Silicon Photomultiplier Performance Tests in Magnetic Resonance Pulsed Fields

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Abstract—We are investigating the performance of silicon photomultipliers (SiPM) in a magnetic resonance imaging (MRI) environment with a view to exploiting the potential advantages in a combined positron emission tomography / magnetic resonance imaging (PET/MR) system. SiPMs are attractive for PET because they are compact, provide high gain at low bias voltage, and fast response times. Importantly for combined PET/MR applications, they offer performance comparable to that of the photomultiplier tube (PMT) and are insensitive to magnetic fields. To study the effect of MR static magnetic, pulsed gradient and radio-frequency fields on SiPM performance, we have acquired data whilst switching magnetic fields with the SiPM positioned in the MR system. The wide ‘dynamic range’ of SiPMs facilitates their use in both low light level and PET applications; we have acquired ‘single photoelectron’ and $^{22}$Na energy spectra to demonstrate SiPM performance for this application. As expected, no degradation in inherent SiPM performance in an MR environment is observed.

Index Terms—SiPM, silicon, photomultiplier, MRI, PET.

I. INTRODUCTION

POSITRON emission tomography is an especially sensitive and quantifiable approach to molecular imaging which can provide information about metabolic function by means of labeling physiologically significant molecular compounds with radioactive isotope markers. By monitoring uptake and activity levels, valuable information about blood flow, oxygen consumption, glucose consumption, and protein synthesis can be measured, targeting early changes in tumor related metabolic activity for detection of cancers and many other applications. However, detailed anatomical reference images can be vital to understanding PET tracer data in most studies [1], which has motivated the development and rapid assimilation of combined PET/CT systems into mainstream medicine [2]. Replacing CT with MRI as an alternative provider of high resolution morphological data offers several advantages. (i) Flexibility: In addition to normal imaging protocols, MRI provides a host of applications such as diffusion tensor imaging, functional MRI, angiography, spectroscopy, etc. which can be harnessed for new research methods using the hybrid system. (ii) Safety: MRI provides high resolution soft tissue images for anatomical reference without the use of ionizing radiation. MRI has no known adverse effect on living organisms. (iii) Simultaneous acquisitions: Unlike PET/CT which images slices sequentially as the patient physically moves through the two scanners, PET/MR machines should be capable of measuring two sets of data which are spatially and temporally registered. (iv) Improved contrast: MRI sequences can be tailored to provide optimal contrast between significant tissue types, unlike CT. Until recently, APDs with relatively low signal to noise characteristics and slow timing remained the only magnetic field insensitive replacement for PMTs in PET [3], [4], [5], leading several PET/MR designs to rely on shielding and/or elongated light guides to continue benefitting from the excellent performance of PMTs [6], [7], [8]. However, a new family of CMOS manufactured photodiodes known as ‘silicon photomultipliers’ (SiPMs) are currently being considered for a wide variety of applications (including PET) due to promising similarities between their characteristics and those of the classic PMT: high gain, fast response, single photon sensitivity, wide spectral range, etc [9]. If proven stable and reliable, these characteristics will undoubtedly be exploited in a new generation of pre-clinical solid state PET detectors [10], [11]. Significant progress toward this goal has already been made by the University of Pisa [12] in collaboration with FBK-irst. In accordance with these PET projects, we hope to show that SiPMs will also be an appropriate choice of detector for combining PET with magnetic resonance imaging, a relatively new phenomenon in the medical imaging community. The feasibility of PET/MR using position sensitive avalanche photodiodes has been proven [13], [14], but the optimization of both modalities in simultaneity is an ongoing field of research. We believe that the high gain, magnetic field insensitivity, simplicity of design, and stability of SiPM detectors may be useful in designing an MR compatible PET camera with minimal front-end electronics and shielding for optimal performance of both systems.

A. Background

a) Silicon Photomultipliers: A SiPM is a photodetector based on the principal of Geiger-mode avalanche photodiodes (APDs); reverse-biased semiconductor diodes which contain a high-field depletion region between p-type and n-type semiconductors when biased beyond the breakdown voltage of the device. When a photon deposits energy in the semiconductor, an electron-hole pair is created. Depending on the structure of the diode, the electron or hole will accelerate across the high-field region with enough energy to generate a rapid ($\sim$ 10 ps)
‘avalanche’ of free electrons with a charge gain of $\sim 10^6$ [15]. A single SiPM detector consists of hundreds of these Geiger-mode APD “$\mu$-cells” wired to a common anode (Fig. 1). The output of the entire detector is therefore the sum of the individual $\mu$-cell responses [16]. The stability of SiPM detectors in B-fields is due to the small length scale of the electron acceleration region (typically $< 1.0 \mu m$); many times smaller than those in PMTs. Over such a short distance, the electron shower cannot be deviated by large B-fields, and the device operates normally. These characteristics make the SiPM an ideal alternative to PMTs. However, in order for SiPMs to be a viable option for combined PET/MRI applications, they must be able to maintain these ideal characteristics under the influence of magnetic resonance imaging conditions.

