1 Feasibility of a brain-dedicated PET-MRI system using 2 four-layer DOI detectors integrated with an RF head 3 coil

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14 Abstract

15 We are developing a PET-MRI system which consists of PET detectors integrated with the
16 head coil of the MRI in order to realize high spatial resolution and high sensitivity in simultaneous
17 measurements. In the PET-MRI system, the PET detectors which consist of a scintillator block,
18 photo-detectors and front-end circuits with four-layer depth-of-interaction (DOI) encoding capability
19 are placed close to the measured object. Therefore, the proposed system can achieve high sensitivity
20 without degradation of spatial resolution at the edge of the field-of-view due to parallax error thanks
21 to the four-layer DOI capability. In this paper, we fabricated a prototype system which consists of a
22 prototype four-layer DOI-PET detector, a dummy PET detector and a prototype birdcage type head
23 coil. Then we used the prototype system to evaluate the performance of the four-layer
24 DOI-PET detector and the reciprocal influence between the PET detectors and MRI
25 images.

26 The prototype DOI-PET detector consists of six monolithic multi-pixel photon counter
27 (MPPC) arrays (S11064-050P), a readout circuit board, two scintillator blocks and a copper
28 shielding box. Each scintillator block consists of four layers of Lu_{1.8}Gd_{0.2}SiO₅ : Ce (LGSO)
29 scintillators and reflectors are inserted between the scintillation crystals. The dummy detector has all
30 these components except the two scintillator blocks. The head coil is dedicated to a 3.0 T MRI
31 (MAGNETOM Verio, Siemens) and the two detectors are mounted in gaps between head coil
32 elements.

33 Energy resolution and crystal identification performance of the prototype four-layer

DOI-PET detector were evaluated with and without MRI measurements by the gradient echo and spin echo methods. We identified crystal elements in all four layers from a 2D flood histogram and energy resolution of 15 - 18 % was obtained for single crystal elements in simultaneous measurements. The difference between the average energy resolutions and photo-peak positions with and without MRI measurements was lower than 0.3 percentage points and 1.7 % for all layers. The results indicated that these performances were sufficient as PET detectors for the proposed PET-MRI system and there was no influence from the MRI measurements on the PET imaging in the simultaneous measurements. The signal-to-noise ratio of the MRI image and static magnetic field of the MRI were also evaluated with and without measurements of the PET detectors. The maximum decrease of the static magnetic field due to the LGSO scintillators was approximately 1 ppm. The signal-to-noise ratio decreased from 242.80 without the PET detector to 44.948 in simultaneous measurements.

Keywords: PET-MRI; PET detector; depth-of-interaction (DOI) detector; MPPC

1. Introduction

Recently, various types of PET-MRI systems are being developed by a number of research groups and several PET-MRI scanners are now commercially available [1-15]. The PET detectors used in these PET-MRI systems have no depth-of-interaction (DOI) capability except for two reports [7,13]. In the case of small ring PET scanners without DOI detectors [1-6,8,9,14] (fig. 1(a)), spatial resolutions are generally degraded in the periphery of the field-of-view by the annihilation photons entering scintillation crystals obliquely. On the other hand, large ring PET scanners (fig. 1(b)), such as whole body scanners [11, 12], suppress the parallax error for much smaller objects than the field-of-view, such as the human brain even without the DOI detectors. The small solid angle of the large ring PET scanners, however, causes low sensitivity and the large ring diameter leads to degraded spatial resolution due to collinearity of the annihilation photons. In contrast, the small ring PET scanners can achieve a higher solid angle and lower influence of the collinearity of the annihilation photons, compared with large ring PET scanners. Therefore, the small ring PET with the DOI detectors improves the performance of PET scanners, especially the scanner sensitivity and spatial resolution, in the PET-MRI system.

