

Home Search Collections Journals About Contact us My IOPscience

SiPM-PET with a short optical fiber bundle for simultaneous PET-MR imaging

This article has been downloaded from IOPscience. Please scroll down to see the full text article. 2012 Phys. Med. Biol. 57 3869 (http://iopscience.iop.org/0031-9155/57/12/3869)

View the table of contents for this issue, or go to the journal homepage for more

Download details: IP Address: 147.46.162.253 The article was downloaded on 28/06/2012 at 11:02

Please note that terms and conditions apply.

Phys. Med. Biol. 57 (2012) 3869-3883

# SiPM-PET with a short optical fiber bundle for simultaneous PET-MR imaging

Seong Jong Hong<sup>1</sup>, Han Gyoo Kang<sup>1</sup>, Guen Bae Ko<sup>2,3</sup>, In Chan Song<sup>5</sup>, June-Tak Rhee<sup>6</sup> and Jae Sung Lee<sup>2,3,4,7</sup>

<sup>1</sup> Department of Radiological Science, Eulji University, Seongnam, Gyeonggi 461-713, Korea

<sup>2</sup> Department of Nuclear Medicine, Seoul National University, Seoul 110-744, Korea

<sup>3</sup> Department of Biomedical Sciences, Seoul National University, Seoul 110-744, Korea

<sup>4</sup> WCU Department of Brain & Cognitive Sciences, Seoul National University, Seoul 110-744, Korea

<sup>5</sup> Department of Radiology, Seoul National University Hospital, Seoul 110-744, Korea

<sup>6</sup> Department of Physics, Konkuk University, Seoul 143-701, Korea

E-mail: jaes@snu.ac.kr

Received 8 December 2011, in final form 27 March 2012 Published 30 May 2012 Online at stacks.iop.org/PMB/57/3869

#### Abstract

For positron emission tomography (PET) inserts to magnetic resonance imaging (MRI) applications, optical fibers have been used for some time to transfer scintillation photons to photomultiplier tubes positioned outside the fringe magnetic field. We previously proposed a novel utilization of an optical fiber for good radio frequency (RF) transmission from body coils to an imaging object. Optical fiber bundles between silicon photomultipliers (SiPM) and scintillation crystals provide an increased spacing between RF-shielded electronics boxes, facilitating RF passage from the body RF coils to imaging objects. In this paper, we present test results of a SiPM-PET system with a short optical fiber bundle for simultaneous PET-MR imaging. We built the SiPM-PET system which consisted of 12 SiPM-PET modules; each module was assembled with a lutetium yttrium oxyorthosilicatecrystal block, a 31 mm optical fiber bundle, a Hamamatsu multi-pixel photon counter S11064-050P and a signal processing box shielded with copper. The SiPM-PET system, with a face-toface distance of 71 mm, was placed inside a 3 T MRI. A small surface coil placed inside the SiPM-PET system was used to receive the signal from phantoms while the body RF coil transmitted the RF pulses. The SiPM-PET system showed little performance degradation during the simultaneous PET-MR imaging and it caused no significant degradation of MR images with turbo spin echo (TSE), gradient echo or 3D spoiled gradient recalled sequences. Echo planar imaging MR images with and without the SiPM-PET inside the

<sup>7</sup> Author to whom any correspondence should be addressed.

0031-9155/12/123869+15\$33.00 © 2012 Institute of Physics and Engineering in Medicine Printed in the UK & the USA 3869

MR scanner were significantly worse than the images obtained with the TSE sequence.

(Some figures may appear in colour only in the online journal)

## 1. Introduction

A whole-body simultaneous positron emission tomography (PET)-MR imager is fast becoming a reality as major works carried out with small animal systems have proven to be very successful (Judenhofer *et al* 2008, Catana *et al* 2006, Lee and Hong 2010, Kwon *et al* 2011, Lee and Kang 2012). Most PET/magnetic resonance imaging (MRI) systems for small animal imaging use radio frequency (RF)-transceiver coils for RF transmission and reception. The latest commercial PET-MR system for simultaneous PET/MR imaging placed a PET system between the body RF coil and the gradient coil, requiring significant modifications to the MRI (Drzezga *et al* 2011), therefore preventing use with existing MRIs. Alternatively, the latest trend in RF coil development is to use body coils for RF transmission and receiver-only coils for RF reception for brain, breast and extremity MR imaging, because of their enhanced signal sensitivity, broader coverage and usage together with the parallel imaging technique (Roemer *et al* 1990, Pruessmann *et al* 1999, Larkman and Nunes 2007).

