

Section B

Explain the fundamental differences between single photon emission tomography (SPET) and positron emission tomography (PET), in particular with regards to attenuation and the radionuclides used.

Describe a PET system based on Bismuth Germanate (BGO) detectors and discuss the advantages and disadvantages of using multiring systems with and without septa.

New PET scanners are now being built using Lutetium Oxyorthosilicate (LSO) detectors. Compare the properties of the two scintillators giving your reasons why you would select one rather than the other.

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1. Introduction

Nuclear medical imaging is based on detecting the nuclear radiation emitted from the body after introducing a radiopharmaceutical into a person in order to tag a specific biochemical function. Depending on the type of technique being used either x-rays, gamma rays, or positrons are emitted and as long as they have enough energy to escape from the body in significant numbers, images can be generated that portray the *in vivo* (inside living human body) distribution of the radiopharmaceutical. In general, nuclear medical imaging can be divided into three categories; conventional or planar imaging, single photon emission computed tomography (SPECT or SPET) and positron emission tomography (PET). In planar imaging, the three-dimensionally distributed radiopharmaceutical is imaged onto a 2D surface, producing a projection image. The tomography process allows 2D slices through the body to be imaged, which could then be reconstructed into a 3D image.

This report aims to discuss the fundamental differences between single photon emission tomography and positron emission tomography, in particular with regards to attenuation and the radionuclides used. A PET system will also be described, based on multi-ring bismuth germinate (BGO) detectors with and without septa. Finally, PET systems using new lutetium oxyorthosilicate (LSO) detectors will be discussed and compared to the scintillation detectors presently used.

2. SPET

Single photon emission computed tomography (SPET) is an imaging technique involving the use of multiple views to compile an image of one or more transverse slices through the patient. SPET uses radionuclides that are attached to molecules called radiopharmaceuticals. This process is known as the ‘labelling’ of molecules, and depending on the radiopharmaceutical used, will target certain parts of the body which the molecule has an affinity to. The SPET technique is based on detecting individual mono-energetic photons emitted at random by the radionuclide to be imaged. It employs a rotating detector (a scintillation camera) that completes a 360° circular or elliptical orbit around the patient lying on the couch. The scintillation

camera halts approximately every 6° to acquire a new view of the stationary source of photons in the patient. A mechanical collimator is used with the detector in order to localise the gamma-emitting activity by restricting the acceptance angle of the photons. This limits the amount of unwanted photons from background radiation and large incoming angles from photon scatters. The disadvantage of this is that the solid angle of detection at each view is very small, and only a fractional amount of the emitted radiation is transmitted through the apertures onto the scintillation crystal. Reconstruction of a SPET image from the detector signal is performed by using a similar algorithm technique to that in x-ray CT, and is known as back projection. In principle, the reconstruction of an image by this method is the inverse of the scanning process. Projection data in terms of x-ray sums are acquired at a given angle of view, but over multiple planes, which are superimposed to produce an image as an assignment of radioactivity to pixels.

2.1 Attenuation

The emitted gamma radiation interacts with the body via photoelectric absorption and Compton scattering processes, producing a significant amount of attenuation of the primary beam of energies. This results in an image which is darker in the centre than at the edges. Since attenuation depends on the properties of the medium between the photon origin and object boundary, it is necessary to know the distribution of the attenuation coefficients corresponding to the energy of the emitted radiation and the source distribution. This would allow a correction for attenuation to be made, but since the source distribution is unknown, obtaining an exact solution for this is very difficult. The function needing to be solved is shown below:

