Cross-Strip Multiplexed Electro-Optical Coupled Scintillation Detector for Integrated PET/MRI

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Abstract—A prototype electro-optically coupled PET brain insert is being developed for 1.5 T or 3 T MRI systems. To reduce the number of electronic channels, we developed and tested a novel silicon photomultiplier (SiPM) multiplexing circuit based on a novel cross strip readout circuit using capacitors that reduces the number of channels by a factor of 4. Next, we developed and tested this multiplexing circuit with a new electro-optical coupling method that uses a small vertical cavity surface emitting laser (VCSEL) to transmit the scintillation signal out over fiber. This multiplexing circuit is specially suited for directly driving VCSEL lasers without using preamps or any other active components. Two detectors were placed inside the MR bores, and operated with and without a continuous RF field. We were able to resolve a 6 × 6 array of 2.63 mm × 2.63 mm × 20 mm LYSO crystals that were directly coupled to a 4 × 4 array of 3.2 mm × 3.2 mm SiPM pixels with a global delay corrected coincidence time resolution 2.50 ± / − 0.015 ns FWHM and a per crystal gain corrected energy resolution of 12.6 ± / − 0.6% FWHM at 511 keV while operating inside an 1.5 T MRI, with or without the RF field on. The two technologies, capacitive multiplexing and electro-optical coupling, have enabled a high-performance clinical PET detector based on silicon photomultipliers, a factor 4 reduction in the number of readout channels, and, when inserted into a 1.5 T MRI system, has low levels of mutual interference between the PET detector and the MRI.

Index Terms—Fiber optics, magnetic resonance imaging, nuclear imaging, nuclear medicine, optical interconnections, positron emission tomography.

I. INTRODUCTION

PET/MRI instrumentation must satisfy two independent goals: achieve state of the art PET performance and integrate this detector with an MRI system without mutual interference. The major focus of the development of MRI compatible PET detector technology until recently has been the incorporation of Faraday shielded avalanche photodiode (APD) solid-state photodetectors that are non-magnetic and insensitive to static magnetic fields [1]–[4]. The silicon photomultiplier (SiPM) photodetector technology [5]–[7] is now replacing APDs used in previous PET/MRI detectors. SiPM devices have shown to be magnetically insensitive while possessing a gain (750e3 − 1.0e6) that rivals a PMT. The signals are large enough that they can be relayed outside the MRI using only flex cables [7] and, despite significant attenuation and dispersion, do not require pre-amplification for non-time-of-flight PET applications.

In all of these designs, the electrical signals are taken out over electrical conductors to a data acquisition system that resides outside of the MRI system. We demonstrated that instead of using coaxial or flex cables, the signal from a SiPM can directly drive a vertical cavity surface emitting laser (VCSEL) without degrading timing or energy resolution [8], a method we refer to as “electro-optical coupling”. We have extended the previous electro-optical coupling work by adapting the single channel design for one SiPM capacitively coupled to one VCSEL laser and creating a cross-strip multiplexing circuit that capacitively couples 16 SiPM devices to 4 VCSEL lasers. By coupling the signal over fiber, we can minimize the electrical footprint of the PET detector. We have designed the electro-optically coupled PET detector to electrically float relative to the RF ground of the MRI (see Fig. 1). In designs that use many electrical connections, it is complex in both size and cost to create high speed electrically ground isolated connections. Also, there is a danger that high speed signals that are transmitted on unshielded cables can radiate electromagnetic interference (EMI) into the RF receive signal pathway of the MRI [7], [9]. The proposed electro-optical coupling approach solves these problems. This paper focuses on development of a new circuit for multiplexing SiPM arrays. Although we apply the capacitive multiplexing to the electro-optical signal transmission scheme, in principle, this technique can be used in any SiPM-based scintillation detector readout scheme.

II. MATERIAL AND METHODS

Two PET detectors using the new cross-strip multiplexing method (see Fig. 2) and electro-optical coupling were constructed and operated in a 1.5 T GE Sigma MRI System. Performance of the two detectors was measured outside the MRI, inside the MRI with the RF off, and inside the MRI with the RF on. Also, MRI images were acquired and compared before and after the PET detectors were insert into the bore of the MRI. We fabricated two detectors consisting of a 6 × 6 array of 2.63 mm × 2.63 mm × 20 mm LYSO crystals (Agile Engineering) coupled to a 4 × 4 array of SiPM devices (SensL Array4) (see Fig. 4). For each of the detectors, SiPM signals were cross-strip capacitive multiplexed into 4 non-magnetic...
In an electro-optical coupled PET detector, the analog signals are relayed over fibers inside the MRI to outside the MRI. The fibers reduce the amount of mutual coupling between the PET electronic circuits and wires.

