Optimization of a Monolithic Detector Block Design for a Prototype Human Brain PET Scanner

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Abstract– We are presently developing a novel PET scanner for human brain functional imaging based on monolithic scintillator crystals read by APD matrices, capable of being inserted into an MRI system. In this work we report on the detailed study that has been made of the design of the detector blocks, aiming at defining the most suitable geometrical and readout configuration for optimizing the overall performance of the entire scanner. Both parallel and trapezoidal geometries have been simulated, featuring two layers of active scintillator material with different or similar thickness and APD readout on the front or back side. Results of this study indicate that a trapezoidal geometry with equal thickness of both layers is the best solution for improving the expected scanner performance.

I. INTRODUCTION

POSITRON emission tomography (PET) is a powerful tool for the imaging of organ function, allowing highly specific metabolic processes at cellular level to be studied and quantified in a growing variety of clinical applications [1].

Over the last years there has been a clear predominance in the use of PET/CT systems over PET-only scanners, based on the advantages of simultaneous anatomical image co-registration and attenuation corrections from the addition of CT to PET, although at the cost of an increase of dose delivered to the patient (up to 20 mSv for a whole body CT scan) [2].

Recent developments in PET detector technology focus on hybrid PET-MRI systems [3] arguing that, compared to CT, MRI provides better soft tissue contrast, no additional dose to the patient and the possibility of attenuation correction based on appropriate MRI sequences or segmentation techniques [4]; moreover, a strong magnetic field could improve PET spatial resolution by confining the path of the positron before annihilation [5].

A clinical field of particular interest for PET/MRI imaging is neurology, profiting from the advantages of both techniques: MRI's soft tissue discernment and PET's growing availability of highly specific radio-pharmaceuticals for the study of brain and central neural system function.

With this application in mind, we are building a prototype PET scanner dedicated to the functional imaging of the human

brain, based on a detector technology compatible with strong magnetic fields [6]. The novelty of our design consists on using monolithic detector blocks instead of pixilated ones, on the arguments of larger active volume and simpler mechanics for similar or better energy and spatial resolutions at detector level [7]. In this work we present the results of the optimization study of the detector block design in order to achieve the best possible overall scanner performance.

II. MONOLITHIC DETECTOR BLOCKS

Monolithic detector blocks present several advantages when compared to pixilated designs. In particular they have a larger active fraction of the total detection volume and a lower manufacturing cost. Moreover, they are capable of providing the coordinates of incidence of a photon in a continuous way with spatial resolutions of the order of 2 mm or better.

Two types of geometries have been studied in this work: single block detectors and full-ring scanners composed of many identical blocks in a modular design.

A. Single detector block geometries

Each detector block considered in this work is formed by two independent LYSO:Ce monolithic crystals disposed in two layers, designated as inner and outer with respect to the main axis of a cylindrical PET scanner. The scintillation light generated by ionization in each crystal is collected by a pair of Hamamatsu S8550-02 APD arrays placed side by side, corresponding to an 8x8 matrix of 64 pixels on the readout surface of the crystal. A dedicated front-end ASIC, described elsewhere [8], reads the 64 pixels and provides the sums of columns and rows as input for the position-determining neural network (NN) algorithms [9]. Based on these general block specifications, several particular cases have been studied for two block shapes (trapezoidal or parallelepiped), two sets of inner and outer layer crystal thickness (07/13 mm or 10/10 mm, summing a total of 20 mm) and four possible sides of APD readout placement (in the front or back of the crystals, with respect to the direction of incoming radiation). An example of a set of the geometries simulated is shown in Fig.1 for trapezoidal blocks with 7 and 13 mm thick layers.

B. Full-ring PET scanner geometries

The global geometrical parameters of the complete scanner were determined by fixing an inner diameter of 40 cm for the detector rings and setting the area of the innermost face of the blocks as being equal to two APD matrices side by side, i.e.

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22.4 mm $\times$ 19.5 mm. As a consequence, the maximum number of detector blocks per ring is 52. Given our available resources we will build a scanner with 4 such rings. An example of a full ring complete geometry is shown on Fig.2 for the FB configuration of the trapezoidal blocks presented in Fig.1.

Fig. 1. An example of the four APD readout configurations studied shown for trapezoidal scintillators with 7 and 13 mm thickness. The inner layer (at the bottom) is the one closest to the axis of the PET scanner.

Fig. 2. Three-dimensional rendering of a full-ring scanner with Front-Back (FB) readout and 07/13 mm thick crystals. The global (left) and local (right) reference axis are also represented.

III. MATERIALS AND METHODS

The performance of the detector blocks has been evaluated through detailed Monte Carlo simulations of single blocks and complete scanners composed of many identical blocks. The following figures of merit have been considered: spatial resolution and linearity of single blocks; sensitivity, single event and PET coincidence counting rate capabilities of complete scanners. All simulations of detectors and physics were done using the GAMOS (Geant-4 Architecture for Medicine-Oriented Simulations) framework [10] based on the Geant4 toolkit [11]. Scintillation photons in LYSO:Ce crystals were generated with a light yield of 32000 ph/MeV as specified in the manufacturer's datasheet. Optical photons were tracked according to the UNIFIED model [12] implemented in Geant4. The natural radioactivity of LYSO:Ce due to the presence of ^{176}Lu was also included in some of the simulations. The details of the simulation strategies used are described in the following paragraphs.

