Simultaneous PET-MR: toward a combined microPET®-MR system

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Introduction

Positron emission tomography (PET) delivers quantitative, functional images with high sensitivity and specificity but does not provide anatomical support for that data. Magnetic resonance provides superb anatomical detail with improved soft tissue contrast and no additional radiation dose in comparison with computed tomography (CT). Consequently, dual-modality PET-MR offers potential advantages over commercially-successful PET-CT.

The performance of photomultiplier tubes used in 'state of the art' PET systems is degraded by static magnetic fields. A number of approaches have been adopted to address that issue: (1) the use of avalanche photodiodes (APDs) to replace PMTs [1,2], (2) field-cycling the MR magnet [3] which precludes simultaneous PET and MR acquisition, and (3) our approach in which both conventional PET and MR systems are modified to accommodate the other with minimal compromise [4].

Methods

A schematic of our system is shown in Figure 1. The novel 1T superconducting magnet has a 36cm bore and an 8cm 'split' to accommodate a complete microPET® Focus120 (Siemens Molecular Imaging) PET detector ring. That ring has a diameter of 15 cm and a axial field of view of 7.8 cm.

Placing the position–sensitive PMTs used in the microPET® system in a region of 'low' (<10 Gauss/1mT) magnetic field is achieved by a combination of magnet design, the use of 'long' (120 cm) fibre optic bundles between the scintillating crystal and the PMT, and additional magnetic shielding of the PMTs. The effect of using 'long' fibre bundles compared to the 'short' (10cm) fibre bundles used in the commercial system, and the performance of the PMT in ~10 Gauss/1mT has been studied.

To investigate as many features of the complete PET-MR system - and importantly the interaction between the PET and MR systems – as possible, a 'dual-module' test has been conducted. Two complete modules each containing four PET block detectors were placed in opposition in the magnet 'split'. The effect of MR radiofrequency (RF) and gradient pulses upon the PET has been monitored by acquiring position profile, timing resolution and crystal energy data whilst MR sequences are running. Interference between the PET electronics and MR signal has been monitored by acquiring MR data with the PET modules *in situ*.

Results

Figure 2 shows position profiles obtained for detectors with 'short' and 'long' fibre bundles. This will result in a increase in the scatter fraction.

The use of 'long' fibre bundles is the major factor determining PET performance – no effect arising from RF or gradient pulses has been observed.

Figure 3 shows MR images acquired using a fast TriPilot FLASH sequence with PET detectors *in situ*, and illustrates the signal to noise achievable with the PET electronics switched on or off.

Conclusion

The 'dual module' test demonstrates that high quality PET and MR data can be acquired simultaneously. With careful system design – particularly with regard to RF screening of the PET components – the interaction between the two systems can be minimized. The complete PET ring is now being constructed.

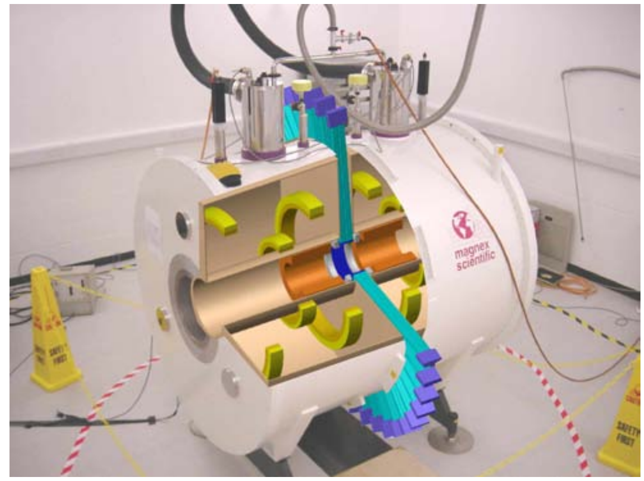

Figure 1: Schematic of PET-MR system. The cut-away shows split gradient coils (brown), scintillating crystal ring (dark blue), fibre bundles (light blue) and screened PMTs (dark blue outside magnet cryostat).

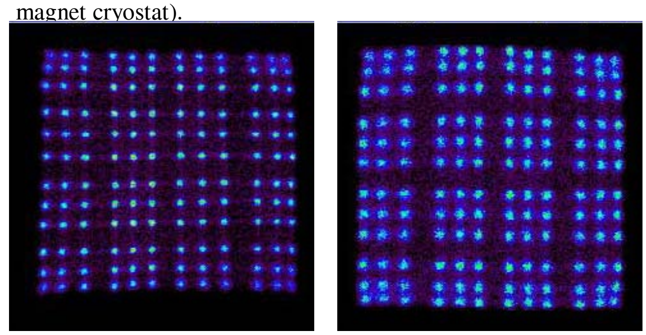

Figure 2: Position profiles obtained using 'short' and 'long' fibre bundles. The energy resolution is degraded from 17.2±0.1 for 'short' to 27.1±0.5 %FWHM for 'long' fibre bundles. The detectors with 'long' fibre bundles measure ~40% of the light measured with 'short' fibre bundles.

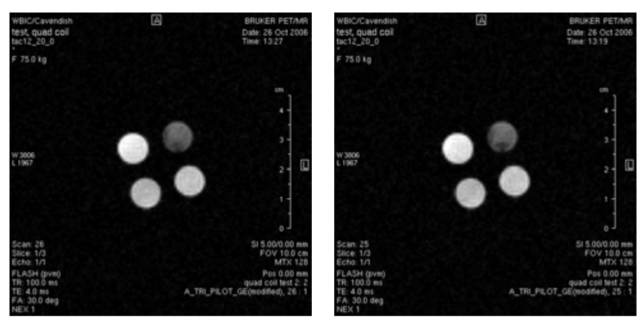

Figure 3: FLASH MR images of a multitube, doped phantom with the PET electronics off (left) and on (right). The signal to noise for the top left tube is reduced from ~53.9 to ~45.7, i.e., noise increases by x 1.2. That noise will be minimized by improved RF screening of the PET modules and electrical connections.

References

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