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A DOI Detector With Crystal Scatter Identification Capability for High Sensitivity and High Spatial Resolution PET Imaging

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Abstract

A new phoswich detector is being developed at the Crump Institute, aiming to provide improvements in sensitivity, and spatial resolution for PET. The detector configuration is comprised of two layers of pixelated scintillator crystal arrays, a glass light guide and a light detector. The annihilation photon entrance (top) layer is a 48×48 array of $1.01 \times 1.01 \times 7$ mm³ LYSO crystals. The bottom layer is a 32×32 array of $1.55 \times 1.55 \times 9$ mm³ BGO crystals. A tapered, multiple-element glass lightguide is used to couple the exit end of the BGO crystal array (52×52 mm²) to the photosensitive area of the Position Sensitive Photomultiplier Tube (46×46 mm²), allowing the creation of flat panel detectors without gaps between the detector modules. Both simulations and measurements were performed to evaluate the characteristics and benefits of the proposed design. The GATE Monte Carlo simulation indicated that the total fraction of the cross layer crystal scatter (CLCS) events in singles detection mode for this detector geometry is 13.2%. The large majority of these CLCS events (10.1% out of 13.2%) deposit most of their energy in a scintillator layer other than the layer of first interaction. Identification of those CLCS events for rejection or correction may lead to improvements in data quality and imaging performance. Physical measurements with the prototype detector showed that the LYSO, BGO and CLCS events were successfully identified using the delayed charge integration (DCI) technique, with more than 95% of the LYSO and BGO crystal elements clearly resolved. The measured peak-to-valley ratios (PVR) in the flood histograms were 3.5 for LYSO and 2.0 for BGO. For LYSO, the energy resolution ranged from 9.7% to 37.0% full width at half maximum (FWHM), with a mean of $13.4 \pm 4.8\%$. For BGO the energy resolution ranged from 16.0% to 33.9% FWHM, with a mean of $18.6 \pm 3.2\%$. In conclusion, these results demonstrate that the proposed detector is feasible and can potentially lead to a high spatial resolution, high sensitivity and DOI PET system.

Index Terms

BGO; contrast; crystal scatter; detector; DOI; GATE; LYSO; PET; phoswich

I. Introduction

SMALL animal PET has been a driving force behind the advances of molecular imaging that allows characterization and understanding of biological processes at the molecular level

[1]–[3]. The use of mice as animal models for applications in pharmacology, genetics, pathology and oncology, demands preclinical PET scanners featuring high resolution and high sensitivity, to visualize subtle distribution and quantify low concentrations of PET probes [4]. Advances in spatial resolution and sensitivity performance of imaging systems can open up applications currently out of the range of PET because of resolution limitations, such as mouse brain imaging and early lesion and metastasis detection in mouse models of cancer [5]. Therefore, high sensitivity and high resolution have been pursued as some of the most important research goals for preclinical PET imaging [6]. For conventional pixelated scintillator detectors, the spatial resolution is determined by the cross section of the scintillator crystal elements [7]. The sensitivity can be increased by employing a compact system geometry to maximize the solid angle coverage, and by using long crystals for higher 511 keV gamma photon detection efficiency.

Unfortunately, long and narrow crystals in a small diameter gantry lead to increased penetration of oblique incident gamma rays before interaction. This causes event mispositioning also called parallax error, degrading the spatial resolution uniformity and distorting the appearance of the source [8]. Therefore, detectors with the capability of encoding the depth of annihilation photon interaction (DOI) are necessary. Much effort has been devoted to develop DOI PET detectors over the past several years [9]–[26]. Among those designs, phoswich detector approaches [24]–[26] obtain DOI information by measuring differences in light decay time between multiple layers of different scintillators. The phoswich detector design has attracted considerable interest and has been employed in several prototype scanners and commercial systems [27]–[29]. Improved spatial resolution uniformity has been achieved in these phoswich DOI scanners compared to scanners of single layer design with equivalent scintillator volume and no DOI capability [30].

