Single Channel Optimization for an Endoscopic Time-of-Flight Positron Emission Tomography Detector

Erika Garutti, Karsten Gadow, Martin Goettlich, Alessandro Silenzi and Chen Xu

Abstract—An endoscopic time-of-flight PET detector combined with an ultrasound probe will be designed and built to develop new biomarkers for pancreas and prostate tumors. The detector will consist of a PET head mounted on an ultrasound endoscope and an external PET detector plate with small crystals individually read out via silicon photomultipliers (SiPM). A 200 ps coincidence time resolution for background rejection outside the 3 cm region of interest is required by the system. Single channel simulation and optimization regarding crystal geometry and photodetector characteristics is discussed. System integration issues are studied with the experience gained on a PET prototype.

I. INTRODUCTION

EARLY detection in pancreatic cancer is crucial for curative treatment. However, existing biomarkers are inadequate for effective early detection [1]. A new European Commission project (Endo-TOFPET-US) [2] is devoted to developing new biomarkers of pancreas and prostate tumors. The objective will be addressed using a new endoscopic approach to allow more sensitive, more precise, less invasive imaging and intervention, possibly with lower radiation dose for the patient. A novel dual-modality endoscopic probe for simultaneous positron emission tomography (PET) and ultrasound imaging is foreseen as the best technological solution.

The system comprises of an ultrasound probe combined with a PET detector head extension and an external PET plate. The endoscope is 15 mm in diameter for pancreas and 25 mm in diameter for prostate with the PET head mounted in front of the US probe. Its partner detector is a 172 × 172 mm\(^2\) plate detector and will be placed near the patient on the opposite side of the organ under investigation. Figure 1 shows the operational concept of the pancreatic version of the detector in which the endoscope is inserted into the patient’s esophagus.

A high precision electromagnetic tracking system will be used to determine the position of the endoscopic probe with respect to the external PET plate for each recorded event in order to carry out the image reconstruction.

A. Endoscopic Probe

We will combine a modified ultrasound endoscope with a PET detector head as shown in the figure 2. The PET detector head consists of scintillating crystal fibers read out by matrices of single-photon avalanche diodes (SPAD) with integrated digital circuits. The pancreatic detector will have 162 (9 × 18) scintillating crystal fibers, each one with a dimension of 0.75 × 0.75 × 10 mm\(^3\). The prostate detector will have 324 (18 × 18) crystals with the same dimension. This makes 162 channels for the pancreas version of the detector, 324 channels for the prostate version of the detector with single sided readout or 648 channels with double sided readout. Each channel consists of 416 (16 × 26) SPAD cells to detect optical photons. The crystal processing and wrapping will be done at CERN while the SPAD array is developed at Delft University of Technology.

B. External PET Plate

The external plate will consist of 4096 single crystals read out by silicon photomultipliers (SiPM). The choice of the dimension and the requirement for the characteristics of SiPMs in order to achieve the desired coincidence time resolution and spatial resolution will be studied based on Monte Carlo simulations. A custom designed readout chip for high time resolution is developed by the University of Heidelberg and LIP. A first prototype of the application-specific integrated circuit (ASIC) has been submitted in July 2011.

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The research leading to these results has received funding from the European Union Seventh Framework Programme (FP7/ 2007-2013) under Grant Agreement n° 256984. (Endo-TOFPET-US)
In order to reject the false background from the activity of organs outside the region of interest (ROI) of 3 cm, 200 ps FWHM (Full-Width Half-Maximum) coincidence time resolution (CTR) is required for the PET system. Several factors may affect the system time resolution. For the scintillator, a high light yield for the 511 keV annihilation photons with fast rise and decay time of the light pulse is desired. The geometry and surface processing of the scintillator should minimize the light loss at the crystal interfaces. The photo-detector should have a high effective photon detection efficiency (up to 30%) and amplification gain with low noise and short rise and decay time of the output pulse. We simulate the performance of a single channel of the detector to provide guidance on optimization of the scintillating crystals as well as readout detectors. Our group is also responsible for detector integration. The work presented here is mainly focused on the former part while some detector integration issues will be discussed.

