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A ROOM TEMPERATURE LSO / PIN PHOTODIODE PET DETECTOR MODULE THAT MEASURES DEPTH OF INTERACTION*

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Abstract

We present measurements of a 4 element PET detector module that uses a 2x2 array of 3 mm square PIN photodiodes to both measure the depth of interaction (DOI) and identify the crystal of interaction. Each photodiode is coupled to one end of a 3x3x25 mm LSO crystal, with the opposite ends of all 4 crystals attached to a single PMT that provides a timing signal and initial energy discrimination. Each LSO crystal is coated with a "lossy" reflector, so the ratio of light detected in the photodiode and PMT depends on the position of interaction in the crystal, and is used to determine this position on an event by event basis. This module is operated at +25°C with a photodiode amplifier peaking time of 2 μs. When excited by a collimated beam of 511 keV photons at the photodiode end of the module (i.e. closest to the patient), the DOI resolution is 4 mm full width at half maximum and the crystal of interaction is identified correctly 95% of the time. When excited at the opposite end of the module, the DOI resolution is 13 mm full width and the crystal of interaction is identified correctly 73% of the time. The channel to channel variations in performance are minimal.

1. INTRODUCTION

It is becoming increasingly important to develop a PET detector module that is capable of measuring the distance that the 511 keV annihilation photons penetrates into the detector ring before interacting (i.e. the depth of interaction) on an event by event basis. This penetration leads to a resolution degradation artifact (known as radial elongation) whose seventy increases as the angle of incidence of the annihilation photon increases (where the angle of incidence is defined as the angle between the direction of travel of the photon and the normal to the front surface of the detector module). While this artifact is barely noticeable in whole body PET cameras (ring diameter ≥80 cm), it causes significant degradation towards the edge of the field of view in cerebral cameras (ring diameter 50–80 cm), dominates the resolution in small animal PET cameras (ring diameter <50 cm), and could be a "show-stopper" for disease specific PET camera geometries where the object to be imaged fills the detector aperture, such as in Positron Emission Mammography cameras [1, 2].

We have previously characterized [3] a PET detector module that measures depth of interaction on an event by event basis using the method shown in Figure 1. The module consists of

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and on the opposite end to a 2 by 2 array of 3x3 mm silicon photodiodes. The PIN photodiodes are Hamamatsu S-2506 (2.77 mm square active area, 100 μm depletion thickness) devices mounted in a special package to allow close coupling to the scintillator crystal. The “lossy” reflector is made by sanding the LSO crystal before applying a white reflective coating. For all subsequent, the photodiode is biased with +80 V and the assembly operated at room temperature (+25 °C). Under these operating conditions, the capacitance is 9 pF and the dark current is <100 pA.

A custom integrated circuit amplifier array [6] with a 2 μs peaking time processes the photodiode signal, and the noise (520 electrons (e⁻) fwhm) is measured using a calibrated test pulse. This is significantly larger than the 330 e⁻ fwhm noise achieved earlier [3] due to the higher operating temperature and shorter shaping time used in the present work.

This test module is illuminated with an electronically collimated beam of annihilation photons from a 68Ge source, as shown in Figure 2. The position of the beam (in 2 dimensions) is varied by moving the entire collimation apparatus, allowing a 2.5 mm fwhm portion of the test module to be excited at an arbitrary depth of interaction. The depth coordinate system is chosen such that 0 mm corresponds to the end of the LSO crystal closest to the photodiode and 25 mm is the end closest to the photomultiplier tube.

3. EXPERIMENTAL MEASUREMENTS

The detector module must be calibrated (on a crystal by crystal basis) in order to determine how the ratio between the photodiode and photomultiplier tube signals depends on interaction depth. To do this, the module is excited from the photodiode end of the LSO crystal (i.e., closest to the patient), but the photomultiplier tube signal produces different measured signals in the 5 photodetectors (4 photodiodes and 1 photomultiplier tube), or

\[ \text{PDT}_i = L \cdot k_i, \]

where \( L \) is the amount of light impinging on the photodetector. In order to compare the measured photodiode and photomultiplier tube outputs and compute a position estimator, it is necessary measure the ratios \( K_i = \frac{PMT(22)}{PD_i(3)} \).

\[ \text{PMT}(22) \]

where \( K_i \) is the amount of light impinging on the photodiode when excited by 511 keV photons at a depth of \( x \) mm. We then obtain the gain normalization factor by dividing

\[ K_i = \frac{\text{PMT}(22)}{\text{PD}_i(3)}. \]

\( K_i \) is the amount of light impinging on the photodiode when the detector is excited at a depth of 22 mm and PD(3) is the 511 keV photopeak centroid in photodiode i when the detector is excited at a depth of 3 mm. The photodetector signals are then be converted into a equal energy scale and directly compared using these \( K_i \).