Inter-crystal scatter (ICS) events, for which the incoming annihilation photons interact with more than one detection element within the same block detector, is another cause of event mis-positioning in addition to the parallax error. As the detection elements become narrower and longer, the fraction of these ICS events increases [31]. With conventional PET detector designs that employ Anger logic positioning schemes [32], such ICS events appear as inaccurate detections. The spatial coordinates corresponding to the energy weighted mean of the multiple interaction sites are different from the location of first interaction. This error in determining the initial interaction location reduces image contrast and degrades spatial resolution. This leads to degradation of the lesion detectability and quantitative characteristics of an imaging system [33], [34]. Therefore, appropriate ICS event identification and correction methods should be pursued if possible. Studies have shown that the capability of rejecting ICS events, or estimating the first interaction site of an ICS event using selection criteria [31], [35], [36], or maximum likelihood based on Compton kinematics [34], [37], yields improved image quality and quantification. However, those approaches require complicated and costly data acquisition systems for measuring individual interactions of the ICS events [22] and significant computational efforts for determining the location of first interaction [37], neither of which are available for conventional Anger logic detectors.

In this work, a phoswich depth of interaction (DOI) detector design composed by two layers of scintillator array made from cerium doped lutetium-yttrium oxyorthosilicate (LYSO) and bismuth germanate (BGO) is proposed. The aim of the detector design is to achieve high sensitivity and high spatial resolution PET imaging. The two layer detector configuration is designed to retrieve DOI information that will improve spatial resolution uniformity across the FOV. Furthermore, this detector allows identification of the majority of the cross layer crystal scatter (CLCS) events (the ICS events that deposit their energy in both layers), allowing a great reduction of this source of error. This new design is expected to be implemented in the next generation small animal PET tomograph being developed at the Crump Institute for Molecular Imaging, at UCLA.

II. Methods

A. Detector Description

The prototype detector configuration in this study was comprised of two layers of pixelated scintillator crystal arrays, a multi-element glass lightguide and a PSPMT.

The top (gamma ray entrance) layer was a 48×48 array of $1.01 \times 1.01 \times 7 \text{ mm}^3$ LYSO crystals (1.09 mm pitch). The bottom (facing the PMT) layer was a 32×32 array of $1.55 \times 1.55 \times 9 \text{ mm}^3$ BGO crystals (1.63 mm pitch). LYSO and BGO scintillator elements were multiplexed in a ratio of 9:4, with each 3×3 LYSO array segment being coupled to a 2×2 BGO array segment. The LYSO and BGO crystal elements were mechanically polished on all sides with the exception of the exit ends which were diffusely ground. The four long sides of each individual crystal were bonded with a specular optical reflector (3M, St Paul, MN). The entrance surface of the LYSO array was covered with four layers of Teflon tape to enhance reflection of the scintillation light onto the PSPMT.

To build a system capable of complete coverage of the whole body of the vast majority of laboratory mice in a single view without any bed motions, an axial FOV around 10 cm is required. In our previous PETbox4 system [38], a 1 mm thick glass lightguide was used to couple a $46 \times 96 \text{ mm}^2$ scintillator array to two axially tiled PSPMTs to obtain a 96 mm axial FOV. Although all crystal pixels were successfully resolved, this simple lightguide suffered from degraded position decoding accuracy and energy resolution for edge crystals and crystals at the junction of the two PSPMTs due to the poorer light collection. Moreover, the transverse dimension of the scintillator array (46 mm) was limited by the effective area of the PSPMT, leading to incomplete angular data sampling and sensitivity loss in the PETbox4 system. In this work, a tapered, multiple-element glass lightguide [39], [40] was used to couple the exit end of the BGO crystal array ($52 \times 52 \text{ mm}^2$) to the photosensitive area of the PSPMT ($46 \times 46 \text{ mm}^2$). The BGO and lightguide scintillator elements not on the edge were multiplexed in a ratio of 9:4, with each 3×3 BGO array segment being coupled 2×2 to a lightguide array segment. The edge lightguide and BGO scintillator elements were multiplexed in a ratio of 3:2, with each 3×1 BGO array segment being coupled to a 2×1 lightguide array segment. The corner lightguide and BGO scintillator elements were 1:1 coupled (non-multiplexed). The complete individual detector module offers an overall dimension of $52 \times 52 \text{ mm}^2$ that matches the external dimensions of the PSPMT package,

which allows continuous positioning of the scintillator arrays for creating flat panel detectors without introducing gaps between detector modules.

The Hamamatsu H12700 PSPMT is used in this study. Compared to the H8500 PSPMT used in our previous PETbox4 system [38], the H12700 offers 45% higher photoelectron collection efficiency (boosted from 60% to 87%). The H12700 can be used as a direct replacement for the H8500 since its external dimensions and anode output characteristics are identical. Optical grease (BC-630, Saint-Gobain Crystals, Hiram, OH) was used for coupling between the two layers of scintillator arrays, the exit face of the BGO scintillator array to the entrance face of the lightguide, and the exit face of the lightguide to the PSPMT.

