Detector technology challenges for nuclear medicine and PET

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Abstract

The challenges facing the development of new detector technology for single photon imaging and positron emission tomography (PET) are considered. There is currently great interest in functional imaging with radionuclides, particularly PET, triggered by new clinical applications and developments in molecular and cell biology. Modality systems that combine radionuclide imaging with CT present new challenges, as do very high resolution systems for imaging small animals. Whilst for PET there are some fairly well defined routes to improving performance, the basic design of single photon systems has remained unchanged for many years. This review outlines the challenges that must be addressed by detector physicists in order to obtain significant advances in performance, and indicates some of the approaches currently being adopted. Emphasis is given to PET which is where the greatest opportunities appear to lie.

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1. Introduction

Radionuclides used in single photon imaging ideally emit a single gamma photon with sufficient energy to escape from the body. The distribution of a radiolabelled pharmaceutical is imaged using a gamma camera, which consists of a large area (typically 50 cm × 50 cm) NaI(Tl) scintillation crystal coupled to a collimator, and read out by photomultiplier tubes (PMTs) (Fig. 1). The performance of a gamma camera is determined as much by the collimator as by the detector itself. The overall sensitivity (i.e. the fraction of disintegrations that result in a correctly registered gamma) is well under 0.05%, and whilst the spatial resolution of the detector itself may be 2–3 mm full width at half maximum (fwhm), the resolution in the final image is usually much greater than this. A gamma camera can be used in a stationary position to obtain planar views of the patient, or rotated to perform single photon emission computed tomography (SPECT). PET is based on the coincidence detection of pairs of 511 keV gamma photons from positron annihilation. Because the line of response joining two detector elements is defined electronically, there is no requirement for a collimator and the sensitivity of the current generation of PET scanners is typically 0.5–5%, i.e. much greater than is possible for single photon imaging. The detector element in the vast majority of installed systems is the block detector—typically a 5 cm × 5 cm × 3 cm block of scintillator (e.g. BGO) is segmented in to an 8 × 8 array read out by four PMTs. Block detectors are arranged around the subject in a ring configuration (Fig. 1). In contrast to the gamma camera, current PET scanner designs, with an image spatial resolution 4–8 mm fwhm, are far from optimal, and new scintillators and scanner configurations promise...
substantial improvements in sensitivity, spatial resolution and image quality.

Clinical nuclear medicine with single photon emitters is well established. Following regulatory approval in the USA for PET applications in clinical oncology, the number of clinical PET installations is now increasing at a dramatic rate and great efforts are being made by scanner manufacturers to develop high quality, rapid throughput PET scanners at an acceptable cost. In parallel with new clinical applications, radio-nuclide imaging is one of several emerging ‘molecular imaging’ techniques, whereby non-invasive imaging methods are combined with specific molecular probes to image molecular and cellular events. For example, techniques for monitoring gene expression in-vivo have been developed. As many of these techniques are currently limited to animal models, this has stimulated the development of high resolution (<2 mm) single photon and PET scanners. A third rapidly developing application is the use of PET in particular to speed up the process of pharmaceutical development.

2. Challenges for improved performance in PET

2.1. Limitations of current systems

Current PET scanners acquire a wholebody scan in typically 30 min, compared to around 1 min for CT or 10 min for MR, and the quality of PET images remains vastly inferior to that of CT and MR. Improved spatial resolution and image SNR will result in the ability to detect smaller tumours and tumours with lower contrast, and will also improve the visual quality of the images which still remains a barrier to the more general acceptance of the technique.

2.2. Factors influencing PET scanner performance

PET images are always count limited, and are smoothed to obtain an acceptable level of noise. Whilst image spatial resolution is key to producing sharper images and detecting small lesions, improving the signal to noise ratio (SNR) is likely to be a more productive way of achieving this than is reducing the intrinsic spatial resolution of the detectors. The image SNR is a complex function of the scanner sensitivity, energy resolution, temporal resolution and count rate capability. Whilst these parameters are all of course influenced by the specification of the basic detector elements, the overall scanner configuration (ring diameter, axial extent, shielding, discriminator thresholds, uniformity of response etc.) is equally important.

