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Performance evaluation of RF coils integrated with an RF-penetrable PET insert for simultaneous PET/MRI

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Abstract

Purpose—An “RF-penetrable” PET insert that allows the MR body coil to be used for RF transmission was developed to make it easier for an existing MR center to achieve simultaneous PET/MRI. This paper focuses on experiments and analyses to study PET/RF coil configurations for simultaneous PET/MR studies.

Methods—To investigate the appropriate RF coil design, a TX/RX birdcage coil and an RX-only phased-array coil (TX from body coil), both fitting inside the PET ring were built and characterized. For MR performance evaluation, B_1 field uniformity and MR image signal-to-noise ratio (SNR) were calculated. PET photon attenuation due to each coil was studied via CT-based attenuation maps and reconstructed PET images.

Results—When using the RX-only phased-array coil (TX from body coil), compared to the TX/RX birdcage coil, the B_1 field uniformity and the MR image (GRE and FSE) SNR increased by $2.4 \pm 4.8\%$, $386.1 \pm 62.3\%$, and $205.0 \pm 56.5\%$, respectively. Although some components of the coil were distributed within the PET FOV, no significant PET photon attenuation was shown in the CT-based attenuation map and reconstructed PET images.

Conclusion—RF coil configurations for an RF-penetrable PET insert for simultaneous PET/MRI were studied. The RX-only phased-array coil (TX from body coil) outperformed the TX/RX birdcage coil with improved MR performance as well as negligible PET photon attenuation.

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PET/MR; PET insert; RF coil; RF-penetrable; instrumentation

Introduction

The integration of positron emission tomography (PET) and MRI has shown to be an important tool for simultaneously acquiring complementary anatomical and molecular information of diseases^{1–4}. As an alternative to the high cost of the commercially available integrated PET/MR systems, PET inserts have been developed by several research groups^{5–8}. While most of the groups are developing PET inserts with dedicated transmit/receive (TX/RX) birdcage coils placed inside the PET insert, we have developed an “RF-penetrable” PET insert which enables the use of the built-in body coil as the RF transmitter; the RF field from the body coil transmits through the PET insert and creates a uniform B₁ field inside the field-of-view (FOV)^{9,10}.

Recently, with availability of clinical simultaneous PET/MRI systems^{11–13}, many neurologic applications, such as, brain tumors, neuro-degeneration, stroke/cerebrovascular disorders, epilepsy, and neurobiology of brain activation, have been studied extensively. PET/MRI allows improved understanding of brain physiology and function and opens new insights into the pathophysiologic inter-relation between different parameters involved in disorders of the central nervous system^{14,15}.

For high-performance “MR-only” clinical neurology studies, the built-in body coil is used as an RF transmitter due to a high B₁ field uniformity; and a high-performance multichannel array coil is used as an RF receiver for high sensitivity and parallel imaging features, allowing accelerated image acquisition and advanced image reconstruction techniques¹⁶. A phased-array RX-only coil is composed of multiple RX-only surface loop coils and is known to provide high signal-to-noise ratio (SNR) in regions of the FOV closer to the elements, but lose sensitivity away from the element position. However, phased-array coils suffer from poor transmit field uniformity and, furthermore, are very complicated to operate as a transmitter; these factors limit utilizing the phased-array coil as a TX or in a combined TX/RX coil^{17,18}.

When a PET system with an RF coil is inserted into an MR system for simultaneous PET/MR imaging, the MR performance as well as the PET performance need to be carefully considered for the coil design. A PET insert, which is a combination of electromagnetic and mechanical systems within the MR bore, can affect the MR performance and high-density components of the RF coil can degrade the PET performance through photon attenuation and scatter.

In conventional PET insert designs^{5–8}, the RF field from the body coil cannot penetrate through the PET insert into the FOV, therefore, combined TX/RX birdcage coils are brought inside the PET ring^{19–24}, which increases PET photon attenuation and reduces the B₁ field uniformity (compared to using the body coil as the RF transmitter). An alternative is to employ a separate TX birdcage coil and a RX-only array coil inside the PET ring^{25,26}, but as

two separate coils are stacked within the PET FOV, this results in more PET photon attenuation, smaller useful FOV, and, again, lower B_1 field uniformity compared to using the body coil as a transmitter. Another approach is to build a combined TX/RX phased-array coil with a Butler matrix circuit^{17,18}, which, compared to the TX/RX birdcage coil, improves the MR receive sensitivity and image SNR; but likewise, the close proximity reduces B_1 field uniformity, and increases PET photon attenuation and complexity.