B. Simulation

To evaluate the characteristics and benefits of the proposed LYSO/BGO phoswich configuration, the Geant4 application for tomographic emission (GATE) Monte Carlo simulation software [41] was used to simulate the data acquired with a prototype two layer detector panel. The detector panel was comprised of a 48×96 array of LYSO crystals coupled to a 32×64 array of BGO crystals (created by continuously positioning two detector modules described in Section II-A). A 10 μCi point source with isotropic emission of single 511 keV gamma photons was positioned 2.5 cm from the LYSO front layer surface, mimicking the emission from center of the FOV of our previous PETbox4 system [38]. The simulation was acquired with an energy resolution of 18%, which is the average measured energy resolution of the PETbox4 [38]. An energy window of 50–650 keV was applied to the singles processing chain at the stage of initial simulation. To confine the investigation to the detector characteristics of crystal scatter on positioning accuracy, no attenuation material was included between the source and the detector, and the phantom scatter was not considered. The lightguide and the scintillation light collection were not simulated. The ROOT format output from GATE [42] was used, which stores information of particle transportation and interactions on an event-by-event basis, allowing event history to be retrieved.

The detected singles events can be classified into six primary categories (as shown in Fig. 1):

1. L: The energy deposited in the detector panel is contributed only from the interaction with the LYSO layer.
2. B: The energy deposited in the detector panel is contributed only from the interaction with the BGO layer.
3. C_1 : The gamma photon deposits its energy in both layers, with its first interaction at the LYSO layer. The energy deposited in the LYSO layer is smaller than that deposited in the BGO layer.
4. C_2 : The gamma photon deposits its energy in both layers, with its first interaction at the LYSO layer. The energy deposited in the LYSO layer is larger than that deposited in the BGO layer.

5. C₃: The gamma photon deposits its energy in both layers, with its first interaction at the BGO layer. The energy deposited in the LYSO layer is smaller than that deposited in the BGO layer.
6. C₄: The gamma photon deposits its energy in both layers, with its first interaction at the BGO layer. The energy deposited in the LYSO layer is larger than that deposited in the BGO layer.

To retrieve the characteristics of the detected single events for appropriate event classification, customized software was developed in C++ to analyze the ROOT output file from GATE. The interaction history of each detected event was investigated and the fraction representing each event type from the total number of interacting gammas was calculated.

C. Measurement

1) Readout—The 64 anode outputs from the PSPMT were multiplexed using a charge division resistor network [43] to four position encoding signals read out from four corner amplifiers. Due to the large difference in scintillation light output and decay time between LYSO and BGO (35000 photons/MeV vs 8000 photons/MeV; 42 ns vs 300 ns), the amplitude of the LYSO signal is 20 ~ 30 times higher than that of the BGO signal. To fit the LYSO signal within the dynamic range of the analogue-to-digital converters (ADC) (VHS-ADC, Nutaq, Quebec City, Quebec) without saturation, overall signal amplification is reduced. As a result, the BGO signal becomes too weak to overcome electronic noise, degrading the position decoding accuracy of the BGO events. In order to simultaneously retrieve accurate information from both the LYSO and BGO signals, a readout circuit was designed and constructed to amplify the detector response by two different factors, as shown in Fig. 2: the signals from the route amplified with higher gain ($\times 6$) were used to detect BGO events, and the signals from the route with lower gain ($\times 1$) were used to detect the LYSO and CLCS events. The two amplified analog signals were filtered by a low pass circuit with a cut-off of -3 dB at 6 MHz. This was done to permit accurate subsequent digital conversion of the signals by two 104 MHz free running ADCs on a signal processing card (VHS-ADC, Nutaq, Quebec City, Quebec). Four identical sets of the signal processing circuits described in Fig. 2 were used to read four corner position encoding signals from the detector. The digital samples were processed in a Xilinx Virtex-4 field programmable gate array (FPGA) (Xilinx, San Jose, CA) in real time, including event triggering, pulse shape discrimination, and event energy and position calculation.