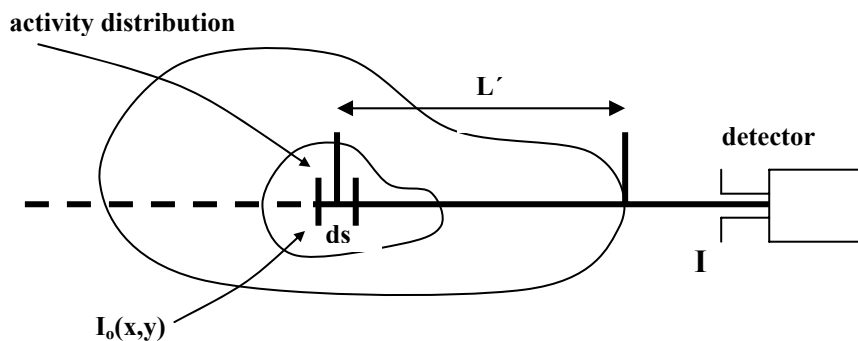


Figure 1. Regions of interest for the attenuation correction [1]

$$I = \int_{(l,\theta)} I_0(x, y) ds \cdot e^{-\int_{(l,\theta)} \mu(x,y) ds} \quad [\text{Equation 1}]$$

where: I = Intensity at detector

I_0 = Intensity at source (function of interest)

μ = Attenuation coefficient (distribution unknown)

The solutions to these integrals are inseparable, but there are a number of ways in which the problem can be tackled. The distribution of attenuation coefficients may not be known, but by assuming it to be constant, an average value can be calculated and used to find the function of interest. This is not ideal, as the distribution of the function of interest will then be calculated wrongly. A comparison with a ‘phantom’ material can be carried out, although the general characteristics of the region of interest will not be uniform, unlike the phantom. Calculations using Monte Carlo methods could also be employed; however this is a slow process. Another way to solve the attenuation correction problem is to carry out transmission tomography first on the patient. The attenuation function can be explicitly found, and used in the SPET calculation to find the source intensity. A lower resolution transmission tomography image is acceptable, although considerations about the extra dose given need to be made.

Besides the attenuation caused by photoelectric absorption events, Compton scattering of the emitted photons within the object introduces another error in the data. Due to the finite energy resolution of the detection system, many of the scattered photons are indistinguishable from the primary photons, which causes blurring and lack of contrast in the image. This is not as prominent as the attenuation losses, but corrections for scatter still need to be made to obtain a better accuracy.

2.2 Radionuclides used in SPET

An ideal radionuclide for any imaging technique must have the best possible balance between a number of physical and biological properties. In the ideal situation the radionuclide must have a short physical half-life in order to minimise the dose to the

patient, along with a suitable biological half-life. It needs to remain in the body long enough to be imaged, but be expelled soon after. Chemical binding between the radionuclide and radiopharmaceutical must be strong, being as pure a gamma emitter as possible, with an energy range between 100-150keV. Also the radionuclide should not be foreign to the body.

In SPET, the most commonly used radionuclide is $^{99}\text{Tc}^{\text{m}}$, with a half-life of 6 hours and gamma energy of 140keV. It emits 10 gamma photons for every one emission from a competing process (characteristic x-rays from internal conversion in this case), but is an anthropogenic element, although has a low toxicity. $^{99}\text{Tc}^{\text{m}}$ is not manufactured in this country, which causes a problem due to the short half-life of this radionuclide. To overcome this, it is kept in its parent form of ^{99}Mo , which beta decays to $^{99}\text{Tc}^{\text{m}}$ with a half-life of 66 hours. The parent-daughter decay process reaches transient equilibrium after approximately 20 hours, and it is then that the two are separated; ensuring technetium activity is at its maximum.

SPET images can be represented as a series of slices, or as a continuously rotating 3D representation on a computer screen, showing the organ and its defects. However, some of the main problems are a lack of variation of spatial resolution as a function of distance from the collimator and the lack of accurate method to correct for attenuation.

