Monte Carlo Study of Signal/Background Characteristics in BNCT-SPECT Systems

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Abstract

Boron Neutron Capture Therapy (BNCT), as an efficient tumor-targeted treatment modality, has demonstrated proven efficacy. Real-time monitoring of boron dose distribution during treatment represents a critical bottleneck urgently requiring resolution in current BNCT clinical applications, with its measurement accuracy directly impacting treatment efficacy and safety. Based on the characteristic emission of 478 keV prompt γ-rays from the 10B(n,)7Li reaction, the utilization of Single Photon Emission Computed Tomography (SPECT) technology to detect these 478 keV γ-rays holds promise for addressing the challenge of 10B dose monitoring. However, the high-background environment induced by the intense neutron-γ mixed radiation field generated during BNCT treatment severely constrains the clinical application of BNCT-SPECT systems. This study addresses the challenges of low efficiency and insufficient statistics inherent in Monte Carlo calculations of pulse height spectra for BNCT-SPECT system detectors by proposing a stepwise sequential calculation method for detector pulse height spectra that discriminates signal versus different background sources, substantially enhancing computational efficiency and precision while enabling quantitative characterization of both target signals and background origins. Simulation results demonstrate: in the BNCT-SPECT system designed herein, the total count rate in the 478 keV energy channel is 6.83 CPS, wherein the 10B(n,)7Li reaction contribution accounts for 53% (comprising 29% from tumor region and 24% from normal tissue); the dominant background originates from Compton scattering of 2.223 MeV γ-rays produced by the 1H(n,γ)2D reaction, contributing 33%; parasitic γ-rays from neutron absorption in detector materials contribute 2%; and other scattered γ-rays contribute approximately 12%. Notably, approximately 71% of the background contribution from 2.223 MeV γ-rays enters the detector via penetration through collimator septa at oblique incidence, revealing that this background component can be further suppressed through optimization of collimator parameters.

Full Text

Monte Carlo-based Study on Signal and Noise Characteristics of BNCT-SPECT Systems

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Abstract

Boron Neutron Capture Therapy (BNCT) has emerged as a promising targeted radiotherapy, with its clinical efficacy well established. However, real-time monitoring of boron dose distribution during treatment remains a major challenge for clinical implementation, as precise dose measurement is crucial for both treatment effectiveness and safety. Based on the characteristic of the ^{10}B(n,α)^{7}Li reaction which releases 478 keV prompt gamma rays, the use of Single-Photon Emission Computed Tomography (SPECT) technology to detect this 478 keV gamma ray is expected to solve the challenge of ^{10}B dose monitoring. However, the strong neutron-gamma mixed radiation field in BNCT creates a high-background environment, significantly limiting the practical application of BNCT-SPECT systems. To overcome the computational inefficiency of Monte Carlo methods when dealing with pulse-height spectrum of BNCT-SPECT detector, we developed a simulation approach that significantly improves both accuracy and computational efficiency while enabling quantitative analysis of signal and background contributions to the PHS of detector. Simulation results indicate that the total count rate in the 478 keV energy window is 6.83 counts per second (CPS), with 53% originating from the ^{10}B(n,α)^{7}Li reaction (29% from tumors and 24% from healthy tissue). The primary background contribution (33%) comes from 2.223 MeV γ rays generated by the ^{1}H(n,γ)^{2}D reaction. An additional 2% arises from neutron-induced gamma rays in detector materials, while the rest of the gamma rays contribute roughly 12%. It is noteworthy that approximately 71% of the background contribution from 2.223 MeV gamma rays penetrates the collimator hole walls via oblique incidence into the detector, revealing that this background component could be further suppressed by optimizing collimator parameters.

Keywords: BNCT-SPECT; Monte Carlo; Pulse Height Spectrum; Source Term

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Over 30 countries worldwide have conducted extensive research on Boron Neutron Capture Therapy (BNCT), with China being an early participant that has developed rapidly, now boasting more than 10 operational, under-construction, or planned projects. BNCT offers significant targeting advantages. The principle involves administering tumor-seeking drugs containing ^{10}B to patients before treatment, which readily accumulate in tumor cells while remaining at low concentrations in normal tissue. Irradiation of the tumor site with epithermal or thermal neutron beams leads to moderation of epithermal neutrons through scattering with elements such as C, H, O, and N in human tissue, converting them to thermal neutrons that are then captured by ^{10}B. This capture releases high Linear Energy Transfer (LET) alpha particles and lithium nuclei with ranges comparable to single cell dimensions. With a thermal neutron capture cross-section of 3,840 barns—far exceeding that of other nuclides—^{10}B can effectively kill tumor cells while largely sparing surrounding normal tissue, making BNCT a highly promising precision radiotherapy technology [1].