A large fraction of data collected does not contribute usefully to the image. The broad backgrounds contributed by scatter and random coincidences can be accurately corrected for, however the noise contribution can not. This has a similar effect to reducing the sensitivity, and images must be smoothed with a corresponding degradation of spatial resolution. Sensitivity, scatter, randoms and deadtime all effect the SNR in the raw data and in turn the SNR in the image. As an attempt to quantify these effects, the concept of noise equivalent count rate (NEC) has been introduced [1]. The NEC is the true coincidence count rate that (in the absence of randoms and scatter) would result in the same level of statistical noise as that actually obtained (Fig. 2). Whilst NEC is obviously not the whole story (for example it takes no account of spatial resolution), it is a reasonable method of comparing performance and predicting image quality for different scanner configurations.
Scanners operate in 2D and/or 3D mode as shown in Fig. 3. 2D mode provides excellent rejection of random and scattered coincidences at the expense of sensitivity. In 3D sensitivity is dramatically increased, but scatter and randoms increase disproportionately and gains in image quality are often small. In addition, in 3D singles rates on each detector can become so high that the detector is saturated at only moderate true coincidence rates. These effects are particularly acute for whole body studies where randoms and scatter are high. These effects are exemplified in the attempts to use gamma cameras mounted in opposition (without collimators) to perform PET imaging. These efforts have been largely unsuccessful, primarily due to the very high singles rates that saturate each detector head. Singles rates can be as high as 1 M cps whilst the true coincidence rate is only a few kcps.

The main challenge in improving PET performance therefore is to develop a detector that will exploit the sensitivity and count rate increases promised by 3D mode acquisition with an ever larger axial field of view, whilst at the same time maintaining random and scatter fractions at an acceptable, or ideally much reduced, level. For scintillation detectors, improvements in deadtime and random rejection translate into a need for high scintillation light yield coupled with short scintillation decay times. High light yield also facilitates the various crystal element/PMT multiplexing schemes that allow readout channels (and cost) to be kept to a minimum. The utility of energy resolution in reducing scatter (from the object) is less clear as energy thresholds are generally set well below the Compton edge in order to accept events which have scattered in the detector but are still in the right place, thus maximizing sensitivity. With detector spatial resolution currently around 4 mm, this is the parameter least in need of improvement.

2.3. Time of flight

If the difference in arrival times of two coincidence gamma photons can be determined with an accuracy of $\tau$, then the position of the initial disintegration along the line joining the detectors can be located with an accuracy of $\frac{c\tau}{2}$ where $c$ is the speed of light. Scanners with a temporal resolution of $\sim 600$ pS, corresponding to a positional uncertainty of $\sim 9$ cm have been constructed using $\text{BaF}_2$ and $\text{CsF}$, but any gains are outweighed by the low sensitivity achieved with these scintillators. LSO scanners have a temporal resolution of 1–4 nS. Whilst between one and two orders of magnitude improvement in temporal resolution would be required to make image reconstruction redundant, smaller improvements can lead to improved SNR, and may prove useful in rejecting scatter coincidences and events originating outside the desired field of view.
3. Current approaches to improving PET scanner performance

3.1. Choice of scintillator

Most current developments of whole body PET systems are based around inorganic scintillators read out with photomultiplier tubes, which currently represents the most cost effective way towards the goals outlined above. The main differences between approaches are in the selection of scintillator material and how the scintillator is coupled to the PMTs [2,3].

Scintillators suitable for PET are listed in Table 1. BGO has the shortest attenuation length and so results in high sensitivity, however the light yield is poor. The attenuation length of LSO is slightly less than that of BGO, however the light yield and decay time are significantly better, with the amount of scintillation light collected in a submicrosecond integration time being an order of magnitude greater than for BGO. Coupled with fast readout electronics, LSO results in improved temporal resolution, energy resolution and deadtime characteristics. GSO has a longer attenuation length, but has a short decay time and can exhibit good energy resolution. Various groups are continuing to search for new crystals with improved parameters, for example LaBr and LaCl [4] exhibit outstanding energy resolution. There is also interest in the use of transparent ceramic scintillators, which have potential advantages in terms of ease of production and homogeneity.

3.2. System configurations

Fig. 4 shows an example of a new whole body system designed to obtain high sensitivity and SNR. The scanner shown consists of five flat panel detectors arranged in an (incomplete) ring. Each panel consists of 10080 4 × 4 × 20 mm$^3$ LSO crystals coupled to an array of 88 PMTs which can identify individual crystal elements via a light sharing scheme. Combined with electronics capable of exploiting the fast detector readout and temporal resolution, such a configuration is designed to obtain whole body studies very rapidly. Fig. 5 shows another configuration in which arrays of small GSO crystals are arranged in a cylindrical configuration and read out with a different light sharing scheme, again with the aim of performing rapid whole body studies in 3D with high image SNR.

![Fig. 4. A high sensitivity PET scanner configuration utilizing flat panel LSO detectors. Courtesy of CPS Innovations.](image)
3.3. New photodetectors

Whilst the past 5 years has seen the adoption of new scintillator materials, these are invariably read out by standard single channel photomultipliers. Detectors based on multi-channel PMTs, position-sensitive PMTs and avalanche photodiodes have been under development for some time but need to offer very significant advantages given the low cost and high reliability of current systems. Likewise detectors based on hybrid photo-detectors and configurations using wavelength shifting fibres have also been developed [5].

