Design, evaluation and comparison of endorectal coils for hybrid MR-PET imaging of the prostate

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Keywords: MR PET, PET-MRI, quadrature coil, endorectal coil, prostate imaging, magnetic resonance imaging

Abstract
Prostate cancer is one of the most common cancers among men and its early detection is critical for its successful treatment. The use of multimodal imaging, such as MR-PET, is most advantageous as it is able to provide detailed information about the prostate. However, as the human prostate is flexible and can move into different positions under external conditions, it is important to localise the focused region-of-interest using both MRI and PET under identical circumstances. In this work, we designed five commonly used linear and quadrature radiofrequency surface coils suitable for hybrid MR-PET use in endorectal applications. Due to the endorectal design and the shielded PET insert, the outer face of the coils investigated was curved and the region to be imaged was outside the volume of the coil. The tilting angles of the coils were varied with respect to the main magnetic field direction. This was done to approximate the various positions from which the prostate could be imaged. The transmit efficiencies and safety excitation efficiencies from simulations, together with the signal-to-noise ratios from the MR images were calculated and analysed. Overall, it was found that the overlapped loops driven in quadrature were superior to the other types of coils we tested. In order to determine the effect of the different coil designs on PET, transmission scans were carried out, and it was observed that the differences between attenuation maps with and without the coils were negligible. The findings of this work can provide useful guidance for the integration of such coil designs into MR-PET hybrid systems in the future.

1. Introduction
Following recent advances, increased interest in the use of simultaneous MR-PET hybrid scanners (Quick et al 2013, Lindemann et al 2018, Shah 2019) has heralded the possibility of obtaining images with the increased sensitivity and specificity of PET together with the superior soft-tissue contrast of MRI to localise and identify prostate tumours from other lesions (Rosenkrantz et al 2015, Eiber et al 2016, Lindenberg et al 2016, Muehlematter et al 2018). As a stand-alone modality, various MR techniques can be used for the detection of prostate cancer lesions with the information obtained then being used for histological verification of suspicious lesions (Hegde et al 2013). However, the use of MR for the detection or exclusion of prostate cancer is unreliable, especially in the case of well-differentiated tumours, small cancerous foci or in the presence of benign enlargement as well as inflammatory or fibrotic changes. Comparatively, recent advancements in molecular PET imaging using prostate cancer-specific tracers against prostate-specific membrane antigen are set to improve prostate cancer diagnosis—especially when combined with MR (Hofman et al 2018).

As the human prostate is relatively flexible and can be distorted and deformed under external conditions, such as patient position or the insertion of a device (Kim et al 2005, Noworolski et al 2008), it is important to
localise the targeted imaging region under the same circumstances when imaging using both MRI and PET, thus making the use of simultaneous MR-PET imaging desirable. Furthermore, as MR and PET represent imaging modalities based on different physical and biological principles, an additional benefit is expected if both modalities are combined.

Since the prostate is positioned in a deeper region of the body (Pinkerton et al 2007), it is difficult to approach using a sensitive local MRI coil, making the acquisition of high-quality images challenging. Consequently, for MR imaging, an endorectal MRI radiofrequency (RF) coil is commonly used, either alone or together with other receive arrays, to image the prostate. Imaging in this way affords increased signal-to-noise ratio (SNR) and better spatial resolution. Moreover, a number of studies have demonstrated improvements in the detection quality and localisation accuracy using the endorectal type of coils relative to other methods (Sosna et al 2004, Beyersdorff et al 2005, Heijmink et al 2007, Yakar et al 2011, Turkbey et al 2014, Shah et al 2015, Costa et al 2016, Gawlitza et al 2017, Martin et al 2018, O’Donohoe et al 2019).

Similarly, from the perspective of PET imaging, conventional and external PET-ring geometries suffer from two significant limitations for prostate imaging—(a) severe attenuation and scatter of photons emitted from the centrally-located prostate, and (b) poor spatial resolution in PET images reconstructed from external ring scanners. Improvements in sensitivity, resolution and efficiency can be achieved by integrating an additional, high-resolution miniature PET detector located close to the prostate in conjunction with external standard resolution detectors (Huh et al 2006, Brzeziński et al 2014, Grkowsky et al 2015, Garibaldi et al 2017). It has been demonstrated in various studies that this asymmetric PET configuration provides higher spatial resolution and magnifies the focused region-of-interest (ROI) (Huh et al 2006, Brzeziński et al 2014, Grkowsky et al 2015). The high-resolution PET detector unit containing the RF coil elements can be inserted into the patient in the same way as the endorectal coil.

