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Development and evaluation of a multi-channel endorectal RF coil for prostate MRI at 7T in combination with an external surface array

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Abstract

Purpose—To develop and evaluate a sterilizable multi-channel endorectal coil (ERC) for use in combination with an external surface array (ESA) for high resolution anatomical and functional studies of the prostate at 7 Tesla.

Methods—A two-loop ERC (ERC-2L) and a microstrip-loop ERC (ERC-ML) were compared at 7T in terms of transmit and receive performance. The best performing ERC was evaluated alone and in combination with the ESA through 1) simulations on both phantom and an anatomically correct numerical human model to assess B_1^+ transmit and specific absorption rate (SAR) efficiencies and 2) phantom experiments to calculate B_1^+ transmit efficiency and SNR. Phantom studies were also performed to look at heating when using the ERC as a transmitter and for comparing the new coil against a single-channel balloon-type ERC (ERC-b). High-resolution MRI acquisitions were performed on a single healthy subject using the two-channel ERC combined with the ESA.

Results—Compared to the ERC-ML, the ERC-2L demonstrated 20% higher SAR efficiency and higher SNR 3 cm from the coil. The presence of a tuned and detuned ERC-2L did not alter the peak local SAR of the ESA alone, however the detuned ERC-2L had 45% less peak local SAR around the rectum compared to the tuned ERC-2L. The receive-only version of the ERC-2L improved the SNR 4.7-fold and 1.3-fold compared to the ESA and ERC-b, respectively. In combination with the ESA, the ERC-2L supported in-plane voxel-size of $0.36 \times 0.36 \text{ mm}^2$ in T2-weighted anatomic imaging.

Conclusions—The reusable ERC-2L combined with an ESA offers a high SNR imaging platform for translational studies of the prostate at 7T.

Keywords

endorectal coil; prostate; prostate cancer; ultra-high field; MRI; ultrahigh field; 7 Tesla body imaging

INTRODUCTION

Anatomic and functional multi-parametric MRI provides valuable information beyond standard clinical tests such as serum prostate-specific antigen measures, digital rectal exams, and trans-rectal ultrasound-guided (TRUS) biopsy (1-3). T2-weighted imaging provides delineation of prostate anatomy and assists in staging, while functional imaging such as diffusion weighted imaging (DWI), dynamic contrast enhanced MRI (DCE-MRI) and 3D spectroscopic imaging (3DSI) improves cancer detection and potential for grading (3-6).

The sensitivity of the MRI for disease assessment can be improved by increasing the static magnetic field strength and by using prostate specific coils. Evidence of this can be found in previous studies showing improved sensitivity and specificity at cancer detection due to increased anatomic detail and spectral resolution when progressing from 1.5 to 3T (7-10). Increased performance can also be expected when using an endo-rectal coil (ERC) compared to the sole use of surface body coils for signal reception (10-12). For example, when using an ERC at 3T, improved image quality and resolution resulted in significantly increased sensitivity and specificity for cancer detection and staging (13).

Performing anatomic and functional studies at ultrahigh field (e.g. 7 Tesla) can in principle provide even further improvements in the identification and quantification of key anatomic and metabolic information necessary to characterize and monitor disease. Ultrahigh field (UHF) MRI promises higher sensitivity, and improved spatial, temporal and/or spectral resolution. However, in order to fully utilize the potential of 7 T MRI, optimized radiofrequency (RF) coils need to be developed as do strategies for addressing the challenges present when performing studies in the human body at these higher frequencies (i.e. shorter wavelengths). The issue of transmit B_1^+ homogeneity and efficiency of 7 T MRI (14-17) in the prostate have been addressed with the use of multi-channel transmit/receive external surface arrays (ESA) (18-21) and accompanying RF shimming techniques (22). In combination with the ESA, a balloon-type ERC (ERC-b) consisting of single loop coil has been demonstrated to greatly improve SNR (23). A two-channel version of a transmit/receive balloon-type ERC was also developed for 7T by placing a microstrip element along the main shaft of the coil, in addition to the loop, which demonstrated improved performance (24). However the balloon type coils used in these previous studies have a limited lifespan (i.e. designed to be single use but have been used multiple times after sterilization at some centers) and were only explored as transceivers while safety and performance in combination with a surface array were not investigated. In this study, we initially evaluated two specific two-channel solid ERC geometries in terms of their transmit and receive performance, the better of which was combined with an ESA for assessing safe operating limits and performance gains. The goal of this work is to develop a sterilizable multi-channel ERC for use in combination with an external surface array (ESA) as a robust and sensitive imaging platform for performing translational prostate studies at 7 Tesla.