Optical fibers have been used for sometime in PET/MRI applications using photomultiplier tubes (PMTs), as they are susceptible to the magnetic field. In these applications, relatively long optical fibers transfer photons from scintillators to PMTs positioned outside the fringe magnetic field. The long optical fibers significantly deteriorated both energy and time resolutions such that practical PET/MRIs were not very attractive (Slates *et al* 1999, Yamamoto *et al* 2004, Raylman *et al* 2006). Optical fibers have been used with position-sensitive avalanche photodiodes (PSAPDs) for other reasons (Catana *et al* 2006), the primary reason being to remove metallic parts from the vicinity of an object being imaged. We also proposed a silicon photomultipliers (SiPM)-PET with a short optical fiber bundle bent  $90^{\circ}$  (Hong *et al* 2011), which employed receiver-only coils for the use of brain, breast and extremity MR imaging. However, the  $90^{\circ}$  bend not only caused diminished energy and time resolutions but also resulted in difficulty extending the PET system in the axial direction.

We previously proposed the novel use of an optical fiber for good RF transmission from the body RF coils to imaging objects (Kang *et al* 2011). In this paper, we present test results of the SiPM-PET system for simultaneous PET-MR imaging with and without 3 T MR (MAGNETOM Trio, Siemens  $AG^8$ ) operation.

#### 2. Materials and methods

## 2.1. Effects of copper shielding on the MR phantom image

In order to study the effects of copper shielding on MR images, an acrylic cylinder containing water, hereafter a water phantom, was wrapped with four different types of copper foils as shown in figure 1. The acrylic cylinder was 60 mm in diameter and 96 mm in length. The thickness of the copper foil was 0.03 mm. Types 1 and 2 copper foils in figures 1(b) and (c) were 60 mm in diameter, but the type 2 was assembled with  $2 \times 4$  patches, each of which had a size of  $46 \times 51$  mm<sup>2</sup>. Types 3 and 4 foils in figures 1(d) and (e) were 100 mm in diameter,



**Figure 1.** Acrylic cylinders wrapped with four different types of copper foil. The acrylic cylinder contained water: (a) the water phantom, (b) the water phantom wrapped with a solid copper sheet with a 60 mm diameter, (c) with  $2 \times 4$  patches, (d) with a solid copper sheet with a 100 mm diameter and (e) with  $2 \times 6$  patches.



**Figure 2.** Schematics of SiPM-PET with short optical fiber bundles: (a) a longitudinal cutaway view and (b) a transverse cutaway view.

and the type 4 was made of  $2 \times 6$  patches. The spacing between the patches was about 1 mm. The MR images were obtained with a turbo spin echo (TSE) sequence (repetition time (TR) = 3000 ms, echo time (TE) = 97 ms, flip angle (FA) = 120°, field of view (FOV) = 49 × 49 mm<sup>2</sup>, slice thickness = 3 mm).

## 2.2. Performance of SiPM-PET modules depending on the fiber bundle

Based on the study of effects of copper shielding on MR images and on previous work on the SiPM-PET with a short optical fiber bundle bent 90° (Hong *et al* 2011), we proposed a SiPM-PET configuration with short optical fiber bundles. The relatively short length of optical fiber bundles is expected not to deteriorate PET performance significantly while optical fiber bundles provide an increased spacing between RF-shielded electronics boxes facilitating RF passage to imaging objects as shown in figure 2.

The length of an optical fiber can be varied to optimize PET performance and MR imaging. The longer the length of the optical fiber, the wider the spacing between RF-shielded electronics boxes resulting in better RF passage but worse energy and timing resolutions. However, the longer optical fiber length does not increase the spacing between crystal blocks.

The SiPM-PET module consisted of a lutetium yttrium oxyorthosilicate (LYSO) crystal block, a double-clad fiber bundle (Kuraray Co. Ltd<sup>9</sup>) and a multi-pixel photon counter (MPPC)

9 http://www.kuraray.co.jp/en/



**Figure 3.** One of the SiPM-PET modules assembled with a RF-shielded box: (a) components and jigs, (b) an optical fiber bundle with a jig, (c) an assembled module, (d) one of the copper shield sides showing  $4 \times 8$  through holes and (e) an assembled module with a RF-shielded box. White paint was applied over the  $4 \times 8$  through holes.

