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Application of scintillating properties of liquid xenon and silicon photomultiplier technology to medical imaging

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ABSTRACT

We describe a new positron emission time-of-flight apparatus using liquid xenon. The detector is based in a liquid xenon scintillating cell. The cell shape and dimensions can be optimized depending on the intended application. In its simplest form, the liquid xenon scintillating cell is a box in which two faces are covered by silicon photomultipliers and the others by a reflecting material such as Teflon. It is a compact, homogenous and highly efficient detector which shares many of the desirable properties of monolithic crystals, with the added advantage of high yield and fast scintillation offered by liquid xenon. Our initial studies suggest that good energy and spatial resolution comparable with that achieved by lutetium oxyorthosilicate crystals can be obtained with a detector based in liquid xenon scintillating cells. In addition, the system can potentially achieve an excellent coincidence resolving time of better than 100 ps.

1. Introduction

Positron emission tomography (PET) is a non-invasive imaging technique that produces a three-dimensional image of functional processes – it does not show anatomic features, but it rather measures the metabolic activity of the cells – in the body. PET is used in both clinical and pre-clinical research to study the molecular bases and treatments of disease. In clinical applications, PET has proven its critical value in several areas such as cancer diagnosing and staging, assessing neurological diseases such as Alzheimer's disease and dementias, myocardium blood flow and viability evaluation in cardiology, as well as an increasing role in radiotherapy treatment planning and chemotherapy monitoring.

The principle of operation (Fig. 1) relies in injecting into the patient a biologically active molecule doped with a radioactive isotope, called tracer (a standard tracer is fluorodeoxyglucose, FDG, formed substituting an atom of oxygen by the isotope ^{18}F in a glucose molecule). The radionuclide decays and the resulting positrons subsequently annihilate with electrons after traveling a short distance within the subject. Each annihilation produces two 511 keV photons emitted most of the time in opposite directions and these photons are registered by a detection system. The signal of each photon from every pair coincidence event is processed individually for spatial, energy, and arrival time information. For a pair coincidence event, if the energy of two photons stays within a preset energy window, of the order of 20–30% full-width-half-maximum (FWHM) for commercial systems, centered on the 511 keV photopeak, and the timing difference stays within a preset time window (typically of up to some 10 ns), a coincidence event will be registered and constitutes a line-of-response (LOR) for image reconstruction.

In this paper we propose a positron electron time-of-flight apparatus based on liquid xenon (PETALO). The system is based in a very homogenous and compact detection cell (the liquid xenon scintillating cell, or LXSC), which captures with high efficiency the copious scintillation light produced in liquid xenon (LXe) as a response to the interaction of a 511 keV gamma. The LXSC is read out by SiPMs connected to specialized ASICs optimized for excellent timing resolution. The high yield available in LXe, coupled to the fact that cryogenic operation of SiPMs eliminates the leading source of electronic noise in this devices (dark current) guarantees good energy and spatial resolution, while the fast scintillation time of LXe, combined with the homogeneity of the cell, guarantees a very low intrinsic coincidence resolving time (CRT).

This paper is organized as follows. In Section 2 we review the properties of the detectors used for PET scanners. In Section 3 we describe signal and noise in PET reconstruction. In Section 4 we discuss the factors that determine the performance of PET scanners. In Section 5 we review the relevant properties of liquid xenon as scintillating material. In Section 6 we describe the characteristics and features of the PETALO concept. In Section 7 we conclude.

2. Detectors used for PET scanners

The physical properties that define a detector suitable for a PET scanner are: attenuation length (λ), which sets the scale of the length

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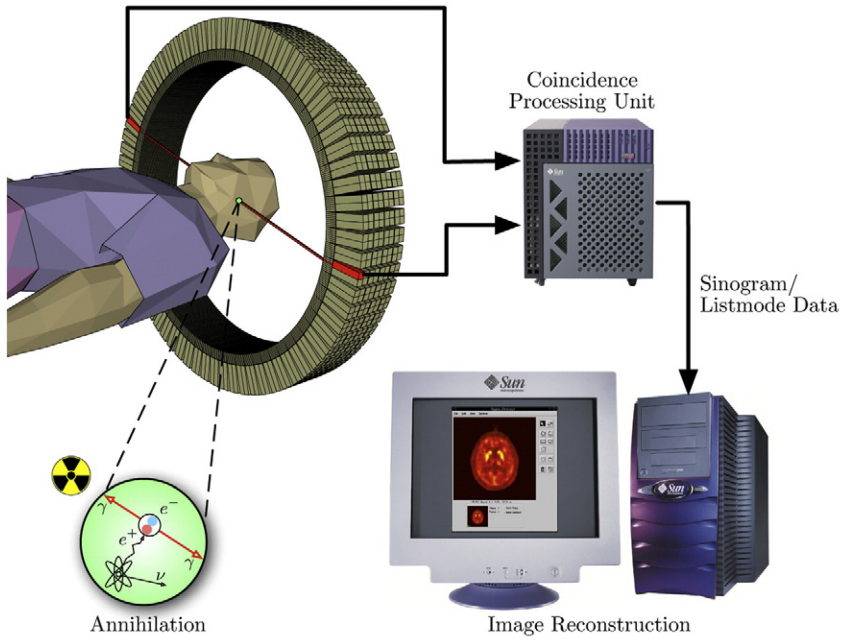

Fig. 1. Principle of operation of PET.

