Current Challenges and Advancement of PET

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1 Abstract

Positron emission tomography (PET) imaging technique is based on detecting two time-coincident high energy photons from the emission of a positron-emitting radioisotope. The physics of the emission, and the detection of the coincident photons, give PET imaging unique capabilities for both very high sensitivity and accurate estimation of the in vivo concentration of the radiotracer. PET imaging has been widely adopted as an important clinical modality for oncological, cardiovascular, and neurological applications. In this term paper, firstly I will discuss about the PET after that physics of PET, new detector technology, application and application development, multi modality imaging, and conclusion of positron emission tomography.

2 Introduction

Positron emission tomography (PET) is a type of nuclear medicine procedure that measures metabolic activity of the cells of body tissues. PET is actually a combination of nuclear medicine and biochemical analysis. Used mostly in patients with brain or heart conditions and cancer, PET helps to visualize the biochemical changes taking place in the body, such as the metabolism (the process by which cells change food into energy after food is digested and absorbed into the blood) of the heart muscle.

PET is a type of nuclear medicine procedure, this means that a tiny amount of a radioactive substance, called a radiopharmaceutical (radionuclide or radioactive tracer), is used during the procedure to assist in the examination of the tissue under study. Specifically, PET studies evaluate the metabolism of a particular organ or tissue, so that information about the physiology (functionality) and anatomy (structure) of the organ or tissue is evaluated, as well as its biochemical properties. Thus, PET may detect biochemical changes in an organ or tissue that can identify the onset of a disease process before anatomical changes related to the disease can be seen with other imaging processes such as computed tomography (CT) or magnetic resonance imaging (MRI).

In clinical and preclinical studies, in living structural imaging technique gives useful information. To reveal the real structures of the physiological time-varying processes that explain disease occurrence, it is necessary to combine morphological information within living molecular imaging. Positron emission tomography (PET) imaging probably offers more translational possibilities than any other modality due to its combination of sensitivity and quantitative accuracy. PET is a noninvasive imaging technique that provides physiological information through the injection of radioactive compounds that is radiotracers, detection of radiation, and reconstruction of the distribution of the radiotracer. PET imaging has evolved from an imaging technique used for research to become a standard component of diagnosis and staging in oncology (such as cancer screening) and it is also used for specific neurological and cardiovascular indications. It provides the clinically useful information that about tissue and organ function, and status, through the use of radio labeled molecular imaging agents. The type of information provided depends on the imaging agent and the disease and can include detection, classification, staging, prognosis, treatment planning.

Merging PET imaging technique with anatomical imaging method such as CT or MRI provides the information about the what is problem is that is from PET and where is problem is that is from CT or MRI. CT or MRI gives the anatomical information that is used to provide the estimates of the quantitative corrections which is needed for this imaging technique. Combination of PET with CT or MRI gives the advancement for this technique which is led for the development and application of new imaging techniques.

3 Physics of PET imaging

The physics that enable PET imaging are based on the emission of a positron by a neutrondeficient radioisotope.fluorine-18 (18F), the most commonly used radioisotope. As the radioisotope decays to a stable state, the emitted positron travels a short distance (typically 1–2 mm) and interacts with an electron; this interaction annihilates both the electron and the positron, thus producing two high-energy photons. The photons travel in opposite directions along an approximately straight line, and can be detected outside the body by the PET scanner. The detectors use scintillator crystals coupled to photo-multipliers.

- The first limitation in PET imaging is the noise, which is driven by the number of individual 511 keV photons detected. In turn, this is determined by the density of the scintillator, the count-rate capabilities of the scanner, and the amount of radiation injected into the patient.

- The second limitation is the spatial resolution, which is determined by the variability in estimating the interaction point of the 511 keV photon in the scintillator. In turn, this is affected by the optics of the scintillator, the number of optical photons emitted, and the fidelity of the photomultiplier tubes and associated electronics. These factors impose the noise and resolution characteristics of the PET imaging process.

