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An educational tool for demonstrating the TOF-PET technique

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Abstract

A detector system for positron emission tomography with time-of-flight capability has been built to serve as an educational tool for undergraduate students. The set-up consists of 48 BaF_2 scintillator crystals, each coupled to a fast photo-multiplier tube, mounted in a circular geometry. The analogue detector pulses are handled by fast constant fraction discriminators. A dedicated unit reduces the 48 channels to eight channels via delay-line encoding, and the signals are then fed to an eight channel fast time-to-digital converter. A VME processor sorts the events and sends them to a workstation where the coincident events are extracted. The time resolution of the detectors together with fast VME based electronics allows for time-of-flight measurements to improve on the signal-to-noise ratio in the reconstructed images. The system can be used for different types of exercises for the students, varying from the fundamentals of scintillator detectors to advanced image reconstruction. The set-up is described and some results are presented.

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1. Introduction

The advent of positron emission tomography (PET) in medical research has provided significant insight into the physiological functions of the human body. PET has also proven to be an important tool for diagnosing disease. The method can provide important information, complementary to other diagnostic tools, such as nuclear magnetic resonance and computer tomography X-ray imaging. The most commonly used material for PET detectors is bismuth germanate (BGO). An important property of this material is its high stopping power for gamma rays.

The commonly used PET technique does not take the time delay between the arrival of the two 511 keV gamma photons from annihilation into account. This is not possible with BGO scintillators, since the time resolution of this material is not good enough. Methods for utilising fast scintillators for time-of-flight (TOF) PET were discussed already in the 1970s, and the first

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systems were built and evaluated in the early 1980s [1,2]. The discovery of the fast light component in Barium Fluoride (BaF_2) [3] was important for this development. Although the TOF principle does not improve much on the spatial resolution of the reconstructed image, the additional position information for each event does reduce the noise far from the point of annihilation. This may improve the signal-to-noise ratio in the final image considerably, depending on the time resolution of the system [4].

To utilise the time-of-flight information of each event, the detector system must have a time resolution that corresponds to a distance shorter than the diameter of the field of interest. This usually means a time resolution in the sub-nanosecond region. Not many detector materials can give this good timing. Fast scintillators like Cesium Fluoride (CsF) and BaF_2 must be used to build these TOF systems. However, these materials are lighter than BGO and this means a reduced stopping power. The lower efficiency demands thicker detectors, and this in turn introduces parallax errors and increases Compton scattering. Apart from improving the signal-to-noise ratio, the TOF systems have the advantage that we can set a narrow coincidence time window. This can reduce the probability of random coincidences, and therefore allow for higher count rates. A major drawback with the TOF method is its more complex data analysis and image reconstruction. The technique also demands good timing properties from the photo-multiplier tubes, and the acquisition electronics. The TOF–PET technique has not yet been able to compete commercially with the standard technique, but the possible development of new fast and efficient detectors gives the technology a high potential for the future.

The purpose of building the present PET array was to give undergraduate students access to a modern medical imaging system. The students are, during laboratory exercises, given the opportunity to study the system in detail. They can look at the individual detectors and their output pulses, study the function of the acquisition system electronics, and investigate the event stream. They can also learn about the principles behind the image reconstruction process. Depending on the course subject and the prerequisites of the students the exercises stress different aspects of PET and TOF–PET.

The aim to provide a detector system for learning about the physics and technology of (TOF-)PET have influenced the design of the system considerable. An open design, and the small number of detectors make it convenient to study individual signals and give the students a good overview of the system. Although the array is not designed to give the same image quality as a commercial PET scanner it well illustrates all the major features of a modern PET-system. The use of BaF_2 detectors and fast electronics gives the system state-of-the-art timing properties, and this allows the students to investigate the time-of-flight properties for a single pair of the detectors, as well as for the whole array.

2. Set-up

The detector array [5,6] (see Fig. 1) consists of 48 cylindrical BaF_2 scintillating crystals, each mounted on a Hamamatsu R2076 photo-multiplier tube. Each crystal has a diameter of 15 mm and a height of 20 mm. Quartz windows allow the short wavelength of the fast component in BaF_2 to enter the photo-multiplier tubes. The