Along each row, the cathodes are connected together and then into two charge splitting capacitors. The splitting capacitors are weighted by the row, and the capacitance is designed to sum to a constant. Along each column, the anodes are connected together and routed into the same network as for the cathodes. The charge from the anode and cathode splitting capacitors are driven into VCSEL lasers (A,B,C,D) each biased with a low input impedance of approximately (25–50 Ohms). The circuit produces four analog modulated light signals that may be taken into a low input impedance amplifier or, in our design, a VCSEL laser. We used linearly weighted capacitors instead of resistors (see Fig. 2), because first, a capacitor is already needed to couple the signal into the amplifier or laser, and second, that a capacitor has a better high frequency response than a resistor. In this paper, we combine both capacitive multiplexing and electro-optical coupling to produce a PET detector. One feature of such a design that is a subject of future study, is that it is RF transparent, since the PET detector floats with respect to the RF coil ground. RF transparency is a topic of Discussion, not investigation for this paper. The bias resistor from the positive HV to the devices should use a value from 5 kΩ to 100 kΩ depending on a tradeoff between noise and count rate. In this circuit the bias resistor was 10 kΩ to allow for sufficient current to bias the detector while not attenuating the high frequency scintillation pulses. The value of the sum of the two linear weighted coupling capacitors (CC) was optimized to be 120 pF.

The coupling capacitor value was optimized by using a circuit simulation tool (LTspice IV, Linear Technologies). The scintillation pulse is modeled by a bi-exponential model with a rise time of 120 ps with an exponential decay of 40 ns. The detector is modeled by a simple current source in parallel with the parasitic capacitance of the device. The current source generates an integrated charge based on the nominal gain of a SiPM (1e6) and scintillation light yield of LYSO (28 photons/keV). The CC value was swept over a range of values from 1 pF to 10 nF. A CC value was chosen that maximizes the sum peak signal, \((A + B) - (C + D)\), while providing enough difference \((A - B)/(\text{sum})\) to be able to resolve crystals (see Fig. 3).

The PET detector signals are relayed over 15 meters of multimode 62.5 μm/125 μm optical fiber, through a RF shield panel into the MRI electronic control room. The 850 nm VCSEL optical signals are converted to analog voltage waveforms by a Finisar 4 Gbps 850 nm photodiode-transimpedance amplifiers.
Fig. 4. The PET detector module uses a (A) $4 \times 4$ array of 3.2 mm x 3.2 mm silicon photomultipliers (not shown) and a 6x6 array of 2.63 mm x 2.63 mm x 20 mm LYSO scintillation crystals with all polished surfaces and a specular reflector surrounding each element. (B) Non-magnetic fiber launch using 850 nm VCSEL lasers with an analog bandwidth greater than 1 GHz and GRIN lenses that was used on the PET detector module. (C) A non-Faraday version of PET detector module with power and electro-optical fibers is shown for visualization purposes.

Fig. 5. The optical signals are relayed to a temperature controlled room next to the 1.5 T Signa GE MRI that houses the MRI electronics. A custom designed optical receiver (top) and two synchronized digital storage oscilloscopes (bottom) trigger and waveform digitize the events from the two PET detectors.

The analog voltage is then pre-amplified by a 3 Ghz differential voltage amplifier (ADL5562), converted to a single ended signal by a Mini-Circuit RF balun (TC1-1-13M+), and driven into the 50 Ohm input of two synchronized digital storage oscilloscope (Agilent DSO90254A) with 2.5 Ghz analog bandwidth and sampled at 20 Gsps (see Fig. 5). The two scopes are synchronized by the external trigger ports using a single coaxial cable and provide 8 channels of simultaneous sampling. A custom written C++ program (vendor supplied API) running on a data acquisition PC controls each DSO over a 1 Gbps
for each excitation.