Despite the clear advantages of using hybrid MR-PET systems, there are technical challenges to be taken into account. One such challenge is the bi-directional interference between PET and MRI (Akram et al 2017, Sander et al 2013), which requires the complete shielding of the inserted PET system. Due to the space limitations imposed through the use of an endorectal PET insert, the shield on the detector is placed close to the RF coil, and consequently, the efficiency of the RF coil is significantly affected, leading to considerable SNR loss. Another consideration is that as the prostate is typically accessed using an endorectal insert via the rectal wall, so the RF magnetic ($B_0$) field in the outer region of the coil is used for imaging. In addition, as tilting occurs due to the orientation of the prostate or position of the patient on the bed, it is not always possible to align the coil and PET module perpendicular to the main magnetic field ($B_0$) direction of the MRI scanner. Thus, the RF coil may be located at a certain angle relative to $B_0$, which could disturb the RF field considerably (Alfonsetti et al 2005).

In this work, we explore various endorectal-style coil designs suitable for use in MR-PET prostate imaging and assess their performance in finite integration technique (FIT) simulations and in MR measurements on a 3 T whole-body MRI scanner. We designed a shielded dummy PET insert which has nearly the same geometry as the prototype time-of-flight MR-PET prostate probe (Garibaldi et al 2017), and evaluated the quality of the images acquired using the selected coils by determining SNR as a function of the coil tilting angle against the $B_0$ direction which, evidently, is the most dominant factor influencing SNR. Although this has not been previously investigated, it is important since this coil will generally not be aligned orthogonally to $B_0$ due to the location and flexibility of the prostate. The chosen designs were also compared by carrying out a transmission scan to assess any quality degradation from a PET standpoint.

2. Material and methods

As described in figure 1, three commonly-used linear coil designs and their combinations for a quadrature drive were selected (Ackerman et al 1980, Meh dizadeh et al 1985, Schnall et al 1989, Gilderdale et al 2000, Zhang et al 2001, Ouhlous et al 2007, Kumar et al 2008, Eryaman et al 2009, Alfonsetti et al 2010, Arteaga et al 2012) and investigated in this study: linear coils of a large single loop (Loop), a microstrip transmission line (MTL) and a figure-of-eight (FO8), and quadrature coils of overlapped loops (L-L) and a FO8/loop (FO8-L), which were constructed by combining the linear coils. These designs were selected since they are aligned perpendicularly to $B_0$ in a normal situation. The quadrature loop/MTL pair configuration was excluded in this study due to the strong coupling between elements, which would require additional lossy decoupling units (trap or PIN-diode circuits). Each coil was constructed using a single-sided, flexible, printed circuit board (PCB) which consisted of a very thin, about 35 µm, copper strip used as the coil conductor and an encapsulation of about 1 mm PMMA used as the thin coil supporter. This was then attached to a 3D printed coil former (Fortus 400mc, Stratasys, Minnesota, USA) in biocompatible polycarbonate. Figure 1 shows photographs of all the coils and their equivalent circuit diagrams, including the component values and dimensions of the coil patterns.
A 2%\text{w/v} agarose gel phantom (10 g) doped into phosphate-buffered saline (500 ml) was prepared which surrounded the whole volume of the coil assembly. A table tennis ball phantom containing a solution of 1.24 g NiSO$_4$ × 6H$_2$O + 2.62 g NaCl per 1000 g H$_2$O (Stumpf et al. 2018) was placed inside the gel phantom and above the coils—approximately 7 mm away from the nearest coil pattern. This distance was always maintained while the coil was tilted. The size of the table tennis ball was assumed to be similar to that of a human prostate and it was further assumed that its location may be approximately 7 mm from the coil placed within the rectal wall.

In order to characterise the performance of the RF coils independently, without PET modules, a shielded, dummy PET detector, shown in figure 2, was built. The dummy PET insert (27 × 25 × 70 mm$^3$) was enclosed using copper tape (3 M, Minnesota, USA), and positioned under the inner coil area. As the distance between the coil and the shield could potentially degrade the efficiency of the coil and limit the possible coil designs, the shape of the coils was curved (assuming that it is located along the inner rectal wall), as shown in figure 2(a). This was done to maximise the distance from the shield while maintaining proximity to the phantom. With this arrangement, the outer region of the coil was used for the imaging experiments. In order to avoid issues arising from the gradient eddy currents during the MR measurements, the copper dummy shield was separated into two areas (gap: approximately 1 mm), but linked via high-value capacitors (1 nF). The simulation model of the L-L coil with the phantom (figure 2(b)), as well as the planned final system assembly and the PET insert drawing with dimensions and components (figure 2(c)), are also shown.

