Role of PET/CT

In Cardiology

Essay

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In Nuclear Medicine

By

Fairoz Mohammed S. Addin Mohammed

Under Supervision of

Dr. Sherif Mohamed El-Refaei

Assistant Professor of Nuclear Medicine

Dr. Mahassen Amin Abo-Gabal

Lecturer of Nuclear Medicine

Faculty of Medicine - Cairo University

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Dedication

This Essay is dedicated to
My Husband Mahfoudh Abdullah
And my daughter Maha.

Abstract:

The integration of positron emission tomography (PET) and multidetector CT (PET/CT) technology provides a potential opportunity to delineate the anatomic extent and physiologic severity of coronary atherosclerosis and obstructive disease in a single setting. It allows detection and quantification of the burden of the extent of calcified and non calcified plaques, quantification of vascular reactivity and endothelial health, and identification of flow-limiting coronary stenosis. PET/CT also has the potential to identify high-risk plaques in the coronary and other arterial beds. Together, by revealing the degree and location of anatomic stenosis and their physiologic significance, and the plaque burden and its composition, integrated PET/CT can provide unique information that may improve noninvasive diagnosis of coronary artery disease (CAD) and the prediction of cardiovascular risk.

In addition, this approach expands the diagnostic capability of nuclear cardiology to include atherosclerosis and may facilitate further study of atherothrombosis progression and its response to therapy, thus allowing assessment of subclinical disease.

Key ward: PET cardiology, PET/CT in Cardiology, Hybrid PET/CT in CAD.

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1 INTRODUCTION

Since the introduction of the first prototype computed tomography (CT) scanner in 1972, Rapid commercial development of tomographic imaging followed has made significant contributions to the diagnosis and staging of disease, and within 3 years of its appearance more than 12 companies were marketing, or intending to market, CT scanners; about half that number actually market CT scanners today. With the introduction of magnetic resonance imaging (MRI) in the early 1980s, CT was, at that time, predicted to last another 5 years at most before being replaced by MRI for anatomic imaging [1].

Obviously this did not happen, and today, with multislice detectors, spiral acquisition, and subsecond rotation times, CT continues to develop and play a major role in clinical imaging, especially for anatomic regions outside the brain. Functional imaging with positron-emitting isotopes was first proposed in the early 1950s as an imaging technique that could offer greater sensitivity than conventional nuclear medicine techniques with single photon-emitting isotopes. The SPECT collimator is eliminated and replaced by electronic collimation the coincident detection of two photons from positron annihilation, greatly increasing the sensitivity of the imaging system. However, other than some early prototypes in the 1960s, instrumentation to image positron emitters did not emerge seriously until the 1970s, and the first commercial PET scanners date from around 1980. PET was initially perceived as a complex and expensive technology requiring both a cyclotron to produce the short-lived PET radioisotopes and a PET scanner to image the tracer distribution in the patient. Consequently, during the 1970s, PET did not experience the explosive growth of CT, nor, during the 1980s, the comparable growth of MRI. In fact, it was not until the 1990s that PET became recognized as an important technique for imaging cancer by mapping glucose utilization throughout the body with FDG. The elevated utilization of glucose by malignant cells allows cancerous tissue to be identified anywhere in the body, even though it may have no anatomic correlate that would allow identification on a CT scan [1].

The recent development of combined PET/CT instrumentation is an important evolution in imaging technology. The development of the first PET/CT prototype was initiated in 1992 with the objectives to integrate CT and PET within the same device,

to use the CT images for the attenuation and scatter correction of the PET emission data, and to explore the use of anatomic images to define tissue boundaries for PET reconstruction. Thus, the goal was to construct a device with both clinical CT and clinical PET capability so that a full anatomic and functional scan could be acquired in a single session, obviating the need for the patient to undergo an additional clinical CT scan. However, by the time the prototype became operational in 1998; neither the CT nor the PET components were state-of-the-art. Nevertheless, the work convincingly demonstrated the feasibility of combining the two technologies into a single device that could acquire co-registered anatomic and functional images without the need for software realignment. As mentioned, a number of important lessons emerged during the clinical evaluation program that followed the installation of the prototype and covered the years from 1998 until 2001 [1].Currently, five vendors offer PET/CT designs. However, P J Ell noted that only 1% of PET/CT scans are used in cardiology with a predicted useage of 15% in near future [2].