In our previous work describing an RF-penetrable PET insert^{27–30}, we employed the body coil as both TX/RX for purposes of studying “RF penetrability”, and have demonstrated that the RF field could be both transmitted and received through the PET ring with ~3.5 dB attenuation for both directions. We also showed that this configuration allows a uniform B_1 field inside the useful PET FOV. The B_1 field attenuation can be compensated with preserved B_1 uniformity by increasing the transmit gain (TG) of the body coil; however, the receive attenuation and longer distance to the body coil result in sensitivity loss that cannot be compensated for with the body coil as the RF receiver.

In this work, we have designed two custom RF coils that are integrated inside the PET ring, including 1) a TX/RX birdcage coil and 2) a RX-only phased-array coil, and compared the MR performances and the PET photon attenuation to find the optimal configuration for high B_1 field uniformity and high SNR MR images, as well as, low PET photon attenuation. Although some of the design parameters are specific to our PET insert design, many of the design concepts will be useful to other simultaneous PET/MR system designers as well as developers of RF coils.

Methods

A. RF-penetrable PET Insert

The design and the basic performance of the RF-penetrable PET insert have been presented in previous work^{9,10,30}. 16 PET detector modules are arranged in a ring pattern with 1 mm inter-modular gaps (Figure 1). The inner/outer diameters of the PET ring prototype are 32/40 cm, respectively, and the PET sensitive axial FOV, where scintillation crystals are present, is 2.8 cm, comprising eight crystal rings with 3.5 mm pitch. Each PET detector module consists of a shielding cage that encapsulates the PET front-end electronics, including the scintillation crystals and silicon photomultipliers that are placed at one edge of the PET detector modules. The total of 128 PET signals per module are multiplexed by a compressed sensing method³¹, with a multiplexing ratio of 8:1. The resulting electrical signals are converted to optical signals by driving vertical cavity surface emitting lasers and transmitted down 16 optical fiber channels. The shielding cages are trapezoidal boxes with 5 cm mid-segment width, 4 cm height, and 20 cm length built from 17.5 μ m thick copper plated FR4s (Figure 1). The PET detectors are powered by non-magnetic batteries (Lithium polymer and Lead acid) as opposed to electrically grounded power supplies to avoid a common electrical ground between the MRI and the PET systems.

Compared to the conventional PET inserts^{19–24}, the RF-penetrable PET insert architecture has several distinguishing aspects. Rather than requiring the RF transmitter to be inside the PET ring, the 1 mm inter-modular gaps allow the RF fields from the built-in body coil to

penetrate into the PET FOV. This RF-penetrability facilitates the concept of an insertable and removable “PET/RF coil” comprising an RF-penetrable PET ring integrated with an RF coil inside the PET ring. In principle, this PET/RF coil can be placed inside the bore of any MR system to achieve simultaneous PET/MR imaging using the RF transmit field from the built-in body coil.

B. RF coils combined with the RF-penetrable PET insert

To investigate the appropriate RF coil design for our RF-penetrable PET insert, various PET/RF coil configurations were built and evaluated. Two different coils, a high-pass quadrature birdcage TX/RX coil³² and an 8 channel RX-only phased-array coil³³, were built, placed inside the RF-penetrable PET insert, and their performance were compared with a commercial 24 cm diameter GE 8 channel RX-only phased-array coil (Brain Array Coil, GE Healthcare) without the PET insert. To align the axial center of the PET sensitive FOV (2.8 cm scintillation crystal, gray in Figure 2(A)) to the MR isocenter, the PET ring was axially shifted. Then, the RF coil was axially aligned at the isocenter of the MR system. Although the PET ring was axially shifted with respect to the RF coil, it was transaxially centered with the RF coil and the MR system axis (Figure 2(A)).