The sharing of the signal between the photomultiplier tube and the photodiode has consequences for the trigger. A trigger threshold at a fixed energy deposit is desired, but only the photomultiplier tube output is available to the trigger. Therefore, a preliminary trigger must be made based on the photomultiplier tube, then the photomultiplier tube and photodiode are read out and converted into a equal energy scale (using the method described above) before a total energy deposit threshold is applied to the sum. In these tests, the photomultiplier tube trigger threshold corresponds to 250 keV energy deposit when the test module is excited at 2 mm depth and 50 keV when excited at 23 mm depth, and an energy threshold of 250 keV applied to the summed signals.

To compute the crystal misidentification fraction, the height of the moveable stage in Figure 2 is adjusted (i.e., in or out of the page) so that the beam of 511 keV photons detector array is incident on either the lower or upper pair of LSO crystals. If a coincidence between the photomultiplier tubes is detected, the 4 signals observed in the photodiode array and in the “test module” photomultiplier tube are simultaneously digitized, read into a computer, normalized, and the photodiode with the largest signal (after applying the gain correction factors \( K_i \)) considered to be the crystal of interaction. This identification is considered “correct” if one of the two crystals in the path of the beam was labeled as the crystal of interaction. When excited at a depth of 3 mm (the patient end), the
The interaction position is measured on an event by event basis by computing a position estimator, defined as the fraction of the summed output from the two photodetectors that is observed by the photodetector, or \( \Gamma = \frac{PD}{PD+PMT} \). Here PD is the largest pulse height (after normalization) observed in the photodiode array and PMT is the re-scaled pulse height observed by the photomultiplier tube. Figure 3 plots this ratio for all 4 photodiode channels with the test module excited at interaction depths of 3 and 22 mm. This plot shows that the centroid of the estimator depends strongly on position and that the estimator distributions are quite similar for all four photodiode channels (suggesting that the channel to channel variations in the optical properties are small). In addition, the width of the estimator (which is related to the error in the position estimation) is significantly greater when excited at a depth of 22 mm than it is when excited at 3 mm.

The collimated excitation beam is scanned along the test module, and at each depth of interaction the centroid and fwhm of the depth estimator \( \Gamma = \frac{PD}{PD+PMT} \) are computed. Figure 4 plots these measurements for all four detector elements as a function of depth of interaction, with the fwhm of the depth estimator represented as error bars on the estimator. The centroid of the estimator \( \Gamma \) is linearly dependent on depth, and the fwhm of this estimator increases with increasing depth (since the noise in the photodiode is constant but the signal decreases with increasing depth). Dividing the fwhm of the depth estimator by the slope yields the depth of interaction measurement resolution, which varies from 4 mm fwhm at a depth of 3 mm (the end near the photodiode / patient) to 13 mm fwhm at a depth of 22 mm. Again, the dependence of the centroids and widths on position is virtually identical in all four channels. This uniformity is very desirable for a module composed of a large number of crystals (64 has been proposed in [3]), as it implies that the only channel specific correction that needs to be applied is the overall gain factor \( K_i \), as opposed to a channel dependent estimator (\( \Gamma \)) versus position (\( x \)) map.

4. Predicted Measurement Resolution

The position resolution \( R \) obtained for a signal \( S \) (defined as the photopack position observed in the photodiode when excited at the photodiode end), photodiode noise \( N \), and “light loss” \( \alpha \) (defined as the ratio between photopeaks observed when excited at either end of the crystal, with \( 0 < \alpha < 1 \)) can be estimated as follows. The average photopack signal observed in each photodetector is assumed to vary linearly with excitation position, ranging from a value of \( S \) when excited at the end of the crystal closest to the photodetector to a value of \( S/\alpha \) when excited at the end furthest from the photodetector. Thus, position dependent signals observed in the photomultiplier tube and photodiode are

\[
PD(x) = \frac{S}{\alpha}[(\alpha - 1)(x) + 1] \quad (4)
\]

and

\[
PMT(x) = \frac{S}{\alpha}[(\alpha - 1)x + 1] \quad (5)
\]

where the position variable \( x \) is normalized (0 \( \leq x \leq 1 \), with \( x = 0 \) at the photodiode end). The centroid of the estimator \( \Gamma(x) \) is

\[
\Gamma(x) = \frac{PD(x)}{PD(x) + PMT(x)} = \frac{(\alpha - 1)(x) + 1}{\alpha + 1} \quad (6)
\]

so the slope of the estimator is given by

\[
\frac{d\Gamma(x)}{dx} = \frac{1 - \alpha}{1 + \alpha} \quad (7)
\]

