Novel Prototype of a Compton Camera Based on a Monolithic GAGG Crystal

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Date: 2024-12-16T00:00:00+00:00

Abstract

Traditional scintillator-based or semiconductor-based Compton Cameras generally employ pixelated detectors to determine the interaction positions of γ-rays within detectors, which suffers from disadvantages including complex fabrication processes, high cost, strong dependence of position resolution on pixel size, and large electronic readout systems. These limitations hinder the miniaturization and commercialization of Compton Cameras. In this study, we utilized monolithic GAGG crystals coupled to silicon photomultiplier (SiPM) arrays to construct two types of detectors: side-readout detectors and back-readout detectors, and developed two position reconstruction algorithms—the Response Function Method and the Photon Distribution Reconstruction Method. The position resolution of both detector types was evaluated, yielding 1.2 mm for the side-readout detector and 1.6 mm for the back-readout detector, respectively. Additionally, we constructed a prototype double-layer Compton Camera based on monolithic GAGG crystals, employing a side-readout detector as the scattering detector and a back-readout detector as the absorption detector. The position resolution for radioactive source localization achieved 5.4 mm with the source positioned 27 mm from the scattering detector.

Full Text

Preamble

Novel Prototype of a Compton Camera Based on a Monolithic GAGG Crystal

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Traditional scintillator-based or semiconductor-based Compton cameras typically employ pixelated detectors to determine the interaction position of γ-rays, which suffers from several drawbacks: complex fabrication processes, high costs,

strong dependence of position resolution on pixel size, and massive electronic readout systems. These limitations hinder the miniaturization and commercialization of Compton cameras. In this study, we constructed two types of detectors using a monolithic GAGG crystal coupled to a silicon photomultiplier (SiPM) array: a side-readout detector and a back-readout detector. We developed two position reconstruction algorithms—the Response Function Method and the Photon Distribution Reconstruction Method—and evaluated their performance. The side-readout and back-readout detectors achieved position resolutions of 1.2 mm and 1.6 mm, respectively. Furthermore, we built a prototype double-layer Compton camera based on a monolithic GAGG crystal, employing the side-readout detector as the scattering detector and the back-readout detector as the absorption detector. This prototype achieved a radioactive source localization resolution of 5.4 mm with the source placed 27 mm from the scattering detector.

Keywords: Compton Camera, Scintillator, Position Reconstruction, SiPM

Introduction

Compton cameras locate γ-ray sources using the principle of Compton scattering and offer unique advantages in detection efficiency compared to other gamma-ray imaging modalities since they require no mechanical collimation structure [?]. They are widely used in biomedical applications and radiation environmental monitoring [?]. In response to substantial market demand, several foreign companies have developed Compton camera products, including the H100 from H3D (USA) [?], the GEGI from PHDS (USA) [?], and the ASTROCAM from Japan's Institute of Aerospace Sciences (ISAS) [?]. ASTROCAM has been field-tested in Fukushima, demonstrating its capability for hot spot detection and radioactive decontamination assessment. Domestic research on Compton camera imaging technology remains in development, with ongoing studies in simulation and image reconstruction algorithms, such as Guo Xiaofeng et al.'s work on a CZT-based Compton camera using Geant4 [?]. Several laboratories have built prototypes, including Liu Yilin at Tsinghua University, who constructed a Compton camera based on 4$×$4 pixel 3-D CZT detectors [?], and Zhang Jipeng et al. at the Institute of High Energy Physics, Chinese Academy of Sciences, who built a double-layer Compton camera based on GAGG crystal and SiPM array [?]. However, domestically developed Compton camera products ready for market remain unavailable.

Currently, both scintillator-based and semiconductor-based Compton cameras obtain γ-ray interaction positions through detector pixelation [?]. Although this approach is conceptually simple, it has clear limitations: reconstruction accuracy depends entirely on the degree of pixelation. Scintillator crystals are prone to cracking and shattering during processing, making it difficult to obtain arrays with small pixel sizes. While semiconductor detectors can achieve small pixels, they are expensive and difficult to promote. Moreover, both scintillator and semiconductor detectors face a serious problem as pixelation increases: an exceptionally large electronic readout system.