Fig. 6. PET Coincidence experiment operated inside a 1.5 T MRI using an 8-channel receive only head coil. The two detectors are powered by a 36 V lead-acid-gel battery regulated to 27.5 V which floats relative to the MRI RF ground. Electro-optical coupling relays the signals to a data acquisition system that sits approximately 15 meters away in an adjacent room.

2-D histogram bin size used. Therefore, we used another method that uses fitted Gaussian functions to the peaks [13], [15]. For each peak in the flood histogram, each peak was fit to a 1-D Gaussian across a row. The distance between each peak on each side (only a single peak for the edge crystals) was averaged and then divided by the sigma 1-D Gaussian fit to calculate the per crystal “figure-of-merit” (FoM) distance over sigma. The mean and standard deviation of the FoM for the entire array was calculated as the array FoM \( (\bar{D}/\sigma_{FoM}) \) and characterizes how well separated the crystal peaks are in the flood histogram.

We characterized the two detectors with and without RF body transmitter using split transmit and 8 channel receive only coil for the MRI acquisition. The detectors were placed as close to the 8-channel receive coil on the bed of the system (see Fig. 6). This paper represents preliminary work to characterized the two PET detectors with and without RF body coils to allow us to understand feasibility of this approach.

To test for RF interference in the PET detector, we used a fast spin multi-echo RF pulse sequence that has a high RF peak power. The parameters for the RF pulse sequence were: \( TR = 15\text{ms}, TE = 68\text{ms}, \) which is T2 weighted. For each excitation, 190° and 12 180° pulses are played out. The 12 180° echo train length is approximately 180 ms long. The pulse sequence has a high RF peak power and a high probability for interfering with PET electronics because of the large 180 degree inversion pulses. The RF pulse was played continuously during the PET acquisition without the MRI acquisition enabled.

C. Simultaneous PET/MRI Characterization of Cross-Strip Multiplexed Electro-Optical Coupled PET Detectors in a 1.5 T MRI With and Without RF

The detectors were Faraday shielded and powered using a battery (see Fig. 6). The detectors were placed inside of a Faraday shield that consisted of a copper screen that was fabricated into a rectangular box. The detectors were placed 10 cm apart to fit within the width of the MRI bed. The ground of the Faraday shield was connected to the negative terminal of the battery over a coax cable. A 33 \( \mu \text{Ci} \) \( ^{22}\text{Na} \) source was placed close to one detector, and flood irradiated the second detector. By operating the detector with a coincidence window of twice the FWHM time resolution, Lutetium background and the Na-22 1.2 MeV peak are rejected.

We acquired timing spectra for each crystal element in one of the arrays. We characterized the coincidence flood image histogram, energy resolution, and timing resolution for one of the two detectors with the 1.5 T MRI. The flood histogram was segmented from a minimum distance to the peak algorithm[11]. Each of the three acquisitions(outside, inside, inside with RF) was performed for a fixed time of 1 hour. The per crystal energy spectra were fit using an automated script written in Matlab, and the gain for each was calculated by the 511 keV photo-peak location. The energy spectra were gain corrected to a common pulse height, and full-width-half-maximum (FWHM) global energy resolution was calculated.

The acquisition acquired 8 waveform samples at 1 Gsps from the two detectors. The time pickoff was linearly interpolated at the 10% crossing threshold from the maximum of the calculated energy waveform after correction for baseline. The energy waveform was calculated by \( (A + B) - (C + D) \) and rejected common mode interference. The global timing spectra was fit after a per crystal delay correction performed using a convex-optimization algorithm [14] to calibrate the timing offsets between the different detectors. In this paper we do not use the peak-to-valley ratio as a flood histogram performance metric because the calculation of the valley is contaminated by scattered events, and the measurement in the valley is based on the 2-D histogram bin size used. Therefore, we used another method that uses fitted Gaussian functions to the peaks [13], [15]. For each row in the flood image, each peak was fit to a 1-D Gaussian across a row. The distance between each peak on each side (only a single peak for the edge crystals) was averaged and then divided by the sigma 1-D Gaussian fit to calculate the per crystal “figure-of-merit” (FoM) distance over sigma. The mean and standard deviation of the FoM for the entire array was calculated as the array FoM \( (\bar{D}/\sigma_{FoM}) \) and characterizes how well separated the crystal peaks are in the flood histogram.