3. PET

Positron emission tomography (or PET) is similar to SPET in that it relies on detecting radiation emitted from a radionuclide attached to a pharmaceutical distributed within the patient. The distribution of radioactivity is again estimated by back projection algorithms, however, the unique characteristic of a PET system is the simultaneous detection of two photons from positron annihilation by coincident detectors. In a typical PET system the patient is surrounded by a ring of scintillation detectors coupled with photomultiplier tubes, as opposed to the single rotating detector used in SPET. When two coincident detectors register an event simultaneously, the positron decay process is assumed to have occurred along the line

between the detectors and registers a count. The advantage of this is that background radiation doesn't interfere, as the two detectors must register simultaneously. Some accidental coincidences may occur between the arrivals of pairs of unrelated photons, caused by the Compton scatter of photons in the body, but huge efforts have been made in recent years to improve the sensitivity and resolution of the PET system, so that more accurate images may be produced.

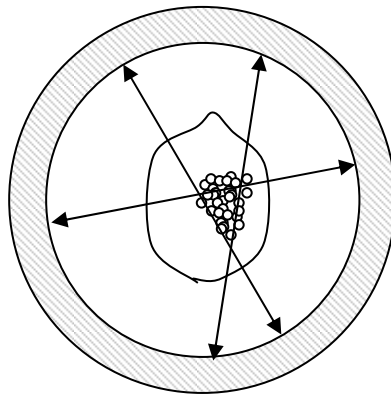


Figure 2. The coincidence detection of a PET system [6]

3.1 Attenuation

One of the major advantages of a PET system, over SPET is that the attenuation procedure is, in principle, exact. However, to achieve quantitative detection a number of problems have to be overcome. A correction for elastic Compton scattering has to be introduced. This can be done to some extent by using an energy discriminator on the detectors. Random coincidences also have to be taken into account, along with an attenuation correction for absorption by tissue. Whereas in the SPET case the distribution of attenuation is unknown, this can easily be resolved by a PET system due to the coincident annihilation photons from the radionuclide. Each 511keV photon follows an exponential absorption determined by the linear attenuation coefficient (a function of photon energy and distance travelled through the tissue), which can be used to calculate the function of interest in equation 1.

3.2 Radionuclides used in PET

The success of PET is largely based on the isotopes used for the imaging process. Common radionuclides are usually of low atomic number and include ^{11}C , ^{13}N , ^{15}O and ^{18}F . These positron emitters have a strong physiological affinity to the human body, with short physical half-lives, so facilitate effective imaging, whilst minimising the dose to the patient. As the radionuclides are all major components of the molecules in living matter, they are closely related to metabolic processes in human physiology, so offer direct study of the metabolism of an organ with zero toxicity. A table of properties for some of the radionuclides used in PET imaging are shown in figure 3.

Radionuclide	Half-Life (minutes)	Maximum energy (MeV)	Range (mm)	
			FWHM	FWTM
^{11}C	20.4	0.96 (100%)	1.1	2.2
^{13}N	10.0	1.20 (100%)	1.4	2.8
^{15}O	2.0	1.74 (100%)	1.5	3.6
^{18}F	109.7	0.63 (97%)	1.0	1.8
^{68}Ga	67.8	1.90 (89%)	1.7	4.0
^{82}Rb	1.3	3.15 (95%)	1.7	5.8

Figure 3. Table of radionuclide properties used in PET [1]

The problem is, however, that such positron emitters require a cyclotron for their production. Due to the short lifetime of the radionuclides, the cyclotron must be located within close proximity to the scanner, so installation of a mini cyclotron and radiochemical facilities along with the support personnel are required for any PET system, which is a major financial investment.

Until the recent developments of the PET system, SPET was the more widely used imaging technique. The radionuclides don't need to be produced on site and multiple slices were more easily obtainable than in PET. But with the developments of the multiring detectors now employed by PET and the wider accessibility of cyclotron accelerators, PET imaging has grown in its use around the country. The major

improvements in detector sensitivity have allowed positron emission tomography to be used extensively in medical imaging. Bismuth germinate (BGO) crystals are used in place of the NaI(Tl) detectors in SPET because BGO has a higher intrinsic efficiency for the higher energy (511keV) photons.