METHODS

ERC designs

Solid ERCs consisting of two loop elements (ERC-2L) were built by placing two rectangular loops on a 2.2 cm diameter 7.2 cm long solid cylindrical core. The short axis of each loop was 1.9 cm with an overlap of 0.3 cm (Figure 1.a). Both a transceiver and an actively detuned receive-only versions of the ERC-2L were constructed following the same geometry. A commercial version with the same general geometry of the ERC-2L is available for 1.5 and 3 T (Sentinelle Endorectal Coils, Invivo, Gainesville, FL).

In addition to the ERC-2L design, a transceiver version of a two-channel microstrip-loop ERC (ERC-ML) was built. The ERC-ML had a 2.2 cm diameter 7 cm long cylindrical core. The ground plane of the microstrip element was placed on the center axis of the core and the loop element was centered around the top conductor of the microstrip element with a short-axis diameter of 1.4 cm (Figure 1.c). The ERCs were positioned inside a sterilizable housing when used in any imaging or heating studies (Figure 1.e).

A previously described single-channel receive-only balloon-type ERC (ERC-b) was used for comparison by replacing the electrical components of a commercially available 3 T coil (Medrad eCoil, Bayer Healthcare, Whippany, NJ) (23). The size of the ERC-b loop was $7 \times 3.5 \text{ cm}^2$.

Phantom Studies

An ~18 liter, 30×45×19 cm (length × width × height) phantom was custom-built (Figure 1.f, CAB & plexiglass body (The Phantom Laboratory, Salem, NY)) and filled with a solution of 4 g/l NaCl and 0.02 g/l MnCl_2 in de-ionized water (23). The phantom had a 30 mm diameter cylinder running lengthwise down the phantom 9 cm from the anterior surface with access from one end which mimicked the position and orientation of the rectum in the human body. During experiments, the ERCs were inside the pseudo-rectum of the phantom in contact with wall on the anterior side held in place by an MRI compatible holder (Figure 1.e). The external transceiver surface array was placed around the outside of the phantom (anterior and posterior sides). In addition, SNR's of the ERC-2L and ERC-b were also tested in a different phantom setup with a larger diameter tube to accommodate the larger form factor of the inflated ERC-b.

MRI system

MRI experiments were conducted on a whole body 7 T scanner (Magnetom Siemens Healthcare, Erlangen, Germany), equipped with sixteen 1kW RF power amplifiers each with independent phase and gain control (Communications Power Corporation, Hauppauge, NY). A 16-channel transceive external surface array (ESA) consisting of 8 anterior and 8 posterior microstrip elements was used in both simulation and experiments (18). Phase-only B_1^+ shimming of the ESA was performed to optimize the transmit efficiency at the location of the prostate in both phantom and in vivo studies (17).

Numerical computations

Geometrically and electrically correct models of the ERC-2L and ERC-ML were generated in SEMCAD X software (Version 14.8, Schmid & Partner Engineering AG, Zürich, Switzerland). ERCs were modeled inside the rectum of the Virtual Family's Duke model (25) (IT'IS Foundation, Zürich, Switzerland). The tissue integrity of the rectum around ERCs was maintained by redefining tissue properties around the coil. ERC conductors were meshed at voxel sizes of $0.4 \times 0.4 \times 1 \text{ mm}^3$ and mesh sizes were gradually coarsened away from the ERCs. EM-field distributions inside the prostate were numerically computed using finite-difference time-domain (FDTD) methods. Relative phases of the ERC channels were optimized for maximum B_1^+ efficiency inside the prostate (22). B_1^+ transmit efficiency ($B_1^+ \sqrt{P_{Coil}}$), peak 1g SAR, and B_1^+ SAR efficiency ($B_1^+ \sqrt{SAR_{peak \ 1g}}$) were computed inside the prostate for both ERC-2L and ERC-ML designs.