We are developing a brain-dedicated PET-MRI system in which four-layer DOI-PET detectors are integrated with the head coil of the MRI. The four-layer DOI detector we previously proposed [16] can identify scintillation crystals in all four layers with photo-detectors attached to only one side of a scintillation crystal block. The one-side readout with the four-layer DOI capability can realize a novel PET-MRI configuration which can achieve high sensitivity with high spatial resolution. A conceptual drawing of the proposed PET-MRI system is shown in fig. 1(c). The four-layer DOI detectors are mounted on the birdcage type head coil which consists of a number of rod-shaped coil elements. The scintillation crystals which are permeable to the RF-pulse from the MRI can be positioned closer to the measured object than the coil elements as shown in fig.1, whereas the photo-detectors and front-end circuits, placed inside the shielding boxes to minimize electric noises from the PET detectors and MRI system, need to be positioned outside the coil elements so as not to block RF-pulses from the head coil elements. Therefore, MRI images can be obtained without interference from the PET detector components.

We have developed a prototype system which consists of a prototype four-layer DOI-PET detectors, a dummy PET detector not including the scintillators and a prototype birdcage type head coil in order to evaluate reciprocal influence between the PET detector and the head coil of the MRI in simultaneous measurements. We evaluated crystal identification performance and energy resolution of the four-layer DOI-PET detector, the signal-to-noise ratio (SNR) of the MRI images and the influence of the PET detector materials on the static magnetic field of the MRI.

4 Fig. 1. Schematic drawings for comparison of (a) a small ring PET-MRI system without the DOI
 5 capability, (b) a whole body PET-MRI system which has a large diameter ring of the PET detectors
 6 and (c) the proposed PET-MRI system which has a small diameter ring of PET detectors with the
 7 four-layer DOI capability.

8 2. Material and methods

9 2.1 Prototype system for evaluation

10 Figure 2 shows a design of the prototype system for evaluation of the proposed integrated
 11 PET-MRI system. The prototype system consisted of the four-layer DOI-PET detector, the dummy
 12 PET detector and the prototype head coil. A photograph of the prototype system is shown in fig.3
 13

14 The prototype DOI-PET detector consisted of six multi-pixel photon counter (MPPC)
 15 arrays (S11064-050P, Hamamatsu Photonics K. K.), two scintillator crystal blocks, a readout circuit
 16 board and a copper shielding box (fig. 4). The dummy detector had all these components except the
 17 two scintillator crystal arrays. The MPPC arrays had 4×4 readout pixels. Dimensions of the readout
 18 pixels were $3 \times 3 \text{ mm}^3$ and each pixel consisted of $50 \mu\text{m}$ cells (3600 cells). The six MPPC arrays
 19 were arranged on a line and soldered onto the readout circuit board. The bias voltages to each
 20 MPPC pixel were controlled by variable resistances (VR) to obtain the same internal gain. The
 21 scintillator blocks consisted of $\text{Lu}_{1.8}\text{Gd}_{0.2}\text{SiO}_5 : \text{Ce}$ (LGSO, Hitachi Chemical Co., Ltd.,
 22 Japan) scintillators arranged in $12 \times 4 \times 4$ layers with reflectors inserted between them. The size of
 23 each crystal element was $2.9 \text{ mm} \times 2.9 \text{ mm} \times 5.0 \text{ mm}$. Reflector arrangements in each layer and the
 24 ideal flood histogram of the four-layer DOI detector are shown in fig. 5. As shown in fig. 5(b), the
 25 crystals in the four-layer DOI detector are represented as spots on one flood histogram by the
 26 four-layer DOI encoding method. Details of the four-layer DOI encoding method are described in ref.
 27 [16]. The multilayer polymer mirror (Sumitomo 3M, Ltd.) which had 0.065 mm thickness and 98 %

reflectivity was applied as the reflector. A half of each crystal block was positioned outside the shielding box where it projected from a hole (83 mm × 15 mm) on the top of the box as shown in fig. 2. The 96ch readout signals from all MPPC pixels were reduced to 4 ch signals with a weighted sum circuit (WSC) and amplified with amplifiers on the readout circuit board. The bottom photo of fig. 4 shows the readout circuit board. The amplified signals were fed to a data acquisition system to be digitalized and recorded using a Windows PC.