b) Magnetic Resonance Imaging: To fully characterize the performance of SiPMs in co-occurrence with MRI, it is necessary to first understand the MRI context. MRI is based on the spin properties of hydrogen nuclei (protons) of which there are large concentrations in the water and fatty tissue of biological systems. In an MRI acquisition the protons which occupy such tissues are stimulated using weaker gradient fields in the imaging field of view. To process these signals, which are applied with a periodicity and duration (RF) signals which can be decoded to obtain information on the spin properties of hydrogen nuclei (protons) of which there are large concentrations in the water and fatty tissue of biological systems. In an MRI acquisition the protons which occupy such tissues are stimulated using weaker gradient fields in the imaging field of view. To process these signals, which are applied with a periodicity and duration (RF) signals which can be decoded to obtain information about the protons’ spatial relationships. The RF radiation is initially produced by a transmitter coil which ideally encloses the object being imaged with a high fill factor. The frequency at which nuclear magnetic resonance occurs is a function of the primary field strength (\( \sim 42.6 \text{ MHz/T} \)). Typical field values for clinical use are between 1 and 3 Tesla. These pulses, which are applied with a periodicity and duration that is unique for each MRI protocol, will induce voltages in conducting materials.

To encode these frequencies, a second set of coils must be used to rapidly modify the primary magnetic field strength using weaker gradient fields in the imaging field of view. To obtain as much frequency-encoded information as possible in a short time-span, the gradient coils must turn on and off very rapidly. It is during this field switching that EM radiation is generated and capable of inducing further noise and eddy currents in nearby conducting materials. Typical switching times for clinical systems are $100 - 200 \mu s$, producing EM noise in the kHz range. Therefore, in addition to operating optimally in a static magnetic field of several Tesla, the SiPM must also be stable in the RF and gradient electromagnetic fields. Our initial investigation of the SiPM has been to analyze detector responses in the presence of each field component of the MRI duty cycle for high and low photon count regimes.

B. Methods
The SiPM pixels tested in this study were produced at FBK-irst. Each pixel consisted of 625 microcells (40 $\mu m \times 40 \mu m$) in an area of 1 mm$^2$ with a (non-optimized) fill factor of 20%. The devices exhibited a gain of order $10^6$, and were operated at a bias voltage of $\sim 3$ V above breakdown ($\sim 33$ V). Newer devices produced at FBK-irst have improved in PDE significantly [17]. In order to best measure the response of the detectors in isolation we constructed a shielded amplifier board with external bulkhead connection leading outside the Faraday cage directly to the SiPM. The SiPM was housed in an unshielded, optically sealed capsule. For low light ‘single photoelectron’ acquisitions, this capsule was also fitted with a pulsed LED, adjusted such that only a few photons reached the detector surface per triggered event. To produce spectra in the PET 511keV energy realm, we have coupled the SiPM to a $1 \times 1 \times 10 \mm$ LSO crystal with no intermediate bonding agent, and exposed the crystal to a $^{22}\text{Na}$ source in the same capsule.

c) Primary Field: To confirm insensitivity of the SiPM to static magnetic fields we have acquired both low and high energy spectra in the 1 Tesla imaging region of a superconducting MRI magnet. All spectra shown were produced by integrating and histogramming the area under each amplified SiPM pulse generated in response to the number of photons incident on the detector surface. These spectra were compared directly to acquisitions measured in terrestrial fields ($< 0.5 \text{ mT}$).

d) Gradient Fields: The combined effects of the gradient coils and primary field were measured by moving the same SiPM set-up as near to the gradient coil windings as possible in order to best simulate the most probable SiPM location in a ring of detectors for future PET/MR systems. Since the induced gradient fields become stronger and less linear near the conductors, we believe it is important to take this displacement into account. We were able to position the SiPM approximately $-2 \cm$ from the windings of our 9 cm gradient set (with the SiPM and electronics positioned inside the gradient bore). When the MR gradients are driven hard, such as in a gradient echo fMRI sequence, gradient switching can occupy up to $\sim 70\%$ of the scan duration. To measure the response of the SiPM only during this noisy fraction of the MRI gradient cycle, we triggered the gradient to ramp up in coincidence with each LED pulse so that the SiPM response occurred during the time required to completely activate the gradient ($\Delta t$). Figure 2 illustrates this method. For the PET-relevant $^{22}\text{Na}$ stimulus, the photon counting is inherently asynchronous with the MR since the positron decay process is random and can not be gated appropriately.