Potential interactions of the ESA and the ERC-2L in both tuned (i.e. transceiver) and detuned (i.e. receive-only) states were investigated by computing the B_1^+ distributions using SEMCAD, and later compared experimentally inside the torso-sized phantom filled with uniform saline ($\epsilon=77.8$, $\sigma=0.82 \text{ S/m}$). Furthermore, RF safety and SAR distributions of the ESA without any ERCs present, and with receive-only and transceive ERC-2L were investigated using XFDTD software (Remcom Inc., State College, PA). EM-field distributions of the ESA and the ERC-2L positioned in the Duke model were numerically computed, again using FDTD methods. The grid size over the ERC-2L was 2 mm while the voxels not intersecting the ERC-2L were 5 mm. Each coil was tuned to the ^1H larmor frequency at 7 T and driven by a 50Ω voltage source port. The receive-only ERC-2L was simulated by terminating the two elements with $50\text{k}\Omega$ and again assessing transceiver performance and local 1g SAR in the region of the rectum.

The per channel time-average power (TAP) limit for safe operation was determined for a local SAR limit of 20 W/kg averaged over 1 g for the transceive versions of the ERC coils and a 10 g for the ESA. Additionally, peak 1 g average SAR around the rectum was computed for the following conditions: 1) ESA excitation in the absence of any ERCs, 2) ESA excitation in the presence of the tuned ERC-2L (transceive), and 3) ESA excitation in the presence of the detuned ERC-2L (receive-only) to investigate the impact of coupling on local E-fields and RF safety performance.

Heating Studies

Potentially high local SAR regions around the ERC-2L were determined using the previously described FDTD computations and further explored through heating studies. High local SAR regions coincided with the locations of the capacitors on the transceive ERC-2L, and those locations were chosen to place fiber-optic temperature sensors (STB Probe, LumaSense Technologies, Santa Clara, CA) the positions of which were confirmed under CT imaging and shown in Figure 2.a. The heating studies were accomplished using ACRNEMA standards driving the coil for 15 minutes with 16-16.7 W total average power (Channel 1: 8.5-8.8W, Channel 2: 7.5-7.8W) with a 90° phase difference between coil channels. Temperature rise during and after the RF exposure was monitored with temperature sensors at 1 Hz. Local SAR at each sensor location was calculated from the

slope of initial temperature rise of the heating curve using the relation $dT/dt = SAR/c$, where c is the heat capacity of the gel and estimated as $4180 \text{ J/}^{\circ}\text{C kg}$ (26). The TAP limit per channel was calculated for a limit of 20W/kg peak local SAR.

Transmit B_1^+ and SNR mapping

Flip angle maps were computed using the actual flip angle imaging method (27) ($TR_1=20\text{ms}$, $TR_2=120\text{ms}$, $TE=3\text{ms}$, $FA=50^{\circ}$, matrix size: $128 \times 128 \times 16$, voxel size: $2.7 \times 2.7 \times 5\text{mm}$).

SNR was computed using the methods of Edelstein et al. (28) with MRI data acquired using a fully-relaxed gradient echo (GRE) sequences ($TR/TE = 10000/3\text{ms}$, acquisition matrix: $350 \times 100 \times 45$, voxel-size: $1 \times 2 \times 3\text{mm}^3$) followed by a noise scan with minimum TR and RF turned off. SNR maps included correction for the transmit field using the calculated flip angle maps. SNR of the ESA and combined ESA + ERC-2L were compared inside the torso phantom.

