Modelling γ transport in a PET/Field-Cycled System

W. B. Handler[1], B. A. Chronik[1]

[1]Physics & Astronomy, University of Western Ontario, London, Ontario, Canada

Introduction: The introduction of clinical PET/CT systems[1] demonstrates the importance of combining modalities to image both anatomical structure and biological function in a single system. MRI is superior to CT for the imaging of soft tissues and therefore a combined PET/MR system would be a more powerful tool for many applications. There are many technical obstacles to combined PET/MR systems: the geometry of traditional MR systems is incompatible with inclusion of a PET detector ring; the presence of the magnetic fields in MRI severely interfere with the operation of standard, photomultiplier tube (PMT) based PET detector systems; the presence of the large amount of structural material in an MR system serves as a source of compton-scattered gamma-rays, thereby degrading PET image quality.

One approach to combining PET with MRI has been to remove the photodetectors from the magnetic field environment and connect them to the scintillators via light-guides[2]. Our approach is different: we propose to construct a field-cycled MRI system around an otherwise normal PET system and operate the two systems in an interleaved fashion. Field-cycled MRI[3] uses two separately controllable resistive magnets in place of the usual single superconducting magnet. A polarising magnet provides a strong field to magnetize the sample, which when switched off is replaced by a readout magnetic field, which creates a much weaker and a more homogeneous field in which to acquire the NMR signal. Because each magnet is relatively easy to design, it is possible to open an annular ring within the field-cycled system to accommodate the PET detector ring. When both magnetic fields are reduced to zero (requiring less than 50 ms), the PET detectors can function normally, allowing PET data to be taken[4]. In this work, we investigate the effect that the field-cycled system elements (magnets, gradients, RF coils) have on the fraction of scattered photons detected by the PET system. A Monte Carlo simulation of the combined system has been performed to investigate these effects.

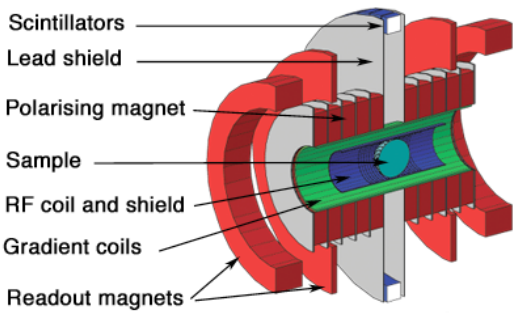
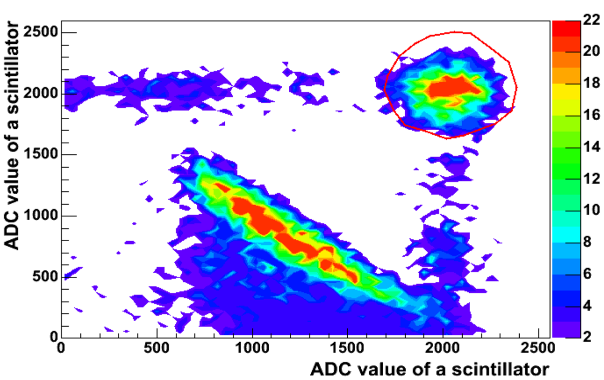


Figure 1 Cutaway of the elements simulated

Methods: The Monte Carlo model was constructed using GEANT4 code in C++[5] and included all geometry of the proposed combined PET/field-cycled system (the RF coil and RF shield, shims, gradients, polarising and readout magnets[6] as shown to the left. Positrons were created with the energy spectrum of the ^{18}F decay[7] within a spherical tissue sample of radius 5 cm and tracked until an annihilation occurred, producing the two back to back 511 keV γ-rays that were tracked through all the physical volumes included. Coincidence hits in a PET ring, composed of BGO (Bismuth Germanate) scintillators, were recorded for later analysis. Each event was examined to determine if the coincidence was caused by two "clean" γ-rays or was caused by an event where one or both of the γ-rays had scattered in transit to the detectors. This was possible as the full history of the transport was written into the data structure. As background, and/or multiple decays have yet to be included no data selection was made based upon time information and the subsequent analysis was performed using energy information only.

Results and Discussions: Data is gated upon a coincidence of two 511 keV γ-rays (see figure 2). Each event in the gate was examined to determine if a γ-ray Compton scattered before being observed in the scintillator. Figure 3 shows the contribution of events that scattered in the sample, the inner elements of the MR system (gradients, shims and RF coil) and the outer elements (the copper magnets and cooling plates). The scattered signal from the inner MR elements is equivalent to the scatter from the tissue, showing that a PET image is feasible in such an environment, though the presence of any additional scattered photons indicates there would be some loss of image quality. To fully understand this problem a phantom source will be simulated both with and without the MR system present and images generated. Key reasons for the development of this Monte Carlo simulation tool are that it allows us to investigate the effect that various shielding strategies, magnet configurations, and RF coil configurations are expected to have on the PET image quality, and it allows us to understand the nature of the effects we can expect to see in the initial prototypes we currently have under construction.



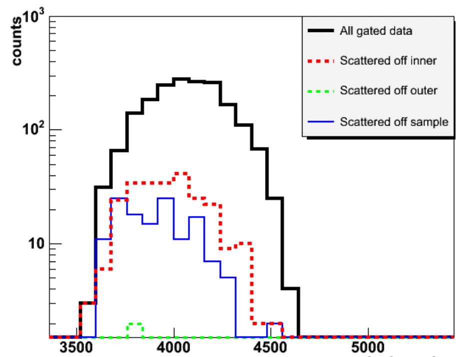


Figure 2 (left) The coincidence spectrum for the two ADCs in each event with the gate shown that was used for the 511 kev γ-ray selection.

Figure 3 (right) The summed adc spectrum for the events gated on in figure 2, showing the contribution of scattered events to the final signal.

[1] Beyer et al, "*A Combined PET/CT Scanner for Clinical Oncology*", J. Nucl. Med., **41**
[2] R.B. Slates et al, "*A study of artefacts in simultaneous pet and mr imaging using a prototype mr compatible pet scanner*" PMB, **44**
[3] A. Macovski & S. Connoly. "*Novel Approaches to low cost mri*" MRM, (1993)
[4] H. Peng, W. Handler & B. Chronik. In preparation.
[5] Agostinelli et al. "*GEANT4- a simulation tool kit*" NIMA (2003)
[6] K.Gilbert, W. Handler & B. Chronik. In preparation.
[7] E. J. Konopinski. "*The Theory of Beta Radioactivity*" (1966)