4. PET Systems using BGO Detectors

Any application that utilises scintillators for photon detection desire materials with high light output, short decay time and excellent stopping power. In addition they also need to be inexpensive and chemically inert. The first PET scanners were developed in the early 1970s and were based on the thallium-activated sodium iodide crystal scintillators used in SPET cameras. They consisted of a single scintillator crystal coupled to a photomultiplier tube, but this design was expensive, and the low density of NaI resulted in a reduced stopping power of the higher energy photons. Bismuth germinate was found to be a denser scintillator, with greater power to stop the annihilation photons, so the BGO block detector became a standard for PET imaging. The current design involves blocks of BGO crystal cut into individual crystal elements bonded to four photomultiplier tubes. This design controls the light distribution along the PMTs and the sum of the four signals is used to form a timing pulse and measurement of the photon energy. The entire module has a decay time of 300ns. This is the time taken after a photon interaction for the BGO crystal to emit its scintillation light. The most modern PET systems employ three continuous rings of these scintillator-PMT blocks, with the cuts in the crystal resulting in a total of 24 rings of detectors.

5. Multi-ring Detectors With and Without Septa

The first multi-ring scanners incorporated lead or tungsten annular shields mounted between detector rings, called septa. Their purpose was to shield the detector rings from photons that scattered out of the transverse plane (i.e. permitting only coincident photons between opposing detectors). It allows two dimensional reconstruction algorithms to be used on a plane-by-plane basis, rather than requiring a full 3D

algorithm. However, restricting the annihilation events to a two-dimensional set of planes made for inefficient use the emitted radiation.

Multi-ring PET scanners with retractable septa were later designed, which could acquire data in either a two or three-dimensional mode. With the septa retracted, the overall sensitivity is increased and a 3D image can be reconstructed. However, this also increases the amount of random coincidences and scatters that are detected, so any estimate of the net benefit of 3D imaging compared to 2D must also take into account these increases, i.e. the increase in signal with the retracted septa is accompanied by an increase in statistical noise. While PET scanning without septa have proved beneficial for the 3D imaging of the brain, there are still advantages for using the 2D with septa mode for whole-body scanning. Recently, however, a number of factors have significantly improved the image quality for 3D whole-body imaging. These factors include advances in reconstruction algorithms, more accurate scatter corrections and the introduction of new, faster scintillators.

6. LSO Detectors

New PET scanners are now being built using lutetium oxyorthosilicate (LSO) detectors. These new, faster scintillators have significantly improved the performance of clinical imaging, due to the shorter decay time of LSO compared with BGO detectors. This in turn reduces the systems dead-time and improves the count rate performance, particularly at high activity levels. The faster LSO scintillator also decreases the amount of random coincidences detected, and its increased light output improves the resolution of the system. Unlike a BGO scintillator, however, the LSO has an intrinsic radioactivity due to the small abundance of ^{176}Lu . Single photon emissions are produced with an energy range of 88-400keV, but such a small radioactive component is of little consequence for the coincidence counting of 511keV photons. A table of properties of BGO and LSO scintillator crystals are shown in figure 4.

Material	BGO	LSO
Density (g cm ⁻³)	7.31	7.41
Effective Z	75	66
Scintillation Decay Time (ns)	300	12/40
Photon Yield (per keV)	4.8	30
Index of Refraction	2.15	1.82
Attenuation length (cm)	1.05	1.16

Figure 4. Table of scintillator properties [1]

7. Conclusion

Traditionally SPET was the more widely used imaging technique. This was due to the fact that radionuclides do not need to be produced on site and the resolution and sensitivity of the system was better than for any other technique. However with the developments of the PET scanner, this became the more desirable imaging process. Attenuation correction calculations became easier, and the advancements with the scintillation detector design and retractable septa have provided a highly efficient system. Work will continue on finding better materials for detectors, along with improving 3D full-body PET scanning.

8. References

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