II. Single Channel Simulation

GEANT4 toolkit [3], [4] is used to simulate the response of a Lutetium aluminum garnet (LSO)-like scintillating crystal to 511 keV gamma rays with the surface and wrapping effect on optical photons. Since the scintillation rise time plays an important role for the timing properties [5], [6], the GEANT4 is modified using equation 1:

\[
I = e^{-\frac{t}{\tau_D}} \cdot \left(1 - e^{-\frac{t}{\tau_R}}\right)
\]

(1)
to take the rise time into account [7]. Where \(I\) is the output light intensity, \(t\) is the time, \(\tau_D\) is the decay time constant and \(\tau_R\) is the rise time constant. On the photo-detector surface, we collected the photon hits with their coordinates \((x,y)\) and time stamps for later analysis.

A detailed SiPM model has been built to apply detector digitization on the photon hits. The SiPM modelling considers the Photon Detection Efficiency (PDE), pixel recovery time, noise performance including dark count rate (DCR), cross-talk probability and after-pulse probability, as well as the pixel jitter. The PDE and noise of the SiPM are taken from experiments using a 3×3 mm² Multi-Pixel-Photo-Counter (MPPC) from HAMAMATSU done by the University of Heidelberg [8]. The pixel recovery time is taken from [9]. The pixel jitter, which is the error in the arrival time estimates of scintillation photons [10], is estimated as 50 ps in FWHM.

In order to optimize single channel performance of the detector, various crystal geometries and detector properties are studied. Simulation results are compared on the basis of figures of merit of the detector, namely, channel sensitivity, coincidence time resolution and energy resolution.

A. Single Channel Sensitivity

The sensitivity of the single channel is calculated as the ratio between the number of events with 511 keV deposited in the crystal and the total number of events. Figure 3 shows that the single channel sensitivity increases with crystal volume. In order to maximize the detector sensitivity, the pancreas endoscopic probe will use the single sided readout scheme. The crystal length is limited by the probe diameter to 10 mm. Our simulation also indicates that the CTR deteriorates with the volume of the crystal for a section of 3×3 mm² as shown in figure 4. Note that the single channel sensitivity increases by 30% when the length of a 3×3 mm² crystal changes from 10 mm to 15 mm, but only increases by 10% when the length changes from 15 mm to 20 mm. In order to maximize the channel sensitivity while maintaining the high coincidence time resolution, we choose the crystal length for the external PET plate to be 15 mm.

The required spatial resolution of the order of 1 mm determines the section size of the crystal used in the probe. We calculate the spatial resolution in FWHM using a semi-
empirical formula proposed in [11]:

\[
\text{FWHM}_{\text{res.}} = 1.25 \sqrt{\left(\frac{d}{2}\right)^2 + (0.0022D)^2 + r^2 + b^2},
\]

where \(d\) is the crystal size, \(D\) is the detector separation photon annihilation acollinearity factor, \(r\) is the effective source size (including positron range), \(b\) is the systematic of the positioning scheme and 1.25 corresponds to the 25% degradation due to tomographic reconstruction. By choosing \(D = 6\) cm and \(r = 0.54\) mm, while considering a tracking precision of \(b = 0.56\) mm [12], we conclude that a crystal size of 0.75 mm will fulfill the design goal of \(O(1\) mm) spatial resolution for the detector.

B. Coincidence Time Resolution

In simulation we estimate the single channel time resolution (STR) by using the spread of the time-of-arrival of the first photon from each event. By assuming a two channel system with identical timing performance, we can get the coincidence time resolution of the system as follows

\[
\text{CTR} = \sqrt{2} \cdot \text{STR}.
\]

This can be considered as the theoretical limit on the time precision for the real detector ignoring effects of the readout electronics. Since the number of detected photons plays an important role for the timing precision, we used our simulation to scan over the important parameters which will affect the number of detected photons, namely, crystal light output and detector PDE. Note that the crystal rise time also affects the time resolution of the system, and recent measurements show that 100 ps rise time is realistic for LSO-like crystal [13].

