Proof-of-principle study of a small animal PET/field-cycled MRI combined system using conventional PMT technology

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1. Introduction

The combined PET/CT scanner has been introduced in 2000 \cite{1,2}, and it is capable of providing both spatial and temporal correlation between functional and anatomical images. Until recently, PET/MRI which may offer several additional advantages compared to PET/CT \cite{3–7}, has stirred much research interest. Two major challenges exist in the development of combined PET/MRI systems. The first is the problem of geometric compatibility. Both PET and MRI are essentially cylindrical systems that enclose the area around the subject to be imaged. Optimization of either modality involves efficient use of the available radial space. Any PET detector configuration would have to be placed within the MRI in some manner, and this may or may not require a significant redesign of the basic geometry of the PET, MRI, or most likely both. The second major challenge is that of magnetic field compatibility of available PET detectors. Typically in PET a scintillating material is used in conjunction with a photomultiplier tube (PMT) with a linear focused structure to detect the gamma rays \cite{8}. It is well known that the behavior of this type of PMT is compromised by even weak magnetic fields \cite{8,9}. In the normal architecture of an MRI system, there are of course very large magnetic fields, which present a major problem for gamma ray detection. In this paper, the focus is on the evaluation of this second problem.

Several different approaches to combined PET/MR have been investigated. One approach involves coupling scintillation crystals to PMTs through long optical fibers (~4 m), allowing the PMT assembly to be placed at a location where the fringe field permits proper functioning of the PMTs, while the scintillator crystals remain located within the bore of the MR scanner \cite{10}. An advantage of this approach is that it does not involve any major changes to the basic MRI system design. A second approach involves the creation of a slightly open-geometry MR scanner, such that the main magnet is divided into two cylindrical sections on either side of its isocenter. A gap is created such that relatively short optical fibers (~120 cm) can be used to couple the scintillators, which remain within the bore of the scanner close to the object. PMT devices are removed radially to a region where the fringe field permits proper PMT function \cite{11}. In both of the

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ABSTRACT

There are currently several approaches to the development of combined PET/MRI systems, all of which need to address adverse interactions between the two systems. Of particular relevance to the majority of proposed PET/MRI systems is the effect that static and dynamic magnetic fields have on the performance of PET detection systems based on photomultiplier tubes (PMTs). In the work reported in this paper, performance of two conventional PMTs has been systematically investigated and characterized as a function of magnetic field exposure conditions. Detector gain, energy resolution, time resolution, and efficiency were measured for static field exposures between 0 and 6.3 mT. Additionally, the short-term recovery and long-term stability of gain and energy resolution were measured in the presence of repeatedly applied dynamic magnetic fields changing at 4 T/s. It was found that the detectors recovered normal operation within several milliseconds following the end of large pulsed magnetic fields. In addition, the repeated applications of large pulsed magnetic fields did not significantly affect detector stability. Based on these results, we implemented a proof-of-principle PET/field-cycled MRI (FCMRI) system for small animal imaging using commercial PMT-based PET detectors. The first PET images acquired within the PET/FCMRI system are presented. The image quality, in terms of spatial resolution, was compared between standalone PET and the PET/FCMRI system. Finally, the relevance of these results to various aspects of PET/MRI system design is discussed.
above approaches, however, the optical fibers cause significant attenuation and temporal dispersion for the scintillation light signal, resulting in degraded energy and time resolution performance.

A third concept is to replace PMTs with semiconductor detectors that are insensitive to magnetic fields [12–14]. In order to minimize interferences from MRI components, the avalanche photodiodes (APD)-based PET detector were located at the edge of MRI scanner, while being connected to the scintillation crystal arrays locating close to the isocenter with short optical fibers (~10 cm). Alternatively, the use of fibers could be avoided by directly coupling the crystals to the APDs which both stay very close to the isocenter, with a properly designed RF shield to isolate PET detectors from the RF transmitter and gradient coils. These designs have exhibited very good performance of PET/MRI acquisitions.