Using the phantom and the dummy PET shield, the coils were tuned to 123.2 MHz (corresponding to the field strength of our 3 T MRI system) and matched to 50  \Omega. The tuning and matching of the coils were achieved using non-magnetic fixed (100 B series, ATC, USA) and variable (Voltronics, USA) capacitors. In this work, trimmers were used in order to minimise the effort of fine adjustment of the coils. However, these can be replaced in a final construction by fixed capacitors, for example, 1 pF and 1.8 pF for replacing the trimmers as shown in figures 1(a) and (b), respectively. To avoid any deleterious effects on the acquisition of the PET data as a result of the insert, all of the capacitors were positioned outside the PET field-of-view (FOV) and only thin and narrow coil patterns (thin copper ~ 35 µm) were placed within the FOV. The responses of these coils were measured on the bench using a vector network analyser (ZNB4, Rohde & Schwarz, Germany). Two types of interfaces containing a transmit/receive (T/R) switch were used to operate the coils either in linear or in quadrature. In order to facilitate quadrature drive (0 and 90° phase), a quadrature hybrid was additionally included in the quadrature version of the T/R switch.

In order to evaluate the performance of the proposed coils, FIT simulations were conducted using CST Studio Suite (CST AG, Darmstadt, Germany). The simulation domain was terminated with a convolution perfectly matched layer. Figure 2(b) shows the simulation model of the L-L coil. All coils were loaded with
Figure 2. Photograph (a) and CST simulation model (b) showing the overall arrangement of the assembly consisting of the shielded dummy PET detector, RF coil, phantoms and phantom holder. The planned future configuration (c) of the final MR-PET assembly placed inside an endorectal tube.

the phantom described above ($\varepsilon = 80$ and $\sigma = 0.5 \text{ S m}^{-1}$) (Stumpf et al. 2018). The coil elements were driven by two voltage sources containing identical amplitude and phase (Collins et al. 2002). To achieve quadrature excitation of L-L and FO8-L, the coils were driven with 0 and 90° phase settings. The results of the transmit (Tx) efficiency and safety excitation efficiency (SEE) generated by the coils (Shajan et al. 2016, Hong et al. 2019) were analysed and compared as a function of the tilting angles relating to the $B_0$ direction. Due to the long computation time arising from the fine mesh requirement to capture the complex model geometry in the tilted case (FIT requires the use of a hexahedral mesh and cannot use conformal tetrahedral meshing), we only evaluated two tilting angles (0 and 20°) for each of the selected coils. The Tx efficiencies of the proposed coils were calculated by dividing the $B_1^+$ field by the square root of the accepted power. The SEE (Shajan et al. 2016, Hong et al. 2019) of the coils was also calculated by dividing $B_1^+$ by the square root of the maximum 1 g specific absorption rate (SAR). The reason why a 1 g SAR was used instead of 10 g is that our phantom of interest (the table tennis ball) was so small it was practically not feasible to determine the 10 g average volume. The averaged efficiencies were calculated within the region of the selected slice of the spherical phantom.

All MR experiments were carried out on a 3 T PRISMA scanner (Siemens Healthineers, Erlangen, Germany) using a 2D gradient echo (GRE) and 3D FLASH sequences. The parameters for GRE were repetition time (TR) = 9 ms, echo time (TE) = 2.23 ms, number of averages (NEX) = 1, slice thickness = 2 mm, acquisition time ~15 s, matrix size = 256 × 256 and FOV = 128 × 128 mm², and for the latter were TR = 8 ms, TE = 1.53 ms, NEX = 1, acquisition time ~1:31 min, and voxel size = $0.5 \times 0.5 \times 2$ mm³. The power required for a 90-degree reference pulse was determined by sweeping the RF power and computing the SNRs in a predetermined ROI. The power required to achieve the highest SNR value for each coil designs was used as the reference voltage.

The coil assembly, including the phantom, was tilted with respect to the $B_0$ direction (in z-axis) by 0, 10, 20 and 30° to replicate the possible circumstances under which the inserted endorectal coil could be located in the subject (Roach et al. 2001, Permpongkosol et al. 2018), and the imaging performance of the coils was evaluated. Profiles in the sagittal direction on the axial images across the phantom, as imaged using all coils,
were analysed as a function of the tilting angles. SNRs were calculated \( \frac{\text{Signal mean}}{\text{Noise standard deviation}} \), that is to say, the signal mean value in the selected ROI covering the entire table tennis ball phantom of those axial images (e.g. yellow dotted circle in figure 5) divided by the standard deviation of the noise image (i.e. the same ROI but with zero Tx power) and compared.

