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Construction and tests of demonstrator modules for a 3-D axial PET system for brain or small animal imaging

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ABSTRACT

The design and construction of a PET camera module with high sensitivity, full 3-D spatial reconstruction and very good energy resolution is presented. The basic principle consists of an axial arrangement of long scintillation crystals around the Field Of View (FOV), providing a measurement of the transverse coordinates of the interacting 511 keV gamma ray. On top of each layer of crystals, an array of Wave-Length Shifter (WLS) strips, which collect the light leaving the crystals sideways, is positioned orthogonal to the crystal direction. The signals in the WLS strips allow a precise measurement of the z (axial) co-ordinate of the 511 keV γ-ray gamma impact. The construction of two modules used for demonstration of the concept is described. First preliminary results on spatial and energy resolution from one full module will be shown.

1. Introduction

PET is generally recognized as the best available molecular non-invasive diagnosis technique sensitive to very low tracer concentrations. It provides access to metabolic and kinetic parameters of a particular molecular process, thus allowing early detection of diseases and prompt follow-up in treatment [1].

The performance of most commercial PET Scanners, both for preclinical studies and for clinical applications, which are presently in regular operation in laboratories and hospitals, are still limiting factors in image quality, and thereby in the interpretation of the results and call for innovation and improvements [1–7].

Important instrumentation limitations in most present day PET scanners are:

- Non-uniform spatial resolution in the detector over the whole Field Of View (FOV) due to Depth Of Interaction (DOI) uncertainty in the scintillation detector.
- Relatively low efficiency of photon conversion due to the correlation between radial thickness of the scintillation crystals and DOI smearing of radial interaction co-ordinate.
- Limited capability to recognize and reject Compton interaction (cascade events) in the scintillation crystals, which lead to smearing of the measurement of the interaction point.

Medical requirements of optimal PET scanning are accurate observation and precise location of small structures e.g. in brain PET or precise location of the position of malignant tumors before hadron therapy treatment (In-Beam PET). Similarly, small animal PET preclinical studies are more and more focused on detecting very small structures.

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This requires very good spatial resolution in the detector close to the limits given by positron range and non-co-linearity of the two back to back photons [8].

Another very important improvement, which is called for, is to have higher sensitivity at lower total dose than obtained with present instruments. In small animal PET, a longer survival of the rodent during an investigation is an important factor, whereas in clinical PET low dose at increased sensitivity is crucial to allow e.g. breast screening using PET.

The other driving force in instrumentation innovation for PET is the possibility of co-registration with other precise morphological imaging modalities like the X-Ray Computed Tomography (CT) or Magnetic Resonance Imaging (MRI).

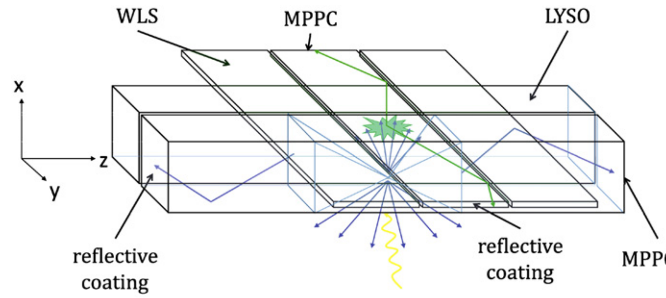

Fig. 2. The principle of an axial coordinate readout using WLS strips.

2. The axial PET concept with wavelength shifter (WLS) strips for an axial co-ordinate measurement

2.1. The concept

The concept of using axially arranged long scintillation crystals with readout on both sides has been considered already more than 25 years ago [9,10]. Previous development work in this collaboration to implement this concept was based on using 10 cm long LYSO crystal bars arranged in axial stacks and readout on both sides by Hybrid Photon Detectors (HPD). The axial co-ordinate was determined from the difference of pulse heights measured on both sides of the long crystal [1,12]. Experimental studies showed that the axial coordinate resolution in this approach could at best be ~6 mm FWHM.

Another approach was proposed in this Collaboration to measure the axial co-ordinate using arrays of WLS strips placed orthogonally in between layers of LYSO crystal bars as shown in Fig. 1.