S11064-050P (Hamamatsu Photonics<sup>10</sup>) assembled using custom-built jigs as shown in figure 3(a) to achieve good optical contacts between various components. For good optical couplings, a 1 mm thick soft polyvinyl chloride foil was inserted between the optical fiber bundle and the crystal block, and between the optical fiber bundle and the MPPC.

The LYSO crystal block consisted of  $6 \times 6$  crystals with a dimension of  $2.47 \times 2.74 \times 20$  mm<sup>3</sup>, each of which was inserted into one of the  $6 \times 6$  holes assembled with 3M-enhanced spectral reflector (ESR) polymer<sup>11</sup> (Hong *et al* 2008a). Since the optical fiber length has to be optimized against conflicting requirements of PET performances and the spacing between RF-shielded boxes, we tested a SiPM-PET module with three different fiber bundle lengths of 31, 56 and 86 mm assembled with double-clad optical fibers of 1.0 and 1.5 mm diameter. The optical fiber bundle, with a dimension of  $30 \times 30$  mm<sup>2</sup>, required approximately 400 optical fibers with a 1.0 mm diameter and approximately 180 optical fibers with a 1.5 mm diameter.

The front-end electronics originally developed for the readout of the  $2 \times 2$  MPPCs was used to read out one MPPC (Yoon *et al* 2012, Ko *et al* 2011). The single MPPC was inserted at the center of  $2 \times 2$  matrices, each of which consisted of four 16-pin connectors. The frontend electronics consisted of a resistive charge division network, and a differential amplifier was enclosed in a shielded box made of copper clad laminates with an 18  $\mu$ m thick copper layer. The spacing between the pads was 0.1 mm. The shielded box, with a dimension of  $34 \times 24 \times 180$  mm<sup>3</sup>, also had  $4 \times 8$  through holes on one face of the box for 32 MPPC pins as shown in figure 3(d). One of the interesting features of the SiPM-PET module is that the SiPM was located outside the enclosed electronics box.

We tested the SiPM-PET module using a data acquisition system which consists of nuclear instrumentation module and versa module eurocard modules. Figure 4 shows the

<sup>10</sup> http://www.hamamatsu.com/

<sup>11</sup> http://www.3m.com/



Figure 4. Data flow diagram to measure energy and coincidence timing resolution of the SiPM-PET modules with short optical fiber bundles.

data flow diagram. Signals from the SiPMs were amplified by a factor of 10 using a CAEN N412 amplifier module, and then forwarded to a CAEN N842 constant fraction discriminator and a CAEN C205A charge-to-digital converter (QDC) module<sup>12</sup>. Digital signals from the discriminator were forwarded to a CAEN N455 coincidence module to generate QDC GATE and time-to-digital converter STOP signal with a 400 ns width. A  $3.7 \times 10^5$  Bq <sup>22</sup>Na source was placed between two SiPM-PET modules to obtain coincidence events. Crystal maps, energy and coincidence timing distributions were then obtained.

# 2.3. Performances of the SiPM-PET during the simultaneous PET-MR imaging

The SiPM-PET system, shown in figure 5(a), had a face-to-face distance of 71 mm and consisted of 12 SiPM-PET modules, each of which was assembled with a  $6 \times 6$  crystal block, a 31 mm-long fiber bundle made of approximately 400 double clad optical fibers with a diameter of 1.0 mm, a MPPC and an electronics box. The SiPM-PET module shown in figure 3 was used to assemble the SiPM-PET. The spacing between the electronics boxes was 15 mm at the inner side and 25 mm at the outer side, as shown in figure 5(a). The SiPM-PET system was placed inside a 3 T MR (MAGNETOM Trio, Siemens AG, See: Footnote 8), as shown in figure 5(b). Signals from the 12 SiPM-PET modules were transferred to the data acquisition system outside the MRI room via nonmagnetic shielded-twist-pair (FTP) cables and an isolation panel on the wall of the MRI room. The SiPMs were believed to be at 20 °C, the MRI room temperature, during the simultaneous PET-MR imaging since it was located outside the enclosed electronics box, as shown in figures 3(c) and (e).

Two different phantoms were used to test the performance of the SiPM-PET. One phantom, hereafter a gelatin phantom, shown in figure 6 was custom-built of gelatin and three different diameters of glass cylinders inside a hexagonal prism which consisted of six  $60 \times 55 \text{ mm}^2$  acrylic rectangular plates and two hexagonal plates with a 60 mm side length. The gelatin phantom had an additional 3 mm vertical acrylic plate indicated by a black arrow



Figure 5. The SiPM-PET and surface coil: (a) a schematic diagram, (b) a photo showing the SiPM-PET and the 3 T MRI and (c) the surface coil.



