PET/MRI insert using digital SiPMs: Investigation of MR-compatibility

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A B S T R A C T
In this work, we present an initial MR-compatibility study performed with the world’s first preclinical PET/MRI insert based on fully digital silicon photo multipliers (dSiPM). The PET insert allows simultaneous data acquisition of both imaging modalities and thus enables the true potential of hybrid PET/MRI. Since the PET insert has the potential to interfere with all of the MRI’s subsystems (strong magnet, gradient systems, radio frequency (RF) system) and vice versa, interference studies on both imaging systems are of great importance to ensure an undisturbed operation. As a starting point to understand the interference, we performed signal-to-noise ratio (SNR) measurements as well as dedicated noise tests on the MRI side to characterize the influence of the PET electronics on the MR receive chain. Furthermore, improvements of sub-components’ shielding of the PET system are implemented and tested inside the MRI. To study the influence of the MRI on the PET performance, we conducted highly demanding stress tests with gradient and RF dominated MR sequences. These stress tests unveil a sensitivity of the PET’s electronics to gradient switching.

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1. Introduction

The combination of Positron Emission Tomography (PET) with its high sensitivity and the possibility for quantitative imaging and Magnetic Resonance Imaging (MRI) has the potential to become the next generation of hybrid imaging techniques [1]. In contrast to the combination of PET with Computed Tomography (CT), MRI offers a better soft tissue contrast and does not use ionizing radiation, thus reducing the overall required radiation dose significantly. To enable the full potential of a hybrid PET/MRI system, both imaging modalities have to work simultaneously, especially to enable a high quality spatial and temporal registration of imaging data at shorter scan times.

An MRI system basically consists of three main components, namely a strong magnet, a gradient system and a radio frequency (RF) system. New detector systems inside the MRI bore, e.g. a preclinical PET insert, have the potential to interfere with all these subsystems of the MRI system and vice versa. Examples for interference phenomena have been reported by several research groups: while Refs. [2–4] observe image degradation on the MRI side caused by the presence of a PET detector, other groups observe a direct influence on the PET performance caused by the RF pulses of the MRI [5] or switching gradients [6]. The first step to enable the usage of a PET detector inside the MRI was the replacement of photomultiplier tubes with solid state photo detectors, e.g. silicon photomultipliers (SiPM). Especially fully digital Silicon photomultipliers (dSiPMs) offer a good timing, energy and spatial resolution as well as a good temperature stability and they are a promising candidate concerning their MR-compatibility [7,8]. However, they tend to generate digital electromagnetic noise patterns which might degrade the MR image quality. Thus, proper PET system design and shielding is required to avoid interference. In this work, we started to investigate the interference phenomena and we tested the dSiPMs and our detector architecture concerning MR-compatibility.

2. Materials and methods

The Hyperion-II PET/MRI insert and all its support electronics are installed (as shown in Fig. 1) on a patient tabletop and trolley, creating a easy to handle and quickly installable system [9]. The PET ring is made up of ten PET Singles Detection Modules (SDM) which are mounted on a MR-compatible gantry, thus creating a PET ring with diameter of around 210 mm [10]. One SDM (Fig. 2)
hosts up to six detector stacks in a $2 \times 3$ arrangement (at this stage only two stacks per module are installed) and the communication and synchronization of multiple modules is done via plastic optical fibers (POF) to avoid galvanic connections between the modules and the Data Acquisition and Processing Server (DAPS, similar to the architecture described in [11]) outside the MR examination room. One detector stack is composed of a crystal array ($30 \times 30, 1$ mm pitch), a 2 mm light guide for light sharing, an 8 dSiPM array (DPC 3200-22-44 by Philips Digital Photon Counting) and a local FPGA [7, 8, 12]. The SDM is housed inside an almost gamma transparent carbon fiber screen which shows good RF shielding properties while being mostly transparent for gradients [13]. The insert is designed to fit into a Philips Achieva 3T MRI system and is equipped for MR acquisition with a dedicated PET transparent T/R mouse proton RF-coil (12 leg birdcage, high pass), which has an inner diameter of 46 mm. Consequently, the combined field-of-view (FOV) in this configuration is $\varnothing 46 \text{ mm} \times 33 \text{ mm}$ (one of the three possible PET rings installed; up to $\varnothing 46 \text{ mm} \times 100 \text{ mm}$ when all detector stacks are installed).