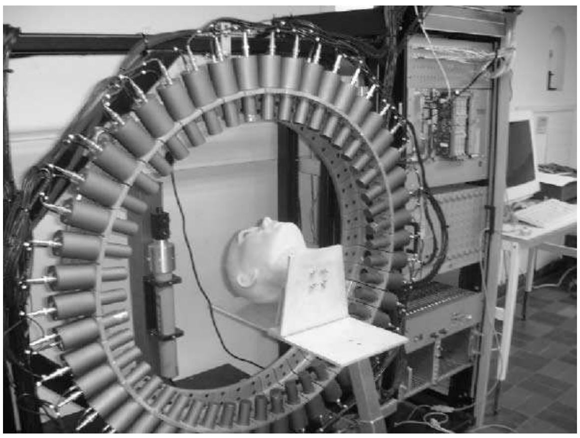

Fig. 1. The PET detector array and supporting equipment. The VME crate to the right handles all signal processing.

inner diameter of the present PET ring set-up is 734 mm. The design of the mechanical support also allows for an alternative two-plane-geometry with an inner diameter of 614 mm. This would give the system the capability of running in a 3D (TOF-)PET mode. However, no 3D image reconstruction code has been developed at this point. Two step-motors allow for rotation and translation of the array, relative to the object under investigation. The sources used in the object have so far been point and distributed sources of ^{22}Na.

The 48 analogue signals are fed from the photomultiplier tubes into 48 fast constant fraction discriminators (CFD) [7]. These discriminators are adjusted to reject energies below the 511 keV photo-peak in the energy spectrum. The 48 digital signals from the CFDs are then merged in a specially designed delay-line unit, creating an 8 channel output with 6 detector signals in each channel. Each detector signal within a channel gets a characteristic delay (which is a multiple of 10 ns). The eight channels, and a common start signal, are then used to fire a time-to-digital converter (TDC). A dedicated C-program in the VME CPU reads the data from the TDC and builds the raw events. The 68030 VME CPU runs the operating system OS9000 suitable for real-time event processing. The data are sent to a Digital Unix workstation via a dual port RAM card. All electronics are VME-based. The workstation receives the data via a PCI-based interface, and processes the raw events with a C-program. This program is linked to a MATLAB user interface, where the user can control the acquisition system, as well as the step-motors moving the detector array. The user interface also includes a picture of the PET-array and provides a feedback by drawing every coincidence line between the two individual detectors event-by-event, see Fig. 2. The event stream is also stored in a file to be read by the image reconstruction codes.

The energy spectrum for a single detector is given in Fig. 3. The ^{22}Na source has two discrete energy peaks; the 511 keV annihilation peak, and a peak at 1275 keV (the 1275 keV peak is small due to the limited efficiency of the BaF_2 crystal at high energies). Due to the small size of the BaF_2 crystals there is a large Compton background next to the 511 keV photo-peak. To avoid most Compton scattered coincidences, an energy thresh-

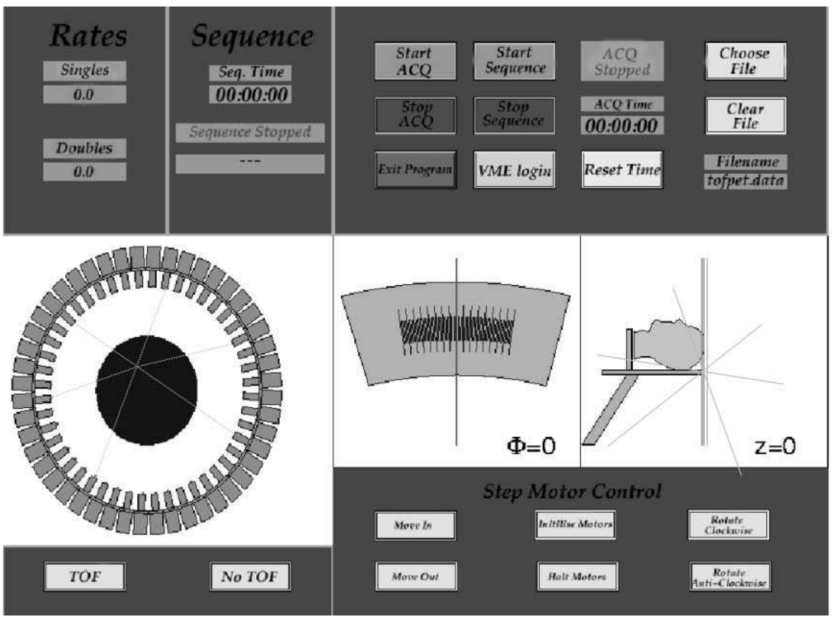

Fig. 2. The TOF–PET user interface.

old was set for every detector according to the figure. Fig. 3 also shows the time spectrum for a pair of coincident detectors. The time properties are slightly different for every detector, but all pairs of detectors give a coincident time pulse with a FWHM between 340 and 500 ps.