4. Challenges and developments in single photon radionuclide imaging

The standard gamma camera configuration has remained essentially unchanged for more than a decade, the current configuration comprising one, two or three rectangular heads that rotate around the patient for tomographic SPECT acquisition. New developments are mainly aimed at small area portable systems for niche applications, such as sentinel node imaging, where the camera can be placed in close proximity to the patient to optimize sensitivity and spatial resolution, which would ideally be around 1 mm or lower [6,7].

Due to the limitations imposed by the collimator the design of whole body systems has become largely an optimization exercise involving trade-offs between the detector and collimator specifications, and the gantry configuration. The sensitivity and spatial resolution of the conventional gamma camera are limited by the collimator and not by the detector itself. A possible route for a step change in sensitivity and hence in image quality is the Compton camera. Localization is obtained from two position measurements and the energy of the Compton recoil electron. A collimator is not required and systems can be light and portable. While Compton cameras are predicted to have much greater sensitivity than conventionally collimated systems, this must be traded off against the large reduction in information obtained as each detected event can only localize a disintegration to the surface of a cone, rather than along a line. Simulation studies have been performed to specify the necessary detector properties for these systems however it is only recently that solid state detectors with the requisite combination of efficiency, energy resolution, spatial resolution and temporal resolution (the systems operate in coincidence) have become available. It is likely that working Compton cameras using silicon microstrips and segmented germanium detectors will be demonstrated in the near future [8,9].

5. Multimodality systems

Functional radiotracer images are inherently lacking in anatomical information, particularly for new specific tracers with high target to background ratios. Whilst effective software image registration techniques exist there is a great deal of interest in performing functional and anatomical studies simultaneously or near simultaneously in order to improve registration accuracy and to resolve the logistical problems associated with software registration. A large fraction of PET and SPECT systems currently being installed now incorporate a CT scanner in the same gantry, so that functional and anatomical images can be performed in rapid succession [10] (Fig. 6). Features identified with PET or SPECT can then be accurately localized via the CT scan. The CT data can also be used to provide a very rapid correction for photon attenuation. At present these combined systems are large, consisting essentially of independent scanners mounted in-line in a common gantry. A major challenge is to further integrate
the active components, and use common detector elements for both acquisitions, resulting in a more compact gantry and the possibility of simultaneous or very rapidly alternating acquisitions. This requires a detector that can switch between very high flux CT acquisition and pulse mode PET/SPECT acquisition.

Combining PET or SPECT with MR poses more technical challenges, as standard PMTs will not function in the high magnetic field (0.1–7 T) within the bore of an MRI scanner. Configurations for near-simultaneous acquisition would allow the PET detectors to be situated away from the highest field region, however true simultaneous acquisition promises more exciting applications. Small prototype systems have been constructed whereby long (~4 m) optical fibres are used to transfer scintillation light from crystals within the magnet bore to PMTs situated away from the magnet [11]. Such systems have so far only produced single planes of PET data. The challenge is to produce a design that can be scaled up and that can perform competitively with a state of the art PET-only scanner.

6. Small animal imaging

PET and SPECT technology development for molecular imaging and drug development is directed at very high resolution systems for imaging of small rodents, primarily mice. Several systems currently operate with a spatial resolution of ~1 mm, and combined PET-CT and PET-MR systems also exist. The challenge for these systems is to further reduce the spatial resolution whilst maintaining the sensitivity such that the SNR per voxel is preserved. PET detector technology for these systems is similar to that for human scanners. The commercially available microPET system [12], developed originally at UCLA, consists of a cylindrical arrangement of $2 \times 2 \times 10 \text{mm}^3$ LSO crystals read out by a short optical fibres to multi-channel photomultiplier tubes. The latest development of this design uses 1 mm square crystals and achieves a spatial resolution of around 1 mm [13]. In many very high resolution systems, the need for compact arrays of small photodetectors has lead to the use of multi-channel and position sensitive photomultipliers and avalanche photodiodes, whereas these have yet to be adopted for full size scanners. Different approaches have also proved successful, notably the ‘high density avalanche chamber’. This uses lead converters contained within a multiwire proportional chamber, and achieves a reconstructed resolution of ~1 mm [14]. Attempting to reduce the spatial resolution below 1 mm introduces the fundamental physical limitations imposed by finite positron range and to a lesser extent photon accolinearity. Also important are the effects of inter-crystal scatter and photoelectron range. However, it seems likely that with adequate sensitivity it should be possible to deconvolve at least some of these effects.

Very high resolution SPECT imaging has taken a slightly different approach and the use of purpose-designed pinhole collimators has resulted in very high resolution, high sensitivity imaging for low energy radionuclides such as $^{125}\text{I}$, using fairly standard detector configurations.

7. Conclusions

This is a very exciting time in the development of PET and single photon radionuclide imaging, with new applications generating ever greater demands. The next 5 years should see significant developments in whole body PET, multimodality PET and SPECT, small area portable single...
photon imaging and high resolution PET and SPECT. The development of new detector technology will play a major role in all these areas.

References