The MINDView brain PET detector, feasibility study based on SiPM arrays

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Abstract
The Multimodal Imaging of Neurological Disorders (MINDView) project aims to develop a dedicated brain Positron Emission Tomography (PET) scanner with sufficient resolution and sensitivity to visualize neurotransmitter pathways and their disruptions in mental disorders for diagnosis and follow-up treatment. The PET system should be compact and fully compatible with a Magnetic Resonance Imaging (MRI) device in order to allow its operation as a PET brain insert in a hybrid imaging setup with most MRI scanners. The proposed design will enable the currently-installed MRI base to be easily upgraded to PET/MRI systems.

The current design for the PET insert consists of a 3-ring configuration with 20 modules per ring and an axial field of view of ~15 cm and a geometrical aperture of ~33 cm in diameter. When coupled to the new head Radio Frequency (RF) coil, the inner usable diameter of the complete PET-RF coil insert is reduced to 26 cm. Two scintillator configurations have been tested, namely a 3-layer staggered array of LYSO with 1.5 mm pixel size, with 35 x 35 elements (6 mm thickness each) and a black-painted monolithic LYSO block also covering about 50 x 50 mm² active area with 20 mm thickness.

Laboratory test results associated with the current MINDView PET module concept are presented in terms of key parameters’ optimization, such as spatial and energy resolution, sensitivity and Depth of Interaction (DOI) capability. It was possible to resolve all pixel elements from the three scintillator layers with energy resolutions as good as 10%. The monolithic scintillator showed average detector resolutions varying from 3.5 mm in the entrance layer to better than 1.5 mm near the photosensor, with average energy resolutions of about 17%.

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1. Introduction

Standard PET/CT (and now also the new PET/MRI) human whole body scanners result in a PET image resolution in the 4–5 mm FWHM range and therefore, are not optimized for imaging small mm-size structures in the brain. Even for processes localized within the amygdala, it is currently not possible to differentiate if they are occurring within the basolateral complex or the central nucleus [1]. With spatial resolutions of ~1 mm it will be possible to answer specific questions relating to brain function in small brain regions.

The main technical goal of the Multimodal Imaging of Neurological Disorders (MINDView) project and, thus, the subject of this contribution, is the achievement of simultaneous high-resolution PET/MRI imaging dedicated to brain examination to study psychiatric disorders in the clinical research setting. The MINDView project will develop a high resolution and high sensitivity brain-dedicated PET system, capable of visualizing neurotransmitter pathways and their hypothesized disruptions in mental disorders...
for diagnosis and treatment monitoring. Furthermore, this compact PET imaging device has to be fully compatible with operation in high magnetic fields in order to allow its integration with most common clinically used MRI scanners. Both the PET system and a new transmitter/receiver (TR) RF coil (custom made by NORAS MRI, Germany) systems are integrated into a single portable and compact design which is dedicated to brain examination. The dimensions of the system are arrived at as a compromise between the PET imager and the RF-coil performances. Based on head size investigations [2] and keeping patient comfort in mind, the geometric aperture of the whole PET/RF device is 26 cm. This is accomplished by using 3 rings in the axial direction, with 20 modules per ring (see Fig. 1) of LYSO crystals (5 cm x 5 cm x 2 cm thick) using Silicon PhotoMultiplier (SiPM) arrays as photosensors. This configuration leads to an axial field of view (FOV) of approximately 15 cm and a geometrical PET aperture of ~33 cm in diameter. The 20 RF coil rungs are integrated in front of the PET system, placed in the junction regions between the modules. In order to minimize distortions in the MRI performance, the distance between the shielded PET system and the rungs of the RF conductor area is about 3 cm [3]. The coincidence of one PET detector module with 10 opposite modules produces a transaxial FOV of about 24 cm, which ensures complete coverage of the human brain [4]. A set of mirrors placed in front of the patient’s eyes (behind the RF coil) will ease the patient’s sensation of claustrophobia. In this initial design, the absolute PET sensitivity for a linear source covering the axial FOV is estimated to be of about 3.3% by taking into account solid angle coverage and linear attenuation coefficient at 511 keV photon energy for 2 cm LYSO thickness [5].