If two photons are detected in a short-coincidence time window (typically 1–10 ns), the joint detection is called a true-coincidence event for the line of response (LOR) joining the two detectors. More accurately, the parallelepiped joining the two detector elements is called a tube of response. The total number of true-coincidence events detected by the two detector elements will be proportional to the total amount of radiotracer contained in the tube of response. This is the key to PET imaging. Based on this relationship, one can process the coincidence events to accurately reconstruct the distribution of the radioisotope. For time of flight (TOF) PET imaging systems, the differential timing of the detection of the two photons is used to localize the annihilation along the LOR.

4 Advantages and Disadvantages of PET

Table 1

Positron emission tomography (PET): new applications and their readiness for implementation

New technology	Problem addressed	Readiness level	Application	Comments
Faster and cheaper scintillators	Increases count-rate performance, light output (with higher energy and better spatial resolution), and usability; enables time-of flight PET imaging	In production, widely available	Cost reduction in clinical scanners; higher image quality in preclinical systems	Radioactive isotopes in the formulation introduce background noise that could limit the signal-to-noise ratio in very low count applications
Subsurface laser engraving of scintillators	Facilitates the manufacture of crystal arrays, reduces costs, increases sensitivity by avoiding the need for reflectors in between crystals	Prototypes are being tested, mainly for preclinical applications	Clinical and preclinical, but more relevant for preclinical use due to the higher complexity of the crystal matrices	Great potential for introducing alternative, affordable coding schemes for depth-of-interaction and light sharing
Solid-state photodetectors	Reduces the cost of the detectors; enables magnetic compatibility for combined PET–MR imaging	In production; several manufacturers offer alternatives and yearly updates	Used in combined PET–MR scanners; they reduce the cost of the PET gantry; new applications may be based on more compact detectors	Semiconductor technology is highly competitive; newer and better devices will appear, and their cost will be reduced over time
Dual-energy CT attenuation correction	Quantitative accuracy of PET imaging	Already developed for CT, but clinical need for combined PET–CT imaging not demonstrated yet	Clinical quantitative imaging	Already applied in preclinical imaging, but no applications yet
Scatter correction in systems with long axial extent	Larger field of view increases the scatter fraction	To be developed	Clinical systems	Monte Carlo simulations and single-scatter approximation approaches will require intensive computation
Clinical 4D (time-varying or gated) imaging	Difficulty of adopting 4D techniques into clinical routines	Widely used for preclinical applications; already adopted in some clinical applications, including cardiology and planning for radiation oncology treatment	Relevant to any imaging requiring suppression of respiratory-motion effects	Clinical applications are limited to simple models due to limitations in the signal-to-noise ratio; movement suppression is slowly being adopted
Multi-isotope PET imaging	Limitations of single-radiotracer imaging	Under development	Simultaneous imaging of different physiological processes	Several approaches have been reported in the literature, but none has been adopted in clinical or preclinical applications
PET–MR imaging	Soft-tissue contrast in anatomical imaging, e.g., brain	Commercially available	Under development	Attenuation correction issues not completely resolved

Figure 1: Advantages and Disadvantages of PET

5 New Detector Technology

Improvements to the quality of PET images are made by increasing the sensitivity and spatial resolution of the image, which requires an array of detectors that can capture the largest possible fraction of the photons emitted by the target object at the highest possible rate and with the best possible spatial accuracy. In addition, the detectors should possess the highest possible energy resolution to reduce the effects of scattered radiation.

5.1 Solid-State Photo-detectors

This include the different type of photo detector silicon photomultiplier, multipixel photon counter, single-photon avalanche diode, and avalanche photodiode .The mass production of semiconductor photo-detectors in radiation detector technology has enabled new imaging modalities, such as simultaneous PET imaging and MRI (20). Different manufacturers name their photo-detectors differently, depending on the manufacturing process and mode of operation. Above mention photo-detectors, The sum of the signals from all individual diodes is a measurement of the amount of light deposited in the array, and this, in turn, is related to the light produced by the scintillator. These new devices have had important impacts on detector design. They have an improved, effective sensing area in comparison with traditional photomultipliers, the time resolution increases their potential in terms of TOF capabilities, their power requirements are lower, their packaging is more robust and compact, and they are compatible with the high magnetic and electric fields present inside an MR scanner.