2) Pulse Shape Discrimination—The four corner position encoding signals amplified by the same factor were digitized and summed in the FPGA, producing an energy pulse for each event. The delayed charge integration (DCI) technique, an algorithm measuring the different light decay constants of two scintillators (LYSO = 42 ns, BGO = 300 ns) to identify event types was applied [24]. For each triggered event, the sum pulse was partially integrated with two intervals: 0–190 ns and 190–800 ns. The ratio of the 190–800 ns integration to the 0–190 ns integration, which depends on the characteristic light decay time of the scintillators, is defined as the DCI ratio in this study. 10k events were acquired from the “ $\times 1$ ” channel shown in Fig. 2 and the DCI ratio histogram was plotted in Fig. 3. The LYSO, BGO and CLCS events were identified based on the DCI ratio: detections with a

DCI ratio less than 0.2 were identified as LYSO events; detections with a DCI ratio larger than 0.8 were assigned as BGO events; detections with a DCI ratio between 0.2 and 0.8 were classified as CLCS events. Based on the event type identified, the FPGA integrates BGO pulses for 800 ns and integrates LYSO and CLCS pulses for 190 ns, for subsequent event energy and position calculations that are recorded to the list-mode file.

3) Flood Image and Energy Spectrum—A 0.25 MBq (6.9 μCi) ^{22}Na point source (Eckert & Ziegler Isotope Products, Valencia, CA) was placed approximately 3 cm from the top face of the LYSO array. The measured count rate with the source was 60.7 kcps. The intrinsic LYSO background for the same experimental setup was measured to be 4.5 kcps without the source. For each detected event, the X and Y coordinates were calculated according to Anger logic [32]. Two-dimensional flood images for LYSO, BGO and CLCS events were acquired. The boundaries were determined for the BGO and LYSO flood images using a semi-automated program to define the crystal LUT that classifies regions in the flood image into the proper crystal of the scintillator arrays. Energy spectra for individual crystals were extracted based on the LUTs and a Gaussian function was fitted to the photopeak of each energy spectra. Energy resolution was measured for every crystal in the detector as the full width at half-maximum (FWHM) of the Gaussian function divided by the energy corresponding to the center of the photopeak, expressed as a percentage resolution. One dimensional profiles were extracted from the LYSO and BGO flood images and the average peak-to-valley ratios (PVR) for the selected profiles were reported.

III. Results

A. Simulation

The fractions of different types of events illustrated in Fig. 1 are summarized in Table I. In singles detection mode, the fractions of L (LYSO) and B (BGO) events are 54.2% and 32.6% respectively. The total fraction of singles CLCS events including C_1 , C_2 , C_3 and C_4 is 13.2%. Considering the coincidence events, the fraction of CLCS events will increase to around 25%, because a line of response (LOR) will be considered as a CLCS event as long as any one of the two single detections is a CLCS event.

Among the four types of the CLCS events, C_1 is the dominant component. This is consistent with the Compton kinetics that gamma rays preferentially scatter in the direction of the incident gamma ray, depositing a relatively smaller amount of energy in the crystal of first interaction, as also observed in [44]. The C_1 and C_4 types of events, corresponding to 77% of the total CLCS (10.1% out of 13.2%), deposit most of their energy in a scintillator layer different from the layer of first interaction. If a traditional Anger logic positioning scheme is applied, those events will yield inaccurate position and DOI information. If these mis-positioned events are included, they will degrade image contrast and spatial resolution. Identification of those events for rejection or correction may lead to significant improvements in imaging performance.

B. Measurement

Flood images and energy spectra of different event types are shown in Fig. 4. The LYSO (Fig. 4(a)) and BGO (Fig. 4(b)) flood images were acquired with an energy window of 250–700 keV, as shown by the gray shaded areas in the energy spectra of LYSO (Fig. 4(d)) and BGO (Fig. 4(e)) events. More than 95% of the LYSO and BGO crystals, including the majority of the edge crystals, were clearly resolved.

The CLCS events were acquired from the path with lower gain ($\times 1$) (Fig. 2), which was also used to acquire the LYSO events. Therefore, the CLCS events use the same energy scale as that for the LYSO events. The CLCS flood image (Fig. 4(c)) was acquired with an open energy window of 100 – 700 keV, as shown by the gray area in the energy spectra of CLCS events (Fig. 4(f)). A distinct pattern can be observed in the CLCS flood image, appearing as a blurred LYSO flood image. This is because the positions of the CLCS events are primarily determined by their LYSO signal component. As mentioned in Section II-C2, the CLCS pulses were integrated for 190 ns to calculate the event position. Because LYSO has much higher light output and shorter decay time than BGO (35000 photons/MeV vs 8000 photons/MeV; 42 ns vs 300 ns), most of the CLCS event signal within the first 190 ns is contributed from the LYSO signal. In the CLCS energy spectra shown in Fig. 4(f), the energies of most CLCS events fall below 250 keV, which agrees with our simulation result that most CLCS events deposit less energy in the LYSO layer (C_1 and C_3 in Table I, corresponding to 83% of the total CLCS events).