The prototype RF head coil was dedicated to a 3.0 T MRI (MAGNETOM Verio, Siemens). The diameter of the coil elements was 27.5 cm. There were eight coil elements and the PET detectors could be mounted on gaps between the coil elements. The four-layer DOI detector and the dummy detector were mounted on left and right sides of the prototype head coil, respectively. The four-layer DOI detector was used for the evaluation of the detector performance and the reciprocal influence between the PET detector and the head coil of the MRI in simultaneous measurements. The dummy detector was used to evaluate the influence of the scintillator blocks on the static magnetic field of the MRI. The detector positions are illustrated in fig. 2. The scintillator blocks of the four-layer DOI-PET detector were 136 mm from the center of the head coil.

Fig. 2. Design and position relationship of the prototype PET-MRI system

Fig. 3. Photograph of the prototype PET-MRI system

Fig. 4. Photographs of the four-layer DOI-PET detector (upper) and readout circuit board for the detector (lower)

Fig. 5. (a) Reflector arrangement and (b) the ideal flood histogram of the four-layer DOI-PET detector.

2.2 Experiment

We evaluated performance of the prototype four-layer DOI detector and influence of PET measurements on MRI images. The PET detectors, the head coil and cables were in the MRI room. The data acquisition system and power supplies for the MPPCs and the preamplifiers were outside it.

The lengths of the cables for signal readout and power supply were 10 m. No temperature control was applied to the PET detectors during the experiment. Figure 6 shows a photograph of the experimental setup. Energy spectra and flood histograms were obtained for 511keV annihilation radiations from a ^{22}Na point-like source with and without MRI measurements. Gradient echo (TR=7300ms, TE1= 2.46ms, TE2=7.38ms) and spin echo (TR=5000ms, TE= 94ms) methods were used in simultaneous measurements.

We also evaluated influence of the PET measurements on the MRI images. A cylindrical phantom (nickel chloride solution, $\Phi 220\text{mm} \times 120\text{mm}$ length) was positioned at the center of the field-of-view. Magnitude images of the cylindrical phantom were measured by the spin echo method (TR=5000ms, TE= 156ms) to evaluate the SNR of the MRI images for the simultaneous measurements. Influence of the PET detector materials on static magnetic field was evaluated using ΔB_0 maps obtained from phase images [17]. In this evaluation, the gradient echo method (TR=7300ms, TE1= 2.46ms, TE2=7.38ms) was used to obtain the phase images.

Fig. 6. Photograph of the experimental setup for simultaneous measurements.

3. Results

Figures 7 show flood histograms for uniform irradiation of the ^{22}Na point source with and without MRI measurements by the gradient echo (GRE) and spin echo (SE) methods. Only 511keV photo-peak events are represented on these flood histograms. The spots corresponding with two $12 \times 4 \times 4$ crystal blocks are shown in one flood histogram. Upper clusters represent interacted crystals in crystal block 1 in fig. 4 and the others correspond to crystal block 2 in the same figure. Figure 8 shows profiles for the crystals in the (a) first and second layers and (b) third and fourth layers for the three flood histograms of fig. 7. The crystals in all four layers clearly appear in all flood histograms. Comparison of the three flood histograms shows no degradation of crystal identification performance for all MRI sequences.

(a) MRI off

(b) GRE

(c) SE

Fig 7. Flood histograms for uniform irradiation of 511 keV annihilation radiations (a) without MRI measurement, (b) with MRI measurement (gradient echo) and (c) with MRI measurement (spin echo). The two crystal arrays are represented in the flood histograms. Upper clusters represent interacting crystals in crystal block 1 in fig. 4 and the others correspond to crystal block 2 also in fig. 4.