MINDView main goals are high sensitivity, spatial resolution of 1 mm in the central brain region and DOI capabilities. To achieve these targets, two major LYSO scintillator crystal configurations for the detector modules are under consideration: (i) a staggered pixelated crystal array and (ii) a monolithic approach (see Fig. 2). There are advantages/disadvantages to both the monolithic and pixelated crystal designs that have been described in the literature [6,7].

Some advantages of the continuous monolithic crystal design are better light collection, higher detection sensitivity per area (no crystal dead area), better spatial uniformity (no sharp discontinuities), continuous positioning (non-pixelation artifacts),

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**Fig. 1.** Left: sketch of the transmitter/receiver (TR) RF coil and the PET inset structure placed around the patient’s head and inside a 700 mm MRI bore. Center: sketch of the simulation for the monolithic approach, 20 modules per ring, 330 mm aperture. Right: photograph of a mock-up of the PET and RF coil for patient comfort studies. The three PET modules in the axial direction are mechanically packed inside a common cassette.

**Fig. 2.** Top-left: sketch of three staggered crystal layers whose individual thickness are 6 mm. There is a half pixel shift of the 2nd and 3rd layers with respect to the 1st in the X direction, and a shift of half a pixel of the 1st and 2nd with respect to the 3rd in the Y direction. Top-right, sketch of the monolithic scintillator with two possible scintillation light distributions. Bottom: Photographs of the 3 staggered crystal arrays coupled to the SiPM array (left) and photograph of the monolithic 50 mm x 50 mm x 20 mm scintillator (right).
Advantages of the pixelated crystal design are intrinsic spatial resolution selected by crystal width, focused light (better signal-to-noise ratio), and improved spatial linearity (especially when compared with the monolithic approach at the borders of the detector modules). However, although pixel sizes up to 0.5 mm × 0.5 mm have been proposed for high resolution PET applications [8], apart from the cost increase, there are fundamental limitations of the spatial resolution affected by the position range or the inter-crystal scattering, but also the achieved energy resolution.

We have decided to test both monolithic and pixelated array configurations at the MINDView PET detector module level prior to choosing the final configuration. Simulations have also been carried out for both configurations at the whole PET device level in order to compare the expected performance of the MINDView brain PET.

Although the total thickness of the crystal (in order to ensure the necessary gamma stopping power) is expected to be in the range of 18–20 mm, scanner sensitivity can be increased by placing detector rings very close to the brain. Such a geometry increases the number of gamma rays which are obliquely incident to the scintillation crystals in the PET detectors. For that reason, DOI reconstruction capability to avoid parallax error is of primary interest to achieve high spatial resolution, especially toward the borders of the FOV. For pixelated crystal approaches, various depth encoding detector designs have been proposed based on stacking single layer crystal arrays [9,10], detecting scintillation photons at two opposite sides of the crystal arrays (dual-end readout detector) [11,12], using pulse shape discrimination (phoswich detectors) [13,14], and light-sharing methods [15,16]. The relative offset method provides another means of obtaining DOI information using staggered crystal layers and a single-end readout, thus reducing an additional photosensor cost and complexity. In this approach, one of two crystal layers is shifted by half a crystal pitch in both the horizontal and vertical directions [17–19]. By combining the above approaches, crystal layer numbers can be further increased [20]. Up to eight-layer DOI detectors have been proposed using the sharing light method and the pulse shape discrimination method [21].

Very few groups have described designs and performance of PET inserts for simultaneous PET and MR imaging, specially developed for human brain studies. A PET insert based on Avalanche Photodiodes (APDs) was designed and clinically used in studies of glioblastoma [22]. In this approach, blocks of 6 arrays of 12 × 12 elements with 2.5 mm × 2.5 mm × 20 mm LSO crystals are read out by 9 APDs. A PET insert with a short axial FOV of 6 cm and an inner diameter of the PET insert of 39 cm has also been presented [23,24]. Although the design proposed in MINDView is based on pixelated LYSO crystals (3 mm × 3 mm × 20 mm) coupled to SiPMs, they locate all the photosensor read-out electronics outside the MRI bore and transmit all SiPMs signals through a 4 m long flexible flat cable. As will be described below, our design features the read-out electronics placed on the scintillator-photosensor module, thus the SiPMs signals do not suffer any significant degradation due to transmission signal losses outside the MRI bore. In the aforementioned design, the spatial resolution varies from 3 mm at the FOV center to up to 5 mm at 10 cm off-center of the FOV due to parallax error, as no DOI correction is applied in that design. An additional development has recently been shown using the four layers of LYSO crystals to return DOI capabilities [15,25]. Pilot tests with one detector module evaluating the MRI and PET performance showed good results with energy resolution differences lower than 0.3% at 15–18%. Some MRI signal degradation was observed due to the lack of proper detector PET shielding.