### 2.1. Influence on the MR performance

To investigate the interference on the MRI system, signal-to-noise ratio (SNR) measurements without and with the PET detector (10 SDMs) are performed. Therefore, a transversal slice of a 50 mm cylindrical phantom (1000 ml demi water, 770 mg CuSO$_4 \cdot 5$H$_2$O, 2000 mg NaCl, 0.05 ml H$_2$SO$_4$ – 0.1N solution) is imaged using spin echo (SE) sequences (TR/TE: 1000/50 ms, voxel size: $0.25 \times 0.25 \times 1$ mm$^3$, flip angle: 90°). To study the noise creation by the PET electronics, dedicated noise scans (TSE sequence, TR/TE: 1044/256 ms, TSE factor: 32, acq. matrix: $1024 \times 1024$, bandwidth per pixel: 180 Hz) are performed with the complete PET detector and with single PET modules. Experiments with subsystems replacing the complete PET detector are conducted to study the RF interference in detail and to identify the noise’s origin. Improvements of the shielding of the power supply unit (PSU) are also realized and tested with these noise scans.

### 2.2. Influence on the PET performance

The PET performance and stability during simultaneous operation was studied with one PET module (equipped with one stack) by single event detection. PET data and system parameters like voltages, currents and temperatures are acquired using a $^{22}$Na point source (activity: 2.8 MBq) over a longer time period (several minutes up to 45 min). During this data acquisition (as shown in Fig. 3), highly demanding RF and gradient stress tests (with various switching directions) are performed in smaller time windows (30 s–2 min).

For the gradient tests, EPI sequences (EPI factor: 49) with maximum gradient strength (30 mT), maximal slew rates and minimal TR with defined switching directions (the individual parameters are listed in Table 1) are used and for the RF test a highly demanding TSE sequence (TSE factor: 16, TE/TR: 21/333 ms, peak $B_1$ amplitude: 20 $\mu$T) is executed.

### 3. Results and discussion

#### 3.1. Influence on the MR performance

The SNR study shows a strong influence of the PET system on the MR performance: while the reference scan without PET detector delivers an MR image with a SNR (calculated according to NEMA standard [14]) of 159, the measurement with the PET detector reduces this value to 109.

### Table 1

| Sequence | Slew rate (mT/m/ms) | TR/TE (ms) |
|----------|---------------------|------------|
| X        | 184.8               | 28/13      |
| Y        | 194.5               | 27/13      |
| Z        | 192.3               | 25/12      |
| XY       | 198.0               | 27/13      |
| XZ       | 197.9               | 26/12      |
| YZ       | 197.6               | 24/12      |
| XYZ      | 198.1               | 26/12      |

Fig. 1. Hyperion-II$^{14}$ PET insert with 10 PET modules (2 stacks each) mounted on the patient table top of a 3T clinical MRI.

Fig. 2. One Singles Detection Module (SDM) hosts up to six detector stacks. Plastic optical fibers are used for synchronization and communication. The module is housed inside a carbon fiber shield.