For detailed understanding of the coil design, we have provided the following details that may be useful to other designers of RF coils, especially those for simultaneous PET/MR systems.

B.1. Custom birdcage TX/RX coil—A custom high-pass birdcage coil to be placed inside the PET insert was built (Figure 2 (B)), comprising two end rings and 16 rungs (19 cm in length) copper taped onto a clear cast acrylic tube (29 cm outer diameter, 3.175 mm thick, 30 cm length). The placement of the rung avoids the PET ring inter-modular gaps to minimize further RF attenuation.

B.2. 8 channel phased-array RX-only coil—Similarly, an 8 channel RX-only phased-array coil was designed to be placed inside the RF-penetrable PET insert (Figure 2 (C)). The phased-array coil was 19 cm in length and 29 cm in outer diameter. Each loop was copper taped over 14 cm (transaxial) $\times$ 19 cm (axial) with approximately 10% of the loop area overlapped with the adjacent loop for decoupling by minimizing the mutual inductance between two neighbor loops.

B.3. Details of coil construction

Pattern layout: The patterns for the birdcage and the phased-array coil were printed and laid out inside the acrylic tube, then copper tape (6.35 mm width, 36 μ m thick foil with 53 μ m conductive acrylic adhesive) was placed on the outer side of the tube following the pattern.

Tuning to the Larmor frequency (127.8 MHz): For tuning the coils to the Larmor frequency of the 3T MR system, the coil was positioned at the isocenter of the PET sensitive FOV (Figure 2(A)) and then the resonant frequency was iteratively tuned to the center frequency with an inductive tuning wand. Non-magnetic multilayer chip capacitors and

variable capacitors (Voltronics, Knowles Capacitor Inc.) were used for rough- and fine-tuning the loop resonance to the Larmor frequency, respectively. For the high-pass birdcage coil, 16 capacitors were uniformly distributed to each ring and no capacitors were on the rungs. The phased-array coil comprised approximately a factor of 3 more capacitors and some were distributed within the PET FOV as well.

Impedance matching: Both phased-array coil and birdcage coil were impedance matched to a 50 Ω source by employing low-pass capacitive dividers at feed ports.

Feedline cables and coil interface: Each coil element was connected to a lumped LC lattice balun (90° phase shift) to mitigate ground loops and an additional quarter wave RG58 cable (90° phase shift) to make a total of a half wavelength of the RF pulse. The cables were connected to the MRI system using a custom coil interface containing 8 commercial low noise preamplifiers (GE healthcare, Aurora, OH) and 2 bias tees, one for preamplifier power and the other for TX/RX bias per channel.

Element decoupling: For the phased-array coil, element decoupling was achieved by overlapping elements to minimize mutual inductance³³ and by the use of low impedance preamplifiers with 180° phase shift respect to the feed port³³.

Low-impedance preamplifier decoupling: All coils are connected to low input impedance preamplifiers to reduce coupling between loops that are not overlapped. The low-impedance can be presented to the coil port either by a direct connection or through a (180×n)° transmission line. We employ a lumped element lattice balun (90° phase shift) to reduce common mode currents and a quarter wavelength cable (90° phase shift) is added for a total of 180° phase shift. This results in coils that behave independently although all coils are simultaneously active and receiving MR signals. As the input impedance of the preamplifier decreases, the decoupling of coupled array elements improves³³.

Passive transmit blocking: To prevent potentially large currents induced during the RF excitation process, a passive protection circuit was applied to the phased-array coil, using a conjugate inductor and antiparallel diode pair.

Active transmit blocking: During RF field transmission, the RF receiver needs to be detuned to avoid the RF induced current damaging the RX circuit. Active transmit blocking was achieved on the phased-array coil by using a series inductor conjugate to the feed port and a shunt PIN diode. During the RF transmit pulsing, the PIN diode is forward biased and the activated parallel LC resonant circuit provides high impedance within the coil loop.

C. MR performance evaluation

The RF-penetrable PET insert combined with different coil types were tested inside a 3T MRI (Discovery 750, GE Healthcare). For the custom RX-only phased-array coil and GE 8 channel RX-only coil configurations, the MR built-in body coil was used for RF transmission. An MR agar phantom (17 cm diameter) was placed at the isocenter of the RF-penetrable PET/RF coil insert for coil analysis. For MR performance evaluation experiments, the configurations listed in Table 1 were tested.

