



Tutorial One

In this tutorial we're going to talk about how PET imaging or Positron Emission Tomography works. Let's start with some basic physics. Let's look at some of the benefits of PET over SPECT imaging. PET scanners provide a higher sensitivity, mainly due to the fact that no collimator is required. Coincidence photons are intrinsically collimated. The sensitivity of a 3D PET scan is typically 2 to 3 orders of magnitude higher than that of a SPECT gamma camera. Spatial resolution of PET is superior to SPECT, although this is tracer-dependent due to the positron range.

This will be covered later. Spatial resolution will also depend on the tracer used and how much can be administered. Images with poor counting statistics require more smoothing, which reduces spatial resolution. Accurate attenuation correction is also possible and superior to that in SPECT imaging. Scatter correction tends to be better as PET imaging has better accounting statistics due to the high sensitivity and therefore a better estimation of scatter.

This means that quantitation is possible, so this can provide absolute measures rather than just the relative measures that are generally encountered in SPECT. There can tend to be a lower radiation exposure to patients due to the shorter half-life of pet radioisotopes as well.

Here are the properties of some positron emitting isotopes. Note that the half-lives are relatively short compared to common isotopes used in general nuclear medicine. For example, technetium 99m with a half-life of 6 hours. All the isotopes emit a positron which annihilate with an electron producing two 511 keV photons, which will come on to later. However, the initial maximum energy of the emitted positron will change between isotopes and the larger the maximum energy of the positron, the further it will travel before it annihilates with an electron.

A positron is a positive electron. The positron will travel a very short distance until it meets an electron and annihilates. The rest mass energy of the positron and the electron are identical. To obey the laws of conservation of energy and momentum two gamma rays or photons are produced from the annihilation, each with an energy of 511keV due to the

conservation of energy and travelling at 180 degrees to each other due to the conservation of momentum.

We'll now look at detection in PET. If both the annihilation photons are detected within a short time interval, this is known as a coincidence event and it is assumed to have originated from a positron electron annihilation. A line can be drawn between the positions at which the two photons are detected, and it is assumed that the positron electron annihilation occurred somewhere along this line.

Physical collimation using a lead collimator as required in SPECT is therefore not required. This means that the sensitivity of PET imaging is much greater than in SPECT.

As with any imaging, there are limits to the achievable spatial resolution. This comes from a number of factors. Firstly, positron range. How far the positron travels before annihilation, which is isotope dependent due to the emitted positron energy. Also acolinearity of the photon pair, which are not quite at 180 degrees due to slight residual momentum of the electron positron pair.

In the real world, though, resolution is governed by other factors such as the detector sampling or depth of interaction effects and the counts that we can achieve in the image, meaning that it may need to be smoothed. Ideally, we can achieve resolution of something like 4 to 5 millimetres, but clinically this will be closer to 8 to 10 millimetres. Nevertheless, this is much better than can be achieved with SPECT, which is closer to 10 millimetres ideally and approaching 20 millimetres clinically, although this is far from that achieved by anatomical imaging of CT or MRI closer to 1 or 2 millimetres.

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The PET scanner consists of rings of small crystals arranged in blocks somewhat like small gamma cameras. Each crystal block consists of a matrix of crystals, which is coupled to photomultiplier tubes which convert the signals from the scintillators into electrical signals. Crystal blocks are joined together to form detector modules, which are themselves joined together into a ring arrangement to form the whole complete scanner.

A good crystal detector for use in PET requires good detection efficiency for 511 keV photons and also ideally high light output to give us better energy resolution and a quick decay time to give us a short coincidence window. Sodium iodide, which is used in gamma cameras, is good for stopping photons used in general nuclear medicine, such as those from technetium 99 with 140 keV.

But most 511 keV photons will be very poorly detected by the scintillator and will require something more dense. A high light output is also desirable. The energy of the incoming photon is measured from the light output of the scintillator and genuine nuclear medicine. This used to determine the energy of incoming photon but in PET all photons are of 511 keV.

Nevertheless, this information is still required to discriminate between true and scattered photons and those that are scattered that we don't really want. The higher the light output of the crystal, the better the energy resolution, and therefore the better the discrimination against scattered events.