Part of the photons produced by an interaction of a 511 keV photon in an LYSO crystal bar are emitted within the angle of total reflection on the crystal bar walls and travel to the ends of the LYSO bar, where they are detected by a suitable photo-detector. Photons traveling in the opposite direction are sent back to the photo-detector by a reflector (Fig. 2).

The photons which are emitted outside the forward cone of total reflection, towards the side of the crystal bar, where WLS strips are placed, leave the crystal bar sideways and enter some of the WLS strips. Choosing adequately the optical properties of the WLS material, these photons can be efficiently absorbed and wave-length shifted into a wavelength domain with little absorption in this material. A fraction of these wave-length shifted photons are transported, again by total reflection, to the end of the WLS strips.

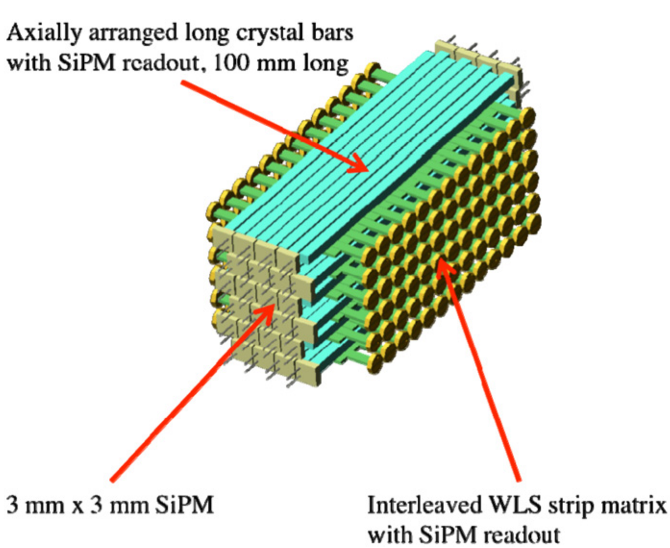

Fig. 1. The concept of an axial PET module with long LYSO scintillation crystal bars and WLS strips. The photo-detectors are silicon photo multipliers.

Fast photo-detectors mounted at the end of the LYSO crystal bars and the WLS strips are used to measure the amplitudes of the light pulses, and thereby the energy deposited in the LYSO bars and the light yield in the WLS strips. The x and y coordinates of each hit LYSO crystal is given by its position in the stack with digital resolution $\sigma_{x,y}=d/\text{sqrt}(12)$ (d is the transverse dimension of the crystal bar). The same applies for all WLS strips which absorb photons from the gamma interaction in the LYSO crystal and re-emit wavelength shifted photons. Since it is expected that in general more than one WLS strip has a detectable signal, the z-coordinate measurement can be performed either by using just the strip with the highest signal or performing an analog interpolation using several strips in a cluster.

In previous measurements [11–15], it has been demonstrated that the light yield from WLS strips is sufficiently big to allow observation in WLS strips of 511 keV photo-absorption events and also recoil electrons from Compton interactions down to energies of about 50 keV. Energy deposition from Compton interactions below 50 keV can still be observed in the LYSO crystal stack. These interactions, having no axial coordinate information, can be removed from the event sample for the image reconstruction.

2.2. Main performance improvements expected from the axial PET concept

One big advantage of this concept is a full 3-D reconstruction of the impact of the interacting 511 keV gamma ray, both for full photo-absorption and cascade interactions. The detector spatial resolution is independent of the origin of the e^+e^- annihilation in the FOV. The spatial resolution in x, y and z in the detector module is simply determined by the choice of the transverse dimensions of the LYSO crystal bars and the WLS strips, and can be chosen to give the best results for a given application. Sensitivity and spatial resolution in the detector are to a high degree uncorrelated, since the number of layers in the direction of the incoming 511 keV photons, determining the sensitivity, can be chosen without affecting the spatial resolution in the x-, y- and z-coordinates, in contrast to conventional radial geometry, where DOI uncertainty increases with the length of the crystal bars.

Photo absorption and Compton cascade events can be fully identified. Part of the cascade events can be used in the image reconstruction, thus increasing the overall sensitivity. Finally using Geiger mode APD arrays (G-APDs) for the readout of the LYSO and WLS strips opens the opportunity of co-registration with an MRI [16].