A. Single detector block studies

The performance of individual blocks was studied using pencil-like beams of 511 keV photons with perpendicular and up to 15° slanted incidence on the bottom face of the detector blocks, mapping its whole volume and generating optical photon data for training and testing of the NN algorithms. The output of each simulation job consisted in the number of

optical photons reaching each APD pixel in a block for all incident gamma interactions depositing more than 350 keV in a given crystal layer. If an incident gamma interacts on both layers of a given detector block, each layer is considered as an independent detector and the event is good if and only if the above condition is satisfied, that is, if the energy deposited in one of the crystals is above 350 keV.

The operation of the APDs was modeled in terms of translating the number of incident optical photons into an electrical charge signal taking into account APD quantum efficiency (0.75 at 420 nm wavelength), gain (50 or 100) and excess noise factor (1.75). Electronics noise was added to the results using either 300 or 600 electrons RMS per readout channel to evaluate the influence of front-end electronics noise in the overall behavior of the detectors.

For all single events acquired, the position of entrance of a gamma at the surface of the detector was estimated using two NNs, one for each orthogonal direction $\mathbf{e}_t$ and $\mathbf{e}_z$ (see Fig.2 for definitions of axis). Each of the NN is composed of 8 inputs and 2 hidden layers of 5 neurons each, previously trained for a given incidence angle according to the procedure described below. It should be emphasized that using this approach there is no need to determine the depth of interaction (DoI) as in standard PET detector modules: the point of entrance of the incoming gamma is referred to the inner surface of the monolithic block thus defining a unique point in 3D space with two orthogonal coordinates along the surface of the block ($\mathbf{e}_t$ and $\mathbf{e}_z$) and a radial coordinate ($\mathbf{e}_r$) corresponding to the distance from the surface to the center of the FoV (see Fig.2 for details).

For each direction of incidence and axis along the detector, five NNs were trained using the randomly shuffled available simulated data in the following way: 50% for training (up to 30 iterations), 15% for testing and 35% for evaluation. Each of the 8 NN inputs was fed with the profiles of the charge collected on the APDs, obtained from summing the data from the 64 pixels in 8 rows and 8 columns, one set for each axis, event by event. The output from the five training and evaluation procedures were finally used to estimate the variance of the results.

The accuracy of the position-determining NN algorithms was evaluated by fitting the error distributions of incidence points along the two orthogonal axis over the inner surface of the detector blocks and expressing the results in full-width at half-maximum (FWHM) and tenth-maximum (FWTM), as well as linearity. The fits were performed following a NEMA-like approach, adjusting a second-order polynomial through the three highest points of the error distributions, using the maximum of the polynomial to find the half-height and calculating the FWHM by linear interpolation on the lateral slopes of the distribution.

B. Full-ring PET scanner studies

The single detector block configurations that performed best were chosen for the simulation of full ring scanner geometries

composed of many identical such blocks. All these simulations were done without optical photon generation and tracking in order to allow an evaluation of the overall efficiency and expected counting rates (single and coincidence events) taking into account only the energies deposited in each LYSO:Ce crystal. A good single event in a detector block was defined whenever there is a deposition of energy between 350 keV and 650 keV on one of the two LYSO:Ce crystals of that detector, whereas a PET event was defined as a coincidence of two good single events inside a 10 ns time window. No APD modeling and no NN training and evaluation were used in any of these studies.

Two different operational scenarios were considered: one with a point source of ^{18}F at the center of the field of view (FoV) for efficiency studies, and one emulating realistic clinical conditions taking into account ^{18}F -FDG uptake in the brain and the heart, as well as natural radioactivity from ^{176}Lu present in the LYSO:Ce crystals. For the efficiency studies a point-like activity of 100 MBq (2.7 mCi) of ^{18}F centered inside a 1.0 mm radius water sphere placed at the center of the FoV was used. The brain and the heart phantoms were defined as ellipsoids with average adult organ dimensions along each main axis filled with a uniform aqueous solution of ^{18}F -FDG. The activity in each phantom was specified by considering a total administered dose of 740 MBq (20 mCi) and 5% physiological uptake in each organ, resulting in a total activity of 37 MBq (1 mCi) for each of the phantoms. The presence of ^{176}Lu in the LYSO:Ce crystals was modeled by considering an average activity of 300 Bq/cm³ for the scintillating crystal material.

IV. RESULTS

The results obtained in this work are presented in the following paragraphs for single detector blocks and full ring scanners.

A. Single detector blocks

The geometrical definition of the detector blocks was studied in terms of shape (parallelepiped or trapezoidal), inner and outer layer crystal thickness (07/13 mm or 10/10 mm) and APD readout placement with 4 possible combinations from “Front” (F) or “Back” (B) readout of each layer.

