Time-of-flight PET with SiPM sensors on monolithic scintillation crystals
Vinke, Ruud

IMPORTANT NOTE: You are advised to consult the publisher's version (publisher's PDF) if you wish to cite from it. Please check the document version below.

Document Version
Publisher's PDF, also known as Version of record

Publication date:
2011

Link to publication in University of Groningen/UMCG research database

Citation for published version (APA):
9 Discussion, outlook and valorization

9.1 Discussion

For high-quality PET images, PET detectors require high detection efficiency, high spatial, energy and timing resolution, and should provide full 3D information about the position of interaction. The last point implies that PET detectors should include the capability of determining the depth-of-interaction (DOI). Unfortunately, these performance parameters often impose conflicting design requirements. For example, high detection efficiency requires thick PET detectors. However, the positioning performance generally degrades with increasing thickness, due to increased scattering of the impinging gamma photons within the detector and, for monolithic scintillation crystals, due to an increasing spread of the scintillation light. High detection efficiency and high spatial resolution is an especially important design conflict for detectors that do not provide information about the DOI: As described in section 2.2.3, the parallax effect severely degrades the image spatial resolution in PET scanners when there is no DOI reconstruction ability, and this degradation worsens for thick detectors that have a large DOI spread.

This work concentrated on the analysis of the performance of monolithic scintillation crystals coupled to fast photosensor arrays for TOF-PET. A statistical reconstruction algorithm based on maximum likelihood estimation (MLE) was developed for estimating the position-of-interaction of gamma photons inside monolithic scintillation crystals (see chapter 4). For a $20 \times 20 \times 12$ mm$^3$ LYSO crystal coupled to a $4 \times 4$ MAPMT with $4.2 \times 4.2$ mm$^2$ anodes, a spatial resolution of 2.4 mm FWHM was achieved using this method.

As discussed in section 3.3.3, monolithic scintillation crystals inherently provide information about the DOI. Several authors have shown that the width of the scintillation light distribution at the photosensor array correlates with the DOI, e.g. [69, 148]. For optimal correlation, monolithic crystals are normally covered with absorptive black coatings to minimize the reflection of the scintillation light at the crystal surfaces. Scintillation photons that undergo multiple reflections lose the information about their position of origin and manifest themselves as background light, widening the spatial distribution of the scintillation light at the photosensor plane, and thus affecting the correlation with the DOI. However, the choice to minimize the reflection of the scintillation light at the crystal surfaces lowers the scintillation light collection efficiency and thus worsens the energy and timing resolution of the detector (see section 3.1). From Figs. 4.11 and 8.7, it is apparent
that the DOI correlates with the width of the light distribution, also for monolithic crystals covered with reflective coatings. The distribution patterns in Figs. 4.11 and 8.7 were obtained by histogramming the detector response according to the reconstructed 3D position-of-interaction. During the position reconstruction, no prior assumptions on the distribution patterns were used (see chapter 4), meaning that the observed correlation is a physical effect and not introduced by the position algorithm itself. Monolithic scintillation detectors can thus be optimized for energy and timing resolution with reflective coatings, while the 3D position-of-interaction of the gamma photons can still be reconstructed using a single photosensor array.

The DOI resolution could be estimated by reconstructing the beams entering the crystal from the side, at a defined distance from the photosensor array (Figs. 4.10 and 5.5). DOI resolutions between 2 mm and 5 mm were achieved for LYSO crystals coupled to a MAPMT array in chapters 4 and 5. The variation of the 3D position resolution (thus also DOI) with distance from the photosensor array results from the variation in the spread of the scintillation light, as discussed at the end of section 3.3.1 (specifically Eqs. 3.11 - 3.13). In chapter 5, it was shown that 3D position reconstruction was even possible for thick monolithic crystals with a cubical shape (with a thickness comparable to the transverse extension), coupled to a single photosensor array. Although the spatial resolution degrades with increasing thickness, it might still be at an acceptable level if maximum detection efficiency is required at minimal detector costs (using a single photosensor array to read out the crystal), e.g. for whole-body PET imaging. For monolithic crystals with a cubical shape, the position reconstruction near the crystal edges is however a challenging task, which needs further investigation. Since the majority of the impinging gamma photons interact at small DOI (due to the exponential attenuation law), the positioning performance of thick monolithic crystals should be improved by placing a semiconductor photosensor array on the front side of the crystal (facing the gamma beam), as discussed in chapter 5. Measurements with this readout geometry are being prepared.

