J-PET: A Novel TOF-PET Detector based on Plastic Scintillators

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***Abstract*—The purpose of the reported research is the elaboration of the method for construction of the cost-effective whole-body single-bed positron emission tomography scanner enabling simultaneous PET/CT and PET/MR imaging. The Jagiellonian Positron Emission Tomograph (J-PET) is built out of 192 scintillator strips arranged axially in three layers forming a cylindrical diagnostic chamber with the diameter of 85 cm and axial field-of-view of 50 cm. The novelty of the concept lies in employing long strips of plastic scintillators instead of crystals as detectors of the annihilation quanta, and in using the timing of signals instead of their amplitudes for the reconstruction of Lines-of-Response. To take advantage of the superior timing properties of plastic scintillators a novel multi-voltage-threshold front-end electronics was developed allowing for sampling of signals in a voltage domain. An axial arrangement of long strips of plastic scintillators, and their small light attenuation allows us to make a TOF-PET scanner with a long axial field-of-view. The presented solution opens unique possibilities of combining PET with CT and PET with MRI for scanning the same part of a patient at the same time with both methods. The relative ease of the cost effective increase of the axial field-of-view makes the J-PET tomograph competitive with respect to the current commercial PET scanners as regards sensitivity and time resolution.**

I. Introduction

Positron emission tomography (PET) is well established medical diagnostics method. It is however very expensive [1], in part due to the very high costs of the currently used PET devices. Hence in practice PET imaging is available as a diagnostic tool only in the developed countries. The high cost of PET is also one of the barrier for production of this modality with the large axial field-of-view which would enable for the simultaneous imaging of the whole human body. The commercial PET scanners are based on the relatively expensive inorganic crystals read out by photomultipliers [2]. Thus one of the way to achieve a low-cost PET with large longitudinal field-of-view seems to be a replacement of crystals to another kind of detectors. For that purpose an utterly new PET concepts are being developed and tested. For example: (i) PET systems based on the modular drift chambers built out of the straw tubes [3], [4] or (ii) PET systems based on large area Resistive Plate Chambers (RPC) [5], [6]. In this article we present the first PET prototype in which inorganic crystals were replaced by the relatively inexpensive plastic scintillators arranged in a way allowing for the increase of the axial field-of-view without increasing of the number of photomultipliers. Moreover, in the presented prototype the reconstruction of the line-of-response and the event selection is based solely on the time measurements of photomultiplier's signals.

II. Materials and Methods

The Jagiellonian Positron Emission Tomograph (J-PET) is built from strips of organic scintillator, forming a cylinder (Fig. 1). Schematic view of the two detection modules is shown in Fig. 2. Light signals from each strip are converted to electrical signals by photomultipliers placed at opposite ends of the strip [7], [8]. The position and time of reaction of gamma quantum in the detector material is determined based on the time of arrival of light signals to the ends of the scintillator strips. The signals are probed at four voltage levels at the leading and at the trailing edges with the electronic accuracy of about 30 ps by a newly developed digital multi-voltage-threshold electronics [9].

The data are collected by the novel trigger-less and reconfigurable data acquisition system [10], [11] based on trigger readout boards [12], [13]. The readout data is streamed to the Central Controller Module and then, further, to permanent

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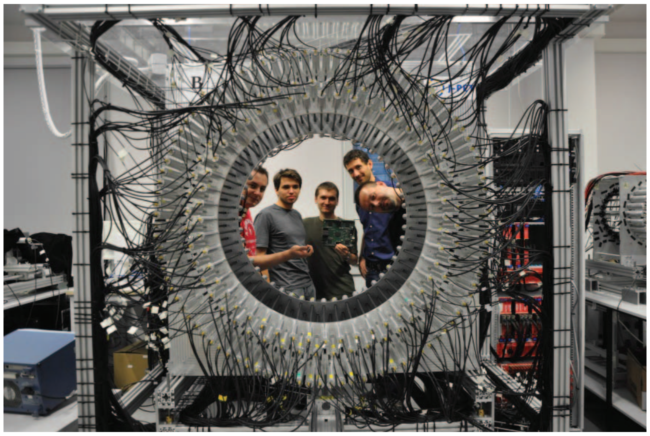

Fig. 1. Photo of the J-PET tomograph with few members of the J-PET group. The active inner part of the detector possesses a cylindrical shape with the length of 50 cm and diameter of 85 cm. The J-PET tomography scanner is made of three layers of plastic scintillators strips (black) and vacuum tube photomultipliers (gray). Figure and caption are adapted from [24].