In order to verify the degradation in the PET performance due to the insertion of the coils, transmission scans were carried out by means of three Ge68-transmission sources (approximately, 150 MBq each) on a dedicated PET Scanner (ECAT Exact HR+, Siemens Healthineers, Erlangen, Germany), independently. All of the coil patterns were attached to the surface of a cylindrical water phantom (200 mm inner diameter) and transmission scans of 120 min were applied. For comparison, reference scans without the patterns were also acquired with identical protocols. The recorded transmission data were iteratively reconstructed (OSEM2D, 6 iterations, 16 subsets) into 63 slices (2.45 mm) with a matrix size of 256 x 256 (1 x 1 mm) representing the attenuation coefficient of the different coil patterns used in this study.

3. Results

Figure 3 displays a comparison of the simulated Tx efficiency maps generated by the proposed coils loaded with the combined gel and ball phantom. This shows the Tx efficiency is higher near the surface of the coil where the phantom was imaged and decreases as it goes further away. The coils, Loop, FO8, MTL, L-L and FO8-L, provided the mean Tx efficiencies of 24.62, 25.81, 20.53, 39.36 and 41.70 \( \mu T/\mu W \), respectively, in the region of interest (white dotted circle) where the ball phantom was located. Figure 3 also shows the calculated SEE values of 3.19, 1.95, 2.16, 3.75 and 3.89, corresponding to each of the coils stated above. The Tx efficiency profiles, normalised by the accepted power (bottom left) and by 1 g peak SAR (bottom right) along the y-direction (dotted black line) on the axial image, are also shown. Overall, the simulation results show that the quadrature coils, L-L and FO8-L, provides higher Tx efficiency and SEE compared to the linear coils. SAR maps of the coils are shown in the supplementary material (stacks.iop.org/PMB/65/115005/mmedia).

This is in agreement with the result in the literature (Chen et al 1983). Among the linear coils, the FO8 and Loop designs provided higher efficiency and the Loop design performed better with respect to SEE when considered close to the coil (< gap + 5 mm). Furthermore, as the distance of the phantom from the coil increases, the efficiencies of the Loop coil are seen to be superior to the other linear coils. For instance, at a...
Figure 4. CST simulation model with two different tilting angles—0 and 20° (top left), associated Tx efficiency maps of L-L and FO8-L coils (right). The unit of the colour bar is µT. The $B_1^+$ values were computed and normalised to $\sqrt{\text{accepted power}}$. White dotted circles refer to the ball phantom location and yellow dotted lines indicate the cross-section of the ball phantom in an axial slice. The profiles of one of the axial ball phantom images are plotted (bottom left) showing the efficiency for both coils decreased due to the tilting.

Table 1. S-parameters ($S_{11}$ and $S_{22}$: reflection loss, $S_{21}$: isolation between two channels) of the coils measured on the bench.

|         | $S_{11}$ (dB) | $S_{22}$ (dB) | $S_{21}$ (dB) |
|---------|--------------|--------------|--------------|
| Linear  |              |              |              |
| Loop    | −31.28       | N/A          | N/A          |
| FO8     | −25.04       | N/A          | N/A          |
| MTL     | −28.37       | N/A          | N/A          |
| Quadrature |           |              |              |
| L-L     | −22.76       | −26.24       | −25.93       |
| FO8-L   | −37.36       | −29.29       | −14.87       |

distance of 12 mm from the bottom of the phantom, the SEE value is found to be nearly the same as for the quadrature coils (L-L and FO8-L) and nearly two times better than the other linear coils. For the 20° tilting condition as depicted in figure 4, both quadrature coils, L-L and FO8-L, show similar Tx efficiency—only a marginal difference between the two implementations can be seen. However, in comparison to the 0° angle, the efficiency of both coils is reduced, with the FO8-L coil being particularly ill affected. In addition, by tilting 20°, the $B_1^+$ pattern of FO8-L shrunk toward the centre of the ball phantom in the x-axis, while that of L-L was maintained.

Measured scattering (S)-parameters are shown in table 1, proving the tuning and the matching conditions of all examined coils, and isolation between the two channels in the case of the quadrature coils (L-L and FO8-L). Figure 5 (top) illustrates how the coil assembly is oriented relative to the $B_0$ directions of 0, 10, 20, and 30°. This contains a blue dotted circle indicating to where the table tennis ball phantom was located, and also black and red dotted lines referring to where one of the axial and sagittal slices of the phantom were selected. In figure 5 (middle), axial and sagittal MR images obtained using 2D GRE and 3D FLASH sequences, respectively, are shown. The selected ROIs (yellow dotted circle) in the Loop 0° angle images were used for the SNR calculation. The higher signal intensity of the images follows the tilted angle slices angles which are also clearly shown in the sagittal images. The shrunken $B_1^+$ pattern in the left to right direction is shown in the axial image acquired using, in particular, the FO8 and MTL coils and was most prominent at the tilting angle of 30°.