Fig 8. Profiles of the third row (arrow a, first and second layers) and fourth row (arrow b; third and fourth layers) of the spots representing crystals in the three flood histograms shown in fig. 7.

Figure 9 shows energy spectra for the 511keV annihilation radiation with and without MRI measurements. In the low energy region, noise components appear in the case of the simultaneous measurements since the noises from the RF-pulse of the MRI were over the threshold level of the data acquisition system. These noise components, however, were rejected by applying an energy window because they did not overlap with photo-peaks.

Figure 10 shows the energy spectra of single crystal elements in each layer with and without MRI measurements. The noise components derived from the RF pulse disappear from the energy spectra in fig. 10. PMT signals of the noise due to the RF pulse do not have position information and the noise events appear at the edge of the flood histogram. The noise components did not overlap with the spots indicating the crystals and they were rejected by setting a region-of-interest around only the spots on the flood histogram. Energy resolutions and photo-peak positions for single crystal elements were obtained from the energy spectra. The energy resolutions are shown in table I and fig. 11. The photo-peak positions are shown in table II and fig. 12. The difference between the average energy resolutions for the crystal elements in each layer with and without PET measurement is lower than 0.3 percentage points. The difference between the average photo-peak positions with and without the PET measurements is lower than 1.7 % for all layers. These results show the influence of the MRI on the energy information of the PET detector is negligible in the simultaneous measurements.

Fig. 9. The energy spectra for the four-layer DOI detector without MRI measurements, with GRE and with SE methods.

Fig .10. The energy spectra of single crystal elements in each layer without MRI measurements, and with GRE and SE methods.

Fig. 11. The energy resolutions of single crystal elements in each layer without MRI measurements, and with GRE and SE methods.

Table I. Energy resolutions of the single crystal elements (% , FWHM)

	MRI off	MRI on (GRE)	MRI on (SE)
1st layer	14.4 ± 0.9	14.6 ± 0.9	14.7 ± 1.0
2nd layer	15.1 ± 1.3	15.2 ± 1.2	15.3 ± 1.3
3rd layer	15.8 ± 1.7	16.0 ± 1.7	16.1 ± 1.7
4th layer	18.0 ± 4.0	18.0 ± 4.0	18.1 ± 3.9

Fig. 12. The photo-peak position of single crystal elements in each layer without MRI measurements, and with GRE and SE methods.

Table II. Photo peak positions of the single crystal elements (ch)

	MRI off	MRI on (GRE)	MRI on (SE)
1st layer	81.7 ± 1.7	83.1 ± 7.2	81.7 ± 7.1
2nd layer	83.1 ± 7.1	84.4 ± 7.2	83.0 ± 7.1
3rd layer	85.0 ± 7.1	86.2 ± 7.1	85.0 ± 7.0
4th layer	87.0 ± 8.0	88.1 ± 8.0	87.2 ± 7.9

Figure 13 shows magnitude images measured for the cylindrical phantom by the fast spin echo method. The left, center and right figures are obtained in the experiments without the PET detectors, with the PET detectors turned off and with the PET detectors in the simultaneous measurements, respectively. Figure 14 shows profiles along the horizontal line (a-b) and the vertical line (c-d) shown in fig. 13. The presence of the PET detectors changes the pixel values at the edges of the phantom image in the horizontal profile. In addition, we can observe the image becomes noisy in the case of the simultaneous measurements. The SNRs of the phantom images by the difference method in the three magnitude images are (a) 247.77, (b) 242.80 and (c) 44.948. The result shows that the current flowing in the PET detector and electronic noises from the acquisition system of the PET degrade the SNR of the MRI images. The noise from the acquisition system enters the MRI room through the signal cables. In addition, even when the PET detectors were turned off, the SNR is slightly degraded due to external noise contamination of the MRI signal, passing through the cables from outside the MRI room.