**Figure 6.** The prototype PET/MRI/US phantom under development: (a) a schematic diagram and (b) a photo showing glass cylinders which contained yellow ink to check leakage. The length of the phantom is 55 mm. The black arrow indicates a 3 mm thick acrylic plate to support glass cylinders.

in figure 6(b) to support the glass cylinders. A MR contrast agent, DOTAREM<sup>13</sup> with a 1:1 mixture with a 0.9% saline solution, was put into the cylinders. The inner diameters of the three cylinders were 1.5, 3.0 and 5.0 mm, respectively. The other phantom, hereafter a cucumber–water phantom, consisted of two syringes with a 10 mm inner diameter filled with water and a cucumber into which a 3.0 mm diameter capillary filled with <sup>18</sup>F ( $1.1 \times 10^7$  Bq) solution was inserted.

The list-mode data from the data acquisition system was sorted into a three-dimensional sinogram, and then rebinned into two-dimensional data using the single-slice rebinning method. All scanned PET data were reconstructed using the maximum likelihood expectation maximization reconstruction with exact position information for each LOR element.

13 http://www.guerbet.com/

Table 1. Summary of pulse sequences.				
Pulse sequence (weighted)	TR (m)	TE (ms)	Flip angle	TI (
TSE sequence (T2)	3000	97	90	_
TSE sequence (T1)	475	15	180	-
GE sequence FLASH (T1)	116	3.83	25	_
GE sequence (T2*)	425	25	20	-
3D TurboFLASH SPGR (T1)	1670	2.3	9	900

We tested the performances of the SiPM-PET with MR pulse sequences currently used in clinical diagnosis. The MR pulse sequences are shown in table 1. To find out possible MRI interferences to the SiPM-PET, event rates over time were measured during various MRI sequences and without a MRI sequence.

# 2.4. Influences of the SiPM-PET on MR images during the simultaneous PET-MR imaging

We tested the influences of the SiPM-PET on MR images during the simultaneous imaging with MR pulse sequences described in subsection 2.3. The same phantoms described in section 2.3 were used. The MR pulse sequences were chosen to see T1, T2 and T2\* effects on gelatin and water in the two phantoms. All MR images were obtained with 3 mm slice thickness,  $256 \times 256$  matrix and 3 averaging except for the T2 sequence, which was obtained with a  $384 \times 384$  matrix and 8 averaging. During the simultaneous PET-MR imaging, the body RF coil transmitted the RF pulses to the phantoms while the surface coil shown in figure 5(c) received the signals from the phantoms. The surface coil was placed inside the SiPM-PET. The image fusion of PET and MRI was performed using software called FIRE (functional image registration) (Lee *et al* 2005).

To study RF interferences to MRI from the SiPM-PET,  $\pm 250$  kHz frequency bands around the Larmor frequency 122.7 MHz of the 3 T were probed after a 'noise sequence' provided by the manufacturer of the MR system (Wehrl *et al* 2011) was applied. The 'noise sequence' was applied for four different cases: without the SiPM-PET placed inside the MRI bore, with the SiPM-PET placed but power off, with the SiPM-PET placed but power on and finally with the SiPM-PET taking data.

To investigate the possibility of functional imaging, an echo planar imaging (EPI) sequence  $(TR = 2000 \text{ ms}, TE = 30 \text{ ms}, \text{ inversion time} (TI) = 900 \text{ ms}, FA = 90^{\circ}, FOV = 280 \times 560 \text{ mm}^2$ , matrix size =  $128 \times 256$ , slice thickness = 4 mm, 10 slices, bandwidth = 2520 Hz/pixel) was applied to a uniform gelatin phantom. The uniform gelatin phantom was made with the same material as the gelatin phantom in figure 6 except that the phantom did not contain the glass cylinders and the acrylic plate at the middle of the phantom. This EPI sequence was slightly modified from the EPI sequence attempted by Wehrl *et al* (2011). A total of 20 volume scans without the SiPM-PET and a total of 165 volume scans with the SiPM taking data inside the MRI bore were obtained.

# 3. Results

#### 3.1. Effect of copper shielding on the MR phantom image

Figure 7 shows, arranged in the same order as in figure 1, the MR images of the water phantom depending on the shielding types. Signal intensities, signal-to-noise ratios (SNR),

(ms)



Figure 7. MR images of the water phantom depending on the shielding types, shown in figures 1(a)-(e). Squares indicate the areas where signal intensities, SNR and CNR were determined.