C.1. B₁ field uniformity—To compare the B₁ field uniformity inside the PET ring, a B₁ field map was acquired using the double angle method³⁴. When the PET was present inside the MR bore, the auto pre-scan feature was applied to automatically control the TG of the built-in body coil to compensate for any B₁ field magnitude loss due to the PET insert. The center slice of a total of 7 slices of the B₁ map (slice thickness: 5 mm, space between slices: 10 mm), which corresponds to the 2.8 cm axial center of the PET sensitive FOV (Figure 2), was analyzed. From the B₁ field map, the flip angle and the uniformities were acquired from the equation below.

$$Uniformity(\%) = \left(1 - \frac{S_{max} - S_{min}}{S_{max} + S_{min}}\right) \times 100,$$

where S_{max} and S_{min} are the maximum and minimum values of B₁ flip angles of the selected ROI, which was the center 70% of the phantom for each slice.

C.2. MR image comparison—MR images of an agar phantom were acquired with the common RF pulse sequences gradient echo (GRE) and fast spin echo (FSE). Again, when the PET insert was present inside the MR bore, the auto pre-scan feature was turned on to compensate for any B₁ field attenuation. The details of the RF sequence parameters are listed below in Table 2.

As processing the raw MR data for image reconstruction biases the noise statistics in a different manner, a pixel-wise SNR mapping approach³⁵ was used for MR image SNR performance analysis. The noise-only data was acquired with the RF pulse turned off and the signal was scaled with the estimated noise from the noise covariance. Then, SNR maps were computed with the noise covariance weighted root sum-of-squares from individual channel images. For the single-channel quadrature birdcage coil, the noise weighted array combining step was simply a scaling by the noise standard deviation calculated from the noise-only data. The central 75% signal area in the center 6 slices of the SNR map that matches the 2.8 cm axial PET FOV were selected for analysis.

D. PET photon attenuation evaluation

To study the PET photon attenuation due to the coil inside the FOV, CT-based attenuation maps and reconstructed PET images of a 6.5-cm diameter resolution phantom were studied with different coil configurations (Table 1. (i), (iii), and (iv)).

D.1. CT-based attenuation maps of each coil—CT-based attenuation maps were acquired using a clinical PET/CT scanner (GE Discovery 690). The CT images were acquired using LightSpeed VCT with 64 slices (slice thickness: 5mm). To evaluate the 511 keV photon attenuation of each coil, the coil was placed at the isocenter of the CT scanner and the acquired CT images in Hounsfield units (HU) were converted to photon attenuation coefficients at 511 keV using a bilinear transformation^{36,37}. HU obtained for various reference tissue substitutes were compared with the known linear attenuation values at 511 keV.

D.2. PET image reconstruction with different coil configurations—For each configuration, a 6.5 cm diameter resolution phantom that consists 5.2 mm, 4.2 mm, 3.2 mm, and 2.8 mm diameter hot rods and 4.2 mm diameter cold rods with no warm background (Figure 3) was scanned. The spacing between the same-sized rods are the same as the diameter of the rod. The PET images were iteratively reconstructed using 3D list-mode OSEM (ordered subset expectation maximization) implemented on GPU (graphics processing unit) for acceleration³⁸. Reconstruction was performed with 10 subsets and 8 iterations with $70 \times 70 \times 28$ (1 mm $\times$ 1 mm $\times$ 1 mm) voxel matrix. An energy window of 410-610 keV and a coincidence time window of 10 ns were applied for acquisition and a Gaussian post-filter was applied to each transaxial slice. In addition to the resolution phantom scan, a normalization scan was acquired using a uniform cylinder with the same dimensions as the resolution phantom (Figure 3. Right). This normalization phantom inherently includes attenuation correction (AC) by assuming similar attenuation coefficients of the plastic and water. All acquired lines-of-response (LORs) in the normalization scans were back projected to generate sensitivity maps and correct for normalization and attenuation. This sensitivity map is integrated into each iteration in the reconstruction algorithm³⁸.