In this paper, we propose a fourth approach based on an FCMRI system that involves a major redesign of the MRI system such that (1) it is geometrically open such that a PET system can be easily integrated with sufficient transaxial and axial FOVs as shown in Fig. 1 and (2) all magnetic fields can be actively cycled on/off during the PET imaging process. This approach has been called field-cycled MRI (FCMRI) or prepolarized MRI, and has been described previously [15,16], but not in the context of system integration with PET. An FCMRI system replaces the superconducting magnet of a traditional MRI with two actively controlled resistive magnets that are switched completely on and off. One magnet provides a strong magnetic field (~1 T) that creates magnetization in the sample during the time that it is active (~1 s). The second magnet creates a weaker (~0.1 T) but homogeneous readout field that is present during the data acquisition. FCMRI still uses gradient and RF coils in the same manner as traditional MRI. Besides the relatively low fringe field, the most critical feature of the system relevant to PET/MRI is that an FCMRI can be switched from an “on” state to a completely “off” (zero field) state in a few tens of milliseconds. This implies that a completely standard PMT-based PET system could be potentially built into an FCMRI system, with the two modalities operated in an interleaved mode. The major question for this system concept is the degree to which large, rapidly changing magnetic fields affect PMT operation in terms of both short-term recovery and long-term stability.

This paper focuses specifically on characterizing the performance of traditional PMT devices as a function of applied magnetic field. For conventional PET detection based on scintillators coupled to PMTs, the static magnetic field is expected to reduce the gain of the devices. This has the additional effect of degrading the energy resolution, the time resolution, and the detection efficiency. Dynamic magnetic fields are expected to affect the recovery of normal PMT performance following magnetic field pulses, as well as to possibly degrade detector stability. Finally, a PET/FCMRI scanner was built using four commercial PMT-based PET detectors for the proof-of-principle study.

2. Methods

2.1. Evaluation of PMTs inside magnetic field

The response of two different PMTs to both static and dynamic fields was measured experimentally. For both energy and timing spectra detectors were measured as a function of static magnetic field strength. The recovery behavior in a dynamic field was measured. In all studies, the detectors were aligned perpendicular to the field direction. A NaI (TI) scintillator (diameter: ~45 mm, thickness: ~30 mm) was used. Two types of PMTs were tested: the Bicron Box and Grid 2M2/2 (hereafter referred to as the B-2M2/2, Saint-Gobain Detectors, Newbury, USA), and the Hamamatsu Photonics K.K., Japan).

The external magnetic field was produced by a pair of coils in a Helmholtz configuration with a radius of 0.16 m. The magnet had a field efficiency of 0.35 mT/A and could be driven up to a maximum of 20 A (approximately 7 mT). The PMT under test was positioned in the center of the magnet. The field homogeneity was simulated and found to vary by less than 2% over the extent of the PMT structure. For the generation of dynamic magnetic fields, the same power supply was used but was connected to the coils through a customized switching box containing a parallel IGBT (insulated-gate bipolar transistor) circuit and a capacitor bank [16]. When the current to the magnet is to be switched off, the IGBT circuit and diodes cut off the oscillations of current at a quarter of the oscillation period as the energy stored in the field is passed into a block of capacitors. The energy is then dissipated through a bank of high power resistors. This allowed the current flowing through the coils to be varied from its full value to zero in a very short time. A dB/dt of ~4 T/s (approximately 5 mT change over 1.25 ms) was generated, as illustrated in Fig. 2.

The standard NIM modules were used for measuring the energy and time spectrum inside static magnetic fields. The CANBERRA 8075 analog-to-digital converter (ADC) and CANBERRA S100 multi-channel analyzer (MCA) were used to digitize signals. For the time spectrum measurements in the static field, the PMT under test was located within the magnetic field while the other PMT detector was placed 0.4 m away. The source was located equidistant between the two detectors. This separation was necessary to ensure that the magnetic field produced by the coils at the other second detector was less than 0.5 mT.

For the energy spectrum measurements in the dynamic field, the entire configuration was the same as in the static magnetic field, except for the introduction of the switching box and trigger. The amplitude of the applied magnetic field was varied from 10 mT to 1.5 T (zero field) at a rate of 4 T/s (approximately 5 mT change over 1.25 ms) was generated, as illustrated in Fig. 2.

Fig. 1. Illustration of the proposed PET/FCMRI prototype (left) and possible data acquisition sequences (right).
Relative time in the following discussion refers to the Agilent 33250A waveform generator, triggered by the TTL output, to indicate when the unit was allowing current to flow. An example of the switching circuit provided a (transistor–transistor logic) (TTL) waveform of current through the coil and the gate input for the analog-to-digital converter (ADC) during the data acquisition.