Since the light collection efficiency was optimized for the monolithic crystals by using reflective coatings, good energy and timing performance was obtained. The energy resolution of ~11% in chapters 4 and 5 is normal for LYSO. The timing resolution of 320-360 ps FWHM for the LYSO-MAPMT detector in coincidence with a BaF$_2$ detector (whereby the BaF$_2$ detector had a single detector timing resolution of ~180 ps FWHM) translates into a single detector timing resolution of 265-312 ps FWHM for the LYSO-MAPMT detector (and a coincidence timing resolution of 374-441 ps FWHM for two such LYSO detectors). These timing resolutions were obtained by digital time pickoff algorithms, triggering on the early part of the rising edge of the timing pulses. Since the emission rate of scintillation photons is highest during the initial moments of the scintillation process, the arrival time variance is lowest for the photons that arrive at the photosensor during
the earliest stage, see [97] for a theoretical derivation. At the same time, the signal slope of the timing pulses is highest at the early part of the rising edge. Triggering on the early part minimizes the electronic noise contribution to the timing resolution. As shown by Eq. 8.1, the signal slope of the timing pulse determines the effect of the electronic noise on the timing resolution.

Chapter 4 presents a method to calibrate and correct for the arrival time variation with the position-of-interaction of the gamma photon. The observation that this arrival time variation with position-of-interaction can be considerable for scintillation crystals, was earlier shown by Moses and Derenzo [89], reporting arrival time variations with excitation depth between 200 and 400 ps for $3 \times 3 \times 30 \text{ mm}^3$ LSO crystal segments, depending on the crystal surface treatment. Figs. 4.12, 8.8 and in particular 5.3 show that there is also an arrival time variation with position-of-interaction for monolithic crystals, but this variation is much smaller than was reported for the $3 \times 3 \times 30 \text{ mm}^3$ LSO crystal in [89]. For the $16.2 \times 18 \times 20 \text{ mm}^3$ LYSO crystal in Fig. 5.3, the arrival time variation with excitation depth was only about 70 ps. This is a very promising result. As discussed in section 8.3.4, it may well be that the scintillation light undergoes far less surface reflections and surface absorptions in monolithic crystals as compared to segmented crystals, due to the larger distance to the crystal side surfaces. This would result in a decrease in the path length (variation) and thus a decrease in arrival time variation. It was already discussed in section 3.3.1 that monolithic crystals may have a better light collection efficiency than segmented crystals, and thus a lower statistical spread in the arrival time of the scintillation photons at the photosensors. The apparently smaller arrival time variation with position-of-interaction for monolithic crystals further indicates that monolithic crystals are better suited for TOF-PET than segmented crystals.

In chapters 4 and 5, the timing resolution for the monolithic crystals could not significantly be improved by applying a position correction to the timing. In chapter 5, it was shown that there was virtually no arrival time variation with excitation depth near the crystal surface opposite to the photosensor plane, where most gamma interactions took place (due to the exponential attenuation law). This was attributed to the reflection of the scintillation light at the crystal surface, boosting the early arrival of scintillation photons at the sensor. It was discussed that the position correction to the timing might be interesting for monolithic crystals read out by a semiconductor photosensor array, facing the gamma beam (this configuration should improve the positioning performance, see discussion earlier in this section). As shown in Fig. 5.3, there is a considerable propagation time variation near the photosensor plane, and this is where most gamma interactions will take place for such a configuration. In this situation, a position correction to the timing might improve the timing resolution.

In chapter 6, the timing resolution of the SiPM sensor was tested. Single
Chapter 9

Photoelectron timing resolutions $\sigma_{t,\text{spe}}$ down to about 100 ps RMS were found for the $1 \times 1$ mm$^2$ Hamamatsu SiPM, whereby the timing resolution $\sigma_t$ for multiple photoelectrons $N_{\text{phe}}$ could be described by Poisson statistics ($\sigma_t = \sigma_{t,\text{spe}}/\sqrt{N_{\text{phe}}}$).