Table 2. Summary of average intensity, SNR and CNR depending on copper shielding type.

Shielding type	Average signal intensity	SNR	CNR
No shielding	237	19.2	28.7
Type-1	155	5.3	7.2
Type-2	238	16.4	23.8
Type-3	155	2.7	2.0
Type-4	237	18.3	27.6

Table 3. Energy resolutions depending on the fiber bundle.

Fiber length: 31 mm			
Fiber diameter (mm)	No fiber	1.0	1.5
Energy resolution (%)	14	18	18
Fiber diameter: 1.5 mm			
Fiber length (mm)	31	56	86
Energy resolution (%)	18	18	19

and contrast-to-noise ratios (CNR) defined by 1 below are given in table 2 (Hendrick 2008):

$$SNR = \frac{S_{syringe}}{\sigma_{syringe}} \quad CNR = \frac{S_{syringe} - S_{bkg}}{\sigma} \quad \sigma = \frac{\sqrt{\sigma_{syringe}^2 + \sigma_{bkg}^2}}{2}.$$
 (1)

As shown in figure 7 and table 2, signal intensities, SNRs and CNRs are strongly dependent on the shielding types. Types 3 and 4 showed the worst and the best image quality, respectively. Type 4, in which the whole piece of the copper foil was broken into the  $2 \times 6$  patches as shown in figure 1(e), gave the best image quality.

# 3.2. Performance of SiPM-PET modules depending on the fiber bundle

Table 3 gives the energy resolutions for various combinations of optical fiber bundle lengths and diameters. We obtained a 14% energy resolution without optical fiber coupling. With the 31 mm fiber bundle, an 18% energy resolution was obtained with the 1.0 and 1.5 mm diameter fiber. As the optical fiber bundle increased from 31 to 86 mm, the energy resolution did not depend strongly on the length of the fiber bundle. We also obtained a 2.0 ns coincidence timing resolution with the 31 mm fiber bundle length and 1.5 mm diameter fiber.

# 3.3. Performances of the SiPM-PET during the simultaneous PET-MR imaging

Figure 8 shows crystal maps obtained with <sup>18</sup>F loaded into two syringes, QDC distributions and energy resolutions with and without MR pulse sequences. As indicated by squares in



**Figure 8.** Flood maps, QDC distributions and energy resolutions from one of the SiPM-PET modules: (a) and (d) 3 T only, (b) and (c) TSE (T2) and (c) and (e) GE (T1).



Figure 9. Count rates over time and during various MR operations.

figure 8, the QDC distributions obtained from the crystals did not contain characteristic lowenergy peaks caused by RF interference. Little degradation of energy resolution due to the MR pulse sequences was observed. Average energy resolutions for 36 crystals were 22.6%  $\pm$ 1.2% for 3 T, 23.0%  $\pm$  1.3% for TSE (T2) and 22.7%  $\pm$  1.2% for GE (T2).

The event rates over time were shown in figure 9. The event rates did not depend on the MRI operation and the MR pulse sequences.

Figure 10 shows the MR image of the cucumber-water phantom (left) and the reconstructed PET image of a capillary tube embedded into a cucumber (right), obtained



**Figure 10.** Reconstructed phantom image of the capillary tube imbedded in the cucumber: (a) MR, (b) PET and (c) overlaid PET-MR image.



**Figure 11.** MR phantom images of the custom-built gelatin phantom, (a) with and (b) without the SiPM-PET, obtained with the TSE (T2) sequence.

during simultaneous PET-MR imaging. An overlaid PET-MR image is shown at the center of figure 10.

## 3.4. Influences of the SiPM-PET on MR images during the simultaneous PET-MR imaging

Figure 11 shows MR images of the gelatin phantom with and without the TSE sequence of a  $3 \times 10^3$  ms repetition and 101 ms echo time. The MR images were from the fourth slice out of ten slices. The signal intensities were higher at the bottom where the small 40 mm diameter receiver coil was located, and they became lower as the distance from the surface coil increased. The average signal intensities over the gelatin phantom were 1171 and 1163 without and with the SiPM, respectively. We also measured the SNRs to be 24.5 without the SiPM-PET and 21.9 with the SiPM-PET. The CNRs were 43.2 and 48.6 without and with the SiPM-PET, respectively. Table 4 summarizes the average signal intensities, SNRs and CNRs obtained with the gelatin phantom. The first/second numbers in table 4 represent the values



**Figure 12.** Signal intensities and SNRs in ten slices over the phantom with/without the SiPM-PET: (a) signal intensity with TSE (T2) and (b) SNR with TSE (T2). The dips in the slice number 7 were caused by the 3 mm thick acrylic plate indicated by the black arrow in figure 6.

