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Basic Science

1 The Physics of PET/CT Scanners

Ruth E. Schmitz, Adam M. Alessio, and Paul E. Kinahan

1.1 What Makes PET Useful?

Positron emission tomography (PET) offers several unique advantages compared with other imaging modalities. PET measures the two annihilation photons that are produced back-to-back after positron emission from a radionuclide-tagged tracer molecule, which is chosen to mark a specific function in the body on a biochemical level (► Fig. 1.1). Hence, PET provides molecular imaging of biological function instead of anatomy. The detection of both annihilation photons in coincidence, providing inherent collimation, yields increased sensitivity over single-photon imaging. Furthermore, PET allows for accurate attenuation correction (AC) either from a dedicated transmission scan or from computed tomography (CT) images with combined PET/CT scanners. Accurate correction for attenuation (and scatter, etc.) allows extraction of accurate quantitative as well as qualitative information from PET images. Only minute amounts of radiolabeled tracer need to be injected because of the high sensitivity of PET. In addition, positron emitters (^{11}C , ^{13}N , ^{15}O , ^{18}F , etc.) are relatively short-lived, which enables optimal use of imaging photons while keeping patient radiation dose low. Furthermore, many of these isotopes can be incorporated into biological substrates (glucose,

H_2O , NH_3 , CO_2 , O_2 , etc.) and pharmaceuticals, without altering their biological activity.

Compared with CT scans and magnetic resonance imaging (MRI), PET images generally appear much blurrier or noisier, due to the relatively limited number of photons that can be collected during an imaging study. In addition, detector spatial resolution is poorer due to the detector physics. X-ray CT scanners can easily resolve points less than 1 mm in size, whereas PET scanners cannot reliably resolve the size of point sources less than 4 to 5 mm at best, and closer to 10 mm in practice. However, this does not impair their high sensitivity to focal tracer concentrations or their usefulness in accurate quantitative functional imaging.

In this chapter, we introduce the physics of PET imaging. Several textbooks provide a more in-depth treatment and are included in the References section.^{1,2,3}

1.2 Radioactive Decay

1.2.1 General Principles

Radioactive isotopes are atoms whose inner core, their nucleus, is unstable, in a state with too much energy. Nuclei consist of a densely packed arrangement of protons and neutrons. By undergoing decay, the nuclei change their composition and properties to arrive in a less energetic and more stable state.

The decay process follows an exponential law: the number of decays per second is always proportional to the number of undecayed nuclei present. The rate of decay is level of radioactivity, also called activity, which is determined by the half-life of the particular nuclide—the time it takes for half of the original nuclei to decay. The most common isotope in PET is fluorine 18 (^{18}F), which has a half-life of 109 minutes. After some time, t , the activity left, $A(t)$, is proportional to the initial number, $A(0)$, and an exponential term involving the half-life, τ , of the nuclide:

$$A(t) = A(0)e^{-t(\ln 2/\tau)}$$

Radioactive rates (or activity) are measured in units of becquerel (1 Bq = 1 decay/s) in the International System of Units (SI) or the traditional curie (1 Ci = 3.7×10^{10} decay/s). A common scale factor used in the clinic is 1 mCi = 37 MBq.

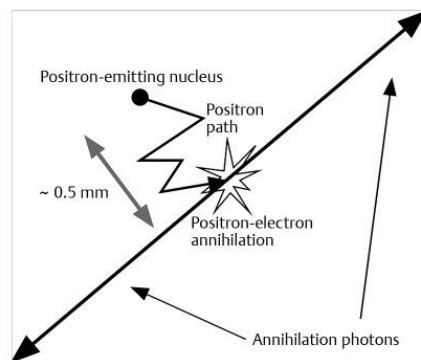


Fig. 1.1 General principle of positron emission tomography imaging: decay of radionuclide, positron (β^+) emission, multiple scatter in tissue, annihilation with electron, and production of two back-to-back 511-keV annihilation photons. (Not to scale.)

1.2.2 Positron Emission and Annihilation

In β^+ (positron) decay (► Fig. 1.1), a nuclide transforms one of its core protons (p) into a neutron (n) and emits a positron (β^+), essentially a positively charged electron, and a neutrino (ν): $p \rightarrow n + \beta^+ + \nu$. The average positron range in matter depends on the positron's energy and material characteristics, such as the density and the atomic number. For [fluorine 18]fluorodeoxyglucose (^{18}F FDG), positron ranges are rather short, typically less than 1 mm.

At the end of its path, the positron, being anti-matter to electrons, will annihilate (recombine) with an atomic electron. In the annihilation, electron and positron convert their mass into energy and produce a pair of 511-keV annihilation photons traveling in opposite directions. The 511-keV photon energy (E) comes from Einstein's famous equation $E=mc^2$, where m is the mass of the electron or positron (a very small number) and c is the speed of light (a very large number, which is then squared). This annihilation radiation is what is detected in PET and what is used to form images of tracer concentration in the body.

1.2.3 Interaction of Photons with Matter

The dominant annihilation photon interaction in human tissue is *Compton scatter*. The photon interacts with an electron, ejecting it from its atomic shell. The photon experiences a loss of energy and an associated change of direction, typically out of the active detector range, and so is unavailable for image formation.

Compton scatter and other interactions lead to an attenuation of the annihilation photons along a straight line. In other words, the number of photons that are observed in a straight line from where they were produced decreases exponentially with increasing length of the material traversed. The thickness of soft tissue required to reduce the intensity of a 511-keV photon beam by one-half is approximately 7 cm, as opposed to 3 to 4 cm for lower energy X-rays. Thus, for approximately 14 cm of soft tissue, the 511-keV annihilation photon flux would be reduced to one-fourth of its original intensity; through the abdomen, the photon flux can be reduced to 1/50 of its original intensity. Thus, attenuation is often the dominant factor in PET image quality, especially for thicker patients.