The calculated relative SNR values of each coil with different tilting angles are also shown in figure 5 (bottom). The SNRs of all coils were decreased with larger tilting angles and the SNRs of the quadrature coils were higher than those of the linear coils. For most of the coils, there is no significant loss in SNR in the axial slice when the coil is tilted to an angle of up to 20°. However, the SNR of FO8 drops considerably at a tilting angle of between 10 and 20°. In a sagittal slice, the coils of L-L, FO8-L and Loop coils lose the SNR rapidly when the tilting angle is increased, and it is more significant between 20 and 30° for the FO8-L and Loop coils. The SNR of those coils generating the transverse $B_1^+$ field, FO8 and MTL, does not decrease.
Figure 5. Top: diagrams showing the position of the coil and phantom with tilting angles to $B_0$. Middle: axial and sagittal MR images respectively acquired using the 2D GRE and 3D FLASH sequences. ROIs in the dotted yellow circle for SNR measurement are also included. Bottom: the calculated relative SNRs of the linear (MTL, FO8, Loop) and quadrature (L-L and FO8-L) coils versus the tilting angles (0° to 30°) relative to $B_0$.

substantially, but rather constantly. Overall, the overlapped loop coil is shown to have superior performance in SNR, compared to all the other coils tested.

Figure 6 shows attenuation maps obtained after the transmission scans with and without the coil patterns (a. with, b. without, c. difference) present. It confirms that the differences in the attenuation maps are negligible among the different coil pattern configurations.

4. Discussion

In this study, we built five of the most possibly used RF coils for endorectal imaging applications and evaluated their performance. Due to the particular constraints associated with endorectal hybrid imaging, we aimed to investigate three important aspects. First, we imaged a phantom located at the outer region of the coils. Second, the coils were examined in terms of the limited available space due to the shielded PET insert.
Third, we evaluated the SNR obtained from the five different coils by considering the four different tilting angles to $B_0$ in which the human prostate could be situated.

Using simulations, it was shown that the Tx efficiency near the surface of the coil, where the phantom was imaged, is higher and decreases as it goes further away. Overall, the simulation results show that the quadrature coils, L-L and FO8-L, provide higher Tx efficiency and SEE compared to the linear coils, as expected. Among the linear coil designs, the FO8 and Loop designs provided higher efficiency and the Loop design performed better with respect to SEE. For the $20^\circ$ tilting condition, as depicted in figure 4, both quadrature coils, L-L and FO8-L, show a similar Tx efficiency—with only a marginal difference between the two implementations. However, in comparison to the $0^\circ$ angle, the efficiency of both coils, more significantly so for the FO8-L coil, was found to be reduced. In addition, when tilting at $20^\circ$, the $B_1^+$ pattern of FO8-L was shrunken toward the centre of the ball phantom in the $x$-axis, while that of L-L was maintained. We believe the reason for the shrunken $B_1$ pattern, as shown in figure 4 FO8-L coil with the $20^\circ$ tilting, was due to differences in the characteristics and geometry of the proposed endorectal coils. More specifically, we believe that this arises from the different flux outside the coil geometry of the FO8 coil compared with that of the Loop coil. While the flux lines are confined within the coil geometry for the FO8, they are close to the outside of the coil conductors for the Loop coil. However, in order to fully understand this, further investigation is required.

In MR measurements at higher tilting angles, the $B_1$ direction of the coil became parallel to the $B_0$ direction, leading to significant signal loss. In all cases, a marginal decrease in SNR (less than 5%) could be seen between the tilting angle of $0^\circ$ and $10^\circ$. However, SNR dropped quite dramatically when the actual tilting angle was at $30^\circ$, and became even worse when the images were acquired with higher tilting angles (data not shown). The overlapped loops (L-L) had the benefit of being driven in a quadrature mode, and their performance in terms of simulated Tx efficiency, SNR and penetration was substantially better compared to the other coils at any tilting angle.