If the photomultiplier tube noise is assumed to be much less than the photodiode noise, the width of the estimator \( \Delta\Gamma(x) \) is

\[
\Delta\Gamma(x) \approx N \frac{d\Gamma}{dPD} = \frac{PMT}{PD \left( PD + PMT \right)^2} = \frac{N}{S} \frac{\alpha[(\alpha - 1)x + 1]}{[\alpha + 1]^2} \quad (8)
\]

and the position resolution \( R(x) \) is given by

\[
R(x) = \frac{\Delta\Gamma(x)}{\frac{d\Gamma}{dx}} = \frac{N}{S} \frac{\alpha(\alpha - 1)x + 1}{1 - \alpha^2} \leq \frac{N}{S} G(x, \alpha) \quad (9)
\]

Note that since \( x \) has been rescaled to be between 0 and 1, equation 9 gives the position resolution \( R(x) \) as a fraction of the crystal length.
The influence of the depth of interaction measurement resolution on the reconstructed point spread function (PSF) has been previously reported for a PET camera with 60 cm ring diameter and 3 mm wide BGO crystals [7]. Briefly summarized, Monte Carlo simulation predicts that a depth measurement resolution of 10 mm fwhm reduces the blurring due to radial elongation by a factor of two, while a depth measurement resolution of 5 mm fwhm effectively eliminates radial elongation. While these results were obtained with a depth independent measurement resolution, they suggest that the measurement resolution achieved by the present detector module will substantially reduce the radial elongation artifact for this PET camera geometry.

Large values of $\alpha$ are not necessarily desirable, even though equation 9 predicts that they improve position resolution at the photodiode end of the detector without harming the position resolution at the photomultiplier tube end, as large values of $\alpha$ impair the ability to correctly identify the crystal of interaction. When the interaction occurs at the photomultiplier tube end of the detector, the 511 keV signal observed by the photodiode is $S/\alpha$ and the photodiode signal to noise ratio is $S/N\alpha$. The dependence on the fraction of misidentified events on the signal to noise ratio in the photodiode in a 64 element detector has been reported previously [8]. Briefly summarized, when the signal to noise ratio is greater than 2:1, the fraction of events that are misidentified due to photodiode noise is negligible, but the misidentification fraction rises rapidly as the signal to noise falls below 2:1. This implies that $\alpha \leq S/2N$ in order to ensure proper identification independent of interaction position.

This LSO detector module has a light loss ratio $\alpha=5$, which is significantly larger than optimal and explains the large fraction of misidentified events. However, there is potential for improving the detector performance significantly, mainly by improving the reflector on the LSO crystal. The reflector material used was developed for BGO, and this work shows that a good reflector for BGO is not necessarily a good reflector for LSO! The maximum photodiode signal observed in the LSO module (1200 e$^-$) is significantly less than the 2000 e$^-$ predicted based on the relative light outputs of BGO and LSO the relative photodiode quantum efficiencies at the different emission wavelengths. In addition, the photodiode noise can be reduced to 2/3 of its present value with a better photodiode/amplifier combination. If these improvement goals are met (i.e. signal $S=2000$ e$^-$ and noise $N=350$ e$^-$), a detector with outstanding performance would result, as $\alpha=5$ would meet the $\alpha \leq S/2N$ criterion (giving a very low misidentification fraction) and give excellent depth of interaction measurement resolution (0.9 and 4.6 mm fwhm when excited at the photodiode and photomultiplier tube end of the crystal respectively, based on equation 9).

5. CONCLUSIONS

A PET detector module that uses an array of silicon photodiodes to both identify the crystal of interaction and measure the depth of interaction has been characterized. The module
uses LSO scintillator crystals in order to obtain sufficiently high signal to noise ratio to allow room temperature operation. A four element test module was constructed, and the average signal observed by the photomultiplier tube depended on the depth of interaction, varying smoothly over a 5:1 range. The signal observed in the photodiode averaged 1200 e	extsuperscript{-} per 511 keV photon interaction when excited at the photodiode end of the module, but was not discernible from the noise (a position independent 520 e	extsuperscript{-} fwhm) when excited at the photomultiplier tube end.

The ratio of the signals observed by the photodiode and the photomultiplier tube was used to measure the depth of interaction on an event by event basis. The accuracy of this depth measurement ranged from 4 mm fwhm at the photodiode end of the module (i.e. the patient end) to 13 mm at the photomultiplier tube end. All four detector elements have similar optical and electrical properties (little difference between elements is seen in Figures 3 and 4), suggesting that channel to channel variations will not be a major concern when constructing a detector module with a large number of elements.

The prospects for a very high performance detector module (0.9 to 4.6 mm depth of interaction measurement resolution and negligible misidentification) based on this concept is excellent. For this to occur, the maximum signal observed in the photodiode must be increased from 1200 e	extsuperscript{-} to 2000 e	extsuperscript{-} and the noise in the amplifier / photodiode combination reduced to 350 e	extsuperscript{-} fwhm. The improvement in the photodiode signal can be achieved by optimizing the reflector on the LSO crystal — the present coating was optimized for BGO and has a significantly lower light output than expected based on the relative light outputs of the two scintillator materials. Similarly, the improvement in noise is possible using standard, existing technology.

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