



Spatial resolution limits for the isotropic-3D PET detector X'tal cube



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ABSTRACT

Positron emission tomography (PET) has become a popular imaging method in metabolism, neuroscience, and molecular imaging. For dedicated human brain and small animal PET scanners, high spatial resolution is needed to visualize small objects. To improve the spatial resolution, we are developing the X'tal cube, which is our new PET detector to achieve isotropic 3D positioning detectability. We have shown that the X'tal cube can achieve 1 mm³ uniform crystal identification performance with the Anger-type calculation even at the block edges. We plan to develop the X'tal cube with even smaller 3D grids for sub-millimeter crystal identification. In this work, we investigate spatial resolution of a PET scanner based on the X'tal cube using Monte Carlo simulations for predicting resolution performance in smaller 3D grids. For spatial resolution evaluation, a point source emitting 511 keV photons was simulated by GATE for all physical processes involved in emission and interaction of positrons. We simulated two types of animal PET scanners. The first PET scanner had a detector ring 14.6 cm in diameter composed of 18 detectors. The second PET scanner had a detector ring 7.8 cm in diameter composed of 12 detectors. After the GATE simulations, we converted the interacting 3D position information to digitalized positions for realistic segmented crystals. We simulated several X'tal cubes with cubic crystals from (0.5 mm)³ to (2 mm)³ in size. Also, for evaluating the effect of DOI resolution, we simulated several X'tal cubes with crystal thickness from (0.5 mm)³ to (9 mm)³. We showed that sub-millimeter spatial resolution was possible using cubic crystals smaller than (1.0 mm)³ even with the assumed physical processes. Also, the weighted average spatial resolutions of both PET scanners with (0.5 mm)³ cubic crystals were 0.53 mm (14.6 cm ring diameter) and 0.48 mm (7.8 cm ring diameter). For the 7.8 cm ring diameter, spatial resolution with 0.5 × 0.5 × 1.0 mm³ crystals was improved 39% relative to the (1 mm)³ cubic crystals. On the other hand, spatial resolution with (0.5 mm)³ cubic crystals was improved 47% relative to the (1 mm)³ cubic crystals. The X'tal cube promises better spatial resolution for the 3D crystal block with isotropic resolution.

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1. Introduction

Positron emission tomography (PET) has become a popular imaging method in metabolism, neuroscience and molecular imaging. For dedicated human brain and small animal PET scanners, high spatial resolution is needed to visualize small objects. In current small animal PET scanners, spatial resolution is limited to 1–2 mm [1]. The spatial resolution of PET imaging is limited by several factors such as detector size, annihilation photon non-collinearity, positron range, parallax error, inter-crystal scattering, crystal decoding error, and sampling error of image reconstruction. Therefore, spatial resolution is represented as a convolution of these components [2].

Typical PET detectors have been designed with a 2D array of segmented scintillator crystals that are coupled to photomultiplier tubes on one side. Conventionally, the crystal cell, in which the photon interaction takes place, is determined by Anger-type calculation, which is weighted by each interaction intensity from the photomultiplier tubes. The parallax error caused by the thickness of the crystals degrades spatial resolution at the peripheral regions of the field-of-view (FOV) as shown in Fig. 1(a). Therefore, a depth-of-interaction (DOI) measurement is essential to achieve uniform and high spatial resolution (Fig. 1(b)). Many types of DOI detectors have been developed with the concept of extension of 2D detectors based on segmented crystals [3–9]. To eliminate the parallax error completely, the DOI resolution needs to be the same as the width of the segmented scintillator crystals.

We have developed a novel, general purpose isotropic-3D PET detector X'tal cube [10–13] (Fig. 1(c)) with effective readout of scintillation photons from six sides of the crystal block. In order to fabricate the 3D crystal block efficiently and precisely, we applied

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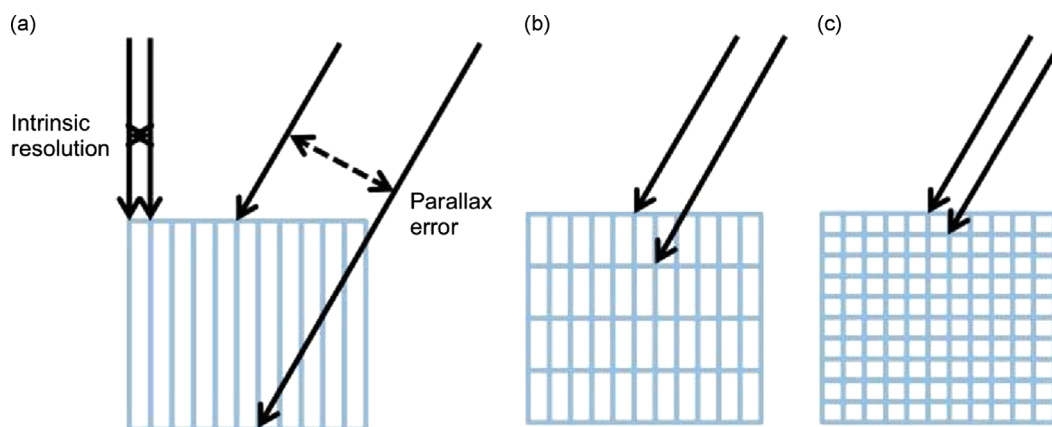