This paper presents the main features of the PET detector insert for the MINDView project in terms of its overall design, electronic readout, and MRI compatibility. In addition, the main parameters of the PET detector module insert, (such as expected spatial and energy resolutions, DOI capability and sensitivity) are discussed in terms of the different approaches considered so far for the construction of the first MINDView prototype.

2. Methods and materials

2.1. Crystal configuration

In this work we have studied the performance of a MINDView PET module with DOI capabilities by using the relative offset method applied to a 3 staggered layers of 1.5 mm pitch pixelated LYSO (Proteus, Ohio, US) configuration with 35 × 35, 36 × 36 and 37 × 37 elements, respectively. Thus, the largest array coupled to the photosensor by means of an acrylic layer of 1.5 mm thickness and optical grease, covers an area of 55.5 mm × 55.5 mm, being slightly larger than the photosensor area (5 cm × 5 cm). The LYSO pixels were polished and later covered by reflective material (Enhanced Specular Reflector, ESR, 3 M™). The three layers of LYSO scintillators were 6 mm thick each (Fig. 2).

For continuous (monolithic) crystal designs, several DOI schemes have also been proposed [26–28]. In continuous crystals, it is possible to determine the DOI by determining the width of the light-distribution. The authors have successfully developed and implemented a novel design for an inexpensive DOI system [29] based on the estimation of the width of the light-distribution. For the monolithic approach of the MINDView PET module, we have considered a 2 cm thick LYSO scintillator (Proteus, Ohio, US), whose base matches the photosensor area of approximately 5 cm × 5 cm too (Fig. 2). The crystal was entirely painted black (except the surface coupled to the SiPM array).

2.2. Photosensors

Due to the incompatibility of PMTs with magnetic fields such as those presented in MRI systems, SiPMs have been suggested to replace PMTs in the design of PET systems. SiPMs are also proposed for Time of Flight applications due to their fast response. However, there are not many developments using large area arrays of SiPMs achieving good performances. In this paper we have used SiPM photosensor arrays of the type MicroFB-30035-SMT (SensL, Cork, Ireland) for the staggered crystals approach and the MicroFC-30035-SMT [30] (SensL, Cork, Ireland) for the monolithic based detector. The MicroFC SiPMs improve the dark count rates from 744 kHz/mm² (MicroFB) to 30 kHz/mm². These SiPMs have an active area of 3 × 3 mm² with a total outside dimension of 4 × 4 mm².

The SiPM array active area covers 50.2 mm × 50.2 mm and is optically coupled to the LYSO crystal by silicone optical grease BC-630 from Saint Gobain (refractive index:1.47). In order to avoid SiPM signal degradation and instability, thermal control is envisaged by forced air cooling, assuring a stabilized temperature of about 20 °C within the photosensor area. Initial tests with dried cold air generated in vortex tubes have shown to efficiently keep this temperature with oscillations below 0.5 °C (Fig. 3) when the room temperature is about 26 °C. Without cooling, the detector block temperature would increase to about 45 °C.