Considering the interaction between the shield and the coil might also be interesting in terms of looking at the effect as a function of the distance between the shield and the coil. However, since the shielded PET detector was situated in a fixed position, we placed the coil as far away as possible from the shield. All high-density materials, e.g. capacitors and cables, were also placed outside the PET FOV. Very thin copper about 35 $\mu$m and PMMA about 1 mm were the only materials used within the PET FOV. It was found that the materials (patterns of the coils and thin PMMA) in front of the PET detector have a negligible absorption effect for 511 keV annihilation photons, as confirmed by the difference images in figure 6(c) (no difference) acquired using the transmission scans and by the transmission coefficient of 511 keV gamma rays which is approximately 99.7% for the 35 $\mu$m thick copper strip (Hubbell and Seltzer 2004). As the additional pattern for quadrature coils is not the dominant source of degradation to PET performance, it was considered to be negligible. Although the interference of the PET components, e.g. SiPM in combination with PET readout electronics was not demonstrated here, and was, therefore not included in this manuscript, they have been tested in a hybrid MR-PET system and reported on (Kolb et al 2012, Ko et al 2016). Furthermore, it was
found that the mutual interference between MR and PET can be minimised, while maintaining the full imaging performance of both techniques, by using effective shielding (Ko et al 2016).

The main purpose of the slit, which was linked by high-value capacitors, was to avoid any potential eddy current interference from the large conducting area during MR experiments (i.e. rapid gradient switching), and this was found to be sufficient for the MR experiments conducted. Traditionally, minimisation of eddy current interference with the use of an RF shield is achieved either by using a number of slits and inserting capacitors for connection between the slits or by using overlapped shield segments on a double-sided PCB. Here, the proposed shield design is a prototype and can be optimised depending on rejecting the interference from PET and MR eddy currents. A change in the location of the slits simultaneously may change the coil characteristics and require retuning and matching of the RF coil. Therefore, in this study, we fixed the shield shape and location of the slit using a 3D printed dummy model and tried to maintain identical condition for all coils to ensure a fair comparison. Alternatively, a copper or aluminium foil of lower conductivity can be used, as long as its thickness is a couple of skin depths at RF and much less at the frequencies of the gradients and PET detectors (Carlson 1994, Lee et al 2018).

In this work, we have focused on the ‘simpler’ but ‘popular’ designs for the transmit/receive endorectal coil. However, carrying out a further optimisation using a novel design concept (Chu et al 2006) would be useful and may improve the image quality in terms of SNR and B1 homogeneity.

5. Conclusions

In conclusion, the performance of different endorectal coil designs for hybrid MR-PET imaging was compared with respect to the tilting angle relative to B0. It was found that the MR image quality obtained with the L-L coil was excellent and superior to that of the other coil designs. The findings of this work may also be of particular interest as guidance for applications in which the MIR coils are not perpendicular to the B0 field direction. Although the setup used here may not represent the final design exactly, particularly in relation to the PET detector, the key dimensions and important considerations were maintained. Furthermore, the negligible effect of the different coil designs on the attenuation maps was observed. Based on these results, we intend to exploit this knowledge in future studies integrating with MR-PET hybrid systems in the future.

Acknowledgments

We thank Claire Rick for English proofreading and Aliaksandra Shymanskaya for the phantom construction. We also acknowledge the reviewers for their constructive comments.