Current methods for measuring ^{10}B drug concentration distribution primarily rely on physical measurement [2], chemical measurement [3], and nuclear measurement techniques [4]. These are mostly offline, invasive, and indirect methods that cannot accurately reflect the true boron concentration distribution and its temporal changes during treatment, leading to uncertainties in clinical BNCT treatment dosing (discrepancies between prescribed and delivered doses) that severely impact treatment precision and efficacy. Therefore, there is an urgent need to develop methods and equipment for accurately measuring boron concentration/dose distribution in tumors and normal tissues during BNCT treatment. To address this challenge, Kobayashi [5] first proposed the concept of a BNCT boron dose online monitoring system based on SPECT principles in 2000. The system leverages the characteristic that ^{10}B emits 478 keV gamma rays in situ after neutron absorption, combining Single-Photon Emission Computed Tomography (SPECT) technology to monitor these gamma rays and reconstruct the spatial distribution of boron dose. BNCT-SPECT systems offer significant advantages including real-time, external, and non-invasive monitoring, making them a research hotspot. However, compared with conventional SPECT, BNCT-SPECT faces unique challenges: (1) **Strong background environment**: During BNCT treatment, SPECT operates in a strong mixed neutron-gamma radiation field. Signals in the 478 keV energy channel recorded by the SPECT detector may originate not only from ^{10}B-produced events but also from gamma rays generated by neutron reactions with other nuclides in the patient, treatment room walls, treatment couch, collimator, detector, shielding, and other materials that deposit approximately 478 keV energy in the SPECT detector, requiring suppression of these background sources to improve signal-to-noise ratio. (2) **Insufficient spatial resolution**: The target gamma ray emission rate corresponding to clinical boron concentrations of 20–40 ppm is about one order of magnitude lower than that of conventional SPECT imaging agents such as ^{99} Tc (typical activity 740 MBq), requiring the system to balance detection sensitivity and spatial resolution. (3) **Energy discrimination from 511 keV**: High-energy gamma rays such as 2.223 MeV (from the ^{1}H(n,γ)^{2}D reaction) and others produced during treatment can generate 511 keV photons through positron annihilation, with energies close to the target signal of 478 keV. In summary, BNCT-SPECT systems must extract the relatively weak 478 keV target characteristic signal from a complex radiation background within limited time while maintaining adequate spatial resolution, imposing stringent requirements on the system. Consequently, although multiple international teams have developed BNCT-SPECT system prototypes [6-9] and achieved certain theoretical results [10-14], the image signal-to-noise ratio has not reached ideal levels, with low signal-to-noise ratio becoming the primary obstacle limiting practical application of these systems.

To address the strong background problem in BNCT-SPECT systems caused by neutron-gamma mixed radiation fields, Altieri et al. [15] in 2018 analyzed the spatial distribution characteristics of neutron and gamma flux fields in treatment rooms and calculated neutron capture reaction rates for nuclides such as ^{10}B and ^{1}H as well as detector materials. To simplify calculations, particle transport was terminated when particles entered treatment room walls, a treatment that neglected contributions from wall-scattered neutrons and gamma rays produced by neutron capture in building materials, leading to underestimation of background.

In 2022, Isao Murata [9] simulated and analyzed the response to 478 keV gamma rays from the ^{10}B(n,α)^{7}Li reaction, gamma rays produced by neutron capture in detector materials, and residual gamma rays at the detector. However, the study did not differentiate contributions from various background sources, and the calculation methodology was not described in detail.

Despite the significant technical challenges in developing BNCT-SPECT technology, its application prospects are broad. This study focuses on investigating signal/background characteristics of BNCT-SPECT using Monte Carlo methods, proposing an efficient calculation method for signal and background components in the pulse height spectrum (PHS) of BNCT-SPECT systems. This method enables quantitative analysis of target signals and background sources, and investigates possible measures for background suppression. The paper is structured as follows: first, we detail the Monte Carlo neutronics benchmark model and simulation methodology for the BNCT-SPECT system; second, based on simulation results, we analyze detector signal characteristics and quantitatively evaluate contributions from various background sources; finally, we summarize the main findings.