Fig. 1. Illustrations of effect of parallax error for three types of PET detectors: (a) 2D block detector (non DOI). (b) DOI detector and (c) isotropic resolution detector (X'tal cube).

a sub-surface laser engraving (SSLE) technique [14] to a monolithic crystal block instead of gluing segmented small crystals. The SSLE technique provides micro-crack walls. We have shown that the X'tal cube can achieve 1 mm^3 uniform crystal identification performance with the Anger-type calculation even at the block edges. We plan to develop the X'tal cube with even smaller 3D grids for sub-millimeter crystal identification. However, widespread application of the SSLE technique still requires meeting some challenging tasks.

2. Materials and methods

2.1. X'tal cube

The X'tal cube is composed of the 3D crystal block with isotropic resolution and arrays of multi-pixel photon counters (MPPCs) (Hamamatsu Photonics K.K., S10931-50P) [15]. Fig. 2 illustrates the X'tal cube with the 3D crystal block of $(1\text{ mm})^3$ cubic crystals. This X'tal cube consists of the LYSO cubic crystal block of $(18\text{ mm})^3$ in which the 3D grids of 1 mm pitch were fabricated by the SSLE technique. The SSLE technique is an internally focused laser processing technique which carves a groove into a monolithic scintillator block by irradiating a short pulse laser beam. The laser beam is focused by an objective lens at an internal point in the scintillator block which is normally transparent to its wavelength. When the laser beam is highly focused into the material (laser pulses of 10 ns duration at a repetition rate of 10 kHz), the intensity at the focal region becomes high enough to induce micro-cracks inside the material locally. Most of these micro-cracks are tens of micrometers in diameter. A dense arrangement of multiple micro-cracks works efficiently as a scattering wall for the scintillation photons. Therefore, fine segmentation can be formed in the monolithic scintillator block without inter-pixel gaps. Micro-crack walls are constructed by scanning the internal focused points, using an XYZ stage controlled by a personal computer. To stop 511 keV gamma rays from a positron emitter efficiently, crystal thickness is 18 mm. The 3D crystal block is composed of a $18 \times 18 \times 18$ array of lutetium yttrium oxyorthosilicate (LYSO) scintillator crystals with $(1\text{ mm})^3$ cubic crystals. 4×4 MPPCs are optically coupled to each surface of the 3D crystal block. To determine the positions of interaction, we adopted the Anger-type calculation for the signals of the MPPCs.

2.2. Simulation setup

Monte Carlo simulation has become a very useful tool for prediction of PET performance. In nuclear medicine research, the

Geant4 application for tomographic emission (GATE) [16,17] has been used for simulating several PET scanners and reliability and usefulness of the GATE platform have been validated. GATE provides the ability to model and account for the effects of detector size, annihilation photon non-collinearity, positron range, parallax error, and inter-crystal scattering.

For spatial resolution evaluation, a point source emitting 511 keV photons was simulated by GATE for all physical processes involved in emission and interaction of positrons. We simulated two types of animal PET scanners as shown in Fig. 3. The first PET scanner had a detector ring of 14.6 cm diameter composed of 18 detectors. The second PET scanner had a detector ring of 7.8 cm diameter composed of 12 detectors. The second PET scanner had the minimum gap between X'tal cubes with respect to thickness of the MPPC array. The energy windows of both PET scanners were set to 400–600 keV with a 15% energy resolution [12]. The simulation traced only gamma rays and neglected optical photon processes. Therefore, the simulation described only the scintillation block without the MPPC array. Also, we obtained only the interacting 3D positions in the $(18\text{ mm})^3$ monolithic LYSO crystal, because GATE cannot describe the complete crystal structure of the X'tal cube.

We evaluated spatial resolution using the ^{22}Na point source (1 kBq) to ignore influence of finite source dimension. Also, this point source covered with a cylindrical aluminum sleeve (3 mm length and 3 mm diameter) to absorb positrons from ^{22}Na . These simulations were repeated while varying the radial offset of the point source to demonstrate the spatial resolution at different locations across the FOV as shown in Fig. 3. Also, in the FOV center, the point source was placed at a 1.5-mm radial offset from the FOV center to avoid singularity. Measurement times of the ^{22}Na point source were 20,000 s for each position in order to obtain sufficient statistical accuracy.

2.3. Image reconstruction

The simulated data were projected onto the sinogram and reconstructed using 2D filtered backprojection. Pixel pitch was 0.1 mm for both PET scanners. FOVs of both PET scanners were equal to the respective ring diameters; this was because a DOI-PET scanner is expected to image as large as possible FOV for the ring diameter. The spatial resolution was characterized as the average value between radial and tangential widths of the reconstructed image, defining the width as its full width at half maximum (FWHM). Also, for evaluating the effect of DOI resolution, we defined the weighted average spatial resolution as

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