One-dimensional profiles across one row of the flood images are shown in Fig. 5. The average PVR of these selected profiles were 3.5 for LYSO (Fig. 5(a)) and 2.0 for BGO (Fig. 5(b)).

The energy resolutions calculated from individual crystals are shown in Table II. The average detector energy resolution derived by averaging those of the individual crystal spectra was $13.4 \pm 4.8\%$ for LYSO and $18.6 \pm 3.2\%$ for BGO (FWHM $\pm$ 1SD). Crystal energy spectra representing the average, best, and worst energy resolution are shown in Fig. 6 (LYSO) and Fig. 7 (BGO). These spectra include the 511 keV as well as the 1275 keV photopeaks present in ^{22}Na . The two peaks visible in a single photopeak in the worst LYSO and BGO energy spectra were due to the poorer spatial separation for the events detected in the edge crystals. Future improvement on energy resolution is possible if this edge crystal compression effect can be reduced.

IV. Discussion

A new phoswich detector is being developed, aiming to improve the sensitivity and spatial resolution for preclinical PET. BGO and LYSO, the most common scintillator materials for PET detectors, are employed in the phoswich detector configuration in this work. Both BGO and LYSO have high stopping power, resulting from their high effective atomic Z (75 and 62) and high density (7.13 g/cm^3 and 7.3 g/cm^3). Compared to equivalent size of detectors made from lower stopping power scintillators such as GSO, the detector made from BGO or LYSO yields higher sensitivity, reduced DOI effect resulting from the reduced crystal

penetration, and reduced ICS events. The reduction of ICS events might lead to improvements in local image contrast.

In this work, the LYSO/BGO phoswich design has several particular advantages. Due to the large difference on light output and decay time of LYSO and BGO signals, three different types of events (LYSO, BGO and CLCS) can be identified with high accuracy. As a result, the DOI information can be retrieved accurately for parallax error correction, leading to improved spatial resolution uniformity. In addition, the capability of identifying the majority of the CLCS events should lead to improved event positioning accuracy and local contrast resolution. Studies have shown that the capability of rejecting ICS events, or estimating the first interaction site of an ICS event, yields improved image quality and quantification [31], [34]–[37]. Furthermore, this design is cost effective, as it only requires traditional Anger logic and single end readout of the scintillation light. The delayed charge integration method for event type discrimination is simple and robust, and can be easily implemented in most digital or analog electronic systems.

Although rejecting ICS events increases the event positioning accuracy, it will inevitably lead to significant sensitivity loss [45]. Alternatively, estimating the first interaction site for ICS events has been proved to yield improved image quality and quantification [31], [34]–[37]. In this work, our simulation and measurement results indicate that the proposed detector design might enable the identification of the first interaction sites of the CLCS events. The simulations show that the first interaction sites for most CLCS events could be obtained if the LYSO signal component from a CLCS event can be extracted separately for event position calculation. As shown in Table I, the C_1 and C_2 types of events, corresponding to 89% of the total CLCS (11.7% in 13.2%), encounter their first interactions in the LYSO layer. Our simulation results are also consistent with the studies of other groups showing that the “minimum DOI” [36] or “maximum Z” [31] crystal positioning scheme yields higher position detection accuracy for ICS events. In addition, our physical measurement indicates the potential to extract the LYSO signal component from the CLCS signal, due to the large difference of light output and decay time between BGO and LYSO. As shown in Fig. 4(c), even without correcting the BGO signal component, a distinct pattern is observed in the CLCS flood image, appearing as a blurred LYSO flood image. This pattern indicates that the BGO signal has relatively small effect on mispositioning CLCS events (with the integration time for CLCS events set to be 190 ns). On the other hand, accurately utilizing the LYSO signal component in a CLCS event might be challenging, as it requires the capability of resolving very low energy LYSO events (< 250 keV) and appropriate modeling and correcting the weak BGO signal component. The first interaction identification algorithm for CLCS events is currently under investigation.