e) Radio-Frequency Field: In this experiment we moved the SiPM and electronics back to the center of the imaging field of view so that the unshielded SiPM could be placed in the center of a 3 cm diameter ‘birdcage’ style radio-frequency transmitter. These conditions may over-emphasize the effects
of the RF since (in a completed system) the detectors would reside significantly outside the RF coil to maintain a high resolution PET field of view inside the RF coil. The transmitter was then pulsed at near-maximum amplitude with the appropriate 42.6 MHz required for imaging at 1 T. A train of RF of pulses 100 μs long with 25 ms spacing were generated to occur in coincidence with the SiPM acquisition for low light measurements, and repeat continuously throughout the \( ^{22}\text{Na} \) acquisition.

C. Results

f) Primary Field: In the primary field alone, no change in energy resolution, gain, or other characteristic can be identified in either the low or high energy results. This result is entirely consistent with the literature and fundamental operation of the detector, since the high field ‘avalanche region’ between the semiconductor layers is not long enough for electrons to be deviated by any external field.

g) Gradient Field: Nearly identical single photoelectron spectra were acquired with the x-gradient channel active and inactive, demonstrating that it is possible to perform precise measurements with a SiPM even with 100% overlap between the gradient switch time and photon counting. Figure 3 is a comparison between low energy spectra acquired in only the static field (blue) and in coincidence with the gradient switching (green), demonstrating the unchanged sensitivity of the devices. The normalized \( ^{22}\text{Na} \) spectrum in Fig 4 shows this same stability at the same energy levels used in PET. However, some interaction between the SiPM and MR systems can be identified in the SiPM output as shown in Fig. 6 despite the unchanged performance of the SiPM. The pickup coil response shown in green is proportional to the rate of change of magnetic field. Identifying and minimizing this interaction is an area of continued investigation, though it appears to be entirely related to the orientation and wiring of support electronics rather than the SiPM itself.

An interesting secondary effect of SiPM operation in close proximity to the gradient set was a decrease in gain due to heat dissipated by the gradient coils. Power is converted to heat in all MRI coils (which often require water cooling), and over time, they can re-radiate heat into the bore of the magnet. Figure 5 shows two 511 keV peaks acquired before (blue) and after (green) a \( \sim 4 \, ^{\circ}C \) temperature increase due to this effect.

h) Radio-Frequency Field: Pick-up of the RF fields in our experimental arrangement was extremely high, such that we were unable to acquire either low or high energy spectra during the RF pulse durations. Figure 6 shows the SiPM trace (purple) flooded by RF noise during transmitter activation and also generating normal pulses corresponding to 1 – 5 photons during the inter-pulse interval. Faraday cage shielding can be added to reduce this interaction, though the amount of conducting material introduced to the bore of the magnet must be minimized in order to preserve homogeneity in the MRI.
Fig. 6. Some interference between the SiPM associated electronics and MR system has been detected. Two screenshots of SiPM output (purple) and pickup coil response (green) when SiPM challenged with a gradient (top) and RF pulse (bottom) are shown. The pickup coil response is proportional to the rate of change of magnetic field (and therefore signal to noise of the images).

II. CONCLUSION

As expected there is no direct evidence for degradation of inherent SiPM performance in the MRI environment. However, it is evident that interference between the PET and MR systems must be eliminated if there is to be no impact on acquisition times. A key issue in these tests has been differentiating interaction between: (i) the SiPM device and the MR system, and (ii) the SiPM support electronics and the MR system. In this study we have attempted to minimize any interaction with the support electronics by traditional means of shielding. The unshielded terminals and wire bonds which bias and readout the SiPM, in addition to multiple cable connections near the MHz and kHz sources, and possible earth are all possible sources of signal degradation. Identifying these points of interaction and optimizing a PET/MR system design that facilitates uncompromised data acquisition - without the requirement for ‘gating’ or ‘filtering’ the PET acquisition during fast MR data acquisition - is ongoing. It also appears that a method of temperature stabilization or compensation may be necessary in PET/MR applications due to residual heat from the gradient coils; another area of continued investigation. A dedicated MR gradient/SiPM assembly is under construction for further tests.

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