Available online at www.sciencedirect.com

SCIENCE DIRECT®

Nuclear Instruments and Methods in Physics Research B 241 (2005) 727–731

NIM B
Beam Interactions with Materials & Atoms

www.elsevier.com/locate/nimb

Preliminary report on the development of a high resolution PET camera using semiconductor detectors

Yohei Kikuchi [a,*], Keizo Ishii [a], Hiromichi Yamazaki [a], Shigeo Matsuyama [a], Takashi Yamaguchi [a], Yusuke Yamamoto [a], Takemi Sato [b], Yasushi Aoki [b], Kenichi Aoki [b]

[a] *Graduate School of Engineering, Department of Quantum Science and Energy Engineering, Tohoku University, Aramaki-Aza-Aoba 01, Aobaku, Sendai, Miyagi 980 8579, Japan*

[b] *Sumitomo Heavy Industries, Ltd., Kitashinagawa 5-9-11, Shinagawaku, Tokyo 141 8686, Japan*

Available online 29 August 2005

Abstract

We are developing a PET camera using small semiconductor detectors, whose resolution is equivalent to the physical limit of spatial resolution. First, a coincidence system of 16 Schottky CdTe detectors of 0.5 mm width obtained a resolution of <1 mm and it was confirmed that the Schottky CdTe detector is suitable for high resolution PET. Next, the performance of a pair of 32 channel CdTe arrays (1.2 mm width per channel) was investigated for the development of the prototype of high resolution PET. The time resolution between opposing detector pair was 13 ns (FWHM) when high voltage (700 V) was applied. The image of a 0.6 mm diameter point source was obtained in an experiment with opposing detector arrays using four channels, indicating that, a higher resolution can be achieved with the 32 channel CdTe array.

PACS: 87.58.Fg; 87.57.−s

Keywords: PET; Semiconductor PET; Spatial resolution; CdTe; Image

1. Introduction

Essentially all PET cameras use scintillation detectors. MicroPET [1] is one of the best spatial resolution PET cameras for animals, and has achieved a spatial resolution of 1.8 mm with LSO

* Corresponding author. Tel.: +81 22 217 7933; fax: +81 22 217 7929.

E-mail address: youhei.kikuchi@qse.tohoku.ac.jp (Y. Kikuchi).

0168-583X/$ - see front matter © 2005 Elsevier B.V. All rights reserved.
doi:10.1016/j.nimb.2005.07.124

scintillation detectors coupled to position sensitive photomultipliers.

The intrinsic spatial resolution of a PET camera is mainly affected by three factors [2]. One factor is the detector size [3], and the other two are caused by angular fluctuation, which are proportional to the gantry diameter, and by the extended positron range that depends on the emission energy of the positron emitter [4]. The physical limit of spatial resolution is determined by latter factors.

Although, scintillation type PET cameras cannot achieve spatial resolution of <1 mm, the calculated physical limit of spatial resolution for ^{18}F used as label nuclide is 0.4 mm for 10 cm diameter gantry, and 0.7 mm for 20 cm diameter gantry. These values indicate that detector size dominates the spatial resolution in conventional PET cameras.

We propose to use semiconductor detectors instead of scintillation detectors in order to pursue the physical limit of spatial resolution because semiconductor detectors can be miniaturized easily and signals are output from each small detectors individually, allowing to develop a semiconductor high resolution PET with small diameter gantry (about 10 cm) for small laboratory animals. CdTe detectors have been applied to γ-ray cameras [5], and are thus expected as detector for PET. We investigated the performance of a Schottky CdTe detector and confirmed that this detector is suitable for a PET camera.

The spatial resolution of CdTe detector pairs was measured, and images of a ^{22}Na point source and a finely structured phantom were obtained. From these results, it could be confirmed the spatial resolution of <1 mm can be achieved by using small size semiconductor detectors for PET.

2. The performance of CdTe detectors

2.1. Schottky CdTe detector

CdTe (cadmium telluride) is a compound semiconductor, and is a promising detector for a PET because the detector material is sensitive to 511 keV annihilation photons and can be used at room temperature. Moreover, because this detector type is commercially available, a steady supply is guaranteed.

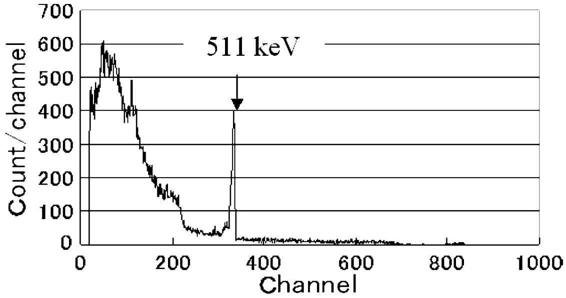

Fig. 1. ^{22}Na Energy spectrum measured with CdTe detector at 200 V bias. The peak near channel 300 corresponds to 511 keV annihilation photon.