A short decay time is also desirable. Once a single photon has been detected, there is a finite coincidence window in which we are looking for the second coincidence photon. The faster the decay time of the crystal, the smaller the coincidence window can be. The smaller the window, the less likely it is that we'll detect undesirable, scattered or random events.

PGO was the detector of choice for a long time, but more recently LSO crystals have become popular due to their faster decay time and better energy resolution.

It's important to look at the types of events that we can detect with a PET scanner. A true coincidence event is the detection of both photons originating from the same positron electron annihilation event. A random coincidence event is the detection of two photons, which originate from separate positron electron annihilation events, but just happen to arrive at the detector at the same time.

We don't want these. Similarly, we don't want scattered coincidence events. These are the detection of two photons from the same event, but where one or both of the photons have been scattered within the patient.

If we look in more detail at the events we see, firstly, the singles rate is the number of individual photons detected per second. The number of true coincidences is far less than this. Often only one of a pair of photons is detected. Although the true rate is much less than the singles rate, it is still proportional to the singles rate and therefore proportional to the total activity.

In the field of view. The randoms rate is proportional to the singles rate squared as we are essentially detecting singles in pairs of detectors and therefore proportional to the total activity in the field of view squared. The randoms rate therefore becomes more significant with increasing levels of activity. So increasing activity does not always improve scan quality.

The randoms rate is also proportional to the width of the coincidence timing window, i.e. the longer the timing window, the more random events will be detected.

Faster detector materials such as LSO allow shorter timing windows, therefore better rejection of randoms. The amount of scatter depends on the energy resolution of the crystals, as mentioned previously. The number of scattered events is affected by acquisition mode. In 2D, we may see only 10% scatter, whereas in 3D mode we might see 30% or more. It is also affected by the amount of scattering material in the field of view, i.e. larger patients will produce more scatter than smaller ones.

Looking at the evolution of PET scanners, they started out as single slice imaging machines and then progressed to multiple slice machines, but still imaging in what is known as 2D mode, where we have lead septa, similar to collimators on the gamma camera between individual slices to block out photons going between one plane and another. These have now evolved to 3D scanners where the lead septa are completely removed and coincidence events can be recorded between any two detectors in any plane.

If we look in more detail at acquisition in 2D or 3D mode in PET. In 2D mode, we have only moderate amounts of random scattered events detected perhaps 10% of each at the expense of running at only a moderate sensitivity. With the removal of the septa, we achieve much higher sensitivity, but at the cost of increasing the randoms and scatter detected. Nevertheless, these days all scanners scan in 3D mode.

As an example of the two acquisition modes. Here we have a patient which is being scanned in both 2D and 3D modes. In 2D mode we acquire less photons and you can see that the image is noisier compared to the 3D mode scan, which even though it has been acquired with more random scatter events, is of a higher quality.

In PET, the combined coincidence photons have to travel through the full thickness of the patient regardless of where the original positron annihilation occurred. Therefore, attenuation in PET can be considered to be a fairly straightforward process. As long as we know the thickness of the patient. In comparison, exact attenuation correction in SPECT imaging is not as straightforward as the origin of the photon is unknown. This difference is largely why PET imaging is a more quantitative technique.

The process of attenuation correction relies on information obtained by performing a transmission scan. In older PET scanners a transmission source, usually of germanium 68, was housed inside the PET gantry and rotated around the patient. The maximum activity of the source was limited due to the proximity to the detector crystals. As the total activity of the sources was limited, the transmission scan would significantly increase the overall scan time and would have to increase as the sources got older and also require frequent replacement of the sources. The dose to the patient from a whole body transmission scan performed in this way, though, is essentially negligible. This process uses CT in combined PET CT scanners as will be covered in the next tutorial.

When we have good attenuation correction, the images may be represented quantitatively in terms of the radioactivity concentration, i.e. in kilobecquerels per millilitre. This is usually converted into what is known as the standardized uptake value or SUV by normalizing the injected activity and patient weight as shown. If we had a uniform distribution of activity throughout the patient, the image would have an SUV of one or over the real patient. We will have accumulation and wash out of tracer, giving characteristic patterns, showing more or less uptake in different tissues. We can use the SUV for any particular tissue as a way of quantifying its glucose metabolism. The absolute level of SUV may then be clinically useful compared to the general population or in patient follow ups.