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A new PET detector concept for compact preclinical high-resolution hybrid MR-PET

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ABSTRACT

This work presents a new PET detector concept for compact preclinical hybrid MR-PET. The detector concept is based on Linearly-Graded SiPM produced with current FBK RGB-HD technology. One 7.75 mm x 7.75 mm large sensor chip is coupled with optical grease to a black coated 8 mm x 8 mm large and 3 mm thick monolithic LYSO crystal. The readout is obtained from four readout channels with the linear encoding based on integrated resistors and the Center of Gravity approach.

To characterize the new detector concept, the spatial and energy resolutions were measured. Therefore, the measurement setup was prepared to radiate a collimated beam to 25 different points perpendicular to the monolithic scintillator crystal. Starting in the center point of the crystal at 0 mm / 0 mm and sampling a grid with a pitch of 1.75 mm, all significant points of the detector were covered by the collimator beam.

The measured intrinsic spatial resolution (FWHM) was 0.74 +/- 0.01 mm in x- and 0.69 +/- 0.01 mm in the y-direction at the center of the detector. At the same point, the measured energy resolution (FWHM) was 13.01 +/- 0.05 %. The mean intrinsic spatial resolution (FWHM) over the whole detector was 0.80 +/- 0.28 mm in x- and 0.72 +/- 0.19 mm in y-direction. The energy resolution (FWHM) of the detector was between 13 and 17.3 % with an average energy resolution of 15.7 +/- 1.0 %. Due to the reduced thickness, the sensitivity of this gamma detector is low but still higher than pixelated designs with the same thickness due to the monolithic crystals. Combining compact design, high spatial resolution, and high sensitivity, the detector concept is particularly suitable for applications where the scanner bore size is limited and high resolution is required — as is the case in small animal hybrid MR-PET.

1. Introduction

While the first simultaneous MR-PET human systems were realized with photomultiplier tubes (PMT) located outside the main magnetic field B_0 [1], subsequent systems were based on avalanche photodiode (APD) technologies [2,3]. However, silicon photomultipliers (SiPM) provide a higher internal gain, better gain stability, and better timing performance. Consequently, recent hybrid human and small animal systems were developed with SiPM based PET detectors [4–10].

In PET scanner development, there is a strong demand for higher spatial scanner resolutions, especially in small animal applications such as mouse brain imaging. The scanner resolution can be improved with a higher intrinsic spatial resolution of the installed detectors. The intrinsic spatial detector resolution defines how accurate the x- and y-position of the scintillation event can be detected. Traditionally, intrinsic detector resolution has been improved, reducing the pixel size of pixelated scintillator crystal arrays. At the same time, sensor pixel sizes have decreased due to new technologies and production methods of SiPM. This was driven by the demand for a higher granularity of readout pixels to increase the intrinsic resolution. Hence, the demand for a reduction of the necessary readout channels exists to compensate for the increased number of resulting readout channels.

While most SiPMs do not provide spatial information encoding, some approaches for implementing this functionality have been recently developed. Specifically, several position-sensitive SiPM (PS-SiPM) methods were developed. Methods with highest encoding density are implemented with four different output channels. Two channels are required

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for each x- and y-direction encoding. With these 4 channels it is possible to encode a sensitive sensor area of, e.g., 4 mm by 4 mm or 8 mm by 8 mm in size. In general, it is possible to distinguish between two basic position-sensitive encoding principles.

The first principle is based on a position encoding where the microcells of a pixel or defined area of the sensor are connected to the different output channels. Here, the number of output channels connected to a single channel varies linearly according to position. The advantage of this implementation method is that the microcells can be connected directly to the output channels to encode the signal. An implementation of resistors to encode the position is not necessary. This avoids potential loss of sensitive sensor area. A drawback of this method is that the intrinsic spatial resolution is limited by the quantity of microcells connected to the readout channels. This is why a position encoding with a granularity of a single microcell position cannot be realized. Currently, there are two developed sensor types published as Sensitivity-Encoded Silicon Photomultiplier (SeSP) [11,12] and Interpolating Silicon Photomultiplier (iSiPM) [13]. SeSP sensors can be implemented with a single metal layer by connecting the microcells in rows or columns. A bridge where readout channel signals cross each other is only required at the edges. Contrarily, iSiPM sensors are not arranged in a structured pattern. As a result, an implementation of iSiPM can be more challenging and requires an additional metal layer. Otherwise, the more random or *salt-and-pepper* distribution of iSiPM qualifies the sensor type better for detectors with monolithic crystals rather than SeSP.

A second type of PS-SiPM contains resistive networks distributing the generated charge of the microcells. This concept was already reported for APD [14] and was widely used with position-sensitive PMT [15,16]. A SiPM technology with a charge distributing resistive network was developed and published as Position-Sensitive Solid-State Photomultipliers (PS-SSPM) [17]. This concept can theoretically encode each single SiPM microcell. However, a resistor between each microcell is required. Consequently, additional space for a resistor layer is demanded by this sensor type. Moreover, the timing resolution may be decreased by the resistive network. Therefore, the timing resolution of PS-SSPM is lower as, e.g., for SeSP devices (cf. [11,12] and [17]). Another method to reduce readout channels while providing a high granularity was published as Linearly-Graded SiPM (LG-SiPM) [18] and is described in the next section. The sensor concept was characterized with 0.8 mm [19] and 0.53 mm [20] pitch Cerium-doped Lutetium Yttrium Orthosilicate:LYSO crystal arrays demonstrating a high intrinsic spatial resolution.