In vivo Imaging

A complete protocol including the following anatomic and functional scans has been completed on a single healthy subject to date with signed written consent under an Institutional Review Board approved protocol. T2-weighted anatomic scans were acquired using multi-slice turbo spin-echo (TSE) sequences with the ESA ($TR/TE = 6000/72\text{ms}$; echo-train length, ETL= 9; field-of-view, FOV = $220 \times 220\text{mm}^2$; voxel-size = $0.7 \times 0.7 \times 3\text{mm}^3$; voxel-volume = 1.47mm^3 , scan duration = 3:06), and the ESA combined with the ERC-2L ($TR/TE = 6000/74\text{ms}$; ETL = 7; FOV = $160 \times 160\text{mm}^2$; voxel-size = $0.4 \times 0.4 \times 1.5\text{mm}^3$; voxel-volume = 0.24mm^3 , scan duration = 3:06). Diffusion weighted images were acquired with $TR/TE = 5000/45\text{ms}$; FOV = $220 \times 220\text{mm}^2$; b-values = 0, 800 (6 directions); voxel-size = $1.7 \times 1.7 \times 4\text{mm}^3$ (ESA), $1.25 \times 1.25 \times 3\text{mm}^3$ (ESA + ERC-2L); acceleration factor (R) = 2 in AP (ESA), 3 in RL (ESA + ERC-2L); number of averages (NSA) = 7 (ESA), 5 (ESA + ERC-2L); scan duration = 4:55 (ESA), 3:50 (ESA + ERC-2L). For the diffusion scans, apparent diffusion coefficient (ADC) and fractional anisotropy (FA) were calculated by scanner software.

Three-dimensional (3D) spectroscopic imaging data was acquired with the ESA combined with the ERC-2L using a semi-LASER sequence with GOIA refocusing pulses (29,30) ($TR/TE = 1700/65\text{ms}$, acquisition matrix = $16 \times 16 \times 8$, elliptical sampling, voxel-size = $4 \times 4 \times 5\text{mm}^3$, NSA = 1, scan duration = 19.5 minutes).

RESULTS

Phantom and simulation results

Decoupling of the ERC-2L elements was accomplished by overlapping the loops, however due to their small size, only modest decoupling performance was achieved ($S_{21} = -10.4\text{dB}$). On the other hand, decoupling the ERC-ML channels resulted in an S_{21} of -17.7dB by symmetrically placing the loop with respect to the microstrip. Numerically computed transmit/receive and safety performance of ERC-2L and ERC-ML are listed in Table 1, and

SNR distributions are shown in Supplemental Figure 1. Both ERC's have very similar transmit efficiency and SNR performance inside the prostate, however worst-case peak 1g averaged SAR of ERC-ML is 53% higher than ERC-2L, and B_1^+ SAR efficiency of ERC-2L is 22% higher than ERC-ML. Even though average SNR performance is similar for both coils, the ERC-2L has 4.7% and 7.0% better SNR at a distance of 3 and 4 cm from the coil anteriorly (Supplemental Figure 1.c). Whereas the ERC-ML has 3.5% and 0.4% higher SNR at distances of 1 and 2 cm from the coil, respectively (Supplemental Figure 1.c). In subjects with larger prostates and for better coverage of the anterior part of the prostate, improved SNR of the ERC-2L at further depth (i.e. >3 cm from the coil) is beneficial. Owing to its better transmit safety performance and higher SNR at depth, we decided to further evaluate the ERC-2L.

Simulation of the transceive ERC-2L demonstrates a peak 1 g average TAP limit of 1.19 W per channel. The results of heating studies of the same ERC-2L coil configuration are shown in Figure 2b. Based on these experimental results, the calculated TAP limit is 1.52 W per channel. The total average power limits for the ESA and ERC-2L based on 10g and 1g averages are 38.4 and 2.38 W, respectively (Table 2).