2.3. Readout electronics

Two approaches were originally considered for the PET readout electronics. One is based on a custom designed
Application Specific Integrated Circuit (ASIC), in which each SiPM array is controlled by means of 3 scalable ASICs with up to 8 outputs [31–33]. However, we recently found that because the ASICs linearly combine all the SiPM signals to return each output, this plays against accurate spatial resolution results. The second approach, (selected for this project), makes use of a special charge division network providing information for each row and column output (R&C) of the SiPM array. The signals of all 12 SiPMs belonging to every row or column of the SiPM array are summed and later amplified before digitalization. In these measurements, particular charge division networks were used. A diode-based readout (AiT, Newport News, VA, USA) with diodes placed between the photodetector and a fast transimpedance amplifier was used for the MicroFB type SiPM array. When using the MicroFC array coupled to the monolithic crystal a resistive readout was used, alternatively to the diodes-based readout. Fig. 4 depicts the principal idea of these readouts. The number of signals to be digitized is 24 (12 rows + 12 columns). These 24 signals in addition to the summed trigger signal are transferred to the data acquisition (DAQ) system by means of coaxial cables. The 24 row and column signals are digitized with custom ADC electronic boards with 12 bit precision, capable of processing up to 66 channels (i.e. two detectors can be connected to a single ADC board). A main trigger board with programmable coincidence windows (3, 5, 7 and 9 ns) manages the detectors whose signals are to be digitized. Each board is equipped with a gigabit Ethernet input/output which connects to a 10 GB switch.

2.4. Description of simulations

The expected performance of the approach using monolithic 20 mm thick crystals was evaluated using GATE [34,35] Monte Carlo simulations (“Standard” model of GEANT4.9.5 [36]), including the definition of the most suitable geometrical and readout configuration for overall PET performance optimization. 20 LYSO detector blocks of 50 mm × 50 mm × 20 mm were considered per each of the three rings with an aperture of 330 mm and an axial coverage of 150 mm. An energy resolution of 15% and a coincidence window of 5 ns were also considered. The spatial resolution and sensitivity performance are investigated and compared based on a combination of the industry accepted NEMA [37] standards. 3D images were reconstructed from the simulated datasets using a filtered back projection (FBP) algorithm. The reconstructed image size was $33 \times 33 \times 33$ voxels with voxel size of $0.3 \times 0.3 \times 0.3$ mm$^3$.

The RF coil was not included in the simulations. It is not expected to have any effect in terms of the resolution performance of the PET detector modules and only minor effects on the sensitivity performance considering the minimum surface area within the active PET ring that the RF coil occupies.

The spatial resolution was evaluated using an $^{18}$F source placed at the center of the axial FOV but at two different transaxial positions (1 cm and 10 cm off-center). The point source was enclosed within a sphere of 8 mm in diameter of PMMA. The number of random coincidences in every acquisition was less than 5% of the number of prompt coincidences and $2 \times 10^6$ coincidences were simulated at each source position. A 350–650 keV energy window and 5 ns timing window were used for data acquisition. The spatial resolution on each position was subsequently calculated by determining the full width at half maximum (FWHM) of the resulting point spread function obtained by interpolation between adjacent pixels in the radial, tangential and axial profiles, after Gaussian curve fitting.

The sensitivity of a scanner represents its ability to detect the original positron–electron annihilations taking place in the target volume. In the NU2-2001, the absolute sensitivity of a scanner is expressed as the rate of detected coincidence events in counts per second (cps) for a given source activity, in MBq. Therefore, a sensitivity phantom consisting of a 700 mm long line source surrounded by five concentric stacked aluminum sleeves of 0.25 cm thickness was used. This phantom was placed at the centre of the scanner using only the line source and the first aluminum tube. The number of metal sleeves around the source was progressively increased in order to amplify the photon attenuation and therefore to
allow investigation of the system sensitivity. The simulated line source was uniformly filled with $^{18}$F with an activity of 1.44 MBq. An energy window of 410–650 keV was used. The time acquisition was chosen to ensure that at least 10,000 true coincidences per slice were collected. Two simulations were performed for each aluminum sleeve, one at the centre and one at 10 cm from the centre of the transaxial FOV.

3. Results

3.1. PET insert simulations

The monolithic design is the one expected to provide the best compromise between resolution, sensitivity and cost in terms of the individual detector performance. Therefore only results of the complete system simulations in terms of resolution and sensitivity for this design are shown.