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References

Ackerman J J, Grove T H, Wong G G, Gadian D G and Radda G K 1980 Mapping of metabolites in whole animals by 31P NMR using surface coils Nature 283 167–70
Alfonsetti M, Clementi V, Iotti S, Placidi G, Lodi R, Barbieri B, Sotgiu A and Alecci M 2005 Versatile coil design and positioning of transverse-field RF surface coils for clinical 1.5-T MRI applications Magn. Reson. Mater. Phys. 18 69–75
Arteaga de Castro C S, van den Bergen B, Luijten P R, van der Heide U A, van Vulpen M and Klomp D W J 2012 Improving SNR and B1 transmit field for an endorectal coil in 7 T MRI and MRS of prostate cancer Magn. Reson. Med. 68 311–18
Beyendorff D, Taymoorian K, Knösel T, Schnorr D, Felix R, Hamm B and Bruhn H 2005 MRI of prostate cancer at 1.5 and 3.0 T: comparison of image quality in tumor detection and staging Am. J. Roentgenol. 185 1214–20
Brzeziński K, Oliver J E, Gillam J and Rafecas M 2014 A study of a high-resolution PET system using a silicon detector probe Phys. Med. Biol. 59 6117–40
Shah N J 2019 Apparatus and method for shielding MRI RF antennae from the effect of surrounding objects US Patent US005304932A
Chen C N, Houli D I and Sank V J 1983 Quadrature detection coils—a further 2 improvement in sensitivity J. Magn. Reson. 54 324–7
Chu D, Matias R M and Stormont R S 2006 Three concentric coil array US Patent US20060255804A1
Collins C et al 2002 Different excitation and reception distributions with a single-loop transmit-receive surface coil near a head-sized spherical phantom at 300 MHz Magn. Reson. Med. 47 1026–8
Costa D N, Yuan Q, Xi Y, Rosfky N M, Lenkinski R E, Lotan Y, Roehrborn C G, Francis F, Travalinii D and Pedrosa I 2016 Comparison of prostate cancer detection at 3-T MRI with and without an endorectal coil: a prospective, paired-patient study Urol. Oncol. 34 2557–13
Eiber M et al 2016 Simultaneous 68Ga-PSMA HBED-CC PET/MRI improves the localization of primary prostate cancer Eur. Urol. 70 829–36
Eryaman Y, Oner Y and Atalar E 2009 Design of internal MRI coils using ultimate intrinsic SNR Magn. Reson. Mater. Phys. 22 221–8
Garibaldi F et al 2017 A novel TOF-PET MRI detector for diagnosis and follow up of the prostate cancer Eur. Phys. J. Plus 132 396
Gawlitzka J et al 2017 Impact of the use of an endorectal coil for 3 T prostate MRI on image quality and cancer detection rate Sci. Rep. 7 40640
Gilderdale D J, Desouza N M, Coutts G A, Chui M K, Larkman D J, Williams A D and Young I R 2000 Design and use of internal receiver coils for magnetic resonance imaging Br. J. Radiol. 72 1141–51
Grković M et al 2015 Evaluation of a high resolution silicon PET insert module Nuclei. Instrum. Methods Phys. Res. A 788 86–94
Hegde J V, Mulkern R V, Panly L P, Fennessy F M, Fedorov A, Maier S E and Tempany C M C 2013 Multiparametric MRI of prostate cancer: an update on state-of-the-art techniques and their performance in detecting and localizing prostate cancer J. Magn. Reson. Imaging 37 1035–54
Heijmink S W T P et al 2007 Prostate cancer: body-array versus endorectal coil MR imaging at 3 T—comparison of image quality, localization, and staging performance Radiology 244 184–95
Hofman M S, Hicks R J, Maurer T and Eiber M 2018 Prostate-specific membrane antigen pet: clinical utility in prostate cancer, normal patterns, pearls, and pitfalls Radiographics 38 580–17
Hong S M, Choi C H, Shah N J and Felder J 2019 Design and evaluation of a 1H/31P double–resonant helmet coil for 3T Phys. Med. Biol. 64 035003
Hobbell J H and Seltzer S M 2004 X-ray mass attenuation coefficients NIST Standard Reference Database 126 (https://doi.org/10.18434/T4D01F)
Huh S S, Clinthorne N H and Rogers W L 2006 Investigation of an internal PET probe for prostate imaging Nuclei. Instrum. Methods Phys. Res. A 579 539–43
Kim Y, Hsu I C J, Pouliot J, Noworolski S M, Vigneron D B and Kurhanewicz J 2005 Expandable and rigid endorectal coils for prostate MRI: impact on prostate distortion and rigid image registration Med. Phys. 32 3569–78
Ko G B et al 2016 Simultaneous multiparametric PET/MRI with silicon photomultiplier PET and ultra-high-field MRI for small-animal imaging J. Nucl. Med. 57 1309–15
Kolb A et al 2012 Technical performance evaluation of a human brain PET/MRI system Eur. Radiol. 22 1776–88
Kumar A and Bottomley P A 2008 Optimized quadrature surface coil designs Magn. Reson. Mater. Phys. 21 41–52
Lee B J, Watkins R D, Chang C M and Levin C S 2018 Low eddy current RF shielding enclosure designs for 3T MR applications Magn. Reson. Med. 79 1745–55
Lindemann M E, Steburn V, Tischkach A, Kirchner J, Umatalu I and Quick H H 2018 Towards fast whole-body PET/MR: investigation of PET image quality versus reduced PET acquisition times PLoS One 13 e0206573
Lindenbergh L, Ahlman M, Turkbey B, Mena E and Choyke P 2016 Evaluation of prostate cancer with PET/MRI J. Nucl. Med. 57 1115–1165
Martin G V, Kudchadker R J, Bruno T I, Frank S J and Wang J 2018 Comparison of prostate distortion by inflatable and rigid endorectal MRI coils in permanent prostate brachytherapy imaging Brachytherapy 17 298–305
Mehdizadeh M, Molyneaux D A and Holland G N 1985 Quadrature surface coils for magnetic resonance imaging US Patent 4918388A
Muehlematter U J, Nagel H W, Becker A, Mueller J, Vokinger K N, de Galizia Barbosa F, Ter Voert E E G T, Veit-Haibach P and Burger I A 2018 Impact of time-of-flight PET on quantification accuracy and lesion detection in simultaneous 18F-choline PET/MRI for prostate cancer EJNMMI Res. 8 11
Noworolski S M, Crane J C, Vigneron D B and Kurhanewicz J 2008 A clinical comparison of rigid and inflatable endorectal-coil probes for MRI and 3D MR spectroscopic imaging (MRSI) of the prostate J. Magn. Reson. Imaging 27 1077–82
O’Donohoe R L, Dunne R M, Kimbrell V and Tempany C M 2019 Prostate MRI using an external phased array wearable pelvic coil at 3T: comparison with an endorectal coil Abdom. Radiol. 44 1062–9
Ouhlous M, Moelker A, Flick H J, Wielopolski P A, de Weert T T, Pattynama P M T and van der Lugt A 2007 Quadrature coil design for simultaneous dynamic PET/MRI for assessment of prostate cancer Magn. Reson. Med. 65 3569–78
Pauliuk H, Gravestock B, Stolker J C, Eiber M and Quick H H 2019 Impact of receiver coil selection on prostate cancer detection using PET/MR Hybrid Imaging Br. J. Radiol. 92 20190080
Pinkerton R G, Near J P, Barberi E A, Menon R S and Barthra R 2007 Transceive surface coil array for MRI of the human prostate at 4T Phys. Med. Biol. 52 3248–47
Pinkerton R G, Near J P, Barberi E A, Menon R S and Barthra R 2007 Transceive surface coil array for MRI of the human prostate at 4T Magn. Reson. Med. 57 455
Quick H H et al 2013 Integrated whole-body PET/MR hybrid imaging: clinical experience Invest. Radiol. 48 280–9
Roach M, Kurhanewicz J J and Carroll P 2001 Spectroscopy in prostate cancer: hope or hype? Oncology 15 1415–16
Rosenkrantz A B, Koesters T, Vahle A, Friedman K, Bartlett R M, Taneya S S, Ding Y and Logan J 2015 Quantitative graphical analysis of simultaneous dynamic PET/MRI for assessment of prostate cancer Clin. Nucl. Med. 40 e226–240
Sander C Y, Keil B, Chonde D B, Rosen B R, Catana C and Wahl L L 2015 A 31-channel MR Brain array coil compatible with positron emission tomography Magn. Reson. Med. 73 2363
Schnall S D, Lenkinski R E, Pollack P E, Imai Y and Kressel H Y 1989 Prostate: MR imaging with an endorectal surface coil Radiology 172 570–4
Shah N J 2019 Hybrid MR-PET: Systems, Methods and Applications (London: Royal Society of Chemistry) (https://doi.org/10.1039/9781788013062)
Shah Z K, Elias S N, Abaza R, Zynger D L, Derenne L A, Knopp M V, Guo B, Schurr R, Heymsfield S B and Jia G 2015 Performance comparison of 1.5-T endorectal coil MRI with 3.0-T nonendorectal coil MRI in patients with prostate cancer Acad. Radiol. 22 467–74