A bold and novel alternative is to use a monolithic scintillator crystal for position reconstruction. Compared to pixelated scintillator arrays, monolithic crystals offer simpler structure, lower cost, and easier handling. Few reports exist on monolithic scintillator crystals in Compton cameras, primarily because confirming where γ-rays interact within a monolithic crystal is challenging. However, with increasing SiPM applications, this problem appears solvable. Traditional Compton cameras use photomultiplier tubes (PMTs) as photoconversion devices, but PMTs are expensive, bulky, and require high voltage, which cannot meet commercial demands for low cost and miniaturization [?]. In contrast, SiPMs are compact, offer high gain, operate at low voltage, are insensitive to magnetic fields, and provide high photon detection efficiency. Most importantly, they can be assembled into detection arrays of any shape [?].

In this study, we constructed two detector types using monolithic crystals coupled with SiPM arrays to reconstruct γ-ray interaction positions. The VMEDAQ system captured interaction event data. We developed two position reconstruction algorithms to determine interaction positions within the crystal and experimentally evaluated both detectors' position resolution performance. We then employed these detectors as scattering and absorption detectors in a Compton camera, developed a new prototype, and experimentally tested its performance.

II. Equipment

A. Experimental Equipment

Scintillators such as NaI, CsI, cerium-doped gadolinium aluminum gallium garnet (GAGG), LYSO, and $LaBr_3$ are commonly used in Compton cameras due to their relatively high detection efficiency and fast time response [?]. This study used GAGG crystals, which offer several advantages over other materials: (1) unlike NaI and $LaBr_3$, GAGG is non-hygroscopic and requires no packaging; (2) unlike LYSO and $LaBr_3$, GAGG produces no internal background radiation; and (3) GAGG has higher light yield and shorter decay time than CsI, which improves signal-to-noise ratio and energy resolution [?, ?].

Traditional Compton cameras use pixelated crystal configurations, requiring detectors to be attached to the crystal back to match each pixel for position information [?]. However, as a scattering detector, the back-readout configuration inevitably affects outgoing γ-rays after interaction, resulting in poor reconstruction. This study employed a monolithic crystal, enabling a novel side-readout detector configuration (S-type, [Figure 1: see original paper]b). We also built a back-readout detector (B-type, [Figure 1: see original paper]c) for use as an absorption detector.

The S-type detector uses a 27\$$\times 27\times 3mm$^{3}\$ GAGG crystal coupled with four 1\$$\times 8 SiPMarrays([Figure1: seeoriginalpaper]a), witheachSiPMreadoutseparately, yielding32signals.The$

typedetectorusesa27×27×2mm^{3}$ GAGG crystal coupled with an 8$×8SiPMarray.TheeightSiPMchannelsineachhorizontalrowofthe8×8arrayaresummedintooneoutput,yie 60035pixelSiPM,whichhasa3×3mm^{2}$ pixel area containing 10,998 micropixel APDs, each 20 µm$×$20 µm. We built a double-layer Compton camera using an S-type detector as the scattering detector and a B-type detector as the absorption detector ([Figure 1: see original paper]e). The distance between detectors is adjustable; this experiment used a 20 mm separation. The VMEDAQ system filtered coincidence events and logged signal amplitudes.

B. The S-Type and B-Type Detectors

Figure 2 Figure 2: see original paper shows Cs-137 spectra measured by the S-type detector with and without a reflective layer (polytetrafluoroethylene). More fluorescent photons are collected with the reflective layer, so the total energy peak channel is significantly higher. The total energy peak resolution is 8.2% with the reflective layer, closely matching the crystal manufacturer's specifications, and drops to 9.4% without it. The reflective layer has minimal effect on energy resolution but substantially impacts position resolution.

Because the reflective layer is not perfectly smooth, fluorescent photons undergo diffuse reflection during transmission, destroying their position information. To study this effect, we irradiated the center of the GAGG crystal vertically with an α source and recorded the 32-channel SiPM outputs with and without the reflective layer. A similar procedure was performed for the B-type detector.

For the S-type detector, 32-channel SiPMs are distributed on the crystal's four sides. Due to central symmetry, Figure 3 Figure 3: see original paper shows spectra for only eight channels on one side. Spectra from geometrically symmetric left and right channels are similar, allowing the eight-channel spectra to be grouped into four categories. With the reflective layer, differences between the four groups are minimal, introducing significant uncertainty in position reconstruction. Without the reflective layer, clear differences emerge, greatly facilitating interaction position determination.

For the S-type detector, the reflective layer slightly improves energy resolution but severely degrades position resolution. Therefore, we adopted a design without the reflective layer to achieve better position resolution.