Each reconstructed image was analyzed in terms of contrast-to-noise ratio (CNR), defined as follows³⁹:

$$CNR = \frac{\text{mean}(ROI_{rod}) - \text{mean}(ROI_{background})}{\sqrt{\text{std}(ROI_{rod})^2 + \text{std}(ROI_{background})^2}}$$

For the CNR measurements of hot rods, the background ROIs were selected as the reconstructed cold background regions, and vice versa for the cold rod measurements. CNR for the cold rods is negative by definition, therefore, the absolute value was plotted for ease of display.

RESULTS

A. MR performance evaluation

A.1. B₁ field uniformity—The flip angles acquired from all configurations were nearly 90°; although configurations (iii) and (iv) deviated from 90° away from the 2.8 cm axial FOV PET, the flip angles around the sensitive PET FOV, where the PET scintillation crystals are located, were comparable (less than 6%) between all configurations (Figure 4). However, the standard deviation, which was calculated from different axial slices, showed significantly higher values when the PET insert was present.

Even though there were some B₁ field distribution discrepancies along the axial slices between PET/RF coil configurations, the B₁ field uniformities at the axial center slice, which also corresponds to the axial center of the PET sensitive FOV, RF coil and MR system, were all comparable with values of 92.5%, 92.2%, 91.0%, 92.9% for configurations (i, ii, iii, iv), respectively.

A.2. MR image comparison—The absolute mean SNR values for each configuration have been added to the Supporting Table S1 in the Supplementary material. As expected, the SNR results acquired with the built-in body coil as a receiver (configuration (i)) showed the worst SNR values of 155.2 ± 5.5 and 121.5 ± 6.4 for GRE and FSE sequences, respectively (Figure 5). Compared to the conventional PET/RF coil configuration, which places a TX/RX birdcage coil inside the PET (configuration (iv)), the RF-penetrable PET insert with the RX-only phased-array coil (body coil as transmitter) (configuration (iii)) showed $386.1 \pm 62.3\%$ and $205.0 \pm 56.5\%$ higher SNR for GRE and FSE sequences, respectively. Although configuration (iii) showed the best performance for the case of the PET insert present, the SNR was $4.0 \pm 10.6\%$ and $23.5 \pm 17.9\%$ lower and transaxially asymmetric compared to the typical MR-only configuration (without PET), which employs the built-in body coil as the RF TX and the commercial 8 channel phased-array coil as the RF RX (configuration (ii)). In configuration (iv), the FSE MR image showed some phase-related artifacts with phase shifted images overlapped (supporting figure S1). Details of the commercial 8 channel phased-array coil compared with the custom 8 channel phased-array coil, as well as other results, will be discussed in the Discussion section.

B. PET photon attenuation evaluation

B.1. PET photon attenuation properties of the RF coils under study—The linear attenuation coefficients at 511 keV of both the TX/RX birdcage and the phased-array coils were processed from the acquired CT images (Figure 6). The highest attenuating components were the SMA cable connectors, preamplifiers and variable capacitors with the maximum value of 0.27 cm^{-1} , which corresponds to a higher attenuation coefficient than dense bone; however, most of these components were at least $\sim 8 \text{ cm}$ away from the axial center of the PET FOV. At the PET sensitive axial FOV, there were only thin acrylic tube with 0.09 cm^{-1} attenuation coefficient, corresponding to an attenuation coefficient of soft tissues, and copper tape with negligible attenuation. For the phased-array coil, more components were arranged on the coil, but the high-density ceramic chip capacitors were sparsely distributed and rarely located within the sensitive PET axial FOV.

B.2. PET image reconstruction with different coil configurations—Figure 7 shows the sum of all reconstructed PET image slices of the resolution phantom and the CNR of each rod with different coil configurations. Each image was reconstructed with ~ 10 million LORs. All cold and hot rods were resolvable and all of the features were visible. For each rod size, the CNR values for different coil configuration showed no significant difference (Table 3).