The simulations resulted in a spatial resolution (FWHM) at 10 mm off-center in the transverse direction of 1.42 mm, and 1.75 mm for the axial direction, assuming a DOI resolution of 6 mm. These values increase up to 2.79 mm (radial) and 1.87 mm, at 100 mm off-center. The sensitivity of the system without absorbers, called absolute sensitivity, was 3.55% and 4.32%, at 0 and 10 cm, respectively.

3.2. Radio frequency coil design

In order to protect the electronic readout of the PET system against the harsh RF environment from the MRI system signals, the PET modules will be equipped with the RF shield. An open RF coil design would minimize the afore-mentioned claustrophobia. Possible open designs, like shortening the RF shield of the RF resonator, have been investigated using CST Microwave Studio 2013 electromagnetic simulation package. These simulations have shown that assuming a coil diameter of 270 mm, a minimum region of 30 mm from the RF coil rungs to the PET shield surface has to be left free (Fig. 5).

Reducing this distance to 20 mm, the B1 field strength is reduced to about 50% (see Fig. 6). Furthermore, the simulations predict that it is possible to use a shortened shield, although this will lead to a strong coupling with the environment.

3.3. Detector module performance

In this section we show the detector module performance obtained for the different configurations considered for the brain PET scanner. The detector assembly was run at a stable temperature of 21 °C and at 30.5 V bias voltage (~6 V overvoltage), with 170 ns wide integration gate to ADC, and DC coupling of the signal.

Measurements were carried out using a $^{22}$Na point source placed away from the detector. For the three staggered layers and R&C readout electronics, the measured energy resolutions as a function of the layer were about 11.8%, 9.6% and 10.2%, for the first, second and third layers respectively. These values are the average of measurements carried out at different positions of the detector block, including the edges, with a maximum variation of 1%. Fig. 7 shows the energy spectra for one pixel in each layer. A second peak is clearly visible in the spectrum for the third crystal layer, and starts to also be visible in the central one and is due to scatter events in the entrance scintillation layer 1 (and/or 2) followed by absorption of the scattered photon in layer 2 (and/or 3). The image obtained with this stack is shown in Fig. 8, where the three layers are clearly resolved, thus allowing a DOI resolution equivalent to layer thickness, i.e. 6 mm.

In contrast to the multiple-crystal-arrays approach, a pilot study was carried out using the monolithic 2 cm thick crystal with the readout providing both X and Y light distribution projections. These projections are processed using autocorrelation and rise-to-the-power [38,39] approaches to determine the planar impact position as well as the DOI. This method has allowed us to estimate the photon impact DOI using N/I (integral of the light distribution projections over its maximum value) as the DOI estimator. The sigma of the DOI resolution is a free parameter ($\sigma_{\text{int}}$) in the following equation [25] used to characterize the distribution of DOIs:

$$\text{DOI}(z) = A e^{-\alpha z} \left[ \text{Erf} \left( \frac{b-z}{\sqrt{2\sigma_{\text{int}}}} \right) - \text{Erf} \left( \frac{a-z}{\sqrt{2\sigma_{\text{int}}}} \right) \right]$$

where $\alpha$ is the attenuation coefficient, $A$ is a normalization factor, and $a$ and $b$ are the lower and upper limits of the distribution. The estimated sigma of the DOI resolution is about 1.7 mm and remains constant for most of the detector block regions. In Fig. 9 we show a DOI distribution measured for a centered ROI.

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**Fig. 5.** MINDView geometry simulation: RF coil of 36 cm and 27 cm, axial length and diameter, respectively, with twenty 4 mm wide rungs. The copper shield (35 μm thick) has a diameter of 33 cm and 20 cm axial length. Coronal (left) and transversal (right) view of the B1 strength variation.
Fig. 6. Simulation results for the B1 field. The plots show the field intensity in the axial direction as a function of the distance, for a rung to shield distance of 30 mm (left) and 20 mm (right).