The magnetic field was repeatedly pulsed on for 40,000. A phantom consisting of three radioactive sources was used (Fig. 3). Each is of $22\text{Na}$ radiation point sources. This was also limited to test for significant differences in the measurement variance. Relative efficiency was calculated as the number of events within plus and minus two standard deviations of the 511 keV photopeak position, after the removal of background counts. Relative efficiency was calculated as the ratio of the efficiency at any given magnetic field to the efficiency measured for zero magnetic fields. The data from the dynamic field measurements were analyzed in the same manner as the static field measurements, with the following exceptions. First, only the energy spectra were measured for the dynamic field experiments. Second, relative gain for the timing experiments was defined as the ratio of the measured gain at any particular time to the gain measured 5 ms after the end of the magnetic field pulse. For the long-term stability study, two conditions (field-on and field-off) were compared using an F-test to test for significant differences in the measurement variance.

### 2.2. Implementation of a proof-of-principle PET/FCMRI system

Four PMT-based PET block detectors with the Anger logic from a decommissioned PET scanner (CPS Accel®) were used. Each PET block consists of an $8 \times 8$ BGO ($\text{Bi}_2\text{Ge}_3\text{O}_{12}$) scintillator crystal array. The pitch size was $6.45 \times 6.45 \times 25.5 \text{mm}^3$. They were expected to exhibit the same response to both static and dynamic magnetic fields, as they have the same internal structure (linear focused) as that of the conventional PMTs studied in Section 2.1. The PET data acquisition system was built using 16 channels CAEN nuclear detection system (Viareggio, Italy).

The built partial PET ring was inserted into the gap within a prototype FCMRI system, as shown in Fig. 3. Only the polarizing magnets of the FCMRI system were in existence for this proof-of-concept study and the current was applied through polarizing magnets to simulate the FCMRI operation. When no current is applied, the PET system in PET/FCMRI is essentially a standalone PET system. The partial ring covers the full angle from $0^\circ$ to $180^\circ$ by mechanically rotating the sample holder six times.

Single-slice rebinning (SSRB) and filtered back-projection (FBP) imaging reconstruction algorithm was used for the image reconstruction [17]. The size of sinogram was 66 (angular direction) by 32 (radial direction). The radial sampling was 2.8 mm, yielding a sampling frequency of 3.57 samples/cm near the center of the field of view (FOV), which implies that the expected spatial resolution is about to be approximately 8.4 mm (three times the radial sampling). This was limited by the large size of detector used that is dedicated for human body scanner. The angular sampling was 1.72°, which is slightly coarser than commonly used in most PET systems ($\sim 1°$). This was also limited by the partial ring geometry and the large size of detector used in the study. A $32 \times 32$ image matrix was used and the pixel size was chosen to be the same as radial sampling (2.8 mm).

The total PET imaging time was 6 h (due to the weak radioactive phantom and partial ring configuration), yielding a total number of counts for each PET image of approximately 40,000. A phantom consisting of three $22\text{Na}$ radiation point sources was used (Fig. 3). Each is of $\sim 2.0\mu\text{Ci}$ activities and was separated from other two sources by 10 mm. The size of the point source itself was 1 mm. The samples were mechanically rotated six times and each rotation corresponded to 1 h data acquisition. The step size of rotation is $30^\circ$. After the PET image acquisition was studied when no current was applied to polarizing magnets, the PET image was acquired with the operation of FCMRI system. The sequence configuration is shown in Fig. 3. The magnetic field was measured to have an efficiency of 1.42 mT/A and provided up to a maximum of 43 mT for this experiment. A parallel IGBT circuit described earlier was used to generate the dynamic magnetic field waveform ($\sim 0.30\text{T/s}$). The magnetic field pulse has a period of 1.6 s and a duty cycle of 50%. The width for PET trigger was
450 ms. The trigger for PET acquisition was applied 10 ms following the moment when the field decays to zero. This was set according to the evaluation of two PMTs in the previous section, which indicates that these detectors could recover to normal performance within 2–3 ms. Other imaging parameters were exactly the same as those used for standalone PET system. The PET system performance in terms of energy resolution, time resolution, sensitivity, and spatial resolution were compared between the two configurations. Intensity profiles of three spheres in each reconstructed image were derived and each profile was fitted with a Gaussian distribution on the top a linear background. The average value of three FWHMs was used to characterize the spatial resolution for each system.