Initial simulation studies with the ESA as the sole transmitter allowed determination of ESA TAP limits and to assess the impact of the ERC-2L on local SAR near the rectum. Numerically computed 10g averaged SAR reaches 20 W/kg when driving all elements of the ESA with a TAP of 2.43 W per channel in the absence of the ERC-2L. Under this condition, the peak 1 g average SAR around the rectum is 2.36 W/kg. In the presence of a tuned ERC-2L, the peak 1 g average SAR around the rectum increases to 4.30 W/kg. Despite this increase, the presence of the transceive ERC-2L did not pose an additional safety risk above that of the ESA alone in this specific situation with respect to local energy deposition (i.e. local SAR deposition near the rectum <20W/kg). The peak 1 g average SAR around the ERC-2L drops to 2.42 W/kg when the transceive ERC-2L is replaced with the detuned receive-only ERC-2L, close to that realized in the absence of an ERC (i.e. 2.36 W/kg). Amount of coupling between the ESA and ERC-2L, and local SAR amplifications near the prostate depend on the ESA B_1^+ shim settings and patient anatomy. In human studies we used the receive-only ERC-2L (detuned during ESA excitation) to alleviate potential safety risks in this regard.

Numerically computed (Figure 3.a-c) and measured (Figure 3.d-f) axial B_1^+ maps of ESA-only excitations inside the torso-sized phantom are shown in three configurations: without an ERC (Figure 3.a, d), with a detuned ERC-2L (Figure 3.b,e), and with a tuned ERC-2L (Figure 3.c,f). Accuracy of the numerical computations can be observed qualitatively when compared with the experimental results. The presence of a detuned ERC-2L has a minimal effect on the transmit field in its vicinity. However, with a tuned ERC-2L, the B_1^+ field near the coil is altered locally as a result of inductive coupling.

SNR maps of the ESA and the ERC-2L are shown in Figure 4.a-b respectively. SNR profiles along the AP dimension intersecting the middle of the ERC-2L and along the RL dimension 2 cm on the anterior side of the ERC-2L are plotted in Figure 4.c and d, respectively; for the ESA, ERC-2L and combined ESA+ERC-2L. At a distance between 3.5-4 cm anterior to the

ERC-2L, the ESA begins to outperform the ERC-2L. Mean SNR of the ERC-2L inside a $5 \times 3 \times 2.1 \text{ cm}^3$ (31.5ml) ROI representing the prostate is 4.7-fold higher than the ESA (sum of magnitudes, absolute SNR [$\times 10^5 \text{ Hz}^{0.5}/\text{ml}$]: ERC-2L = 6.41 ± 3.64 , ESA = 1.37 ± 0.22).

SNR maps of the ERC-b and the ERC-2L are shown in Figure 5.a-b, respectively. SNR profiles along the AP dimension intersecting the middle of the ERC-2L and along the RL dimension 2 cm on the anterior side of the ERC-b (in red) and the ERC-2L (in blue) are plotted in Figure 5.c-d, respectively. Inside a $5 \times 3 \times 2 \text{ cm}^3$ (30 ml) ROI centered 2 cm anterior to the ERC, the SNR of the ERC-2L is 30.6% higher than the ERC-b (absolute SNR [$\times 10^5 \text{ Hz}^{0.5}/\text{ml}$]: ERC-2L = 5.03 ± 2.16 , ERC-b = 3.85 ± 1.79).

***In vivo* study**

T2-weighted TSE images of the prostate acquired using the receive-only ERC-2L in three orientations (sagittal, coronal and transverse) are shown in Figure 6. T2-weighted transversal MRI of the prostate acquired using the combined ESA + ERC-2L are shown in Figure 7.a and d, respectively. Nominal voxel volume of the image acquired using the ERC-2L ($0.24 \mu\text{l}$) is 6.1-fold smaller than the ESA-only image ($1.47 \mu\text{l}$).

The higher SNR provided by the ERC-2L enables increased functional imaging resolution. Apparent diffusion coefficient (ADC) maps (Figure 7.b, ESA; Figure 7.e, ERC-2L) and fractional anisotropy (FA) maps (Figure 7.c, ESA; Figure 7.f, ERC-2L) are shown. Voxel volume of the maps acquired using the ERC-2L ($4.7 \mu\text{l}$) are 2.5-times smaller than the ESA voxels ($11.6 \mu\text{l}$).

Results from a three-dimensional spectroscopic imaging study are shown in Figure 8. High quality spectra obtained throughout the prostate are obtained with the semi-LASER acquisition. The long TR of 1700 ms in this acquisition is necessary to avoid exceeding SAR limits. The actual voxel size when accounting for the point spread function resulting from applying a hamming window spatial apodization and elliptical sampling is $\sim 150 \mu\text{L}$.