To provide whole-body mouse imaging with about the same volume resolution as obtained in human body scans, submillimeter spatial resolution should be pursued [4], [6]. In the proposed detector, two layers of the scintillator arrays with different crystal size were used, with the LYSO crystal (1.08 mm pitch) smaller than the BGO crystal (1.63 mm pitch). This approach fully takes advantage of the higher light output of the LYSO scintillator. Utilizing finer pixelated detector as an accessory for a coarse pixelated scintillator scanner has been proved to significantly improve the spatial resolution of the overall images, as shown in

insert applications [46], [47]. Therefore, we expect the proposed detector in this work would also benefit from the finer pitch of the LYSO layer and potentially achieve submillimeter spatial resolution, should the detector and system response be appropriately modeled for image reconstruction.

The preliminary results presented in this work are meant to demonstrate the feasibility of the proposed approach. Important detector parameters such as crystal thickness and imaging system geometry configuration need to be determined with further investigation of how well the image reconstruction algorithm can compensate for the non-ideal detector response. The optical scintillation light transmission and collection in this detector, including the crystal surface treatment and the choice of the reflectors, are under optimization. The timing resolution of the BGO-BGO layers for this new detector is expected to be close to that of our previous PETbox4 system (4 ns FWHM) [38], which employs BGO scintillator detectors, PSPMT and the same electronics. The timing resolution of the LYSO-LYSO, LYSO-BGO layers might be better than that of the BGO-BGO layers due to the higher light output and shorter decay time of the LYSO signal. Currently more detector modules are under development. More detailed performance evaluation of this new detector including timing resolution, intrinsic spatial resolution, etc., will be reported in the future. We have also performed preliminary simulation studies on energy window optimization, similar to what we have done for our previous PETbox4 system [48]. We conclude that a tight energy window will be optimal for the LYSO events if this detector is employed in a compact geometry system. In that case, the effect of the intrinsic activity of ^{176}Lu will be minimized by the increased low level discriminator (LLD). The choices of the energy window for the three types of events might be different, which also depend on the performance of the realistic detector on decoding low energy events. The detailed energy window optimization is beyond the scope of this work and would be presented in a future publication. The current readout of the proposed detector utilizes eight ADC channels for simultaneously acquiring all three types of events from four sets of the signal processing circuits that are described in Fig. 2. A readout circuit is currently under construction in which the front end readout will be multiplexed by factor of two before digitization to allow operation with only four channel readout per detector. This concept will be similar to the circuit design described in the OPET application [49].

V. Conclusion

In conclusion, this paper reports on the design and initial feasibility study of a new DOI detector for implementation in a next generation small animal PET system at UCLA. Both simulations and physical measurements demonstrate that the proposed detector is feasible and can potentially lead to a high spatial and contrast resolution, high sensitivity, and DOI PET system.

The individual modular detector design also provides flexibility in the configuration of large area detector plates and multiple-detector systems. Besides preclinical PET imaging, the proposed detector may also be used in neuro-imaging and other specialized imaging system like PEM where high spatial resolution and high sensitivity are also desired [6].

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Fig. 1. Illustration of different types of events: L: LYSO events, B: BGO events; C₁ ~ C₄ represent four types of cross layer crystal scatter (CLCS) events.

Fig. 2.

One of four identical signal processing circuits used for simultaneously acquiring LYSO, CLCS and BGO scintillation events. The input is from one corner of the charge division resistor network. TA is a transimpedance amplifier with a conversion gain of 750 mV/mA. LP is a low pass filter with a cut-off of -3 dB at 6 MHz.

Fig. 3.

DCI ratio histogram for events acquired from the "x1" channel shown in Fig. 2.

Fig. 4.

Flood images and energy spectra of the three types of events: LYSO flood image (a) and energy spectra (d); BGO flood image (b) and energy spectra (e); CLCS flood image (c) and energy spectra (f). The gray areas in energy spectra represent the events used to plot the flood images.

Fig. 5.

(a) Horizontal profile across one row of the LYSO flood histogram shown in Fig. 4(a); (b) Horizontal profile across one row of the BGO flood histogram shown in Fig. 4(b).

Fig. 6. Energy spectra of LYSO events representing the average (a), best (b) and worst (c) energy resolution.

Fig. 7. Energy spectra of BGO events representing the average (a), best (b) and worst (c) energy resolution.

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TABLE I

Fraction of Different Types of Events Illustrated in Fig. 1

Type	L	B	C ₁	C ₂	C ₃	C ₄
fraction	54.2%	32.6%	9.8%	1.9%	1.2%	0.3%

TABLE II

Energy Resolution of the LYSO/BGO Phoswich Detector

Scintillator	Mean (%)	Best (%)	Worst (%)
LYSO	13.4 ± 4.8	9.7	37.0
BGO	18.6 ± 3.2	16.0	33.9

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