3. Results

3.1. Evaluation of PMTs inside magnetic field

The response of energy and time spectrum under two magnetic field strengths are shown in Fig. 4. The relative gain, energy resolution, relative efficiency, and time resolution as a function of
the static magnetic field strength for the two PMTs are shown in Fig. 5. The relative gain decreases monotonically with increasing magnitude of the applied field for both detectors. The relative gain falls below 0.5 for a field of approximately 2.5 mT for the H1161, and for approximately 4.6 mT for the B-2M2/2. The decrease in relative gain is attributed to the effect of the Lorentz force on the motion of the photoelectrons within the PMT structure. As the magnetic field is increased, the Larmor radius of the electron trajectory decreases, causing larger numbers of electrons to miss dynodes along the cascade stages [8,9,18].

The energy resolution degrades monotonically with increasing static magnetic field for both PMTs. The B-2M2/2 degrades by only 12% for fields up to approximately 4.2 mT, and then quickly degrades for higher field exposures. The H1161 degrades by only 4% for fields up to approximately 2.1 mT, and then degrades more rapidly with increasing fields. For a field exposure of 6.3 mT, the B-2M2/2 energy resolution has increased from 0.074 to 0.137, a degradation of 85% and the H1161 energy resolution has increased from 0.106 to 0.138, a degradation of 30%. This is also attributed to the decrease in the total number of photoelectrons hitting the anodes, corresponding to the decrease in the gain of PMTs.

The relative efficiency for static magnetic field exposure is shown in Fig. 5 for both devices. The B-2M2/2 displayed an essentially constant relative efficiency for the entire measurement range. The efficiency of the H1161 was approximately constant for fields up to 3 mT, and varied for fields above that value. A significant decrease in relative efficiency was observed for the H1161 for magnetic fields around 6.3 mT, which is possibly due to difficulty in distinguishing the 511 keV photopeak from the low-energy Compton scattering edge; therefore, some true events are treated as background counts and the apparent relative efficiency is artificially decreased.

The time resolution measured using both detectors are approximately the same at zero field (B-2M/2: 7.7 ± 0.3 ns, H1161: 7.6 ± 0.3 ns, NaI crystal used), and the resolution degrades monotonically with increasing field. The degradation for both devices is relatively slow for fields up to approximately 4.5 mT, but increases to approximately 27.0 ± 0.9 (B-2M/2) and 29.5 ± 0.9 ns (H1161) for magnetic fields around 6.3 mT. The deterioration in time resolution then increases rapidly with fields of 5 mT and higher. For a particular scintillator, the increase of the time resolution is attributed to the following two reasons: first, the time resolution is inversely proportional to the total number of photoelectrons (also known as the slope of the rising edge of the signal) [19], which decreases as the magnetic field strength increases as discussed above; secondly, the electrons in a single radiation event do not arrive at the anode at the same time. Due to the change of trajectories inside the PMT caused by the magnetic field, the transit time spread will be broadened and will also result in an increase in the overall time resolution [8].

Fig. 6 shows the relative gain and relative energy resolution of both devices as a function of time, relative to the end of the 5 mT magnetic field pulse. The most important observation here is that both devices recover to their zero-field gain and energy resolution values within 2–3 ms following the end of the ramp-down of the

![Fig. 5.](image) Relative gain (a), energy resolution (b), efficiency (c), and time resolution (d) as a function of static magnetic field strength for two PMTs.
field pulse. Furthermore, the relative gains and energy resolutions remain stable over the period following the end of the field pulse. For times before the start of the magnetic field ramp-down, the relative gains and energy resolutions of both devices are consistent with their static field values at 5 mT. At the time during which the magnetic field is changing, the device response vary between the 5 mT and zero-field values because the measurements during this interval were effectively subjected to an averaging over the changing magnetic field (i.e. convolution of different responses to various magnetic field strengths).

The stability of the PMTs gain and energy resolution is summarized in Fig. 7. These results seem to indicate that there is no significant effect on stability of the devices when subjected to repeated magnetic field pulses. An $F$-test for significant differences in variance between the field and no-field conditions was performed on the data for each device. For the relative gain, $F$ values of 1.33 for the B-2M2/2 and 0.79 for the H1161 were obtained. For the energy resolution, $F$ values of 1.06 for the B-2M2/2 and 0.93 for the H1161 were obtained. For 14 degrees of freedom, the critical $F$ value is 3.52 ($p$-value of 0.01); therefore, we conclude there is no significant difference between the variances as measured in the field versus no-field conditions for either device.