Table 4. Summary of average intensity, SNR and CNR in the gelatin phantom depending on pulse sequences (without/with SiPM-PET).

Pulse sequence (weighted)	Average signal intensity	SNR	CNR
TSE sequence (T2)	1171/1163	24.5/21.9	48.6/43.2
TSE sequence (T1)	413/421	14.8/17.5	28.2/32.8
GE sequence FLASH (T1)	363/372	16.5/13.2	32.3/25.7
GE sequence (T2*)	34.3/38.1	1.5/1.7	2.4/2.8
3D TurboFLASH SPGR (T1)	125/158	17.2/15.1	34.0/29.8

 
 Table 5. Summary of average intensity, SNR and CNR in the cucumber–water phantom depending on pulse sequences (without/with SiPM-PET).

Pulse sequence (weighted)	Average signal intensity	SNR	CNR
TSE sequence (T2)	N A/850	N A/3.6	N A/7.3
TSE sequence (T1)	214/214	3.7/3.9	7.0/7.3
GE sequence FLASH (T1)	305/271	3.5/3.7	6.9/7.2
GE sequence (T2*)	286/276	3.8/5.1	7.4/10.1
3D TurboFLASH SPGR (T1)	94/93	4.4/4.4	8.4/8.3

taken without/with the SiPM-PET, respectively. Signal intensities and SNRs in the ten slices over the gelatin phantom are shown in figure 12. The results in figure 12 and in table 4 show that no significant degradation occurred due to the presence of the SiPM-PET. The dips in the signal intensity and SNR at the slice number 7 were caused by the acrylic plate indicated by the arrow in figure 6(b).

The MR phantom images of the cucumber–water phantom were similar with and without the gradient echo (GE) sequence with a 116 ms repetition and 3.83 ms echo time. The MR images were again taken from the fourth slice out of ten slices. Table 5 summarizes the average signal intensities, SNRs and CNRs obtained from water inside the syringe of the cucumber–water phantom. The average signal intensities over water inside the syringe of the cucumber–water phantom were 305 and 271 without and with the SiPM-PET, respectively. We also obtained the SNRs to be 3.5 without the SiPM-PET and 3.7 with the SiPM-PET. The CNRs were 6.9 and 7.2 without and with the SiPM-PET. Signal intensities and SNRs over the ten slices were similar with and without SiPM-PET. No significant degradation was again observed. The average signal intensity, SNR and CNR without the SiPM-PET are not given



**Figure 13.** MR phantom images of the uniform gelatin phantom and signal intensity over image number; (a) MR phantom image without the SiPM-PET inside the MRI, (b) MR phantom image and (c) signal intensity over image number during the simultaneous PET/MR imaging.

in table 5 for the TSE sequence with T2-weighting because the water phantom was moved during the MR imaging. The movement was caused by incomplete filling of water.

Test results of RF interferences to MRI from the SiPM-PET within  $\pm 250$  kHz frequency bands around the Larmor frequency 122.7 MHz produced all zeros for all four different cases: without the SiPM-PET placed inside the MRI bore, with the SiPM-PET placed but power off, with the SiPM-PET placed but power on and finally with the SiPM-PET taking data. To verify the zero values, the same 'noise sequence' was applied keeping a MRI room door opened. This produced spikes, 64 at -200 kHz, 229 at -10 kHz and 175 at 190 kHz, measured in arbitrary unit.

Figures 13(a) and (b) show EPI images of the uniform gelatin phantom without and with the SiPM-PET taking data. Both images were significantly worse than the images in figure 10. The signal intensity in the region of interest indicated by the red circle in figure 13(a) was independent of the image number without the SiPM-PET. However, the signal intensity was slowly degraded by -0.9% over 165 images (5 min 30 s) with the SiPM-PET taking data inside the MRI, as shown in figure 13(c).

# 4. Summary and conclusion

The proposed SiPM-PET inserted into the 3 T MRI bore did not degrade the MR images during the various MRI sequences excluding an EPI sequence, and the MR operation also did not deteriorate the SiPM-PET performance.

We believe that the type 3 produced worse image quality than type 1 because more magnetic flux passed through the cylinder, inducing more current in the shielding. Type 4 gives the best image quality because of the better RF transport to the water phantom and the longer distance between the water phantom and the copper shielding.