DISCUSSION

In this study, we evaluated two specific two-channel ERC geometries (ERC-2L vs ERC-ML) in terms of their both transmit and receive performance, for the purpose of developing a robust prostate imaging platform for translational prostate studies at 7T. The developed ERC is planned to be used in transmit mode in the future, therefore transmit characteristics of the ERCs were also taken into consideration to determine the better of the two designs. The ERC-2L was chosen over ERC-ML due to its better safety performance (i.e. lower peak 1g-averaged SAR) and its higher SNR at distances $>3\text{cm}$ away from the coil for improved coverage in subjects with large prostates and on the anterior part of the prostate.

The quadrature configuration of the ERC-2L improved the performance compared to a single channel ERC-b by approximately 30%. A similar performance improvement was previously observed by de Castro et al. (24) on a two channel balloon-type transceiver ERC, consisting of a microstrip element on the shaft and a larger loop. In both two channel ERC designs, coil elements were decoupled geometrically from each other and the anticipated

performance gains from the extra elements were realized. In addition to introducing a multi-channel solid ERC for prostate studies at 7T, we investigated its performance against and in combination with an external surface array (ESA) which is arguably a preferable platform for prostate studies at 7T (23). In terms of practicality, the ERC-2L design offers advantages over balloon-type ERCs. The ERC-2L can and has been used in repeat studies with standard sterilization procedures used for cleaning ultrasound probes. As such, the ERC-2L is a more cost-effective and practical solution at 7T for which disposable balloon coils are not manufactured and therefore would have to be made in-house for each study. While some centers have approval to use the balloon coils for several studies, their lifespan is limited and obtaining approval for using these coils multiple times is not always possible in some institutions. In addition, the solid housing can potentially house more complex electronics and coil geometries as it is not mechanically compromised when inserted into the rectum. The ERC-2L will also reduce physical compression and distortion of the prostate due to its smaller deployed form factor in the rectum (32). Reduction of prostate deformation during imaging is beneficial when the MRI data needs to be registered with different imaging modalities for treatment planning (32,33).

There are several limitations of our study. The first is that, only a single healthy subject and no patients were imaged with the complete anatomic and functional protocol described with the current ERC-2L and ESA coil configuration. While the ERC-2L was able to produce high quality spectroscopy and diffusion data, patient studies are needed to evaluate the benefits of the improved data on assessing prostate cancer in the presence of possible B_0 homogeneity and motion issues. While it may distort the prostate more, inflation of an ERC-b has been shown to reduce motion (34) and potentially limit migration of gas between the coil and the anterior wall of the rectum thus improving the chances of achieving and maintaining the desired static magnetic field (B_0) homogeneity throughout a study. As both spectroscopy and diffusion require high B_0 homogeneity to maximize spectral resolution and decrease spatial distortions, this is an important consideration. A second limitation is that the estimates of local SAR for determining TAP limits were based on a single human body model. The amount of coupling between the ESA and ERC-2L and local SAR amplifications near the prostate, depends on the shim settings of the ESA, placement of the ERC and specific patient geometries. To minimize potential safety risks, the detuned receive-only rather than the transceiver ERC-2L was used in human studies. To further ensure patient safety, the SAR near the ERC was based on a 1 g average as opposed to the standard 10 g average as a means to derive TAP limits. Further assessment of subject dependent B_1+ shimming and its impact on local coupling and heating could be assessed by monitoring temperature through embedded fiber-optic sensors on the surface of the coil and/or using an RF radiometry method (31). Temperature monitoring of the coil surface may also identify additional constraints which need to be considered beyond SAR when using the ERC coils as transceivers.

In this work, we developed a two-channel solid ERC (ERC-2L) for use in combination with an external surface array. The proposed design improved the SNR compared to a single channel ERC-b by 30% and about 5-fold compared to the ESA alone inside the prostate. The improved SNR of the ERC-2L in combination with the ESA enabled high quality anatomical and functional imaging in vivo at finer resolutions compared to the ESA alone on a healthy

subject. The ERC-2L may provide a safe, viable, cost-effective option for prostate studies at 7T.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.