Figure 2(c,d) shows Cs-137 spectra for the B-type detector with and without the reflective layer. As with the S-type, more fluorescent photons are collected with the reflective layer, yielding a higher total energy peak channel. However, the B-type detector's energy resolution is worse than the S-type's regardless of the reflective layer, primarily because the B-type uses more SiPMs, increasing the impact of response inconsistency. Additionally, the B-type uses additive rather than individual readout, and differences between additive circuits further reduce energy resolution. The reflective layer degrades energy resolution from 10.2% to 14.4%—a significant reduction.

While wrapping the reflective layer might seem beneficial for the B-type detector, this study prioritizes position resolution. Figure 3(c,d) shows the eight-channel output spectra in the horizontal direction (vertical direction is similar) with and without the reflective layer when the B-type detector center is irradiated with an α source. Due to the fixed irradiation position, each output's energy spectrum should theoretically follow a Gaussian distribution. However, with the reflective layer, each channel's spectrum deviates significantly from Gaussian. In this experiment, the trigger was set to "or" mode. Without the reflective layer, triggering occurs through fixed channels; with it, many channels may trigger, resulting in non-Gaussian spectra that could devastate position reconstruction. Therefore, the B-type detector also adopts a configuration without the reflective layer.

To improve energy resolution, we first measure the total energy deposited in both scattering and absorption detectors, select events with total energy at the full-energy peak, and calculate the absorption detector energy by subtracting the scattering detector energy from the total. Future improvements could optimize the B-type detector's additive circuit or switch to individual readout.

C. Interaction Position Calibration Experiment

We performed interaction position calibration using an α source (Am-241, [Figure 4: see original paper]b) and PCB masks ([Figure 4: see original paper]a). The PCB mask thickness is 2 mm, with constant hole diameter and center spacing within each mask. Multiple masks with different hole diameters and spacings were used, such as a mask with 1 mm holes spaced 3.4 mm apart (PCB (1.0, D3.4)). The α source was placed sequentially in each hole, and events from each hole were considered to originate from the same position, with x,y coordinates taken from the hole center and z fixed at 1.5 mm for the S-type detector and 1.0 mm for the B-type detector ([Figure 4: see original paper]c).

III. Algorithm

This study developed two reconstruction algorithms to determine the γ-ray interaction position (x, y, z) in the detector: the Response Function Method and the Photon Distribution Reconstruction Method. In both algorithms, z is treated as a fixed value. The Response Function Method works only for S-type detectors, while the Photon Distribution Reconstruction Method works for both types.

A. Response Function Method

Assuming a homogeneous scintillator with uniform 4π fluorescence photon emission, SiPMs closer to the interaction point receive more photons. For example, with α sources placed sequentially at eight test points in the black box in Figure 5: see original paper, E_{up} represents energy deposited in the upper SiPM and E_{down} represents energy deposited in the lower SiPM. As the interaction position changes, the ratio E_{down}/E_{up} varies according to a specific response function, shown in Figure 5: see original paper. Repeating this operation yields response functions for the entire detector (eight horizontal and eight vertical). The x,y values of any event can then be reconstructed using these functions. This algorithm only works for S-type detectors because similar response functions cannot be obtained for B-type detectors.

B. Photon Distribution Reconstruction Method

When a single fluorescent photon reaches a SiPM, it produces an output pulse of a certain amplitude. When multiple photons arrive, the output signal is the superposition of all individual photon signals:

$$V_i = V_{i0} \cdot N_i$$

where V_i is the output signal amplitude of SiPM i, N_i is the number of fluorescent photons received by SiPM i, and V_{i0} is the output amplitude for a single photon. The number of photons received by SiPM i correlates strongly with the solid angle subtended by SiPM i at the interaction position:

$$N_i = N_0 \cdot E \cdot \Omega_i$$

where N_0 is the number of fluorescent photons per unit solid angle per unit deposited energy, and E is the energy deposited at that point. Assuming isotropic fluorescence emission at the interaction position and neglecting attenuation from crystal self-absorption, N_0 is a fixed value determined by the crystal's light yield. Ω_i is the solid angle subtended by SiPM i at the interaction position, as shown in [Figure 6: see original paper].

The ADC value of SiPM i's output signal is:

$$ADC_i = a_i \cdot V_i + b_i$$

where a_i and b_i are scaling coefficients between signal amplitude and ADC channel, obtained experimentally as fixed values. Substituting the previous equations:

$$ADC_i = a_i \cdot V_{i0} \cdot N_0 \cdot E \cdot \Omega_i + b_i = K_i \cdot E \cdot \Omega_i + b_i$$

For unit energy deposition in the crystal, the ADC value received by SiPM i per unit solid angle can be calculated using this formula. K_i and b_i are conversion coefficients measurable experimentally.