Using the relatively short optical bundle of 31 mm length, we achieved about 20% energy resolution and 2.0 ns time resolution from the SiPM-PET modules. The energy resolutions in figure 8 are worse than the energy resolutions in table 3, probably because of non-optimal optical contacts along the light path. Energy resolutions of the SiPM PET modules, up to the 100 mm long fiber bundle, did not seem to depend strongly on the fiber length and the coincidence timing resolution should be adequate for coincidence detection.

The SNRs and CNRs in table 4 were better with the SiPM-PET for the TSE sequence with T1-weighting and the GE sequence FLASH with T2\*-weighting, probably due to statistical fluctuations in measurements.

Even though we have used the 31 mm long optical fiber bundles with 1.0 mm diameter optical fibers, neither the length of the optical fiber bundle nor the diameter of the fiber have been optimized. The longer optical fiber bundle which increases the trans-axial spacing between shielded electronics boxes might be needed for more demanding MR applications such as EPI and MR spectroscopy.

We have attempted an EPI sequence slightly modified from the EPI sequence by Wehrl *et al* (2011). MRI images with and without the SiPM-PET inside the MRI were significantly worse than the images obtained with the TSE (T2) sequence, something which requires further study. On the other hand, the signal intensity was slowly degraded by -0.9% over 165 images (5 min 30 s) with the SiPM-PET taking data inside the MRI. Even though we obtained zero values for the test MRI from the SiPM-PET within  $\pm 250$  kHz frequency bands around the Larmor frequency 122.7 MHz, further test is needed with a finer scale as shown by Werhl *et al* (2011).

A PET-MR system which did not use copper shielding for PET electronics was recently reported (Vaska *et al* 2011). The authors did not observe MR interference and claimed the distance between an imaging object and PET electronics which was an important factor. The SiPM-PET system that we developed also increases the distance between the imaging object and electronics by the optical fiber length. This increase might have contributed to the absence of MR interference and it would also help improve the field uniformity near the imaging object, resulting in insignificant differences in MR images with and without the SiPM-PET system. However, we have not attempted to remove the copper shielding because the body coil of the 3 T MR system generates far more powerful RFs than local transceiver coils.

SiPMs are well known for their strong dependence on temperature and the need to operate at a stable temperature (Hamamatsu Photonics 2007). Enclosing the SiPMs with electronics in the shield box requires cooling for stable operation. Since the SiPM-PET system that we developed places the SiPMs outside the copper shielding box, the SiPMs are at the stable temperature of the MR room and thus they do not need to be actively cooled. It is worthwhile to note that the event rate of the SiPM-PET without the MR pulse did not change much without cooling during various measurements, which often requires more than an hour.

A SiPM-PET system with an optical fiber bundle can utilize the body coils for RF transmission and receiver-only coils for RF reception. Since the receiver-only coils are simpler than the transceiver coils, the SiPM-PET system with optical fiber bundles can be beneficial for multi-channel phased arrays for brain imaging.

We plan to obtain small animal images by replacing the  $2.47 \times 2.74 \times 20$  mm<sup>3</sup> crystal blocks by those with  $1.5 \times 1.5 \times 7.0$  mm<sup>3</sup> crystals. Unlike the small animal PET systems

developed so far (Judenhofer *et al* 2008, Catana *et al* 2006), the SiPM-PET that we developed can be used with existing small coils such as the surface coil, which we used for this experiment.

In conclusion, the SiPM-PET system which we developed performed well and it can be easily extended for PET/MR imaging of the brain, breast and other extremities.

# Acknowledgments

The authors would like to express their sincere gratitude to Ms Hanna Lee at Eulji University for developing the gelatin phantom. The present study was supported by Eulji University in 2011 and by grants from the Korean Research Foundation funded by the Korean Government (2010-0012393, 2011-0005082) and the Atomic Energy R&D Program (2008-2003852) of the Korean Science and Engineering Foundation (KOSEF).