(a) Without PET detectors (b) With PET detectors turned off (c) Simultaneous measurements
Fig 13. Magnitude images measured for the cylindrical phantom by the fast spin echo method. The left, center and right figures were obtained in the experiments (a) without the PET detectors, (b) with the PET detectors turned off and (c) with simultaneous measurements.

(a) Horizontal profiles (b) Vertical profiles
Fig 14. Profiles along the horizontal line (a-b) and the vertical line (c-d) shown in fig. 13

In order to evaluate influence of the PET detectors on the static magnetic field of the MRI (B_0), ΔB_0 maps were calculated with and without the PET detectors. The ΔB_0 map shows the difference of the B_0 from the standard magnetic field in parts per million (ppm). The ΔB_0 maps were calculated from two phase images obtained by the gradient echo method for the cylindrical phantom. The ΔB_0 maps without and with the PET detectors are shown in figs. 15(a) and (b). The difference between the ΔB_0 maps without and with the PET detectors is shown in fig. 15(c). Differences of ΔB_0 are small on the right side; this is near the dummy detector in which no LGSO crystals were included. On the other hand, the differences of ΔB_0 increase on the left side of the phantom image; this is near the DOI-PET detector with the LGSO crystal arrays and the LGSO includes gadolinium which is a magnetic material. The maximum decrease of the static magnetic field due to the LGSO scintillators is approximately 1 ppm. In fact, compared with both sides of the profiles of the magnitude images (fig. 14(a)), the increase of the normalized pixel values on the left side with the four-layer DOI-PET detector including the LGSO crystals is larger than that on the right side without the LGSO crystals.

Fig. 15. ΔB_0 map (a) without the PET detectors, (b) with the PET detectors and (c) the difference between the ΔB_0 maps with and without the PET detectors.

4. Discussion and conclusion

We evaluated the influence of the MRI measurement on the DOI-PET detector in simultaneous measurements. We also evaluated the influence of the PET detectors on the MRI measurement in the simultaneous measurements.

We were able to identify all crystal elements in all four layers from a flood histogram and we obtained energy resolutions of 15 - 18 % for each single crystal element even in the simultaneous measurements. Additionally, we observed no degradation of the crystal identification performance and the energy resolutions of the DOI-PET detector in the simultaneous measurements. These results indicated that these performances are sufficient as PET detectors for the PET-MRI and there is no influence by the MRI measurements on the PET imaging in the simultaneous measurements using the gradient echo and the spin echo methods.

On the other hand, influence on the MRI measurement was observed in the simultaneous measurements in terms of the SNR of the MRI images and the uniformity of the static magnetic field by the external noises and the magnetism of the LGSO crystals in the scintillator arrays. The SNR ratio dropped from 242.80 without the PET measurement to 44.948 with the PET measurements due to noise contamination from outside the MRI room. The detector position which is close to the elements of the head coil and the opening spaces on the top of the shielding boxes were the primary reasons for the SNR degradation. We expect that the contamination of the external noises can be reduced by inserting band-pass filters into the cables at a penetration panel. Additionally, we are investigating effective shielding methods for the shielding boxes, especially regarding the holes on the top of the shielding boxes. The maximum decrease of the static magnetic field by the LGSO scintillator arrays was approximately 1 ppm. The influence of Gd on the static magnetic field can be avoided by using non-magnetic scintillators, such as LSO and LYSO.

No temperature control on the MPPCs was applied during the experiment with

the presented system. Internal gain of the MPPCs is strongly affected by temperature variation. We think that it was negligible in this experiment, because we performed calibrations just before the experiment and the temperatures of the MPPCs did not change during the experiment which continued for a few hours. However, the room temperature may change in long time operations. We will add a simple gain control system to the detector modules which will adjust the supply voltage by monitoring temperatures around the MPPCs.

As a next step, we are planning a one-ring prototype which consists of eight PET detectors for evaluation of PET imaging performance. Then those modifications will be applied in order to improve the MRI images in simultaneous measurements.

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