2 INSTRUMENTATION AND PRINCIPLES OF PET/CT IMAGING

2.1 Positron Annihilation and Tomography (PET) imaging

2.1.1 Introduction

Positron emission tomography (PET) is a noninvasive modality that produces tomographic images of the distribution of a radionuclide-labeled tracer injected in the body [3]. As the name suggests, PET imaging is based on radionuclide that decay by positron emission (Figure 1. For example, fluorine-18 (^{18}F) decays to oxygen-18 (^{18}O), emitting a positron (β) and a neutrino (νe):

$$^{18}F \rightarrow ^{18}O + \beta^{+} + \nu e \qquad \text{Equation 1}$$

A positron is classified as antimatter and is analogous to an electron, having identical mass but opposite electrical charge. The positron is ejected from the nucleus and rapidly loses its kinetic energy through collisions with numerous nearby electrons. Since antimatter and matter are mutually unstable, the positron and an electron then undergo a process known as *positron annihilation* and are converted into

a pair of gamma-ray photons. Because total energy is conserved, each gamma ray has energy of 511 keV, which is equivalent to the mass of the positron and of the electron according to the well-known relationship $E = mc^2$. Because total momentum is conserved, the two gamma rays travel in opposite directions with a relative angle very close to 180 degrees.

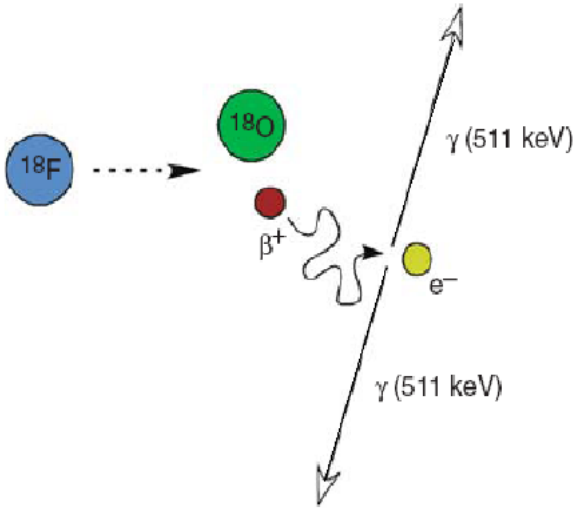

Figure 1: Annihilation radiation (pair of 511 keV gamma rays) results from the interaction of an electron e- and positron β+ emitted by a PET radionuclide (^{18}F in this example, which decays to ^{18}O).

2.1.2 PET Detectors

Detectors used in PET scanners [4, 5] are designed for optimal detection of 511 keV coincident gamma rays under clinical imaging conditions.

- First, the incident gamma ray is absorbed in a scintillation crystal and produces energetic electrons, which in turn produce a cascade of visible photons. This flash of visible light exits the crystal and is shaped by a light guide before reaching an array (often a 2x2 block) of photomultiplier tubes (PMTs).
- The PMTs convert the flash of light into electronic pulse signals that are processed by front-end amplifiers and other electronics.
- The integrated signal from the group of PMTs is measured and is proportional to the total energy deposited in the crystal.
- The scintillation event is rejected if the detected energy is outside the allowed range for 511 keV gamma rays, set by the lower level discriminator (LLD) and upper-level discriminator (ULD) values (approximately 400 keV and 650 keV,

respectively). By restricting the acceptable energy range, the numbers of scattered gamma rays have energies lower than 511 keV is minimized.

- A position-weighted signal is processed in order to determine the crystal of interaction, which specifies the detected location of the event. The spatial resolution of the detector is limited largely by the physical size of the individual scintillation crystals.
- After being processed by the front-end electronics, the electronic signals associated with the scintillation events are then processed by coincidence electronics.
- The coincidence electronics sample all detectors and accurately determine the time of each detected event, within time resolutionτ. The timing pulses from all electronics banks then are compared. If two time pulses overlap, the two associated events are considered to be simultaneous and are designated as a coincidence pair. The definition of simultaneity is limited by the coincidence timing window 2τ, which is in the range of 4 to 16 ns on clinical PET scanners.
- The coincidence timing window is set to be as small as possible so that nearly all true coincidence events are detected and as many "random" coincidence events as possible are rejected. Random coincidences arise when two 511 keV gamma rays originating from different positron decay events are detected by chance within the coincidence timing window. For a pair of detectors having "singles" count rates of $S1$ and $S2$ for individual 511 keV gamma rays, the random coincidence rate R for the pair of detectors depends linearly on the coincidence timing window and is given by:

$$R = 2\tau\, S_1\, S_2$$

Equation 2

- The random coincidence rate is proportional to the square of the source activity, whereas the true coincidence rate is linear with respect to the source activity. Thus, the relative contribution of random coincidences rises for increasing injected dose. Both random and scatter events lead to erroneous back projection (Figure 2) and should be minimized through appropriate detector design and configuration