### 3.2. Implementation of a proof-of-principle PET/FCMRI system

The performance of a single PET block detector within the magnetic field sequence running was investigated, including the energy resolution, time resolution, and the flood map using the Anger logic-positioning scheme. The waveform of the magnetic field sequence and the trigger signal for the PET detection has been shown in Fig. 3. The average energy resolution over 64 crystals is $24.2 \pm 0.3\%$ (inside FCMRI) and $24.0 \pm 0.3\%$ (outside FCMRI). The average time resolution (measured against the B-2M2/2 PMT) is $9.7 \pm 0.2\,\text{ns}$ (inside FCMRI) and $9.5 \pm 0.2\,\text{ns}$ (outside FCMRI). In addition, no significant difference was observed in the flood position map between the two scenarios. These results agree with our predication, as the PMT-based detectors are able to recover to the normal performance 2–3 ms after the recovery of magnetic field demonstrated in the earlier sections.

![Fig. 6. Recovery of the relative gain (left) and the relative energy resolution (right) within the dynamic magnetic field ($-4\text{T/s}$). The definition of relative time is illustrated in Fig. 2. For both PMTs, the gain and energy resolution recover to their zero-field values within 2–3 ms following the end of the ramp-down of the field pulse.](image)

![Fig. 7. Stability of the relative gain (left) and the energy resolution (right), for situations without and with magnetic fields. The time interval between each data point is 5 min. An $F$-test shows no significant differences in variance between two conditions for both devices.](image)
The reconstructed phantom images with and without FCMRI are shown in Fig. 8. In both constructed images, the phantom consisting of three point sources of 10 mm equidistant separation can be resolved though not completely separated, which implies that the system exhibits the spatial resolution of approximately from 8 to 10 mm, which is limited by the coarse radial sampling (2.8 mm). In addition, as the phantom was located in the cylindrical sample holder made of Teflon (~20 mm thickness), the fact that no attenuation and scatter correction were made also contributed to the partial overlapping of distributions of three point sources. The streak artifacts introduced by the FBP method and poor angular sampling (1.72°) are noticeable.

The total counts and the average of full-width-half-maximum values of the transaxial spatial resolution are shown in Fig. 9. The quantitative analysis indicates that there is no statistical difference between the two configurations, which means that no degradation of PET images is introduced by the dynamic magnetic field pulses of the FCMRI system. For the standalone PET system, the count is 40,026 ± 200 and the spatial resolution is 8.67 ± 0.49 mm. For the PET/FCMRI case, the count is 39,668 ± 200 and the spatial resolution is 8.78 ± 0.33 mm. The spatial resolution here is mainly due to both the limited radial sampling (large crystal pitch) and the limited angular sampling (rotating the phantom only six times to cover 180° coverage).

4. Discussion

The goal of this work was to carefully characterize the effects of static and dynamic magnetic fields on the operation of conventional PMT detectors, and prove the feasibility of a PET/FCMRI system implementation using such detectors. If successful, such solution would be an alternative to other PET/MRI solutions based on solid-state detectors such as avalanche photodiodes (APDs) or silicon photomultiplier tubes (SiPMs).

The gains of two devices under test decrease rapidly by 70–80% for static magnetic fields up to 5 mT. Both energy and time resolution begin to suffer appreciably for magnetic field exposures around 5 mT. It has been reported that when using a 3 m long optical fiber to couple the scintillation light to a PMT device is located outside the magnetic field, the effective gain of a PET detector can decrease by 17% [20]. However, it should be kept in mind that the light loss along the transmission inside the optical fibers (i.e. loss of photons) would cause more degradation of the energy resolution, compared to the loss of photoelectrons studied in this paper (due to the Lorenz force). The reason is that for a PMT device, the fluctuation of the signal (related to the energy resolution) results mainly from the Poisson statistics of the effective light signal out of scintillation crystals (~1000 visible photons), rather than the multiplication gain that is quite high (~1e5–1e6). This also explains that at 5 mT in our study, the energy resolution maintains around 13% even when the gain decreases by nearly 80%. For a PET system, degradation of energy resolution results in the inclusion in the photopeak of more low energy scattered events. Degradation of time resolution increases the number of random coincidence events. Both of these effects result in the reduction of the noise equivalent count (NEC) and hence a degradation of PET image quality [21], though such degradation might be more tolerable for the small animal imaging versus the whole body imaging [22].
On the other hand, we did not study the performance of two PMTs within the field strength beyond 7 mT, as we believe that the degradation (such as gain, energy resolution, and time resolution) would make PMTs unable to maintain satisfactory performance for a PET system. This is in consistency with one of the earliest PET/MRI designs where PMT-based PET detectors were located at a fringe magnetic field of 5 mT [10]. For most superconducting MRI scanners used today, our results indicate that the scintillator placed inside the scanner bore must be coupled to PMTs located in the fringe field of the magnet via the optical fiber transmission technique, as deployed by other groups [10,11]. It must be noted that the distance from the imaging region to a region of fringe field less than 5 mT will be strongly dependent on: (1) the main field strength of the magnet; (2) the size of the magnet bore; and (3) the presence of active shielding of the main magnet. For example, in a typical 3.0T, whole-body actively shielded the magnet system, the distance from the isocenter to the 5 mT line is approximately 3.4 m along the length of the magnet bore (on-axis), and 2.2 m transverse to the bore direction (off-axis). However, on an unshielded 4.0T, whole-body system, the 5 mT line is approximately 8.0 m on-axis and 5.5 m off-axis. An actively shielded 9.4T small animal imaging system (33 cm magnet warm-bore), the 5 mT line is approximately 2.8 m on-axis and 1.7 m off-axis. However, the use of optical fibers several meters long will inevitably result in the light loss and degrade the performance of PET detectors. This also explains why the PMT technology is less appealing compared to APDs and SiPMs when integrating a PET system with a conventional superconducting MRI scanner.