1.1 BNCT-SPECT Principle

The ^{10}B absorbed dose in different tissues during BNCT (characterized as local boron dose) is an integrated quantity reflecting cumulative dose from the start of irradiation to the present moment. The expression involves neutron flux, boron concentration, ^{10}B neutron absorption cross-section, compound biological effect value, and reaction energy, with 5.76$\times$10^{7}$ as the unit conversion coefficient. Due to continuous metabolism of ^{10}B in the patient, the boron concentration varies dynamically during irradiation, making real-time measurement difficult through simple experimental means. After ^{10}B absorbs neutrons, it emits alpha particles and Li nuclei (Equation 2), with 93.7% of Li nuclei in an excited state that decays by emitting 478 keV gamma rays. Therefore, during BNCT treatment, the 478 keV gamma ray emission rate directly correlates with the ^{10}B neutron capture reaction rate, providing a foundation for real-time treatment monitoring. BNCT-SPECT leverages this characteristic by applying SPECT technology to detect the spatial distribution of 478 keV gamma rays released from the ^{10}B(n,α)^{7}Li reaction, reconstructing three-dimensional boron dose images and enabling quantitative localization of ^{10}B dose distribution during treatment. The working principle is illustrated in Figure 1 [Figure 1: see original paper]-1.

1.2 Monte Carlo Simulation Benchmark Model

Based on literature [9], this study constructed an optimized BNCT-SPECT Monte Carlo neutronics benchmark model (Figure 1-2) with respect to collimator parameters and shielding design. The benchmark model consists of several core components: a 20 cm diameter head, a 3 cm diameter tumor, SPECT detection system (including SPECT collimator, detector, and shielding), and treatment room walls. The collimator is spaced 5 cm from the head. These components collectively constitute the environment in which SPECT operates during BNCT treatment, providing a realistic simulation environment for subsequent signal/background characteristic studies.

1.2.1 Head Model

This study aims to investigate signal/background characteristics of BNCT-SPECT systems, employing a simplified head geometry (Figure 1-3) comprising tumor and brain tissue. Geometric parameters: the head is a 20 cm diameter sphere centered at the coordinate origin (0,0,0); the tumor is a 3 cm diameter sphere centered at (0,0,5.5). Boron concentration distribution: based on clinical case statistics, tumor region boron concentration is set at 24 ppm, while normal tissue boron concentration is 6.85 ppm (tumor-to-normal tissue boron concentration ratio T/N = 3.5). Neutron source parameters: parallel neutron beam irradiation (beam spot diameter 10 cm) with incident neutron energy of 10 keV and neutron flux of 1×10^{9} n/(s · cm^2).

1.2.2 SPECT Collimator

The SPECT collimator geometry configuration is shown in Figure 1-4. Perpendicular to the neutron incidence direction, the collimator contains an 8×8 array of 64 collimator holes. Parameters are as follows: collimator material is tungsten with total length 26 cm, hole diameter 3.5 mm to meet spatial resolution requirements at the millimeter scale, and inter-hole septal thickness of 0.5 mm.

1.2.3 SPECT Detector

Primary detector options for BNCT-SPECT systems include scintillators (BGO, GAGG, NaI) and semiconductors (HP-Ge, CZT). Performance comparisons show: (1) scintillators are superior to semiconductor detectors which are vulnerable to neutron activation; (2) for 478 keV gamma ray detection efficiency, GAGG and BGO are optimal, followed by CZT, NaI, and HP-Ge; (3) for energy resolution, HP-Ge and CZT are optimal, with GAGG (4–5%@662 keV) showing good recent progress; (4) for engineering applicability, HP-Ge requires liquid nitrogen cooling, severely limiting its use. Considering these factors, GAGG was selected as the detector material in this study due to its advantages of high detection efficiency and good energy resolution. Specific design: [text appears truncated in original]

1.2.4 SPECT Shielding

The BNCT-SPECT detection system operates in a strong neutron-gamma mixed radiation field, requiring multiple shielding layers around the collimator and detector. The SPECT shielding in this study comprises a four-layer composite structure from outermost to innermost (layers 1-4 in Figure 1-5): (1) neutron moderation layer: 5 cm polyethylene to moderate fast neutrons; (2) neutron absorption layer: 1 cm lithium fluoride to absorb moderated neutrons, reducing gamma rays released from neutron capture in shielding and collimator; (3) gamma shielding layer: 10 cm or more of tungsten to absorb and shield stray gamma rays; (4) neutron absorption layer: 2 cm of ^{6}Li to absorb remaining thermal neutrons, reducing neutron flux into the detector.