Figure 5 shows the CTR limit of a two channel detector system for different detector PDEs and crystal properties. The line shows the system design goal of the CTR if we estimate the contribution from electronic noise to be 100 ps. The result indicates that we need our detector PDE to be larger than 25% in order to achieve the CTR of 200 ps design goal.

In order to be triggered on the first photon, a fast readout electronic system with fast time-discriminators in parallel is required. The external PET plate and the endoscopic probe use different approaches to realize the high time precision.

The endoscopic probe uses digital SPAD arrays with integrated time-to-digital converters (TDC). Multiple TDCs will be used in order to reject noise of the SPAD i.e. dark events from real photons while still being able to determine the arrival time of the first few photons. Dark events may deteriorate the time resolution of the system in two ways. A pixel with significantly high dark count rate may occupy all TDCs before a real event happens. This problem can be solved by using the integrated circuit to mask out noisy pixels. Besides this, random dark events may still deteriorate the precision of the arrival time measurement if they are mixed with photons from the real event. A check on the total number of activated TDCs in a small periodic interval is therefore implemented. A predefined threshold for the number of activated TDCs will be applied to determine whether a real event has triggered the chip. A real event differs from dark events by the large amounts of instantaneous photons. Records will be discarded if the total number of activated TDC is lower than the threshold, the TDCs will be reset for further measurements. When the threshold is crossed, the recorded arrival times will be kept and read out for further analysis.

The external PET plate uses SiPMs for read out. A double-trigger system will be implemented to realize triggering on the first photon from each event. The system consists of two trigger thresholds. The lower threshold at one photon equivalence (often at 0.5 p.e.) enables the data acquisition (DAQ) system to record the arrival time of the first photon. A higher threshold is used to validate whether this is a real event or it is triggered by dark events. In practice, once the lower threshold is passed, the triggering time stamp will be recorded and a time window for event validation will be opened. Only if the higher threshold is also crossed within the same time window, the time will be kept as the arrival time of the first photon by the DAQ system. Otherwise the system will discard the information and wait for the next trigger.

C. Energy Resolution

The energy resolution of the system in FWHM is determined from the simulation and shown as a function of detector PDE in figure 7. In order to reject Compton events from photoelectric events, the system needs an energy resolution in FWHM better than 20%.
TABLE I
SIMULATION VALIDATION

<table>
<thead>
<tr>
<th></th>
<th>$N_{det}$</th>
<th>$\sigma_{FWHM} [%]$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Simulation</td>
<td>544 ± 54</td>
<td>11.8 ± 0.4</td>
</tr>
<tr>
<td>Experiment</td>
<td>539 ± 25</td>
<td>14.1 ± 1.0</td>
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Validation measurements for the Monte Carlo simulations with respect to energy resolution have been carried out. Figure 8 shows the measurement setup for the simulation validation. A $3 \times 3 \text{mm}^2$ MPPC (S10362-33-050C) with 50 $\mu\text{m}$ pixel size is coupled with an LSO crystal with dimension of $3 \times 3 \times 15 \text{mm}^3$ to measure the spectrum from a $^{22}\text{Na}$ source. The crystal is wrapped in enhanced specular reflector foil. Optical grease is used to couple the MPPC with the crystal. Optical grease is a clear, colorless, silicone optical coupling compound that features good light transmission of about 95% for wavelengths between 280 nm and 700 nm with refraction index of 1.465. Signals from the MPPC are read out and integrated by a charge to digital converter (QDC) from Lecroy. A comparison between the spectrum obtained from the experiment using an MPPC operated at 0.66 V over its breakdown voltage and simulation result with detector PDE equals 13% is given in figure 9. Table I summarizes the result of the comparison, where $N_{det}$ is the number of fired pixels corresponding to the center of the photo-peak, and $\sigma_{FWHM}$ gives the energy resolution of the photo-peak. As expected the energy resolution from the experiment is deteriorated by the electronic noise of the DAQ system. Worse agreement between experiment and simulation exhibits for higher over voltage of the MPPC, this is probably due to the lack of detailed experimental input of cross-talk and after-pulse, since they are more device specific and become dominant at higher over voltage. More measurements regarding the noise performance of the devices will be carried out in order to extend the regime of validity of the simulation.