Regarding the shape, a trapezoidal geometry corresponds to a larger active volume and, therefore, to a higher overall probability of interaction of up to 10% when compared to a parallelepiped shape. The main drawback comes from a loss of spatial resolution closer to the edges associated with non-linearity effects, both which are more pronounced for the outer layer crystals. In any case, having extra material at the slanted edges of a trapezoidal block results in an advantage compared to having a smaller active volume.

Concerning the thickness of each crystal layer, the 07/13 mm configuration yields similar probabilities of interaction for both layers whereas the 10/10 mm solution results in an approximate 2:1 ratio between interactions in the inner and the outer layer,

respectively, in good agreement with direct calculations of the interaction probabilities from the linear attenuation coefficient of LYSO:Ce at 511 keV. Moreover, it was observed that for the 07/13 mm configuration some APD readout configurations gave bad NN results for the outer layer crystal, with spatial resolutions worse than 2 mm FWHM and large non-linearity effects.

The more evident effect of the placement of the APD readout in the “Front” (F) or “Back” (B) of the LYSO:Ce blocks is the enhancement of Compton scattering on passive materials, most importantly on the ceramic encasing of the APDs when located before the active crystals. All those configurations with “Back” readout have higher interaction rates for singles with more than 350 keV deposited energy while the worst behaving under these conditions is FF (“Front-Front”).

The placement of the APDs has a tangible effect on the overall NN performance, with similar results for 7 or 10 mm thick layers with “Front” or “Back” readout but with significantly worse results for 13 mm thick “Back” readout, probably due to the larger light spread in a thicker crystal before reaching the APD pixels when compared to a thinner one. Apart from the configurations with 13 mm “Back” readout, all detector blocks fulfill the conditions of spatial resolution better than 2 mm FWHM and 5 mm FWTM (from the fits to the distributions of errors in the position accuracy profiles). A full set of results for the “Front-Back” (FB) trapezoidal configuration with two layers of 10 mm thick crystals is shown as an example in Fig. 3 and Fig.4 for, respectively, perpendicular and 15° incidence.

Finally, it was observed that, in general, reducing the electronic noise and increasing the APD gain leads to better overall FWHM and FWTM values on all configurations.

B. Full-ring scanners

Following the results of the previous section, only the trapezoidal detector block configurations were considered in the simulations of the full rings.

Regarding the first simulated scenario, the average sensitivity found for a 100 MBq (2.7 mCi) ^{18}F point source located at the center of the FoV was 3.1% for all APD readout schemes except FF “Front-Front”, which was smaller to the enhanced Compton scattering on passive materials previously observed for that particular configuration. Moreover, it was observed that having both detector layers of equal thickness (10 mm) resulted in an overall increase of 12% in PET coincidence rates compared to having 7 and 13 mm thick layers, an effect that could be explained by the fact that a thicker inner layer has a higher probability for photoelectric absorption of a Compton scattered gamma than a thinner one.

Concerning the scenario of scanner operation under clinical conditions, several estimations of counting rates were obtained for 37 MBq (1 mCi) of ^{18}F -FDG distributed in a uniform human brain phantom. The expected single event rates per detector block were found to be between 14 and 42 kHz for all interactions (maximum values for outer and inner layer,

respectively) and below 20 kHz for good single events with energy deposited between 350 and 650 keV, whereas the expected PET coincidence rates for a full scanner with 4 rings were of the order of 400 kHz. The cumulative effects of ^{18}F -FDG uptake in the heart and the natural radioactivity of the LYSO:Ce crystals increment the single event rates by up to 12% with less than 3% variation in the PET coincidence rates.

Fig. 3. Spatial resolution and linearity performance of NN algorithms for a trapezoidal 10/10 mm detector block with “Front-Back” APD readout. Data points refer to spatial resolution FWHM (blue) and FWTM (red) as well as linearity (green), for the inner (top row) and outer (bottom row) layers along e_t and e_z axis at perpendicular incidence. Error bars represent the variance from five independent NN training and evaluation procedures. Values in mm.

Fig. 4. Spatial resolution and linearity performance of NN algorithms for a trapezoidal 10/10 mm detector block with “Front-Back” APD readout. Data points refer to spatial resolution FWHM (blue) and FWTM (red) as well as linearity (green), for the inner (top row) and outer (bottom row) layers along e_t and e_z axis at 15° slanted incidence. Error bars represent the variance from five independent NN training and evaluation procedures. Values in mm.

V. CONCLUSIONS

We have performed a Monte Carlo optimization study of the monolithic detector blocks for our prototype human brain PET

scanner. The main conclusions can be summarized as follows: trapezoidal blocks lead to higher overall efficiency with no significant degradation on NN algorithms close to the crystal edges; the thickness of both crystal layers should be 10 mm; APD readout configuration “Front-Front” should be discarded. With all the results taken into account, the final decision was made on a trapezoidal 10/10 mm block with “Front-Back” APD readout, which apart from its overall behavior also represents a good solution from the electromechanical point of view. Finally, the combined results from the full ring scanner simulations were used to specify the bandwidth requirements of the readout electronics architecture of the prototype human brain PET scanner presently under development at CIEMAT.

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