Additionally, a 1.5 cm thick ^{6}Li layer is placed in front of the collimator entrance (marked as 5 in Figure 1-5). This lithium layer minimally affects 478 keV gamma ray transmission while efficiently absorbing thermal neutrons, reducing thermal neutrons entering through collimator holes and thereby decreasing gamma background from (n,γ) reactions in the detector.

Treatment room walls consist of three layers: 1 cm cadmium to absorb low-energy neutrons, 3 cm lead to shield gamma rays, and 20 cm concrete.

1.3.1 Physical Processes of Target and Background Signals

Analysis shows that contributions to the detector's 478 keV energy channel can be categorized into four gamma ray sources: **Source 1**: 478 keV characteristic gamma rays from ^{10}B(n,α)^{7}Li reactions; **Source 2**: 2.223 MeV gamma rays released by ^{1}H neutron absorption in the head; **Source 3**: Parasitic gamma rays from neutron capture by nuclei such as Ga and Gd in detector materials; **Source 4**: Gamma rays from neutron capture in shielding, collimator, treatment room walls, and other materials. In this study, the target signal is the 478 keV gamma ray produced by tumor ^{10}B (Source 1), while the remaining three categories represent background signals.

The detection of Source 1 involves: (1) epithermal neutrons incident on human tissue and moderation; (2) moderated thermal neutrons reacting with ^{10}B nuclei in tissue to release 478 keV gamma rays; (3) 478 keV gamma rays reaching the detector without interaction with tissue or collimator; (4) 478 keV gamma rays depositing full energy in the detector through photoelectric effect and other mechanisms, registering counts in the 478 keV channel of the pulse height spectrum.

Source 2 contributes to the detector's 478 keV channel through: (1) epithermal neutrons incident on human tissue and moderation; (2) ^{1}H elements in the head releasing 2.223 MeV gamma rays via ^{1}H(n,γ)^{2}D reaction; (3) 2.223 MeV gamma rays entering the detector through: direct incidence through collimator holes, oblique incidence through penetration of collimator septal walls, or transmission through SPECT shielding; (4) depositing partial energy (~478 keV) in the detector before escaping, registering counts in the 478 keV channel.

Source 3 contributes through: (1) epithermal neutrons incident on human tissue; (2) neutrons scattering with elements in tissue and exiting the head; (3) stray neutrons reaching detector boundaries via collimator holes, septal wall penetration, or SPECT shielding transmission; (4) stray neutrons continuing transport in the detector and being captured by detector materials to produce parasitic gamma rays; (5) parasitic gamma rays depositing ~478 keV energy in the detector before escaping, registering counts in the 478 keV channel.

Source 4 contributes through: (1) epithermal neutrons incident on human tissue; (2) neutrons scattering with elements in tissue and exiting the head; (3) stray neutrons being captured by materials including shielding, collimator, and treatment room walls to release stray gamma rays; (4) stray gamma rays penetrating SPECT shielding or collimator to reach detector boundaries; (5) depositing partial energy (~478 keV) in the detector before escaping, registering counts in the 478 keV channel.

1.3.2 Calculation Methods for Different Source Terms

Based on the benchmark model described above, this study employs the MCNPX 2.5.0 general-purpose particle transport code developed by Los Alamos National Laboratory [17] for simulation calculations. The code integrates multiple advanced nuclear reaction physics models and cross-section databases, enabling accurate simulation of particle transport processes and various interactions. The ENDF/B-VI evaluated nuclear data library is used to ensure data reliability.

The transport process in real BNCT treatment involves complex physics including neutron-photon coupling transport in complex treatment room geometries, large aspect ratio collimators, thick shielding around detectors, and photon energy deposition in detectors. Additionally, SPECT detector shielding must effectively block most stray neutrons and gamma rays, resulting in inefficient Monte Carlo calculation of stray radiation contributions to pulse height spectra with insufficient statistics. Based on the physical processes by which different gamma sources contribute to the detector's 478 keV channel, we employ a stepwise successive calculation method that generates intermediate gamma/neutron source terms to significantly improve simulation efficiency and accuracy.