Shajan G, Mirkes C, Buckenmaier K, Hoffmann J, Pohmann R and Scheffler K 2016 Three-layered radio frequency coil arrangement for sodium MRI of the human brain at 9.4 tesla Magn. Reson. Med. 75 906–16

Sosna J, Pedrosa I, Dewolf W C, Mahallati H, Lenkinski R E and Rofsky N M 2004 MR imaging of the prostate at 3 tesla: comparison of an external phased-array coil to imaging with an endorectal coil at 1.5 tesla Acad. Radiol. 11 857–62

Stumpf C, Malzacher M and Schmidt L P 2018 Radio frequency modelling of receive coil arrays for magnetic resonance imaging J. Imaging 4 67

Turkbey B et al 2014 Comparison of endorectal coil and nonendorectal coil T2w and diffusion-weighted MRI at 3 tesla for localizing prostate cancer: correlation with whole-mount histopathology J. Magn. Reson. Imaging 39 1443–8

Yakar D, Heijmink S W T P J, Hulsbergen-van de Kaa C A, Huisman H, Barentsz J O, Fütterer J J and Scheenen T W J 2011 Initial results of 3-dimensional 1H-magnetic resonance spectroscopic imaging in the localization of prostate cancer at 3 tesla: should we use an endorectal coil? Invest. Radiol. 46 301–6

Zhang X, Ugurbil K and Chen W 2001 Microstrip RF surface coil design for extremely high-field MRI and spectroscopy Magn. Reson. Med. 46 443–5