J-PET: A Novel TOF -PET scanner using Organic Scintillators

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ABSTRACT

Positron Emission Tomography (PET) is one of the most advanced nuclear medicine imaging techniques that have potential to identify many diseases (like cancers, heart diseases, neurological disorders and other abnormalities) in vivo in the earliest stages. However, production of PET modalities for covering the whole human body is economically unrealistic when applying the current technologies. In order to achieve a goal of more economical PET scanner with large geometrical acceptance and improved time resolution, the Jagiellonian Positron Emission Tomography (J-PET) Collaboration is realizing a new project aiming at construction of TOF-PET detector using plastic scintillators instead of crystals. Novelty of the J-PET scanner lies in: (i) application of plastic scintillators as well as in (ii) its front-end electronics which allows signal sampling in voltage domain, (iii) a trigger-less data acquisition system, and (iv) the new time and hit-position reconstruction methods. Moreover, the proposed solution enables to increase the axial field-of-view of the tomograph by extending the length of the plastic scintillator strips without changing the number of photomultipliers and electronic channels.

KEY WORDS: J-PET detector, plastic scinitllators, TOF resolution, axial resolution.

1. INTRODUCTION

Currently, Positron Emission Tomography plays an important role in medical diagnostics as well as in monitoring therapy effects particularly in oncology, cardiology, neurology, gastrology and psychiatry. Detailed description of the PET method and its applications can be found. The cost of PET scanners is about few million Euro. One of the reasons which makes scanners so expensive is the application of inorganic scintillator crystals used as the radiation detectors.

In order to decrease the costs and enable a more cost effective production of the whole-body PET, an application of another kind of detectors were proposed. For example straw tubes drift chambers or large area Resistive Plate Chambers (RPC). In this article we describe a new method of Positron Emission Tomography aiming at reduction of production costs and at a cost effective increase of the axial field-of-view by the application of plastic scintillators. In the article detection technique used in current PET modalities along with the new concept of J-PET tomograph is presented.

Current and new Solutions:

Current status: At present all commercial PET scanners use BGO (GE Healthcare), LSO (Siemens) or LYSO (Philips) as a radiation detectors. In all current scanners photoelectric interaction of gamma quanta in crystals is used. Interaction of gamma quanta with the scintillator cause the transfer of their energy to electron, which in turn, ionize and excite the atoms or molecules inducing the light flashes. Scintillation crystals used in current PET scanners are usually the blocks of about 5 x 5 x 2.5 cm^3 dimensions. In addition they are divided into smaller sub-crystals and their size ranges from 4 x 4 x 20 mm^3 of LSO to 6.3 x 6.3 x 30 mm^3 of BGO for the PET/CT scanners. These sub-crystals are coupled to photomultiplier tubes larger than their size and hit-position of gamma quanta is estimated by the amplitude distribution of the light pulses with spatial accuracy equal to the size of small crystal element. Determination of Line-of-response (LOR) is based on Anger logic which limits the resolution of obtained image. In some recent PET designes small silicon digital photomultipliers (SiMPs) or avalanche photo-diodes are used in order to have one-to-one correspondence between photomultipliers and detectors.

The PET imaging has been substantially improved signal to noise ratio by introducing the concept of time-of-flight (TOF), time difference between the arrivals of gamma quanta to the detectors on both sides of a single LOR. PHILIPS made the first commercial TOF-PET scanner in 2006 using LYSO scintillator crystal and obtained 585 ps (FWHM) for time-of-flight resolution. With prototype constructed using LSO crystals SIEMENS achieved the time resolution of about 540 ps in 2008, which corresponds to the spatial resolution of about 8 cm along LOR. TOF-PET scanner built by GE using LSO crystals obtained a time resolution of 544 ps. But a more recent TOF-PET scanner

(Vereos) developed by PHILIPS in 2009 with improved electronics and new digital SiPM, achieved time resolutions of about 345 ps and spatial resolution of about 5.2 cm along LOR. As regards the scanning area, PET scanners built by SIEMENS and PHILIPS have detection rings of width of about 18 cm with internal diameter ranging from 70 cm to 90 cm which allow the scanning of the patient's body over a length of 190 cm. However, the whole body scan is done by moving the patient many times (~ 17 times) inside the tomograph. In order to scan the entire human body in a single attempt PET scanner with large field of view (FOV) is required. With the current solutions it is economically unrealistic to produce such a scanner for a common use because the number of crystals, photomultipliers and electronics modules increase linearly with increasing longitudinal field of view.