5.2 Digital photon counter (DPC)

It is a variant of conventional silicon photomultiplier (Si-PM) detectors, and it operates in a digital mode. The DPC treats the signal from individual photodiodes as the information and, instead of adding them in analog mode, counts them to produce a digital code that is related to the number of photons .In principle, this method should produce noise-free data that are precisely timed with respect to a master clock, thus producing high-resolution measurements of time. These novel characteristics of the DPC have created high expectations about the performance of the PET scanners that use them, and the first commercial clinical systems are appearing.

5.3 CZT Photo Detector

This detector based on the cadmium zinc telluride which does not use scintillators because they directly convert ionizing radiation to charge production, providing higher energy and spatial resolutions than scintillator-based PET detectors. Accordingly, new designs for detectors begin with new scintillator materials because this is the preferred method for detecting annihilation photons. Scintillation crystals produce fluorescent light after being excited by a 511 keV photon that has been scattered by or absorbed into the crystal. The intensity of the pulse of visible light contains information about the energy of the original photon that caused the scintillation. Therefore, annihilation photons can be characterized by detecting and counting the visible light photons with a photo detector. A good scintillator crystal should have high stopping power for 511 keV annihilation photons and high efficiency for converting the energy from the annihilation photon into detectable light; it is also desirable that the resulting light pulse occurs within a very short time, and the intensity should be proportional to the energy of the annihilation photon.

The optical properties of the crystal are also important: The crystal must be transparent to the wavelength of the fluorescent light to avoid self-attenuation, the index of refraction must match that of the photo-detector to optimize the collection of light, and the crystal should be rugged enough to be processed and manipulated during manufacturing.

Modern scintillators produce shorter and more intense light pulses compared with earlier crystals, such as BGO (bismuth germanate), which used to dominate PET-detector technology. BGO has a high density and high atomic number, so the 511 keV photons have a large probability of interacting with the crystal, but the BGO low light production limits the energy resolution of the detector, thus resulting in more scatter in the final image, and its long decay time precludes its use in applications with high count rates. Newer, more luminous, and faster scintillator crystals, such as LSO (lutetium oxyorthosilicate) and LYSO (lutetium yttrium oxyorthosilicate) (4, 5),

have been used to build detectors that have better energy resolution and faster light decay times that can reduce undesired effects, such as the piling up of events on the detector, degradation of event timing, or background scatter. These new scintillators also enable clinical TOF PET imaging (6–8), and they are being used in layered combinations (known as phoswichs)—of LSO, LYSO, LuYAP (cerium-doped lutetium yttrium perovskite), GSO (gadolinium oxyorthosilicate), LaBr3:Ce (cerium-doped lanthanum bromide), Ce:GAGG (cerium-doped gadolinium aluminum gallium garnet) (9), and others (10)—in small-animal PET scanners to implement depth of interaction (DOI) corrections on long crystals (11), which increases spatial resolution and reduces parallax error (12).

6 Drawback of Scintillator

A drawback of using scintillator crystals made from lutetium (176Lu) is that it is a radioactive contaminant that produces a natural background coincidence count rate, which increases the noise in low count-rate applications (13). Another drawback to using modern scintillators is their high cost. Alternative ceramic scintillators may ameliorate this limitation (for example, BaAl4O7:Eu2+); although they are too slow for PET time-coincidence imaging, they compete with currently used scintillator crystals when used for single-photon imaging (14). The mass production of transparent, ceramic scintillators may make them more affordable than current scintillators, and this may open new opportunities to confront the design of next-generation PET scanners.

In next update i will discuss about the how can we improved data acquisition system in PET.