However, the CdTe detector is afflicted with a long output signal risetime, which leads to poor time resolution, and PET detectors require good time resolution. Although this problem is reduced when high detector bias voltage is applied, leakage current of the detector increases concurrently.

By contrast, Schottky CdTe detector can be used at high bias voltage with low noise [6]. Therefore, we propose the use of Schottky CdTe detector produced by evaporating Pt and In as electrodes onto surfaces of CdTe wafer. We investigated the spectral performance and the time resolution of Schottky CdTe detector ($5.0 \times 5.0 \times 0.5$ mm^3) with a ^{22}Na point source (0.6 mm diameter, 2.0 MBq).

The ^{22}Na energy spectrum obtained is shown in Fig. 1. When the applied bias voltage is 200 V, a sharp peak followed by Compton edge and tail corresponding to 511 keV annihilation photons is measured.

Moreover, the time resolution measured with the ^{22}Na point source in coincidence between a detector pair was 8 ns at full width at half maximum (FWHM) in the coincident spectrum.

2.2. Intrinsic spatial resolution

By moving a positron emitter source between a pair of coincidence detectors, the count rate profile as a function of source position is obtained. The width of the profile compared to the point source indicates the intrinsic resolution defining the limit of resolution of the PET system. The profile of a pair of 0.5 mm wide Schottky CdTe detectors was obtained with the ^{22}Na point source, resulting in a FWHM of 0.8 mm at 10 cm distance between the detector pair. Because the profile includes the point source size, the intrinsic spatial resolution of the CdTe detector pair is <0.8 mm.

3. Coincidence system

To evaluate the spatial resolution of PET image with CdTe detectors, a coincidence system of a detector array with opposing detector pair and a rotating sample stage between the detector pair was built including signal processing electronics and a PC based data acquisition.

For the hot measurements, a pair of detector units including eight CdTe detector crystals were mounted on opposing sides of a movable stage pair, and annihilation photons emitted from the sample were measured.

In the detector unit, four 0.5 mm wide CdTe detectors are aligned with 2.4 mm spacing in transaxial direction, and four additional detectors are aligned behind. Data in the gap between detectors is acquired by moving the unit in 0.6 mm steps in transaxial direction. Angular data sample is acquired by rotating the sample.

Fig. 2 shows images reconstructed from data measured by this coincidence system with 10 cm distance between opposing detector units. The left image shows the reconstruction of the 0.6 mm diameter ^{22}Na point source with FWHM of the image profile being 0.9 mm. The right side shows the image of a finely structured phantom with several ^{22}Na hot spots of 0.6 mm and 1.0 mm diameter. The larger hot spots could be reconstructed and identified well; the small hot spots are at least visible in the image.

Conventional PET cameras cannot resolve this complex hot spot image due to their poor spatial resolution. By contrast, the coincidence system achieved high spatial resolution, attained by using small CdTe semiconductor detectors.

4. 32 Channel CdTe array

We are developing a semiconductor PET with small gantry (<10 cm) that will be applied to small laboratory animals like mice. When using an animal, it is necessary that detectors be packed densely unlike the above-mentioned coincidence system because efficient data acquisition is needed.

We suggest a multi-channel detector array obtained by dividing the surface electrode of a semiconductor crystal, is suitable for PET application. Thus we investigated the performance of a pair of 32 channel CdTe arrays (size per channel $1.2 \times 1.15 \times 4.5\ mm^3$) shown in Fig. 3.

The ^{22}Na energy spectrum was measured with two detector arrays. The detector response to

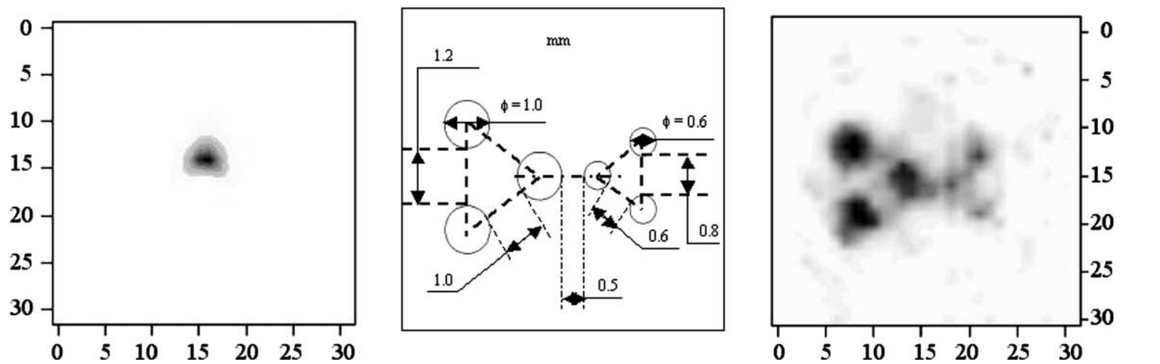

Fig. 2. Reconstructed image from data measured with the coincidence system. The coincidence system covers a field of view of 9.5 mm. The image was obtained with EM algorithm reconstruction method. The image size is $9.6 \times 9.6\ mm^2$, at pixel size of $0.3 \times 0.3\ mm^2$. Left: Image of ^{22}Na point source with 0.6 mm diameter. Center: Schematic arrangement of hot spots in finely structured phantom. Right: Reconstructed image of finely structured phantom.