DISCUSSION

An RF-penetrable PET insert for simultaneous PET/MRI that allows the MR built-in body coil to transmit the RF field through the PET ring into the imaging FOV has been built (Figure 1). This facilitates the concept of a removable PET/RF “coil” that could in principle be inserted into any MR system. Two RF coils were manufactured to investigate potential RF coil configurations for the RF-penetrable PET insert: a TX/RX birdcage coil, which is

used by most conventional PET inserts, and a RX-only phased-array coil, which is most practical when using the built-in body coil for RF transmission.

Due to the axial asymmetric positioning of the scintillation detectors inside the shielding cages, ~40% of the RF coil extended out axially from the PET (Figure 2) and resulted in axially non-uniform B_1 field and MR image performance. During the auto pre-scan stage, the TG of the MR system is set based on the axial center slice observation, which is aligned with the center of the PET sensitive FOV. This leads to erroneous flip angle distribution for axial positions outside the shielding cages of the PET insert. The B_1 field amplitudes in configurations (iii) and (iv) in Figure 4 show an asymmetric flip angle distribution with over-flipped spins outside the PET ring axial FOV, resulting in a larger standard deviation over all slices. Similarly, a phase-related artifact was observed in FSE images due to phase overlapping problems of over-flipped spins (Figure 5 and Supplemental material). FSE generally utilizes multiple RF pulses where the errors from the axial B_1 inhomogeneity aggregate, whereas a GRE image only uses a single RF pulse. The axial B_1 uniformity can be improved as more portion of the RF coil overlaps with the PET ring; the maximum B_1 uniformity will be derived when the axial length of the RF coil matches the PET ring. However, in realistic situations, considering the extra components of the PET ring and the patient's head/neck, the PET scintillation crystals typically have to be placed at the one edge of the PET ring closer to the patient's head/neck. Therefore, a perfect overlap between the RF coil and the PET ring is challenging.

In addition to the B_1 field uniformity drop from the axially asymmetric design, the presence of PET ring (of shielding cages) itself reduced the B_1 field uniformity due to the field-distorting effect of the PET insert (Figure 4). When transmitting the RF field from the built-in body coil, as it has to divert through the inter-modular gaps as well as the open ends of the PET insert, the mean B_1 field uniformity dropped from 93.9% to 90.7% (3.4% relative mean change) compared to the “no PET” case (Figure 4). Furthermore, the custom TX/RX birdcage coil, which has a smaller diameter, showed a mean uniformity drop from 93.9% to 88.6% (5.6% relative mean change) compared to the “no PET” case (Figure 4). Of the two different sources (field-distorting effect and small diameter geometry) of B_1 field non-uniformity, the uniformity drop resulting from the smaller diameter coil was worse. However, even with a reduced uniformity, the diminished B_1 field uniformities were not significant and were still both higher than 90% at the axial center of the PET sensitive FOV.

To facilitate the comparison of the custom-built RF coils to a “gold standard”, a commercial 8 channel phased-array coil, which is commonly used in MR-only brain imaging, was used as a reference coil. The SNR (GRE and FSE) and B_1 field uniformity acquired with the custom 8 channel phased-array RF coil were $4.0 \pm 10.6\%$, $23.5 \pm 17.9\%$, and $2.4 \pm 4.8\%$ worse compared to the commercial 8 channel phased-array coil, respectively (Figures 4 and 5). However, this reduced performance is understandable since (1) the design of the commercial array coil is a well-studied industry product with careful selection of materials, (2) the smaller diameter of the commercial coil fits the head in closer proximity, and (3) the PET insert was not present for any of the studies with this commercial coil. When a coil is placed within a PET ring, the PET ring as a shield is loading the coil and reduces the coil's quality factor, yielding a lower efficiency of the coil. The commercial 8 channel array coil also has

integrated low-noise preamplifiers very near each coil element whereas the custom coil had quarter wavelength lumped LC baluns and another quarter wave of cable before the preamplifiers. Further optimization with our coil design will likely lead to a comparable MR performance compared to the commercial array. The slightly reduced B_1 field uniformity of the custom phased-array RF coil compared to the commercial coil seen in Figure 4 can be explained by the axially asymmetric positioning described earlier.