Before describing the experiment, we explain how to calculate the solid angle. Taking the crystal center as the origin (0,0,0) of the xyz coordinate system, the spatial coordinates of the 32 SiPMs are known quantities. For SiPM i, the coordinates of points A (x_{Ai}, y_i, z_{Ai}), B (x_{Bi}, y_i, z_{Bi}), C (x_{Ci}, y_i, z_{Ci}), and D (x_{Di}, y_i, z_{Di}) are defined, where y_i is fixed for a given SiPM. For an interaction at any position (x,y,z) in the crystal, the solid angle Ω_i relative to the interaction position is obtained by integrating $\phi\{Ai\}$ *and* {Ai}. In the x coordinate system, the relative positions of points ABCD are transformed to A(x_{Ai}-x, y_i-y, z_{Ai}-z), B(x_{Bi}-x, y_i-y, z_{Bi}-z), C(x_{Ci}-x, y_i-y, z_{Ci}-z), and D(x_{Di}-x, y_i-y, z_{Di}-z). While changes slightly during integration from A to D, this change is negligible unless the interaction position is very close to the SiPM. Therefore, is considered constant in the integration.

The solid angle calculation involves:

$$\Omega_i = \int_{\phi_{Ai}}^{\phi_{Di}} \int_{\theta_{Ai}}^{\theta_{Bi}} \sin\theta \, d\theta \, d\phi = \int_{\phi_{Ai}}^{\phi_{Di}} ((-\cos\theta_{Bi}) - (-\cos\theta_{Ai})) \, d\phi$$

where:

$$\sin\phi_{Ai} = \frac{y_i - y}{\sqrt{(y_i - y)^2 + (x_i - x)^2}}$$

$$\cos\phi_{Ai} = \frac{x_i - x}{\sqrt{(y_i - y)^2 + (x_i - x)^2}}$$

$$\cos\theta_{Ai} = \frac{z_{Ai} - z}{\sqrt{((y_i - y)\cdot\tan\phi_{Ai})^2 + (y_i - y)^2 + (z_{Ai} - z)^2}}$$

Defining $k_{Ai} = \frac{y_i - y}{z_{Ai} - z}$, we get:

$$\cos\theta_{Ai} = \frac{1}{\sqrt{(k_{Ai}\sin\phi_{Ai})^2 + 1}}$$

Similar transformations apply for $\phi\{Bi\}$ *and* ϕ\{Di\}. Substituting into the integral:

$$\Omega_i = \int_{\phi_{Ai}}^{\phi_{Di}} \left(\frac{1}{\sqrt{(k_{Ai}\sin\phi_{Ai})^2 + 1}} - \frac{1}{\sqrt{(k_{Bi}\sin\phi_{Bi})^2 + 1}} \right) d\phi$$

Using integral tables, the solution becomes:

$$\Omega_i = \arctan\left(\frac{\sqrt{A_i + 2\sin^2\phi_{Ai}}}{2\cos\phi_{Ai}} \right) - \arctan\left(\frac{\sqrt{A_i + 2\sin^2\phi_{Di}}}{2\cos\phi_{Di}} \right) - \arctan\left(\frac{\sqrt{B_i + 2\sin^2\phi_{Ai}}}{2\cos\phi_{Ai}} \right) + \arctan\left(\frac{\sqrt{B_i +}}{2} \right.$$

The solid angle for any interaction position relative to SiPM i can now be calculated. Next, K_i and b_i are measured experimentally. An α source was placed at various positions in the crystal, the solid angles for all 32 SiPMs were calculated theoretically, and the corresponding ADC values were recorded experimentally. While K_i and b_i could theoretically be obtained from two experiments, we used multiple iterative experiments to reduce error. The iteration criterion minimizes the difference between the experimental energy spectrum center (blue line in [Figure 7: see original paper]) and the theoretically calculated expected value (red line in [Figure 7: see original paper]) for all 32 SiPMs. [Figure 7: see original paper] shows this comparison for a specific experimental point. The optimized K_i and b_i ensure the total difference between theoretical and experimental values across all 32 SiPMs is minimized, though some individual SiPMs may show slight deviations.