(across the photon line of flight) that the detector has to have in order to stop most of the incoming radiation; density (ρ), which is related with the total size and weight of the detector; photon yield per keV (Υ), which must be as high as possible to record large signals; refraction index (r), which must be as close as possible to the refraction index of the material of the photodetectors, which is glass in the case of photomultipliers (PMTs), and Si_3N_4 and SiO_2 layers for passivation and protection in the case of semiconductor devices such as avalanche photodiodes (APDs) and silicon photomultipliers (SiPMs); energy resolution at 511 keV (σ_E), which must be as good as possible; transverse spatial resolution (σ_T) (relative to the photon line of flight), which in turn depends on the photon yield and the granularity of the readout sensors; longitudinal spatial resolution (σ_L), important to minimize the so-called parallax error and crucial to achieve a good CRT; and scintillation decay time (τ), which must be as short as possible, to maximize the number of events acquired per unit time and to minimize the window used to correlate events in different crystals. In addition, if the system has very good time resolution (in the range of few hundred picoseconds) TOF measurements are possible.

Most modern PETs are built as rings using monolithic or segmented solid scintillating detectors read out by light sensitive detectors such as PMTs, or SiPMs) (Fig. 2). Many of the high-end scanners use lutetium oxyorthosilicate (LSO),[1] which is characterized by an attenuation length of 12 mm to 511 keV photons, a light yield of 26 photons per keV, and a scintillation decay time of 40 ns. These three parameters, essential for the performance of a PET scanner, result in an energy resolution of typically 10–15% FWHM, a spatial resolution of the order of 2–3 mm FWHM for the transverse coordinates. This is often achieved using segmented detectors read out by SiPMs. For example, for detectors of 3×3 mm^2 cross section, the rms resolution in the transverse coordinates is $\sim 3/\sqrt{12}$ mm, or about 2 mm FWHM. The error in the longitudinal coordinate goes as $\sim t/\sqrt{12}$, where t is the detector thickness. For $t \sim 20$ mm (enough to stop around 80% of the incoming gammas), the rms error is 5.8 mm (13.3 mm FWHM). Finally, the coincidence resolving time measured in the laboratory is of the order of 200 ps (for systems at ambient temperature) and 500–600 ps for commercial devices.

[1] LYSO (or lutetium-yttrium oxyorthosilicate) is also widely used. Both materials are very similar. In the rest of this document we will refer, for definiteness, to LSO.

3. 3 Signal and noise in PET reconstruction

The reconstruction of the image in a PET system requires crossing many LORs which in turn define one emission point in the area under study. LORs are formed by detecting the coincidence of two photons. Three types of coincidences, illustrated in Fig. 3, are relevant:

- True coincidences occur when both photons from an annihilation event are detected by detectors, neither photon undergoes any form of interaction prior to detection, and no other event is detected within the coincidence time-window (Fig. 3a).
- A scattered coincidence is one in which one of the detected photons (sometimes both) has undergone at least one Compton scattering event prior to detection (Fig. 3b). Since the direction of the photon changes due to the scattering process, the resulting coincidence event will, most likely, produce a wrong LOR. Scattered coincidences add a background to the true coincidence distribution, decreasing contrast and causing the isotope concentrations to be overestimated. They also add statistical noise to the signal. The number of scattered events detected depends on the volume and attenuation characteristics of the tissue being imaged. The best way to reject scattered coincidences is to build a PET system based on detectors with excellent energy resolution, since the scattered photons have also lost a fraction of their energy, and can therefore be rejected by imposing that the measured energy is inside a narrow window around 511 keV.

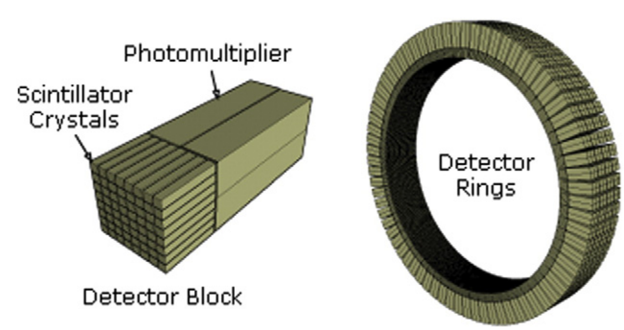

Fig. 2. A PET ring made of segmented solid scintillators.