The J-PET tomograph: In order to achieve a goal of more economical PET scanner a new project to build a TOF-PET detector using plastic scintillators has started in 2011 at Jagiellonian University in Krakow. Novelty of the concept lies in employing predominantly the timing of signals to obtain the hit-position and hit-time of gamma quanta instead of using their amplitudes. Because of their low density (1.03 g/cm^3) and small atomic number so far plastics were not considered as a potential candidate to be applied in PET scanner design. Small density implies a small efficiency to detect a gamma quanta, while, small atomic number makes interaction of gamma quanta via photoelectric effect less probable or even negligible. Thus, we use a Compton scattering to detect the annihilation gamma quanta and take advantage of large acceptance and better TOF resolution. Moreover, an axial arrangement of scintillators in the J-PET tomograph enables to build a diagnostic chamber with several independent detection layers. In J-PET solution, the longitudinal extension of diagnostic chamber does not entail an increase in the number of photomultipliers and electronic channels. This feature, in contrast to crystal-based PET scanners, allows for building single-bed, whole-body PET scanners without significant increase of costs with respect to scanners with short AFOV. Furthermore, J-PET diagnostic chamber is free of any electronic devices and magnetic materials, thus it is possible to build a PET/MRI and PET/CT in a way different from the currently developed configurations.

One of the possible arrangements of strips is visualized schematically in Fig.1. Light signals from each strip are converted to electrical signals by two photomultipliers placed at opposite edges shown in Fig.2. Method of reconstructing the hit-time and hit-position in J-PET scanner, is different from the one used in current PET modalities. In case of J-PET modality distribution of light signals produced by interaction of gamma quanta via Compton scattering is continuous, as a consequence amplitude of signals varies even if they originate from the same interaction point. The hit- time and hit-position reconstruction method is based on comparison of detector signals with the library of synchronized model signals registered for a set of well-defined positions of scintillation points. The time of the interaction is determined as a relative time between the measured signal and the most similar one in the library. Schematic view of the method for TOF calculation using two strip J-PET module is shown in Fig.2. The module was built from two strips of plastic scintillators with dimensions 5 x 19 x 300 mm^3, optically connected at both sides to photomultipliers, from which signals were sampled by means of the Serial Data Analyzer with 100 ps intervals. We have obtained an axial resolution of about 1.0 cm (sigma) for hit position reconstruction and TOF resolution of about 330 ps (FWHM). This TOF resolution is better than resolutions so far achieved with crystal tomographs. The achieved results are promising and still there is a room for improvement as we have installed a dedicated front-end electronics which will allow us to sample the signals in voltage domain (using multi-threshold sampling) with the electronic time resolution below 20 ps. Furthermore, results of simulations have proven that using the array of SiPMs instead of vacuum tube photomultipliers can improve time resolution by a factor of about 1.5.

2. SUMMARY

In this contribution we have presented novel solution for the detector system with superior time resolution which may be applied in the Positron Emission Tomography. Especially promising is a possibility of producing long diagnostic chamber in a way which does not entail an increase in the number of photomultipliers, hence, enabling simultaneous imaging of the physiological processes throughout the whole body of the patient.

Furthermore, it is worth to stress that the J-PET tomograph is in fact a multi-purpose detector which allows to perform studies of the decays of positronium-atoms in the field of fundamental physics and in the field of life and material sciences. The positronium atoms inside the tomograph may be produced in a effective way using a sodium ^{22}Na isotope and the aerogel or polymer materials.

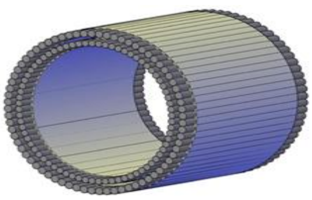

Fig.1. One of the possible arrangements of scintillation elements (rectangular bar in blue) and photomultipliers (cylindrical tubes in grey) for the diagnostic chambers of J-PET

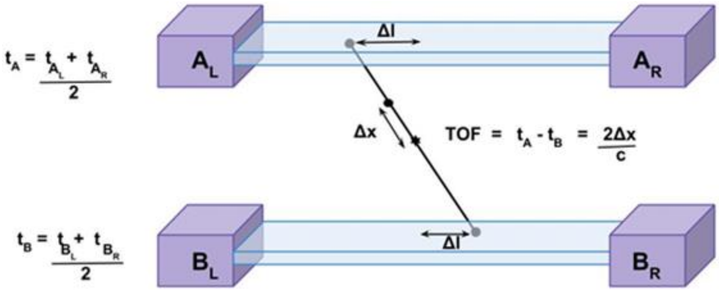

Fig.2. Schematic view of the method for TOF calculation

Determination of hit position versus the center of scintillator (Δl) is based on time difference measured on both sides of the scintillator strip (L and R), and the position (Δx) along the line-of-response is estimated from time difference measured between two modules. A_L and A_R is a pair of photomultipliers connected to left and right ends of scintillator A, respectively. Similarly, B_L and B_R are the photmultipliers connected to two ends of strip B.

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