For an FCMRI system, all of the static field effects discussed above are relevant depending on the mode of operation. In addition, a serious concern regarding the effect of rapidly changing magnetic fields on PMT performance was temporal stability. Specifically, (1) detector recovery immediately following magnetic field pulses and (2) long-term detector stability during extended trains of dB/dt pulses, are both of importance. In terms of short-term recovery following the application of rapidly changing magnetic fields, both the gain and energy resolution were observed to recover completely within 2–3 ms of the end of the pulse. It was also found that this recovery was stable over repeated sets of pulses, with no significant difference between the variances as measured in the field versus no-field conditions for either device. The temporal recovery of both standard PMT devices immediately following large magnetic field pulses indicates that an interleaved implementation of PET and an open-geometry FCMRI should be possible, which is also supported by the preliminary imaging experiments.

In this proof-of-concept study, the built PET partial ring shows no significant difference in its performance between PET/FCMRI and PET standalone configurations, including energy resolution, time resolution, flood map, reconstructed spatial resolution, and counts. Employing clinical PET detectors for the small animal system configuration (small ring diameter) is a major limitation in this study, which not only resulted in the poor spatial resolution but also made the spatial resolution quantization difficult. Nevertheless, the results support the hypothesis that an interleaved PET/MRI dual modality system using conventional PMT technology can be achieved using the FCMRI architecture. Currently, we are implementing a full-ring PET system based on detectors in the Siemens Inveon® small animal systems and we are also investigating its performance with gradient field and RF field in operation.

In terms of the operation mode, the FCMRI system would be operational for a time on the order of one second, followed by removal of all magnetic fields. Then PET data collection would take place for a time also on order of one second. This process would repeat itself for whatever time required for adequate collection of both image data sets, with each modality operating with approximately 50% duty cycle. The actual duty cycle of each modality can obviously be changed; for example, it may be determined that it is most advantageous to operate the PET for 80% of the time, and FCMRI for 20%. Taken to the extreme, it would also be possible to not interleave PET and MRI acquisition at all, but instead to acquire the entire PET data set at the start, and then we can follow with the acquisition of the complete FCMRI data set. The object under investigation would not be moved between scans, and this way we can maintain system's ability to achieve excellent spatial coregistration between the modalities. The question whether to interleave the acquisition of the two modalities would depend on the degree of temporal coregistration desired.

The primary disadvantage of the PET/FCMRI system concept is the reduction in imaging efficiency of both the PET and MRI system due to the interleaved operation, as well as the low field strength limiting the MRI performance. The primary advantage of the PET/FCMRI system is that the PET system can make use of technology maturity, high gain, and signal-to-noise ratio, low cost, and well-studied readout multiplexing scheme. In addition, as the PET detectors are located outside the gradient and RF coils, less performance degradation caused by the eddy current and RF interferences is expected. Note that it may also be possible to improve the imaging efficiency of a PET/FCMRI system through additional system modifications. Active magnetic shielding could be added to the MRI system, in the immediate vicinity of the PMT-based detectors [23]. This approach may be able to reduce the magnetic field at the location of the detectors to an acceptably low level such that simultaneous operation could be achieved. In addition, PMT devices with alternative internal dynode structures such as multi-channel plate or mesh [24] are available and can provide significantly increased tolerance to magnetic fields. These could be used in combination with the active shielding approach to further increase the efficiency of PET acquisition in a combined system.

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