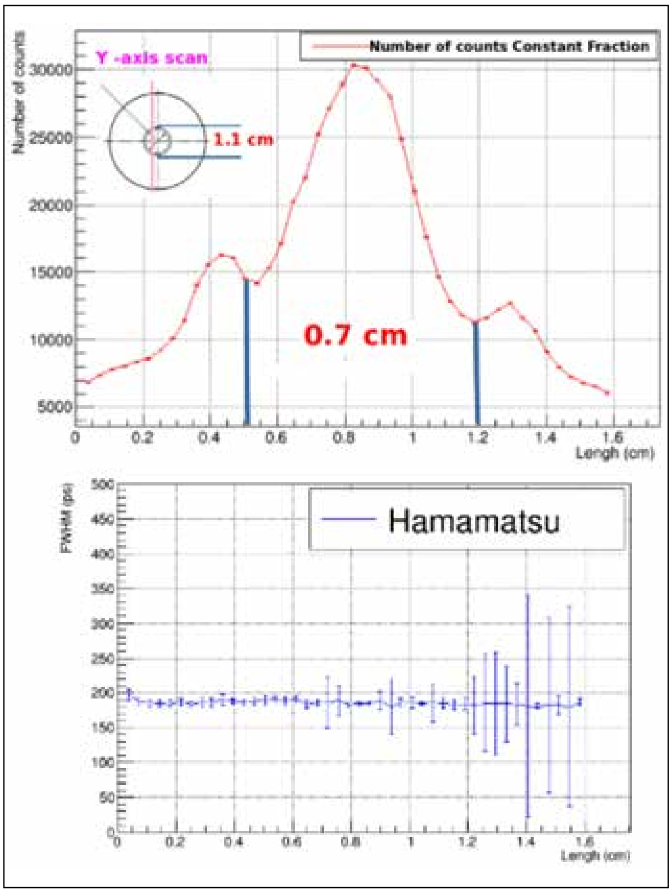

Figure 5. Hamamatsu MCP-PMT scanner (Y-axis).

Conclusions

In this work two different MCP-PMTs (Hamamatsu R3809 and Burle 85001) were studied using a digital ring sampler chip DRS4 to read out single photon output signals. After the waveforms were digitized and analyzed the best time resolutions achieved were (161 ± 2.21) ps and (220 ± 2.63) ps FWHM for the Hamamatsu and Burle MCP-PMT respectively. No significant variances were observed for the FWHM over the active surface of the MCP-PMTs. With these results using a DRS4 the FWHM is limited to (160 - 220) ps. It is necessary a more suitable compromise between efficiency and the FWHM time resolution for the MCP-PMTs before studying with ^{22}Na source different Cherenkov crystal materials as (PbF_2 , PWO_4 and lead glass) using gamma rays with an energy of 511 keV originating from the point of annihilation.

References

[1] HUMM JL, ROSENFELD A & DEL GUERRA A. From PET detectors to PET scanners. Eur J Nucl Med and Mol Imag. 2003; 30(11). http://www.iss.infn.it/topem/TOF-PET/from%20pet%20detectors%20to%20pet%20scanners%20(EJNM).pdf.

[2] ZHOU L, DEFRISE M, VUNCKX K & NUYTS J. Comparison between parallel hole and rotating slat collimation: analytical noise propagation models. IEEE Transactions on Medical Imaging. 2010; 29(12). http://ieeexplore.ieee.org/stamp/stamp.jsp?arnumber=5523954&tag=1.

[3] LECOQ P. Development of new scintillators for medical applications. Nucl Instrum and Meth in Phys Res A. 2016; 809. https://www.sciencedirect.com/science/article/pii/S0168900215009754.

[4] DOLENEC R, CHANGANI H, KORPAR S, et. al. Time-of-flight with photonics multi-channel MCP-PMT using MCP signal. IEEE Nuclear Science Symposium Conference Record. 2009. http://ieeexplore.ieee.org/stamp/stamp.jsp?arnumber=5402278.
[5] KORPAR S, DOLENEC R, KRIŽAN P, PESTOTNIK R & STANOVNIKD A. Study of TOF PET using Cherenkov light. Physics Procedia. 2012; 37: 1531-1536. http://www.sciencedirect.com/science/article/pii/ S1875389212018718.
[6] DOLENEC R, KORPAR S, KRIŽAN P & PESTOTNIK R. Cherenkov TOF PET with silicon photomultipliers. Nucl Instrum and Meth in Phys Res A. 2015; 804: 127-131. https://www.sciencedirect.com/science/article/pii/S0168900215011146.
[7] KRIŽAN P, ADACHI I, FRATINA S, et. al. Tests of the Burle 85011 64-anode MCP-PMT as a detector of Cherenkov photons. Nucl Instrum Meth in Phys Res A. 2006; 567: 124-128. http://www.sciencedirect.com/science/article/pii/S0168900206008898.
[8] BITOSSI M, PAOLETTI R & TESCARO D. Ultra-fast sampling and data acquisition using the drs4 waveform digitizer. IEEE Transactions on Nucl Science. 2016; 63. http://ieeexplore.ieee.org/stamp/stamp.jsp?arnumber=7488281.
[9] RAN A, BIN C, Y-FEN L, WEI K & RITT S. A new positron annihilation lifetime spectrometer based on DRS4 waveform digitizing board. Nucl Instrum and Meth in Chinese Physics C. 2014; 38. http://staff.ustc.edu.cn/~bjye/ye/papers/1674-1137_38_5_056001.pdf.
[10] HU W, CHOI Y, HONG K, et. al. A simple and improved digital timing method for positron emission tomography. Nucl Instrum and Meth in Phys Res A. 2010; 622: 219-224. http://med.stanford.edu/miil/publications/files/151_PUB.pdf.
[11] SURTI S. Update on time-of-flight PET imaging. J Nucl Med. 2015; 56: 98-105. https://www.ncbi.nlm.nih.gov/pmc/articles/PMC4287223/pdf/nihms652988.pdf.
[12] AUFFRAY E, FRISCH B, GERACI F, GHEZZI A, et. al. A comprehensive systematic study of coincidence time resolution and light yield using scintillators of different size and wrapping. IEEE Transactions on Nuclear Science. 2013; 60: 3163. http://ieeexplore.ieee.org/stamp/stamp.jsp?arnumber=6566141

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