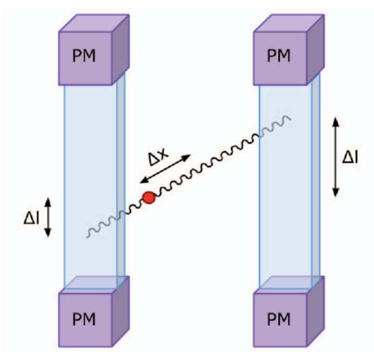

Fig. 2. Schematic view of the two detection modules [19]. A single detection module consists of a scintillator strip read out by two photomultipliers labeled PM. Red dot indicates a place of e+e- annihilation. In the first approximation the hit distance from the center of the scintillator (ΔL) is determined based on time difference measured at both ends of the scintillator strip. The position (Δx $\equiv$ LOR) along the line-of-response is determined from time difference measured between two modules. In practice more advanced method of hit-time and hit-position were developed which take advantage of the variation of the signal shape as a function of the hit-position [17], [18], [19]. Figure and caption are adapted from [24].

storage [11]. For the data processing and simulations, a dedicated software framework was developed [14], [15], [16]. The hit-position and hit-time are reconstructed using the dedicated methods based on the compressing sensing theory [17], [18] and the library of synchronized model signals [19], [20], [23]. Events corresponding to the scattering of gamma quanta in the patient or in the detector are suppressed based on (i) the correlation between distance between detection modules and the time-of-flight value [21], [22] and on the (ii) time-over-threshold (TOT) measurement, where TOT denotes a width of the photomultiplier's signal corresponding to the time interval between the leading and trailing edge at a given voltage level. Finally for the image reconstruction novel algorithms are developed and adopted for fast iterations on the graphics processing units (GPU) [25].

In order to compare the performance of the J-PET with the presently available LSO based TOF-PET devices we introduced [27] a figure-of-merit (FOM) which is directly proportional to the geometrical acceptance of the tomograph and to the square of detection and selection efficiencies of image forming events. Morover we assume that FOM is inversely proportional to the coincidence resolving time and the number of steps needed to scan the whole body. Fig. 3 indicates that in order to compensate for the lower efficiency of plastic scintillators and thus to obtain figure-of-merit of the J-PET comparable to the LSO crystal based scanners (with axial field-of-view equal to AFOV = 20 cm) it is required to use either two detection layers or to increase the J-PET AFOV to about 50 cm. Certainly the FOM of the LSO based PET would also grow approximately as square of AFOV but at the same time the cost of such PET detector would increase almost linearly proportional to AFOV, whereas the cost of the J-PET does not increase significantly. It is also worth mentioning that the J-PET detector is built from non-magnetic and low-Z material strips. Therefore it enables combining J-PET with CT and J-PET with MR, so that the same part of the body can be scanned simultaneously with both methods [36], [37], [38], [39].

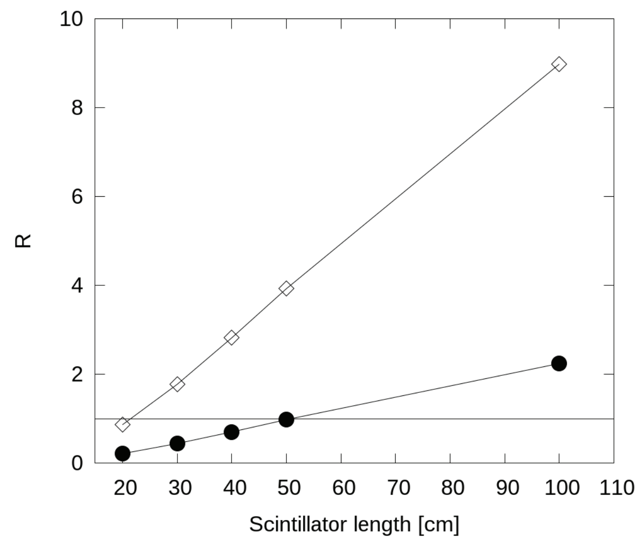

Fig. 3. Ratio of the figure of merits (FOM) for the whole body imaging with J-PET and crystal based PET detectors as a function of the J-PET axial field-of-view. For the comparison the properties of the best presently used crystals (LSO) were used. The shown ratio is defined as R = FOM(J-PET) / FOM(LSO). Horizontal axis of the figure refers to the length of the J-PET detector (length $\equiv$ axial field-of-view). The length of the LSO scanner was fixed to 20 cm, and the diameter of the scanners was fixed to 80 cm. Full dots indicate result determined for a single J-PET layer, and open squares indicate results for J-PET with two layers. The presented result was obtained under the assumptions that coincident resolving time (CRT) of the LSO based detectors is equal to 0.4 ns and that CRT values of the J-PET are as estimated in reference [27].

III. Conclusions

A construction of the full scale J-PET prototype was completed. The system is based on long strips of plastic scintillators which are characterized by better timing properties than the inorganic scintillator crystals used in the state of the art PET scanners [26], [28], [29], [30]. The J-PET with the 50 cm field-of-view possesses time resolution comparable to the currently best TOF-PET modalities with about 20 cm field-of-view [2], [31], [32], [33]. Due to the relatively low costs of plastic scintillators and their large light attenuation length (in the order of 100 cm) it is possible to construct a detector with a long axial field-of-view in a cost-effective way. These features make the J-PET technology promising and can make it competitive to the present solutions as regards the whole-body imaging.

It is also important to stress that J-PET as a detector of low energy gamma quanta characterized with the high angular and time resolution allows experimentation in the field of fundamental physics, biophysics and medical diagnostics e.g. by the studies of discrete symmetries in the decays of positronium [24], [34], by the development and tests of the multi-photon imaging [40], [35], or by the studies of properties of positronium atoms in the living organisms [40], [41].

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