1.3 Data Acquisition

1.3.1 Photon Detection and Scintillation Detectors

The general goal of photon detection is to measure the total energy deposited by the photon when it traverses the detector. For highest sensitivity and accuracy, all of the photon's energy should be deposited, but in practice this is not always possible.

In most PET scanners today, scintillation detectors are used as detection elements. They couple inorganic scintillation crystals that emit visible or near-ultraviolet light after interaction with an incident high-energy (511 keV) photon to photo detectors that detect and measure the scintillation photons.

In scintillation crystals, the incident annihilation photon (nominally 511,000-eV energy) interacts and creates tens of thousands of visible wavelength photons (~1-eV energy each) in a very short flash, or "scintillation." The number of scintillation photons produced in the crystal is proportional to the energy deposited by the annihilation photon.

Scintillators for PET photon detection can be rated on four of their characteristic properties:

1. The *stopping power* is the inverse of the mean distance traveled by photons before they deposit energy in the crystal. This length depends on density and effective atomic number (Z) of the material. A short travel distance is favorable because it will yield more interactions with the 511-keV photons and a better efficiency for detecting them in crystal of fixed size.
2. The *decay constant* describes how long the scintillation flash lasts in the crystal. Shorter decay constants are desirable because they allow for counting higher photon rates and lower background rates.
3. Good *energy resolution*—a small ratio of energy variance over energy—means that there are only small fluctuations in the energy measurement. This gives a means to distinguish against PET photons that have Compton scattered (and lost energy) before being measured. The energy resolution depends on the light output and the intrinsic energy resolution of the crystal.
4. The *light output*, as the name indicates, is the number of scintillation photons produced by each incident photon. Again, this should be as high as possible, allowing the best spatial and energy resolution.

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Table 1.1 Scintillators used in PET scanners

Material	Cost	Light output ^a	Effective density ^b	Decay time ^c
Nal(Tl)	Inexpensive	Highest	Lowest	Long
BGO	Expensive	Lowest	Highest	Long
LSO (or LYSO, or LFS)	Most expensive	High	High	Very short

Abbreviations: BGO, bismuth germanate; LFS, lutetium fine silicate; LSO, lutetium oxyorthosilicate; LYSO, lutetium yttrium orthosilicate; Nal(Tl), thallium-doped sodium iodide.

^aDetermines energy and spatial resolution.

^bDetermines scanner sensitivity.

^cDetermines dead time, random coincidences rate, and time-of-flight resolution.

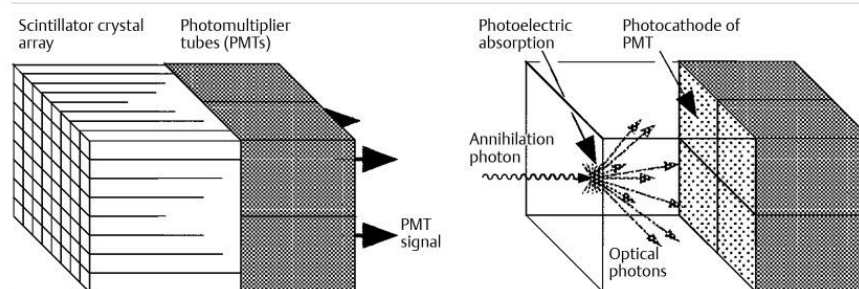


Fig. 1.2 Schematic of a block detector with finely segmented scintillator crystals read out by four photomultiplier tubes.

The most commonly used PET scintillators are listed in ► Table 1.1. Other materials are being evaluated (e.g., lanthanum bromide [LaBr]). Historically, manufacturers have adopted different materials for their systems. Current time-of-flight PET scanners (TOF-PET) use the LSO (lutetium orthosilicate)-type scintillators because of their favorable decay constant, providing very short decay time.

The most commonly used photodetectors for PET are photomultiplier tubes (PMTs). PMTs are vacuum tubes with a photocathode, which produce electrons from incoming light photons that are accelerated and amplified. The resulting electrical current is proportional to the number of initial scintillation photons and therefore to the energy deposited in the scintillation crystal by the PET photon. Recently, a new class of photodetectors is being offered in newer PET scanners called silicon photomultipliers (SiPMs). These are silicon-based solid-state detectors that are smaller than PMTs and offer advantages in size, price, and performance.

By segmenting the scintillator blocks, using many small PMTs or SiPMs, the location of the photon detection can be determined. The most commonly used setup today is the *block detector*

(► Fig. 1.2). Here, small individual scintillation crystals, a few millimeters in size where they face the patient, are tightly packed into blocks, which are typically coupled to four or more small PMTs or SiPMs. To determine the interaction position of the annihilation photon from the spread-out scintillation photon signals, the relative outputs from the PMT or SiPM signals are compared. The calculated location then determines the crystal element to which the photon is assigned.

Spatial resolution (in the detector) of a few millimeters is possible with this scheme, as it is mainly determined by the size of the crystal cross section.

A conventional clinical PET scanner is constructed as a cylindrical assembly of block detectors, stacked together as multiple rings. The sensitive volume inside the detector cylinder that a patient can occupy is called the imaging field of view (FOV), which in human scanners is typically 70 cm in diameter and 16 to 25 cm in axial length (► Fig. 1.3).

1.3.2 Coincident Photon Events

Because of the positron annihilation, we expect to observe two photons at roughly the same time (in