Fig. 3. PET data acquisition scheme: PET data is acquired over a longer time period (green) while for certain smaller time windows (30–120 s) RF and gradient stress tests are performed (red). (For interpretation of the references to color in this figure caption, the reader is referred to the web version of this paper.)
detector (10 SDMs, power on, data acquisition) reveals a SNR degradation by a factor of 2 (SNR: 81) [9]. Fig. 4 shows the corresponding noise spectrum of the RF system without PET detector (black) and with detector (green: powered off and unplugged from the AC outlet, blue: data acq., red: data acq. + 2.8 MBq $^{22}$Na source). The noise floor is strongly increased, when the PET detector is switched on (in the frequency range of the SE images by a factor of approx. 1.9, thus explaining the observed SNR degradation), and features broad peaks which are shifting as a function of time, especially in the heat up phase of the scanner. These peaks are approximately 250 kHz apart, which is also the switching frequency of the last converter in the employed switched mode PSU. Experiments with purely resistive loads replacing the PET modules have identified clearly common mode noise originating from the PSU as main noise source. As a consequence, we improved the PSU’s shielding as shown in Fig. 5 (left: improved version, right: unmodified PSU). Measurements were performed with one PET module (1 stack). A fully equipped PET ring consists of 10 SDMs containing 60 detector stacks and has the potential to disturb the acquisition of the MRI much more.

### 3.2. Influence on the PET performance

A first evaluation of acquired PET data inside and outside the MRI shows no degradation of PET data quality and PET performance: for instance, flood histograms and energy resolution remain unaffected [12]. The singles data rates of the PET modules are stable even during harsh MRI sequences [9]. However, the measurements are performed with one single SDM equipped with one stack. A fully equipped PET ring consists of 10 SDMs containing 60 detector stacks and has the potential to disturb the acquisition of the MRI much more.
prepared energy cut to (511 ± 30) keV, all pixel around main pixel present) during highly demanding gradient sequences with different switching directions (as indicated with color shaded areas) for two different over voltages (OV) (top: OV = 2.9 V, bottom: OV = 2.5 V). Only the measurement with the higher OV shows singles rate drops by up to 5.6% during active z gradient sequences. (For interpretation of the references to color in this figure caption, the reader is referred to the web version of this paper.)

4. Conclusion

We have successfully operated a fully digital PET detector inside a 3T MRI. On the MRI side, we observed a SNR degradation by a factor of 2 which is mainly caused by common mode noise from the switched mode power supply. Improvements on the PSU’s shielding lead to a notable reduction of the noise. In the latest modification no difference between the reference noise floor and the one during PET acquisition is visible. Although we observed no technical problems with the modifications during the operation with one SDM, a final evaluation with a complete PET scanner has to be done. Until this test is conducted, the presented solution can only be seen as intermediate step.

On the PET side, we notice that our PET system works stable even under unrealistic demanding stress tests. Up to now, we do not observe any hiccups in the data communication. However, the performed stress tests reveal a sensitivity for switching z gradients. We observe a ripple on the bias voltage and a broadening effect of the energy resolution for an aggressive chosen bias voltage setting. After application of a narrow energy cut (511 ± 30) keV) around the photo peak, this broadening effect leads to a singles rate drop by approx. 5.6%. The coupling mechanism of the gradients causing the V_B fluctuations and the exact influence of these fluctuations on the detector performance is unclear at this stage. The above mentioned explanation is currently a working theory and has to be proven by further investigations.
strong ripple on time. In time regions with active z gradients (indicated by green boundaries) a different over voltages (OV) (top row: \( \text{OV} = 2.5 \text{ V} \), bottom row: \( \text{OV} = 2.9 \text{ V} \)). (For interpretation of the references to color in this figure caption, the reader is referred to the web version of this paper.)

![Energy histogram](image)

**Fig. 9.** Energy histogram (left: overall, right: photo peak range) for time windows without gradient switching (black, shaded) and with switching z gradients (red) for two different over voltages (OV) (top row: \( \text{OV} = 2.9 \text{ V} \), bottom row: \( \text{OV} = 2.5 \text{ V} \)).

![Measurement of the bias voltage and bias current](image)

**Fig. 10.** Measurement of the bias voltage \( V_B \) (top) and bias current \( I_B \) (bottom) for two different over voltages (OV) (black: \( \text{OV} = 2.5 \text{ V} \), red: \( \text{OV} = 2.9 \text{ V} \)) as a function of time. In time regions with active z gradients (indicated by green boundaries) a strong ripple on \( V_B \) and \( I_B \) occurs. (For interpretation of the references to color in this figure caption, the reader is referred to the web version of this paper.)

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