D. MRI Quality Assessment

We used a 2D gradient recalled echo (GRE) T2* weighted imaging sequence to test for radio-frequency emission image artifacts. We used an 8-channel receive only coil (see Fig. 6). The parameters used for the GRE acquisition were a RF excitation flip angle of 45°, \( TR = 500\text{ms}, TE = 40\text{ms}, \) Pixel BW = 244.141 Hz/Pixel, Field-of-view (FOV) 22 cm \( \times \) 22 cm, slice thickness 5 mm, image size 256 \( \times \) 256, and number of average acquisitions \( (NEX) = 1 \). The phantom was a 20 cm diameter sphere filled with agar. In addition to potentially causing magnetic susceptibility artifacts, the PET detector can emit electromagnetic interference (EMI) that can be picked up by the RF receiver coil. By placing the block detectors next to the RF receiver array, any radiated EMI will cause line or streak artifacts in the MRI image. The spherical phantom filled with agar was imaged before and after the PET detectors were insert(shown in Fig. 6). The variance central 95% of the phantom was selected. A difference was selected in the exact same central area on two...
Fig. 7. Flood histograms (top row), gain calibrated global energy spectra (middle row), and delay calibrated global time resolution characterization (bottom row) of the PET/MRI detector operating outside the 1.5 T, inside the 1.5 T magnet, and inside the 1.5 T magnet with a high duty cycle, RF spin-echo pulse sequence. Due to the nature of the data acquisition configuration and a fixed count acquisition for all three columns, the high duty cycle RF sequence data (right) resulted in fewer counts in the photopeak for the flood image, energy and timing spectra.

TABLE I
DETECTOR CHARACTERIZATION

<table>
<thead>
<tr>
<th>B_0 = 0</th>
<th>B_0 = 1.5T</th>
<th>B_0 = 1.5T, RF ON</th>
</tr>
</thead>
<tbody>
<tr>
<td>Time resolution</td>
<td>2.50 ± 0.02 ns</td>
<td>2.5 ± 0.015 ns</td>
</tr>
<tr>
<td>Energy resolution</td>
<td>12.8 ± 0.2%</td>
<td>12.6 ± 0.2%</td>
</tr>
<tr>
<td>Spatial D/σ FoM</td>
<td>2.5 ± 0.8</td>
<td>2.6 ± 0.8</td>
</tr>
<tr>
<td>Count Rate</td>
<td>222.6 ± 1.9 cps</td>
<td>220.9 ± 1.9 cps</td>
</tr>
</tbody>
</table>

different acquisitions. The variance of the first image is compared to the difference. We also analyzed the signal-to-noise ratio (SNR) and the MRI uniformity. The SNR was measured by selecting a 12 cm diameter ROI across 8 central frames of the 22 cm spherical phantom. The SNR is calculated by dividing the mean value in the ROI by the standard deviation of the ROI. The reported SNR is the mean (n = 8) across the 8 frames with the standard deviation reported as +/− 1 sigma. The uniformity was also calculated by the formula:

\[ \text{MRI-uniformity} = 100 \times \left(1 - \frac{\max(ROI) - \min(ROI)}{\max(ROI) + \min(ROI)}\right) \]

over the same 12 cm ROI across the same 8 central frames (n = 8).
portion of phantom is 8.92%, versus the central 95% area of the difference is 7.19%. Therefore, the mean squared error of the difference image with and without the PET insert present is smaller than the mean squared error of the phantom itself. The SNR and MRI uniformity across a 12 cm diameter ROI over the central 8 frames of the sphere phantom before the PET detectors were inserted was 7.27 ± 0.6 and 43.1 ± 7.4% versus 7.31 ± 0.5 and 41.3 ± 8.7% respectively after the PET detectors were placed and power on. This, there was no significant difference in SNR or MRI uniformity when the 2D GRE sequence was acquired.