The Photon Distribution Reconstruction Method does not directly calculate position from ADC values. Instead, it uses Equation (4) to calculate expected 32-channel ADC values at each interaction position. These values are stored as a dataset. By changing the interaction position, multiple datasets are generated to create a database. When an unknown interaction occurs, the measured 32-channel ADC values are compared against the database to find the most similar dataset, whose position is taken as the reconstructed position. Larger databases yield higher reconstruction accuracy. While experimentally creating a large database is difficult, Equation (4) enables generation of databases of arbitrary size.

The solid angle calculation in [Figure 6: see original paper] is schematically shown for the S-type detector. If the crystal is compressed in the y-direction and elongated in the z-direction, the schematic becomes that of the B-type detector, making Equation (15) valid for both types. Therefore, the photon distribution reconstruction method applies to both S-type and B-type detectors.

IV. Results

A. Response Function Method Reconstruction Results

[Figure 8: see original paper] shows experimental reconstruction results using PCB (1.0, D3.4) with the response function method. Red dots indicate PCB hole positions, black dots show reconstruction position centers, and uncertainties represent the FWHM of reconstruction positions for events from the same hole. Approximately 3,000 reconstruction events were counted per experimental site.

The reconstruction centers closely match the experimental positions, with an average deviation of about 0.3 mm in both x and y directions across 64 points. Notable deviation occurs in the second-to-last data point, showing approximately 5 mm offset in the vertical axis. This occurs because the response function in Figure 5: see original paper changes rapidly around 5 mm, and insufficient experimental points prevent the function from accurately describing this region's trend, resulting in poor reconstruction. Reconstruction uncertainties are relatively large because each SiPM collects only a fraction of fluorescent photons, and statistical fluctuations in photon number are significant. Using more SiPMs during reconstruction could reduce this impact.

B. Photon Distribution Reconstruction Method Results

[Figure 9: see original paper] shows reconstruction results using the photon distribution method for S-type (left) and B-type (right) detectors with PCB (1.0, D3.4). Red dots indicate hole positions, black dots show reconstruction centers, and uncertainties represent FWHM. Approximately 3,000 events were counted per site.

The crystal was divided into a central 6$×$6 grid (36 points) and an outer ring (28 points). For the S-type detector, average deviation is about 0.3 mm in both directions for central points, matching the response function method. For the B-type detector, central point deviation is about 0.4 mm, slightly higher than the response function method. In the edge region, average deviation increases to 0.5 mm for S-type and 0.7 mm for B-type. This deterioration occurs because optical scattering of fluorescence more severely damages position information near the crystal edges.

These results show that edge-region events reconstruct less accurately than central events. Therefore, subsequent position resolution limit experiments focus on central detector events while excluding edge events.

C. Position Resolution Limit Experiments

We investigated position resolution limits using PCB masks with varying hole spacings. To reduce position uncertainty from hole size, hole diameter was reduced to 0.5 mm for masks with spacing below 2 mm. Figure 10: see original paper shows x-direction projection reconstruction results using PCB(0.5,D1.2) for the S-type detector, while Figure 10: see original paper shows results using PCB(0.5,D1.6) for the B-type detector. Both use the photon distribution reconstruction method.

The S-type results show 11 peaks from 8 mm to 22 mm, with average spacing of 1.27 mm, closely matching the PCB's 1.2 mm hole spacing. This indicates the S-type detector's position resolution limit is approximately 1.2 mm. The B-type results show 12 peaks from 5.5 mm to 26 mm, with average spacing of 1.71 mm, slightly larger than the PCB's 1.6 mm spacing, indicating a B-type resolution limit of about 1.6 mm. We conservatively estimate 1.6 mm rather than 1.7 mm because, except for peaks 2-3 and 9-10 which overlap significantly, other peaks are well resolved, suggesting further hole spacing reduction is possible.

Peaks 3 and 9 show significantly lower counts because their positions align exactly with gaps between two SiPMs in the array. The gap width is about 0.43 mm, similar to the PCB's 0.5 mm hole diameter. When interactions occur at peaks 3 and 9, many fluorescent photons escape through SiPM gaps rather than being detected. This problem does not affect the S-type detector, which uses all 32 SiPMs simultaneously for reconstruction. When the interaction position falls in a gap between two SiPMs, the remaining 30 SiPMs can still reconstruct position without significant degradation.