In addition to the limitations of the manufacturing process and the axially asymmetric positioning between the coil and the PET ring, the custom-built phased array coil showed transaxially asymmetric SNR maps (Figure 5). This could be due to an imperfect centering of the coil in the PET ring or the phantom in the coil, which is the dominant effect over the coil loading effect. In fact, the phantom does not load the coil as much as a human head does; therefore, when a real patient head is inserted, the coil loading of a human head (symmetric effect) would dominate and lessen this asymmetric effect. Furthermore, the SNR drop between configurations (either between (ii)-(iii) or (iii)-(iv)) was larger for the FSE sequence than for the GRE sequence. This is likely due to the FSE sequence utilizing multiple RF pulses while the GRE sequence excites only one pulse. In the multiple RF pulse process, any errors (i.e., the transaxially asymmetry and/or axial B_1 inhomogeneity) of the FSE sequence aggregates and leads to worse uniformity and SNR performance especially in the perimeter of the phantom compared to that of the GRE sequence.

Although echo planar imaging (EPI) sequence is a common sequence for brain imaging, it was not performed in the study. As presented from earlier publications^{10,29,30}, the intense gradient field slew rate of EPI sequence induced eddy current on the solid copper shielding cage of the RF-penetrable PET insert and suffered from the Nyquist ghosting artifact. To allow the PET ring operable with EPI, various designs and materials for the shielding cage with sufficiently high shielding effectiveness and low induced eddy current have been investigated⁴⁰. This alternate shielding configuration will allow simultaneous PET/MR studies during EPI sequence.

The PET photon attenuation of the RF coil components was analyzed through the CT-based attenuation maps (Figure 6) and the CNR analyses from the reconstructed PET images (Figure 7 and Table 3). Even though more high-density components were in the custom phased-array coil, most of these components were located axially away from the PET sensitive FOV. Only low-density acrylic tubes, copper tape and small capacitors were arranged at the PET sensitive FOV (2.8 cm axial center) and the effect of PET photon attenuation was negligible between the custom birdcage coil and the phased-array coil (Figure 6). However, the standard CT-based attenuation coefficient measurement used in this study is established for human tissues, which could lead to erroneous estimation for high-density materials³⁶, such as, beam hardening artifacts resulting in an overestimation of the attenuation coefficients. To improve the attenuation measurement for more accurate evaluation, either a 511 keV transmission source can be used or a CT-based attenuation measurement for non-tissue high-density materials needs to be developed.

In the reconstructed PET images, $1\text{ mm} \times 1\text{ mm}$ transaxial pixel size was used to capture the features of the small rods in the resolution phantom. Normalization correction was

performed using a uniform cylinder with the same dimension as the resolution phantom. This approach inherently includes AC but has the disadvantage of exposing the LORs to unequal amounts of activity; the LORs that intersect closer to the center of the FOV see more activity, which results in radial non-uniformity. To correct for this, a radially dependent sensitivity factor based on the phantom geometry was applied to the image reconstruction process. However, even with the sensitivity correction factor, the inaccuracy in normalization and AC contributed to artificially high background signals at the radially outer edge of the reconstructed phantom image (Figure 7). Alternative normalization source geometries are under investigation. In addition, an appropriate AC for RF coils needs to be implemented as attenuation of PET photons can lead to the underestimation of tracer uptake and to image artifacts⁴¹. Since the coil retains its position respect to the PET FOV, a fixed template attenuation map can be generated, stored, and used in the PET reconstruction whenever the coil is used in the scan.

As shown throughout this paper, combining two independent imaging modalities requires careful design considerations. Currently, there are two commercial simultaneous PET/MR systems for whole-body imaging, the GE Signa PET/MR^{11,12} and Siemens Biograph mMR¹³. One major difference between these and the PET inserts that is not performance-related would be the cost. The commercial whole-body PET/MR systems cost ~\$5~6 million, not including room renovation cost, which limits wide dissemination. In contrast, a PET insert, which can be easily inserted into an existing MRI system with a much less cost burden, would increase the dissemination. Apart from the economic analysis, their utility in clinical applications can be compared. The integrated whole-body PET/MR systems obviously have advantage of a dedicated system. However, for neurology studies, a brain PET insert, which has a smaller diameter, can have better photon sensitivity and higher spatial resolution (<3 mm) compared to the PET in the integrated whole-body PET/MR systems (4~5 mm). Finally, we note that the RF-penetrable PET insert technology and concept can also be scaled up to a whole-body ring diameter.