Separate software has been designed to read the event files for image reconstruction. Different algorithms have been implemented for the reconstructions. One difficulty when comparing the TOF–PET technique to standard PET is that the standard *sinogram* transform cannot be used with TOF. To allow for direct comparison, a simple direct projection algorithm has been used. More advanced reconstruction codes will be included in the future. More work is also needed to adapt the reconstruction algorithm to the motorised movement of the detector array.

3. Results

The TOF–PET system has been used to record data and reconstruct several images both with and

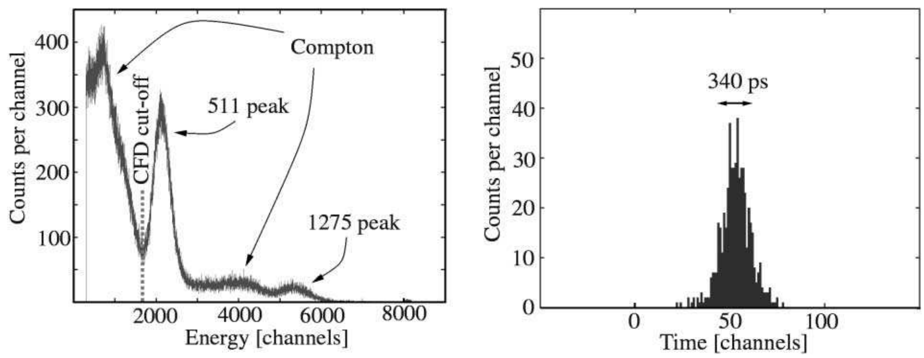

Fig. 3. Energy and time spectra for the BaF_2 detectors from a measurement with a ^{22}Na point source. The energy spectrum shows the strong 511 keV peak and the much weaker 1275 keV peak. The CFDs are used to remove the Compton scattered events at low energy. The time spectrum corresponds to a number of coincident 511 keV gamma rays recorded in two detectors at opposite sides of the source. The best time resolution for a pair of the detectors was 340 ps, and the average was 440 ps.

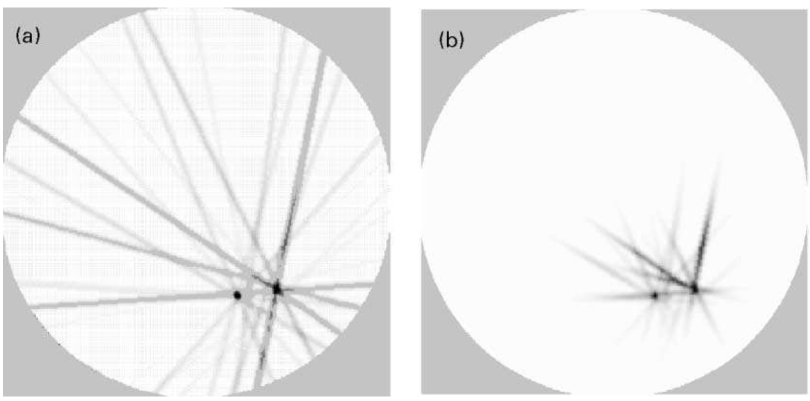

Fig. 4. A reconstructed TOF–PET image of two 70 μCi ^{22}Na ^{22}Na point sources. The field-of-view radius is 150 mm. Time-of-flight information is used for the reconstruction in (b) but not in (a). The noise far from the sources is reduced considerable with the TOF technique. The system is biased towards certain coincident pairs of detectors when the point sources are off-centre, because the detector crystals do not cover the full 2π angle. This problem will be eliminated by coordinating a rotation of the detector system over a number of angles with the image reconstruction algorithm.

without the TOF information. The measured spatial resolution for a 70 μCi ^{22}Na point source in the centre of the detector geometry is 8.5 mm (FWHM) for the PET mode and 7.5 mm (FWHM) for the TOF–PET mode [6]. Examples of reconstructed images is seen in Fig. 4. Two ^{22}Na sources of about 70 μCi are placed outside the centre of the detector ring, and at a distance of 25 mm from each other. The two sources are separated well in both the PET and TOF–PET mode. The most important advantage with TOF–PET is the noise reduction far from the sources. This effect is clearly seen in Fig. 4.

4. Conclusion

A PET detector array with TOF capabilities has been built for educational purposes. The set-up is intended to be simple and illustrative so that students can grasp the basic ideas of (TOF-)PET and the principles for data acquisition and image reconstruction. The system has already been successfully used in two undergraduate courses and plans exists to broaden the range of exercises.

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