V. Compton Camera Prototype and Performance Testing

We built a double-layer Compton camera prototype using monolithic crystals, with an S-type detector as the scattering detector and a B-type detector as the absorption detector ([Figure 1: see original paper]e). The detector spacing was 20 mm. A Cs-137 point source (approximately 1 mm diameter) served as the imaging object, placed at two positions 27 mm from the scattering detector, separated by 8 mm in the same imaging plane. Both detectors used the Photon Distribution Reconstruction Method. [Figure 11: see original paper] shows reconstruction results for both positions. Using FWHM as the spatial resolution metric, the resolution at 27 mm is about 5.4 mm.

Compared to existing Compton cameras, Turecek et al. [**?**] developed a double-layer Compton camera using a 1 mm silicon detector and 2 mm CdTe detector based on Timepix3 pixel array technology, achieving 2.5 mm spatial resolution at 25 mm—among the best results for double-layer semiconductor detectors. The monolithic scintillator-based Compton camera's spatial resolution is lower, primarily because scintillator energy resolution is inferior to semiconductor detectors. However, semiconductor detectors suffer from high processing costs, harsh operating conditions, and complex electronics, making low-cost commercialization difficult. The monolithic scintillator-based Compton camera developed in this study offers low cost, simple structure, and only slightly lower detection performance than semiconductor-based cameras, showing greater potential for low-cost commercialization.

VI. Discussion

Most existing Compton cameras use pixelated crystals to determine interaction positions [**?**, **?**]. While higher pixelation undoubtedly improves accuracy, it introduces several problems. This discussion focuses on scintillator detectors (excluding semiconductor or gas detectors).

First, crystal processing difficulty and cost inevitably increase with pixelation, limiting how far pixelation can be advanced. For the 27$\times 27\times 3 mmGAGGcrystalusedinthisstudy, achieving 1.2$ mm pixels, increasing processing cost by 3-4 times.

Second, each pixel requires independent data acquisition. As pixelation increases, the acquisition system becomes extremely redundant, reducing signal-to-noise ratio and increasing crosstalk. In this study, the S-type detector requires 32 acquisition channels; achieving equivalent resolution with traditional pixelated crystals would require 529 channels (23$\times$23). Some new schemes use one acquisition channel for multiple pixelated crystals with algorithmic position reconstruction, but this cannot guarantee position resolution and increases algorithm complexity [**?**].

Third, pixelated crystals must be coupled one-to-one with the acquisition system, limiting configurations to B-type arrangements. In multi-layer B-type detectors, γ-rays may interact with electronic devices instead of crystals, reducing detection efficiency. Multi-layer S-type detectors would significantly mitigate this efficiency loss.

This study proposes a monolithic crystal interaction position reconstruction method achieving good accuracy with potential applications in many fields. In nuclear reaction and structure studies, track measurement is important but existing detectors (solid-state track detectors, ionization chambers, time projection chambers) primarily respond to charged particles with low γ-ray efficiency [**?**, **?**]. Future work could combine multiple S-type detectors into a telescope system for effective γ-ray tracking. In medical imaging, both SPECT and PET require accurate γ-ray interaction position measurement, currently using semiconductor detectors (CdTe, CZT) or pixelated scintillators (CLYC, CsI) [**?**, **?**]. This study's technology could reduce costs, simplify system structure, and improve accuracy. It also has potential in muon imaging, radioactive surveys, and radiation protection.

This study is preliminary work on monolithic crystal position reconstruction and Compton camera construction. Several improvements remain. First, the S-type detector used a thin 3 mm structure coupling only one SiPM layer, preventing z-direction determination and requiring a fixed value. Future work could increase crystal thickness, couple multiple SiPM layers, and use photon distribution to determine z-position. Second, the current algorithm considers only solid angle, neglecting photon optical transmission properties in the crystal, which degrades edge reconstruction accuracy. Incorporating optical transmission properties could improve accuracy. For the Compton camera, the current two-layer configuration uses an S-type scattering detector and B-type absorption detector; future multi-layer S-type configurations could improve detection efficiency.

VII. Conclusion

This paper presents a prototype Compton camera based on a monolithic GAGG crystal, using an S-type detector as the scattering detector and a B-type detector as the absorption detector. Two position reconstruction algorithms were developed, achieving interaction position resolutions of 1.2 mm and 1.6 mm for the scattering and absorption detectors, respectively. Using a Cs-137 point source, the source position can be determined with 5.4 mm resolution at 27 mm from the scattering detector.

The monolithic crystal-based Compton camera developed in this work offers low cost, simple and compact structure, and moderate imaging performance that meets requirements for radioactive source localization and nuclear medicine applications in the nuclear industry. These characteristics demonstrate the high potential for commercialization of monolithic crystal-based Compton cameras.

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