CONCLUSION

RF coil configurations for an RF-penetrable PET insert for simultaneous PET/MRI were studied. Conventional PET inserts acquire MR images from TX/RX birdcage coils inside the PET insert. However, the RF-penetrable PET insert enables the use of the built-in body coil for RF transmission and a high-performance RX-only phased-array coil inside the PET ring. The custom phased-array RX-only coil (with body coil as TX) outperformed the combined TX/RX birdcage coil in SNR performance by $386.1 \pm 62.3\%$ and $205.0 \pm 56.5\%$ for GRE and FSE images, respectively, with comparable B_1 field uniformity and PET image quality. Hence, the custom phased-array RX-only coil with an RF-penetrable PET insert is expected to enable high-performance clinical and research applications for simultaneous PET/MRI brain imaging.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.

The RF-penetrable PET insert on the patient bed of a 3T MR system. Each of the 16 detector modules comprising the 32 cm inner diameter PET ring are shielded inside 17.5 μm thick copper cages that extend 20 cm axially. The entire insert is powered by batteries. The PET electrical signals are multiplexed and converted to optical signals, which are taken out of the MR system through 20 m length optical fibers.

Figure 2.

(A) Schematics of the simultaneous PET/RF coil insert placed inside an MR system. The PET scintillation crystals are placed at the axial edge of the PET ring prototype, which is placed at the MR system isocenter, therefore, some part of the RF coil (19 cm length, 29 cm diameter) protrudes axially from the PET ring. Photos of a custom (B) quadrature TX/RX birdcage coil and (C) 8 channel RX-only phased-array coil built for the studies reported in this paper.

Figure 3.

Schematic of the resolution and normalization phantom. The resolution phantom consists of 5.2, 4.2, 3.2, and 2.8 mm diameter hot rods and 4.2 mm cold rods in a 6.5 cm diameter cylinder. The gap between rods is the same as the diameter of the rod. The normalization phantom is a uniform cylinder phantom with the same dimension as the resolution phantom.

Figure 4.

The B₁ field maps (center slice), axial slice amplitudes (flip angles) and field uniformities are plotted for all configurations.

Figure 5.

The center slices of GRE and FSE sequences MR images and SNR maps were plotted for all configurations.

Figure 6.

CT-based attenuation maps of the custom TX/RX birdcage coil and 8 channel phased-array RX-only coil. The attenuation maps that align with the sensitive axial FOV of the PET system are shown in the red-dotted box.

Figure 7.
Sum of all 28 reconstructed PET image slices and associated CNR of the rods in the sum image for each coil configuration.

Table 1

PET/RF coil configurations

Configuration	PET configuration	RF TX	RF RX
(i)	Without PET	Body coil	Body coil
(ii)	Without PET	Body coil	GE RX-only phased-array coil
(iii)	With PET	Body coil	Custom RX-only phased-array coil
(iv)	With PET	Custom birdcage coil	Custom birdcage coil

Table 2

MR pulse sequence parameters

Sequence	TR (ms)	TE (ms)	Voxel size (mm ³)	Matrix size	Flip angle (°)	Number of average (NEX)	Echo train length (ETL)	Pixel bandwidth (Hz/pixel)
Gradient echo (GRE)	500	4.9	0.86×0.86×5	256×256	90	1	1	244.141
Fast spin echo (FSE)	3000	62.8	0.86×0.86×5	256×256	90	1	12	122.109

Table 3
Contrast-to-Noise ratio of reconstructed PET images with different coil configurations

Rod size	Hot rods				Cold rods
	5.2 mm	4.2 mm	3.2 mm	2.8 mm	4.2 mm
No PET	2.95±0.48	2.34±0.48	2.00±0.46	2.61±0.59	1.58±0.39
Birdcage	2.50±0.47	2.25±0.49	2.32±0.63	2.39±0.87	1.78±0.62
Phased-array	2.22±0.74	2.16±0.77	2.59±1.03	2.36±0.85	2.24±1.19