We select different physical intermediate processes to generate secondary source terms for different gamma ray sources: (1) For Sources 1 and 2 (gamma rays generated within the head), since the neutron beam directly irradiates the head, neutron-photon coupled transport can obtain gamma source terms with good statistics and high-precision spatial discretization within the head (Source Term 1, 478 keV and Source Term 2, 2.223 MeV), corresponding to completion of step 2 in the physical processes for Sources 1 and 2; (2) For Source 3, we calculate a neutron surface source entering the detector boundary (Source Term 3, corresponding to step 3 in Source 3's physical process). The challenge lies in employing various variance reduction techniques to ensure statistically reliable neutron surface sources; (3) For Source 4, we calculate a high-precision gamma ray surface source entering the detector boundary (Source Term 4, corresponding to step 4 in Source 4's physical process). For these stray gamma rays distributed throughout the BNCT treatment room, variance reduction techniques are also required to ensure statistical reliability of the gamma surface source.

The specific generation methods for Source Terms 1-4 are described below:

Source Terms 1 and 2: We independently developed a source term subroutine based on grid probability sampling with capabilities for custom grid division, energy-space coupled sampling, and spatial distribution adjustment. Based on this subroutine and the spatial distribution of neutron capture reaction rates for various nuclides within the head, we generate Source Terms 1 and 2. Compared with MCNPX's SDEF card source, this subroutine offers higher automation, can accurately reproduce complex source terms with discrete spatial and energy distributions, and has been validated through comparison calculations with SDEF card sources.

Source Term 3: We employ an efficient variance reduction strategy to obtain high-precision neutron surface sources: (1) Weight window generation: using the on-the-fly GVR global variance reduction method [18] combined with MCNPX's built-in weight window generation function, optimized with SPECT detector neutron flux as the objective to obtain efficient weight window parameters that promote neutron transport to the detector; (2) Surface source generation: transport calculation based on optimized weight windows using the SSW card to record neutron information entering the detector boundary and generate neutron surface source files.

Source Term 4: (1) Weight window generation: during neutron-photon coupled transport, photon importance in tumor, head, and detector regions is set to 0, neutron importance is 1, and photon/neutron importance is 1 in other regions. The on-the-fly GVR global variance reduction method combined with MCNPX's built-in weight window generation function is employed with SPECT detector photon flux as the optimization objective to obtain efficient weight window parameters that promote stray photon transport to the detector; (2) Surface source generation: transport calculation based on optimized weight windows using the SSW card to record gamma ray information entering the detector boundary and generate gamma surface source files.

Based on Source Terms 1-4 above, successive calculations obtain the detector pulse height spectra for four gamma sources (Figure 2 [Figure 2: see original paper]-3). We also performed direct Monte Carlo calculation without intermediate secondary source terms to validate the stepwise successive calculation approach. The 478 keV channel count rates (CPS) obtained from direct calculation and stepwise successive calculation are 6.66 and 6.83 respectively, with similar results demonstrating the correctness of the generated intermediate source terms and stepwise successive calculation method. Compared with direct calculation, stepwise successive calculation reduces computation time from several days to several hours at equivalent statistical levels, improving computational efficiency by more than an order of magnitude.

2.1 Primary Source Term Simulation Analysis

Figure 2-1 shows the prompt gamma ray energy spectrum distribution generated by neutron capture reactions with nuclides in human head tissue. The spectrum clearly displays over 30 distinct photopeaks. Among these, the 2.223 MeV gamma ray shows high intensity due to the high concentration of hydrogen (H) elements in human tissue and their large neutron capture cross-section, potentially causing significant background interference for 478 keV gamma ray-based SPECT reconstruction. Despite low ^{10}B concentration in the body, a prominent gamma ray peak at 478 keV is still observed, corresponding to the characteristic gamma ray from the ^{10}B(n,α)^{7}Li reaction, demonstrating the feasibility of online dose monitoring through 478 keV gamma ray detection.

This study simulated the spatial production rates of three important characteristic gamma rays within the head (478 keV, 511 keV, 2.223 MeV) with grid dimensions of 0.5$×0.5×$0.5 cm (Figure 2-2). Table 2 -1 lists the total neutron flux, ^{10}B(n,α)^{7}Li and ^{1}H(n,γ)^{2}D reaction rates in tumor and surrounding normal tissue, providing important source term basis for subsequent analysis.

