Design of a Dynamically-Controlled Resistive Shield for a Combined PET and Superconducting MRI System for Small Animal Imaging

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Introduction

Efforts to combine anatomical and functional imaging modalities have resulted in the development and widespread use of PET/CT systems. PET/MRI would offer improved soft tissue contrast, high-resolution anatomical information and would benefit longitudinal studies due to reduced dose. Combining conventional PET and MRI faces numerous technical challenges, particularly the sensitivity of photomultiplier tube-based (PMT) PET detectors to magnetic fields.

Current approaches to PET/MRI typically (a) employ novel PET detectors immune to magnetic fields, or (b) place conventional PET detectors in a region of minimal magnetic field. An approach of the first type uses avalanche photodiodes (APD), which are unaffected by magnetic fields, in place of PMTs within an MR-compatible PET insert [1]. An example of the second type of approach was recently introduced by Philips, where a conventional human-scale PET system with passive magnetic shielding is separated by several meters from an otherwise normal MRI system. A single bed moves patients between the PET and MR imaging regions, analogous to PET/CT.

In this abstract, we describe an approach in which a resistive electromagnet shield is used to null the field at the PMTs of a conventional PET system in the vicinity of an MRI system. The ability to use commercially available, highly optimized PET systems without major modification is a significant advantage of this approach. The design presented is based on the Siemens Inveon small-animal PET (Siemens Medical, Knoxville, TN) and a Magnex 2.0 T 310-mm-bore superconducting MRI, and is intended for small-animal imaging applications. The ability to have a single bed for the animal that can extend through both the MRI and PET systems without disconnection of monitoring equipment or anaesthetics was considered to be critical.

Methods

As indicated in Figure 1, a targeted shielding coil, placed between the two systems is energized during PET imaging to shield the PET detectors from the B_0 field of the MRI system. During MR imaging, the shield is turned off. A moveable and extendible bed with 30-micron precision moves the animal between PET and MR fields of view along a rigid track without disturbing the animal's position on the bed or disconnecting anaesthetic supply lines.

Proof-of-principle tests have shown that linear and mesh PMTs recovered normal operation within several milliseconds of a magnetic field being turned off with no long-term effects [2]. In addition, the authors have found that the Siemens Inveon PET system suffers no permanent performance degradation after repeated exposure to magnetic fields of 11 mT. The PET detector consists of a ring of 16 modules, each consisting of a row of four LSO crystal blocks (1.59x1.59x10 mm crystals in 20x20 block) coupled to four Hamamatsu R8900 position-sensitive PMTs. During exposure to magnetic fields, the R8900 PMT suffers changes in gain and efficiency of approximately 10% at 0.3 mT (radial, xy) or 1.0 mT (axial, z) [3]. So long as the spatial field profile does not change with time, the PET system can be calibrated to account for this level of gain and efficiency change with negligible affect on performance. Therefore, our goal was to reduce the field at the PET detector ring to below 0.3 mT (radial) and 1.0 mT (axial).

The targeted shield coil was designed using the boundary element method, with data for the MRI fringe field obtained from the manufacturer. A cylindrical ring of null-field targets extended 15 cm axially along the length of the 12-cm-long four-PMT module (1.5 cm buffer on either end) and extended from a radius of 8 cm (the face of the scintillator) to 14 cm (1 cm beyond the PMT). The distance between the center of the MRI system and the PET system was chosen to be 1.7 m. While larger separations would reduce power requirements, the range of bed motion must be short enough for cables and supply lines to remain connected between PET and MRI scans.

Results

The candidate shielding coil has radius 75 cm and length 60 cm and its front end is located 90 cm from the centre of the MRI system. The coil consists of a single layer of 34 variable-separation turns with a 10.7-mm minimum wire separation. The hollow wire to be used has 10x10-mm cross-section with a 5-mm-diameter liquid cooling channel. The characteristics of the shield coil are summarized in Table 1. The proposed design successfully reduces the maximum magnetic field at the location of the PMT detectors to better than 0.13 mT, as required. The power requirements for the coil are essentially equivalent to those of present-day insert gradient coil systems and would be acceptable for this application.

Discussion

The primary challenge in the operation of this system is certainly expected to be interaction between the shield and the superconducting magnet. The allowed distance between the PET and MRI systems directly affects the shield requirements and further tradeoffs are possible. It would also be possible to add a term in the algorithm cost-function that represents coupling between the shield and magnet. The result would be increased power deposition in the shield.

The approach described in this abstract benefits from allowing the use of commercially available PET systems, which include state-of-the-art timing & energy resolution, high sensitivity, and highly optimized event processing hardware. The sensitivity of the PMT-based system to be used (10%) is substantially greater than that of PET systems used in current simultaneous approaches to PET/MRI (0.23% [1]), meaning faster PET imaging time or lower dose for longitudinal studies.

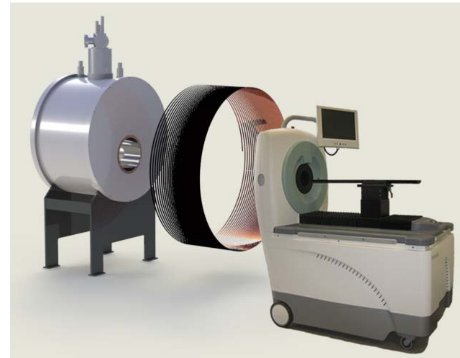

Figure 1. Proposed geometry of the actively shielded PET/MRI system. Bed track (not shown) connects PET and MR imaging regions.

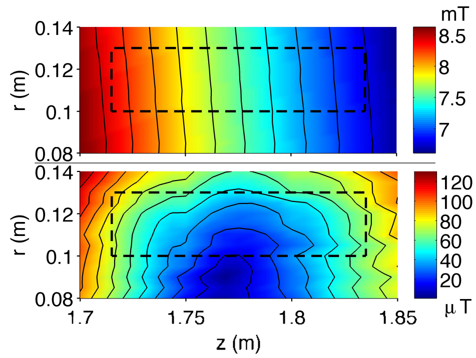

Figure 2. B_0 Magnitude field map in the vicinity of the PET detectors without shielding (top) and with shielding (bottom). The origin of the coordinate system is the MRI isocenter. The dotted line shows the location of the four PMTs that make up one module. To account for positioning error, the shield coil was designed to null the field over the entire region shown. The maximum magnitude field was 8.6 mT without shielding and 0.13 mT with shielding.

Table 1
Electromagnetic and Physical Characteristics of the Shield Coil

Characteristic	
Inductance (mH)	2.1
Resistance (mΩ)	34
Efficiency, *average over null region* (mT/A)	0.016
5 Gauss line, *minimum radius* (m)	2.2
Net force on coil, *z-axis* (N)	720
Mass, *copper / total* (kg)	120 / 200
Wire length (m)	160
Current, *DC* (A)	460
Power (W)	7300
Parallel cooling channels	4
Cooling water flow at 3.5 atm (cc/s)	130
Temperature rise (°C)	14

[1] Judenhofer, M., *et al. Nature Med.* (2008) **14**:459-465
[2] Handler,W., *et al. Phys. Med. Biol.* (2006) **51**:2479-2491
[3] Sakaki, N., *et al. Proc. of the 28th ICRC* (2003) 931-934