Fig. 7. Energy spectra for single pixels in the first crystal array layer (top-left), in the second (bottom-left) and in the third (bottom-right). The energy profiles in layers 2 and 3 also show a smaller distribution at lower photopeak energies due to scatter events on the upper layers.
Fig. 10 right shows the detector spatial resolution as a function of four depth of interaction layers. In this case a $9 \times 9 \, ^{22}\text{Na}$ sources array with 5 mm pitch, collimated through 24 mm tungsten was used. The depicted FWHM values correspond to the central row of sources. The achieved DOI accuracy facilitates the separation of the scintillator block in at least these 4 virtual slabs. The worst resolution is observed in the entrance region, with an image quality improvement towards the region closest to the photosensor array. These images include both electronic collimation and $^{22}\text{Na}$ source spread function correction (FWHM$_{\text{coll}} \sim 0.9$ mm). The intrinsic spatial resolution depends on the DOI layer, but improves from about $2.3 \pm 0.7$ mm (average) in the entrance layer to $1.1 \pm 0.1$ mm in the deepest DOI layer. At the crystal center, the values for the spatial resolution have a less significant variation being 1.7, 1.4, 1.1 and 1.1 mm, for DOI1, DOI2, DOI3 and DOI4, respectively. The energy resolution performance of this detector block has also been determined as a function of the DOI region. In Fig. 11 we have plotted the energy spectra for a centered ROI obtained using a $^{22}\text{Na}$ source located in front of the detector without collimation versus the DOI layer. We observe an improvement of the energy resolution towards deeper DOI impacts since a larger fraction of the scintillation light is transferred to the photosensor array.

### 4. Discussion and conclusions

The aim of the MINDView project is to develop a brain imaging system with the capability of simultaneous acquisition of very high resolution PET and MRI images.

For the RF coil, 20 RF rungs with a bore of 270 mm in diameter and 270 mm length are envisaged. Simulations suggest a safe distance of 30 mm between the two conductors, RF coil and its shielding, thus ensuring a good homogeneity and sufficient amplitude of the B1 field in the system field of view.

The test results of a system based on R&C projections showed good results for both the monolithic and the 3 staggered pixelated array approach. Obviously, the cost is to digitize and to manage 24 signals/event in the R&C approach instead of 5 for the formerly considered ASIC configuration. Multiplexing approaches to reduce the number of readout channels for the whole scanner are under consideration, for example, by combining column and row signals.
from the neighboring modules. The R&C approach also allows one to use alternative methods to determine the impact position of the incident photon, including DOI, based on the knowledge of the X and Y projections of the light distribution. In this way, the authors have developed a procedure to correct for the marked non-linearity and image compression at the edges produced when working with monolithic crystals due to the truncation of the light distribution towards the crystal borders [40,41].

Laboratory results with 3 staggered pixilated layers showed DOI and spatial resolution capabilities that would provide 1 mm final image resolution in the central brain region. However, cost for the 3 staggered pixelated arrays is significantly higher than

Fig. 10. Monolithic 20 mm thick and R&C readout approach. Left, flood images for four “virtual” crystal layers: DOI1 (entrance layer 20–15.5 mm), DOI2 (15.5–11 mm), DOI3 (11–6.5 mm) and DOI4 (6.5–2 mm). As especially visible in DOI layers 2 and 3, these images also show some compression effect. Right, average spatial resolution FWHM for the central row as a function of the DOI and the source position.

Fig. 11. Energy spectra for a centered ROI obtained with a $^{22}$Na source placed in front of the scintillation crystal as a function of the DOI region.
that of the monolithic crystal approach. Moreover, our pilot results show that monolithic crystals combined with novel methods (correlation) to determine the 3D photon impact coordinates could provide sufficient intrinsic resolution at a reasonable cost. These results suggest a detector module for the MINDView brain PET scanner based on a 20 mm thick monolithic crystal with R&C readout approach that will allow an intrinsic spatial resolution of about 1.5 mm on average, enabling discrete depth of interaction readout approach that will allow an intrinsic spatial resolution of about 1.5 mm on average, enabling discrete depth of interaction.

New photosensors are envisaged for this project free of ferromagnetic materials such as Nickel, and of the same active area dimensions. We are currently carrying out experiments with the monolithic blocks but having all surfaces, except the one in contact to the photosensor, covered with ESR reflector films. Preliminary measurements have shown an average energy resolution improvement to 15%. A slightly larger edge effect is in general observed but its dependence with the DOI layer seems to be smaller when compared to the black painted approach.

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