Especially in preclinical hybrid MR-PET scanner design, available space limits the scintillator crystal thickness and scanner geometry resulting in restrictions for scanner sensitivity. If pixelated crystal arrays are designed with smaller pixel sizes, the proportional amount of reflector foil and glue between the sensor pixels is increased. This results in an additional loss of sensitivity — especially for coincidence detection. For example, the amount of nonsensitive area loss for a pixelated array with 0.45 mm × 0.45 mm crystals resulting in an effective pitch size of 0.53 mm as assembled in [20] is 28%. Measuring in coincidence, this leads to a loss of 48% of sensitive area caused by scintillator volume replaced by reflector foil and glue, cf. [21]. For example, the resulting PET scanner sensitivity for small animal scanners which provides very high spatial resolution due to very small pitch sizes of pixelated scintillator crystal arrays is below 1% for a ring with an axial field-of-view of 7 mm, crystal length of 13 mm and events with energies higher than 250 keV [22]. In some preclinical PET studies observing, e.g., mouse brains for neurological research, besides a high spatial resolution a high scanner sensitivity is necessary. This is caused by the limitation of injectable neurotransmitters where a higher concentration of the neurotransmitters would change the studied metabolic function. Thus, the activity is limited by receptor occupancy [23]. Moreover, a higher sensitivity of PET scanners can reduce the radiation exposure of living objects. This is relevant to longitudinal studies where animals have to be scanned repeatedly [24,25].

High resolution below 1 mm with 12 mm by 12 mm large and 5 mm as well as 10 mm thick monolithic crystals coupled to pixelated SiPM was successfully demonstrated in [26]. The SiPM sensor consisted of an 8 by 8 pixelated array resulting in 64 readout channels for an encoding of a sensitive area of 144 mm^2. SeSP and LG-SiPM sensors can encode an up to 64 mm^2 large area with 4 output channels. In addition, in [26], a Depth-of-Interaction (DOI) resolution close to 2 mm was achieved. A spatial resolution close to 1 mm and a DOI resolution of 2.5 mm were also reported in [27]. There, a detector area of 1024 mm^2 was encoded with 64 channels.

Another PET prototype scanner published as Digi-PET [28], realized with monolithic crystals, provides a sensitivity in the same order of magnitude as the prototype published in [22], although the thickness of the monolithic crystals is only 2 mm. Nevertheless, the Digi-PET scanner consists of dSiPM from Philips Digital Photocounting. These sensors are only available in one format which has not been optimized for small animal scanners, especially for hybrid MR/PET scanners with a small bore size. Proof of the high sensitivity still in demand, combined with high timing resolution is evident by the ongoing project called Small Animal Fast Insert for mRi (SAFIR). The project has the goal of developing a PET insert for ultra-high field 7 T Bruker MRI scanner. It should achieve a scanner sensitivity of 6% while providing a spatial FWHM resolution of 1.5 mm [29]. However, this project is not using detectors based on monolithic crystals. Moreover, it is not aiming at sub-millimeter spatial resolution.

This work presents a new detector concept based on LG-SiPM coupled to monolithic crystals. It provides a high intrinsic detector resolution, potential high PET scanner sensitivity, a compact design, and a readout channel reduction method with granularity on microcell level. This ideally qualifies the new detector concept for small animal MR-PET scanners and neuroscience applications.

2. Detector concept

2.1. Linearly-graded silicon photomultiplier

The new detector concept consists of LG-SiPM as the required scintillation light sensor. The sensor concept was first published by Alberto Gola, who first implemented this concept into SiPM, in [18]. Here the essential part of this new detector concept, which is described in detail in [18], is summarized briefly.

Fig. 1 shows the circuit diagram of a 2-dimensional LG-SiPM sensor. In contrast to other SiPM microcells, a microcell of a LG-SiPM consists of a diode D driven in Geiger mode with two equal quench resistors $R_{q,h}$ and $R_{q,v}$. The two quench resistors act as a current divider represented by S_q. The current divider divides the generated photo current of each diode into two equal currents. A main advantage of LG-SiPM is its high granularity, which means that each microcell position can be encoded individually, while reducing the required resistors compared to PS-PSSM. This is realized by organizing microcells into a grid consisting of horizontal and vertical rows. The generated current of each cell is halved in horizontal and vertical parts by S_q and is summed up along each cell position in each horizontal and vertical row. The position of each row is encoded with a second current divider S_h resp. S_v. The resistors $R_{h,A}$ and $R_{h,B}$ resp. $R_{v,C}$ and $R_{v,D}$ vary linearly with the horizontal or vertical position and are connected to the readout channels A and B resp. C and D. As a result, all possible horizontal and vertical positions of the grid of microcells are encoded linearly. Hence, each microcell position can be individually encoded, leading to highest possible granularity. Furthermore, the required quantity of resistors is significantly reduced to two resistors for each horizontal and vertical row. The centroid of a flash of light can be calculated in a horizontal direction with

$$x = \frac{Q_A - Q_B}{Q_A + Q_B}, \tag{1}$$

and in a vertical direction with

$$y = \frac{Q_C - Q_D}{Q_C + Q_D}, \tag{2}$$