### References

- Catana C, Wu Y, Judenhofer M S, Qi J, Pichler B J and Cherry S R 2006 Simultaneous acquisition of multislice PET and MR images: initial results with a MR-compatible PET scanner J. Nucl. Med. 47 1968–76
- Drzezga A, Souvatzoglou M, Beer A J, Ziegler S I, Fürst S, Nokolla S G and Schwaiger M 2011 Integrated simultaneous whole-body MR/PET: first comparison between MR/PET and PET/CT in patients J. Nucl. Med. Suppl. 1 52 79

Hamamatsu Photonics 2007 Cat. No. KAPD0002E01 [Online] http://www.hamamatsu.com/

Hendrick R E 2008 Breast MRI (Berlin: Springer)

- Hong S J, Kim C M, Cho S M, Woo H, Ko G B, Kwon S I, Rhee J T, Song I C and Lee J S 2011 A feasibility study on the use of optical fibers for the transfer of scintillation light to silicon photomultipliers *IEEE. Trans. Nucl. Sci.* 58 579–89
- Hong S J, Kwon S I, Ito M, Lee G S, Sim K S, Park K S, Rhee J T and Lee J S 2008a Concept verification of three-layer DOI detector for small animal PET *IEEE. Trans. Nucl. Sci.* 55 912–17
- Hong S J, Song I C, Ito M, Kwon S I, Lee G S, Sim K S, Park K S, Rhee J T and Lee J S 2008b An investigation into the use of Geiger-mode solid-state photomultipliers for simultaneous PET and MRI application *IEEE Trans. Nucl. Sci.* 55 882–88
- Judenhofer M S et al 2008 Simultaneous PET-MRI: a new approach for functional and morphological imaging Nature Med. 14 459–65
- Kang H, Ko G, Jang S, Kim C, Yoon H, Ito M, Kwon S, Song I, Lee J and Hong S 2011 A study of simultaneous PET/MR imaging using SiPM PET modules with a short optical fiber bundle Suppl. J. Nucl. Med. Suppl. 1 52 446
- Ko G B, Yoon H S, Kwon S I, Hong S J, Lee D S and Lee J S 2011 Development of FPGA-based coincidence units with veto function *Biomed. Eng. Lett.* 1 27–31
- Kwon S I *et al* 2011 Development of small-animal PET prototype using silicon photomultiplier (SiPM): initial results of phantom and animal imaging studies *J. Nucl. Med.* **52** 572–79
- Larkman D J and Nunes R G 2007 Parallel magnetic resonance imaging Phys. Med. Biol. 52 R15-55
- Lee J S and Hong S J 2010 Geiger-mode avalanche photodiodes for PET/MRI *Electronic Circuits for Radiation* Detection ed K Iniewski (Boca Raton, FL: CRC Press)
- Lee J S and Kang K W 2012 PET/MRI Clinical PET and PET/CT: Principles and Applications ed E E Kim, M C Lee, T Inoue and W-H Wong (Berlin: Springer)
- Lee J S, Park K S, Lee D S, Lee C W, Chung J K and Lee M C 2005 Development and applications of a software for functional image registration (FIRE) Comput. Methods Programs Biomed. 78 157–64
- Pruessmann K P, Weiger M, Scheidegger M B and Boesiger P 1999 SENSE: sensitivity encoding for fast MRI Magn. Reson. Med. 42 952–62
- Raylman R R, Majewski S, Lemieux S K, Velan S S, Kross B, Popov V, Smith M F, Weisenberger A G, Zorn C and Marano G D 2006 Simultaneous MRI and PET imaging of a rat brain *Phys. Med. Biol.* 51 6371–79
- Roemer B P, Edelstein W A, Hayes C E, Souza S P and Mueller O M 1990 The NMR phased array *Magn. Reson. Med.* **16** 192–225
- Slates R, Cherry S, Boutefnouchet A, Shao Y, Dahlbom M and Farahani K 1999 Design of a small animal compatible PET scanner IEEE. Trans. Nucl. Sci. 46 565–70

- Vaska P et al 2011 An MRI-compatible pet insert for whole body studies in rodents at high functional and anatomical resolution Proc. IEEE Nucl. Sci. Symp. and Med. Imag. Conf., Valencia, (Spain, MI) C13–1
- Wehrl H F, Judenhofer M S, Thielscher A, Martirosian P, Schick F and Pichler B J 2011 Assessment of MR compatibility of a PET insert developed for simultaneous multiparametric PET/MR imaging on an animal system operating at 7 T *Magn. Reson. Med.* **65** 269–79
- Yamamoto S, Kuroda K and Senda M 2004 Development of an MR-compatible gamma probe for combined MR/RI guided surgery *Phys. Med. Biol.* **49** 3379–88
- Yoon H S, Ko G B, Kwon S I, Lee C M, Ito M, Song I C, Lee D S, Hong S J and Lee J S 2012 Initial results of simultaneous PET/MRI experiments with an MRI-compatible silicon photomultiplier PET scanner J. Nucl. Med 53 608–14