III. SYSTEM INTEGRATION

A. Choice of SiPMs

We summarize key factors for our PET detector regarding the choice of SiPMs which will be used in the system. In order to maximize the detector sensitivity, minimum dead material between crystals is required. Therefore we are looking for surface mounted SiPM matricies with minimum package size. A potential candidate for the PET plate detector is the $4 \times 4$ MPPC array (S11828-3344M) from HAMAMATSU [9] with 200 $\mu\text{m}$ gap between SiPMs. However, due to the relatively large package size of the array module, there will be still a large area of dead material left if they are used as detector unit for the external PET plate. For example, each MPPC array consists of 16 $3 \times 3 \text{mm}^2$ MPPC modules, the total package size of the array is 194.48 (13.6 $\times$ 14.3) mm$^2$. This means 26% of the area within one module is dead material if we integrate the array to the detector system. Note that the loss in sensitivity due to dead material in the detector is not considered in the single channel simulation shown in the former section.

B. System Operation Study

Another aspect concerning the system integration is to compensate the parameter spread among SiPMs when using multiple channel read out. We have investigated the system operation issues by using a TOF-PET prototype developed by DESY and the University of Heidelberg [7]. The system consists of two detector arms mounted on a movable support.
Each arm consists of 4 submodules of detector units. A submodule has $4 \times 3 \times 15$ mm$^3$ Lutetium Fine Silicate (LFS [14]) crystals read out by a $2 \times 2$ MPPC array. Therefore the system consists in total of 32 channels. We get the most uniform time resolution between channels by tuning all SiPMs to equal gain to give the same number of fired pixels in photo-peak, which requires 0.2 V spread on bias voltage for 32 channels, and in turn results in 13% spread in energy calibration constant for 511 keV gamma ray. For a SiPM read out chip for 64 channels, the result from the study infers single channel bias tuning of larger than 1 volt is necessary.

The temperature dependence of the breakdown voltage of the SiPM is also studied with the system. Figure 10 shows the breakdown of a $3 \times 3$ MPPC (S10362-33-050) as a function of the environment temperature. In order to compensate the instability introduced by the temperature change, the detector system will use a layer of Peltier elements for cooling. Temperature controlled feedback will be applied to the bias voltage supply of the detector. And the custom designed read out chip will be optimized to low power consumption ($\leq 20$ mW/ch).

**IV. Conclusion**

We have shown the current design of the Endo-TOFPET-US detector. The Monte Carlo simulation including realistic SiPM digitization has indicated that at least a detector PDE of 25% is required to realize the high time precision of the order of 200 ps. The crystal geometry for the probe detector is constraint to $0.75 \times 0.75 \times 10$ mm$^3$ by simulation due to channel sensitivity as well as the probe size. The crystal size for the external plate is determined as $3 \times 3 \times 15$ mm$^3$ considering the results from simulation. We conclude that according to the simulation with realistic data inputs and SiPM digitization scheme, the aimed time resolution of 200 ps is achievable for the detector under study.

**ACKNOWLEDGMENT**

We would like to thank Excelitas, STMicroelectronics and Zecotek for kindly providing their SiPM samples for us to test. We would like to thank all our colleagues from the Endo-TOFPET-US consortium (see the footnote on page 1). We also thank N. Hegemann and M. Schmit from the University of Hamburg who helped in some of the experiments.

**REFERENCES**