In chapter 7, $3 \times 3 \times 5$ mm$^3$ LaBr$_3$:Ce scintillators were coupled to $3 \times 3$ mm$^2$ SiPMs. For 511 keV gamma photons, excellent coincident timing resolutions of $\sim$100 ps FWHM were obtained, using low-noise and high-bandwidth preamplifiers and digital time pickoff, triggering on the early part of the rising edge of the timing pulses. This world-record timing performance indicates that SiPMs are well-suited for TOF-PET. For monolithic crystals read out by SiPM arrays, the scintillation light is shared over multiple SiPM elements. Each SiPM element introduces dark counts and each associated preamplifier introduces electronic noise. For optimal timing performance, the timing signal from the SiPM array needs to be based on a combination of the signals from the SiPM elements, and will thus contain the noise from multiple SiPM signals (there is no common 'dynode' signal available, as in a MAPMT, see section 3.2). From Eq. 8.1, it is clear that the noise performance needs to be as good as possible for the preamplifier that forms the timing signal from the SiPM array. In chapter 8, a single detector timing resolution of 225 ps FWHM was achieved for a $16.2 \times 18 \times 10$ mm$^3$ LaBr$_3$:Ce(5%) crystal coupled to a $4 \times 4$ SiPM array with $3 \times 3$ mm$^2$ pixel size, whereby the timing signal was based on an analogue sum of the 16 SiPM output signals. This timing resolution was limited by the large amount of dark counts and electronic noise in the timing signal. It is expected that a preamplifier with an optimized noise performance will improve the timing resolution for this detector.

In this work, a TOF-PET detector concept based on monolithic scintillation crystals and fast photosensor arrays was characterized. It was shown that the 3D position-of-interaction (thus including DOI) could accurately be reconstructed with a single photosensor array coupled to the monolithic crystal. The light collection efficiency of monolithic crystals may be better than for segmented crystals, making them more suitable for TOF-PET. The costs of manufacturing are further much lower than for segmented crystals. With digital time pickoff algorithms, good timing performance was obtained. Due to their compactness and transparency to 511 keV gamma photons, SiPM arrays allow flexible readout geometries of the monolithic crystals. Because of the high gain and low transit time jitter, SiPMs have excellent timing performance and can be used for TOF-PET. They are further compatible with magnetic fields, which makes them possible candidates for PET-MRI integrated systems.
9.2 Outlook and valorization

In the near future, thick monolithic crystals will be used with front and double-sided readout by SiPM arrays, using a redesigned multichannel preamplifier optimized for noise and bandwidth. With this preamplifier, it is expected that single detector timing resolutions below 200 ps FWHM (and thus coincidence timing resolutions below 300 ps FWHM) can be achieved for monolithic LaBr$_3$-SiPM detectors. With this preamplifier, the timing performance of monolithic LYSO-SiPM detectors is also expected to be optimized. The position reconstruction near the crystal edges will be investigated further for these detectors.

The TOF-PET detector concept described in this work is interesting for whole-body PET imaging, whereby the excellent timing resolution enables PET images with a very high signal-to-noise ratio. In addition, the developed position-of-interaction algorithm is capable to accurately measure depth-of-interaction (DOI), making sure that there is no parallax degradation, such that the image resolution is optimized over the entire field-of-view (FOV). The absence of parallax degradation also enables the construction of PET systems with detector rings with a reduced diameter (comparable to the transaxial FOV), thereby reducing detector costs (because less detector material is needed), increasing the system sensitivity (more solid angle coverage) and reducing the position blurring associated with the acollinearity (Eq. 2.1). No crystal segmentation procedure is required for this detector concept, while no detector modifications are necessary to enable DOI measurement, such that the costs for the detector parts are significantly reduced compared to the current PET detector technology in commercial scanners. In addition, field-programmable gate arrays (FPGAs) and application-specific integrated circuits (ASICs) allow advanced signal processing at affordable costs and in reasonable time. Further, when using SiPMs for reading out the scintillation light, simultaneous PET-MRI imaging is possible, since SiPMs are insensitive to magnetic fields. This detector concept has the potential to significantly improve PET’s ability to visualize, quantify and characterize disease. This could lead to an earlier diagnosis of cancer, which is essential for an effective cancer therapy.

This detector concept is also interesting for applications involving a small FOV. Dedicated, small detectors for imaging certain organs in close proximity should give superior spatial resolution and signal-to-noise ratio compared to that attainable with whole-body imaging, because of the smaller distance to the detectors. Examples include imagers for the head and neck region, and mammography. Another example is radioguided surgery using intraoperative PET. Here, the PET detectors enable real-time tumor localization and help the surgeon to verify complete resection of the tumor. Since the FOV is much smaller, it is even more important that the DOI of the gamma photons is reconstructed properly in the detectors, as compared to whole-body imaging. Since the detector concept in this work can
be very compact (especially when using SiPM arrays) and is able to accurately reconstruct the DOI, it may also be interesting for these small FOV applications. In addition, because of the excellent timing resolution, these applications may also benefit from the TOF-PET technique.