IV. DISCUSSION

We have developed a capacitive charge multiplexing circuit (see Fig. 2) for an electro-optical coupled PET detector (Fig. 1) that demonstrates excellent SNR outside and inside an operating MRI system (see Fig. 7). Results indicate that our multiplexing method and electro-optical coupling scheme has comparable or better performance 1:1 coupled (no light or electrical multiplexing) SiPM Sensl array (SPMArray2) detector approaches published in literature. For example, Hong et al., used 1–1 coupling in a PET/MR brain scanner system [16], coupled to a scintillation block that has a larger pixel size. In that published work, Hong measured a 18.0 ± / − 0.1% energy resolution and 4.23 ns FWHM timing resolution for the scanner. In our design, we possibly achieved a better energy and timing resolution because of the improvements in the SPMArray4 versus the earlier SPMArray2. Also, we performed single channel characterization in our earlier work [8] with the SPMArray2 using a single 3.0 mm × 3.0 mm × 20 mm LYSO scintillation crystal with both electrical and optical characterization. In that experiment, we measured a / ± / 1.0% energy resolution and a single sided timing resolution of 1.3 ± / − 0.02 ns FWHM for a single crystal. Our current design achieves a 2.5 ns FWHM over an entire scintillation block which compares to an estimated pair-coincidence timing of 1.86 ns (multiplying the single sided timing by √2) for the single channel experiment. It is common for single crystal-single SiPM pixel characterization to outperform position sensitive detector modules because of light loss/crosstalk between channels, and/or charge multiplexing that is introduced when creating the latter. Therefore, our multiplexing method did not degrade the energy or timing performance relative to both our previously published single channel results or other 1–1 coupling designs based on the a similar SiPM array and scintillation crystal block.

In this work, multiplexing is used to reduce the interconnection between the front-end and the back-end data acquisition. There were two kinds of multiplexing used: scintillation light multiplexing by coupling a 6 × 6 crystal array to a 4 × 4 detector array and electrical multiplexing that connects a 4 × 4 detector array to 4 outputs. Both of these techniques reduce detector performance in energy and time resolution, but significantly reduce the number of channels. Another limitation is the complexity of coupling using lasers to relay signals. The VCSEL lasers used in this work are custom built, and are not as easy to use as simple electrical connectors.

For the limited study on how this PET detector design affects the MR image quality, we observed no significant effects.
however we plan to test this observation more thoroughly once we have a full detector ring. One feature of such an approach, that is also a subject of future study, is that it is RF transparent, since the PET detector floats with respect to the RF coil ground. This RF transparency would enable the RF transmit coil to reside outside the PET detector ring. A limitation of this study was the 8-channel data acquisition approach using a digital storage oscilloscope (DSO). Because of the nature of the DSO scope triggering, we were not able to achieve the full sensitivity of the detector when the RF is powered on. The trigger on the digital scope can only sample when a waveform crosses a fixed threshold close to the baseline, therefore, this method was unable to reject the RF leakage using the DSO.

It can be seen from the energy spectra for \( f_0 = 1.5 \) 'RF ON' (see Fig. 7), that after digital subtraction of the differential waveforms, there is no leakage of the RF into the energy channel of the DAQ, however, having rejected those RF signals, the resulting counts are lower. The DSO has only a limited count rate ability to acquire events (about 200–300 events per second). Therefore, the RF pulse reduces the 511 keV photon sensitivity of the acquisition when firing because the DSO cannot trigger on a differential waveform, but must trigger only on the single ended channel that contains some RF contamination. Furthermore, there are two mitigation strategies that can be used to reduce RF leakage and spurious triggering. The first is to move to 3 T where the RF can be filtered from the analog bandwidth of the receiver, and second, to use a free-running ADC acquisition that can digitally subtract any RF leakage and trigger on differential waveforms. At 1.5 T the Larmor frequency is 62.86 MHz which is in the analog bandwidth of the system, whereas at 3 T it increases to 127.7 MHz where it can be effectively filtered out. We are currently working on both remedies.

V. Conclusion

We have demonstrated a working cross-strip capacitive multiplexed, electro-optically coupled, PET/MRI detector that has a one fourth number of channels, that potentially has little influence on the MRI. Future work will test the PET detector configured into a full ring system with a custom designed free-running ADC PET data acquisition system.

ACKNOWLEDGMENT

The authors would like to thank R. Watkins of the Radiological Sciences Laboratory for useful discussions on MRI compatible batteries. Also, Dr. V. Spanoudaki for her knowledgeable expertise on SiPM detectors and data acquisition electronics.

REFERENCES