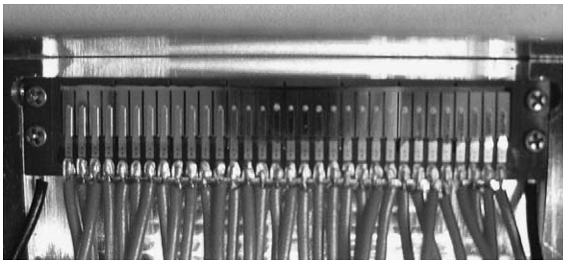

Fig. 3. The picture of 32 channel CdTe detector array. The 32 channel detector array is composed of eight CdTe crystals and each crystal is divided into four sections. Each section works as an individual detector.

511 keV annihilation photons measured for each channel of the detector arrays is practically identical. Therefore, the threshold level of energy discrimination can be set easily and precisely in this type of PET detector array. Moreover, because the noise level of Schottky CdTe detector is very low compared with 511 keV, it is possible to lower the threshold and obtain better efficiency.

Time resolution (FWHM) was measured with two detectors, one is each of the detector arrays and the other is BaF_2 detector. The time resolutions of the two detector arrays were not identical, and the average of resolution per channel in one detector array (detector array 1) was 4.4 ns, and was 9.7 ns in the other (detector array 2) at 700 V bias.

Time resolution of the CdTe detector array pair was 13 ns at 700 V bias. A resolution below 10 ns can be expected when using superior CdTe detector arrays like the detector array 1. A time resolution of <10 ns is sufficient for injecting activity of several hundred decays per second as is actually used in PET scans.

The spatial resolution of the CdTe detector array pair was measured in the same way as described for 0.5 mm wide CdTe detector pair, and a resolution of 1.0 mm at 10 cm distance between detector pairs was obtained. Note that this value includes the size of the point source (0.6 mm), and thus the intrinsic resolution is below 1.0 mm.

The image of the ^{22}Na point source was obtained with opposing detector arrays using four channels. It was assumed that three hot spots existed by moving the point source in steps of 1.5 mm during measurement.

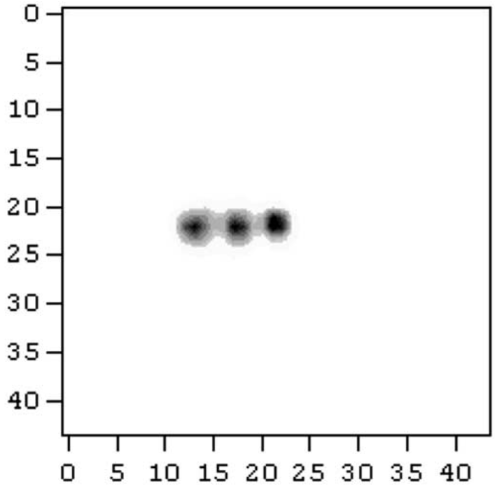

Fig. 4. Image reconstructed from data measured with opposing 32 channel detector arrays. Point source with 0.6 mm diameter was moved by 1.5 mm in two steps. The image was obtained with EM algorithm reconstruction method. The image size is 15.4×15.4 mm^2, at a pixel size of 0.35×0.35 mm^2.

Fig. 4 shows the reconstructed image. Three hot spots are well resolved in the image, and this shows that the detector array conserves high spatial resolution, although the intrinsic spatial resolution is inferior to the detector with 0.5 mm width.

5. Conclusion

The performance of the Schottky CdTe detector is largely sufficient for PET applications, and the spatial resolution of detector pairs as well as the images acquired in coincidence shows that, using small semiconductor detectors makes it possible to achieved high spatial resolution in a PET camera. The detectors can be packed densely in small space by using segmented multi-channel detectors.

We are now developing the prototype of semiconductor PET that can achieve a spatial resolution of <1 mm with small gantry diameter using a multi-channel detector array.

References

[1] S.R. Cherry, Y. Shao, et al., IEEE Trans. Nucl. Sci. 44 (1997) 21.

[2] L. Verger, M. Boitel, et al., Nucl. Instr. and Meth. A 458 (2001) 297.
[3] T. Budinger, K. Brennan, et al., Nucl. Med. Biol. 23 (1996) 659.
[4] S.E. Derenzo, IEEE Trans. Nucl. Sci. 33 (1986) 565.
[5] R. Lecote, D. Schmitt, et al., IEEE Trans. Nucl. Sci. 31 (1984) 556.
[6] K. Matsumoto, T. Takahashi, et al., IEEE Trans. Nucl. Sci. 45 (1998) 428.