Figure 2-2 shows that the ^{10}B reaction rate in the tumor region is higher than in normal tissue. Meanwhile, the spatial distribution of 2.223 MeV gamma ray production rate is almost identical to the thermal neutron flux distribution due to relatively uniform H atom distribution in human tissue. Notably, the total reaction rate of 2.223 MeV gamma rays is 12 times that of 478 keV gamma rays, indicating that prompt gamma rays from hydrogen neutron capture constitute one of the primary background sources.

2.2.1 PHS Simulation Results for Different Gamma Sources

The PHS results for four gamma sources at the detector obtained through successive calculation are shown in Figure 2-3, with Sources 1-4 represented by blue, green, black, and cyan curves respectively. The superposition of PHS from all four sources (red curve) completely characterizes the pulse height spectrum features of the detector across 0–3000 keV energy range, with the 478 keV characteristic peak from $^{10}B(n,\alpha)^{7}Li$ reaction clearly visible. Compared with the blue curve for Source 1, the baseline of the 478 keV peak is elevated and the peak-to-valley ratio reduced, primarily due to superposition of Compton scattering continuum from 2.223 MeV gamma rays. The Compton edge of 2.223 MeV gamma rays is clearly identifiable in the spectrum, and secondary effects (such as positron annihilation) produce a 511 keV characteristic peak, further increasing the difficulty of discrimination from 478 keV.

Table 2-2 shows the CPS and corresponding contribution percentages for each gamma source in the detector's 478 keV energy channel obtained through stepwise successive calculation. Detailed analysis reveals: (1) 478 keV gamma rays from ^{10}B neutron absorption in tumor and normal brain tissue account for 29% and 24% of total CPS respectively; (2) 2.223 MeV gamma rays from ^{1}H neutron absorption in the head contribute up to 33%, representing the primary background source; (3) parasitic gamma rays generated within the detector contribute a relatively small 2%; (4) other stray gamma rays contribute approximately 12%.

2.2.2 Effect of SPECT Shielding on Background

Figure 2-4 compares PHS from Sources 3 and 4 in GAGG detectors with and without SPECT shielding, as well as total PHS. Results show that without shielding, CPS in the 478 keV channel from Sources 3 and 4 are 56.5 and 63.7 respectively, far exceeding the target signal of 2.01. With SPECT shielding, backgrounds from Sources 3 and 4 are reduced by 1–2 orders of magnitude, significantly improving signal-to-noise ratio. The current SPECT shielding design effectively suppresses stray neutrons and gamma rays reaching the detector, controlling contributions from these two major background sources to acceptable levels.

2.2.3 Background Signal at 478 keV from 2.223 MeV Gamma Rays

As analyzed above, 2.223 MeV gamma rays from the $^{1}H(n,\gamma)^{2}D$ reaction represent the primary background source. Their pathways into the detector include: (1) direct incidence through collimator holes; (2) oblique incidence through penetration of collimator septal walls; (3) transmission through SPECT shielding (negligible compared to direct and oblique incidence). To obtain contributions from direct and oblique incidence pathways, calculations were performed with neutron/photon importance set to 0 in the collimator (allowing only direct incidence gamma rays to reach the detector), yielding a direct incidence contribution of 0.65 CPS. Table 2-3 lists the background signal strength in the 478 keV energy channel for different incidence pathways.

Table 2-3 shows that oblique incidence accounts for approximately 71% of the total contribution. In this study, the inter-hole septal thickness is 0.5 mm, which effectively collimates 478 keV gamma rays but is insufficient to shield higher-energy gamma rays. To reduce background from oblique incidence, septal thickness could be increased in future collimator designs to improve signal-to-noise ratio. However, it should be noted that increasing septal thickness will inevitably sacrifice some field of view (FOV), requiring careful balance between signal-to-noise ratio and geometric efficiency in practical design.

3. Conclusion

This study addresses the high-background environment challenge faced by BNCT-SPECT systems by developing a simulation calculation method that discriminates signal and background sources in SPECT detector pulse height spectra, enabling quantitative analysis of pulse height spectra from different gamma sources. Simulation results demonstrate that the current SPECT shielding design effectively suppresses backgrounds from stray neutron and gamma fields. The maximum background originates from 2.223 MeV gamma rays generated within the head, contributing 33% of total CPS, with approximately 71% entering the detector through oblique penetration of collimator hole walls.

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Note: Figure translations are in progress. See original paper for figures.