(a) Schematic drawing of ERC-2L and (c) ERC-ML along with their Semcad models inside Duke model (b,d; ERC-2L, ERC-ML, respectively) are shown. Top and bottom rows in b,d show axial and sagittal views of the ERC models. Prostate region appears in green. (e) Sterilizable solid housing for the ERC and (f) the human torso-sized and shaped phantom with a pseudo rectum are displayed.

Figure 2.

Temperature sensors were placed near the capacitors of the ERC-2L and their locations were shown in (a) the schematic. (b) Temperature curves recorded during the heating study were plotted.

Figure 3.

Numerically computed transmit profiles (B_1^+ maps) inside the torso-size phantom during ESA excitation (a) in the absence of an ERC, (b) when a detuned ERC-2L is present and (c) when a tuned ERC-2L is present. Experimentally acquired transmit profiles are shown in the bottom row while transmitting using the ESA (d) without an ERC, (e) in the presence of the detuned ERC-2L and (f) in the presence of the tuned ERC-2L.

Figure 4.

Absolute SNR maps of (a) the ESA and (b) the ERC-2L are shown. Line profiles of the absolute SNR along (c) the vertical and (d) horizontal annotate lines shown in parts (a, b). SNR of the ESA (green), ERC-2L (blue) and combination of the ESA and ERC-2L (red) are plotted.

Figure 5.

Absolute SNR maps of (a) the balloon-type ERC (ERC-b) and (b) the ERC-2L are shown. Line profiles of the absolute SNR along (c) the vertical and (d) horizontal annotated lines shown in parts (a, b). SNR of the ERC-b (red) and ERC-2L (blue) are plotted. SNR of the ERC-2L is 30.6% higher than ERC-b (calculated inside an ROI of $5 \times 3 \times 2 \text{ cm}^3$ (30ml) 5mm anterior to the ERC).

Figure 6. T2-weighted TSE images of the prostate acquired using the receive-only ERC-2L in (a) sagittal, (b) coronal and (c) axial orientations are shown.

Figure 7. (a, d) T2-weighted anatomical images, (b, e) apparent diffusion coefficient (ADC), and (c, f) fractional anisotropy (FA) maps of the prostate. Images in the top row (a-c) and bottom row (d-f) were acquired using the ESA and the ERC-2L, respectively.

Figure 8.

(a) Representative spectrum from the 3DSI semi-LASER acquisition in the prostate with the combined ESA and ERC-2L on receive. Spatial maps of spectra across a single slice covering (b) 2.9-3.3 ppm which includes signals from total choline, polyamine and creatine and (c) 2.4-2.8 showing the inverted citrate peaks.

Table 1

Numerically computed and experimental results comparing the ERC-2L and the ERC-ML coils.

		ERC-2L	ERC-ML
Simulated	B₁⁺ Transmit Efficiency ($\mu\text{T}/\text{W}^{0.5}$)	2.01 ± 1.35	2.04 ± 1.48
	Worst-case peak 1g SAR (W/kg)	10.9	16.7
	B₁⁺ SAR Efficiency ($\mu\text{T}/\text{SAR}^{0.5}$)	0.61	0.50
	SNR (a.u.)	1.46 ± 0.92	1.51 ± 1.07
Experimental	B₁⁺ Transmit Efficiency ($\mu\text{T}/\text{W}^{0.5}$)	1.53 ± 0.45	1.64 ± 1.00
	SNR ($\times 10^5 \text{ Hz}^{0.5}/\text{ml}$)	5.33 ± 2.56	5.41 ± 2.98

Table 2

Experimental SNR and simulated TAP limits for the ESA and ERC-2L.

	Experimental SNR in ROI	Time-averaged power (TAP) limit: Simulation [Heating Study]	
	$\times 10^5 \text{ Hz}^{0.5}/\text{ml}$	Per channel (W)	Total Coil (W)
ESA	1.37 ± 0.22	2.4	38.4
ERC-2L	6.41 ± 3.64	1.2 [1.5]	2.4 [3.0]