Published in final edited form as:

Magn Reson Med. 2014 July ; 72(1): 291–300. doi:10.1002/mrm.24911.

Nineteen-Channel Receive Array and Four-Channel Transmit Array Coil for Cervical Spinal Cord Imaging at 7T

Wei Zhao^{1,2}, Julien Cohen-Adad³, Jonathan R. Polimeni^{1,2}, Boris Keil^{1,2}, Bastien Guerin^{1,2}, Kawin Setsompop^{1,2}, Peter Serano^{1,2}, Azma Mareyam^{1,2}, Philipp Hoecht⁴, and Lawrence L. Wald^{1,2,5,*}

¹A. A. Martinos Center for Biomedical Imaging, Department of Radiology, Massachusetts General Hospital, Charlestown, Massachusetts, USA.

²Harvard Medical School, Boston, Massachusetts, USA.

³Department of Electrical Engineering, Ecole Polytechnique de Montreal, Montreal, Quebec, Canada.

⁴Siemens AG, Healthcare Sector, Magnetic Resonance, Erlangen, Germany.

⁵Harvard-MIT Division of Health Sciences and Technology, Massachusetts Institute of Technology, Cambridge, Massachusetts, USA.

Abstract

Purpose—To design and validate a radiofrequency (RF) array coil for cervical spinal cord imaging at 7T.

Methods—A 19-channel receive array with a four-channel transmit array was developed on a close-fitting coil former at 7T. Transmit efficiency and specific absorption rate were evaluated in a B_1^+ mapping study and an electromagnetic model. Receive signal-to-noise ratio (SNR) and noise amplification for parallel imaging were evaluated and compared with a commercial 3T 19-channel head–neck array and a 7T four-channel spine array. The performance of the array was qualitatively demonstrated in human volunteers using high-resolution imaging (down to 300 μm in-plane).

Results—The transmit and receive arrays showed good bench performance. The SNR was approximately 4.2-fold higher in the 7T receive array at the location of the cord with respect to the 3T coil. The g-factor results showed an additional acceleration was possible with the 7T array. In vivo imaging was feasible and showed high SNR and tissue contrast.

Conclusion—The highly parallel transmit and receive arrays were demonstrated to be fit for spinal cord imaging at 7T. The high sensitivity of the receive coil combined with ultra-high field will likely improve investigations of microstructure and tissue segmentation in the healthy and pathological spinal cord.

*Correspondence to: Wei Zhao, Ph.D., A. A. Martinos Center for Biomedical Imaging, Department of Radiology, Massachusetts General Hospital, Charlestown, MA. wei@nmr.mgh.harvard.edu.

Keywords

high-field MRI; RF coil; transmit array; receive array; cervical spinal cord MRI

INTRODUCTION

MRI is routinely used for diagnosing pathologies in the spinal cord, such as tumors, vascular malformations, traumas, and neurodegenerative diseases. However, the ability to detect subtle pathological features is limited by the image contrast-to-noise ratio and the spatial resolution of current clinical systems (1,2). Ultra-high field MRI (7T and above) can potentially improve diagnostic power using the increased signal-to-noise ratio (SNR) and improved susceptibility contrast to support higher spatial resolution and higher sensitivity. This strategy of trading increased SNR for increased spatial resolution has resulted in an improved ability to detect demyelinating lesions in multiple sclerosis (MS) at 3T compared with 1.5T (3,4).

At ultra-high fields, however, the difficulty of constructing highly parallel receive coils and the associated lack of commercial coils has limited spinal cord imaging. Additionally, the lack of an RF body coil at 7T requires the integration of a surface transmit coil adjacent to the spine receive array or the sharing of the transmit (Tx) and receive (Rx) function within the same spine array elements (a transceiver array of Tx/Rx elements). Several groups have explored the transceiver array strategy for spine imaging at 7T. Wu and colleagues designed a four-channel Tx/Rx array (5) using microstrip elements (6). The length of the array was 48 cm, and the flat design enabled it to be positioned at any vertebral level. Similarly, an eight-element (two rows of four loops) transceiver array has been described for spine imaging (7). This design combined the elements during transmit using a 180° phase shift between the rows to increase the B_1^+ field amplitude along the spine. Sigmund et al. (8) demonstrated a prototype four-channel transceiver C-spine array (also loop elements) that provided increased resolution ($0.2 \times 0.2 \times 3$ mm) at 7T compared with 3T and exquisite detail of cord anatomy with clear white/gray matter delineation.

Although the transceiver approach simplifies the array structure by combining transmit and receive functionality, there are some potential benefits to splitting these two functions. First, placing the transmit elements closer to the body produces more steeply varying E fields, leading to a higher local-to-global specific absorption rate (SAR) ratio for the array, which must be taken into account to ensure that local SAR limits are observed. Second, the receive efficiency should be the primary target of optimization, because a reduced transmit efficiency can often be overcome simply by providing more RF power to achieve the needed flip angle. These two issues suggest a separation of the transmit and receive function with the Rx elements placed closest to the body, while maintaining a spatial offset for the Tx. Finally, the transmit and receive profile from a given loop have different spatial patterns (9) due to the gyro-tropic nature of the MR phenomena (10). Thus, it is not possible to optimize the position of the elements with respect to the spinal cord for both Tx and Rx performance using a single element for both functions. This separate configuration has generally been adopted for array coils at 3T (11,12) and 7T (13,14).

The separated Tx and Rx functionality has been applied in several experimental designs for spinal cord imaging at 7T. Duan et al. (15) evaluated the strategies of quadrature excitation and two Rx elements with an optimal spatial offset between the Tx and Rx elements for improved efficiency and reduced SAR. Vossen et al. (16) presented a receive coil with eight overlapped elements covering the entire spine (total length 90 cm). A quadrature double-loop transmit coil was placed on the anterior side of the body. The Tx coil could be slid down the body to image the entire spine in stations.

In this study, we improve the separated transmit and receive array strategy for sensitivity and parallel imaging performance in the cervical spinal cord by extending the receive array design to 19 channels of small, overlapping loop coils placed on an anatomically curved former designed to follow the natural morphology of the posterior neck and upper thoraces. For Tx, we used a planar four-channel array of decoupled rectangular loops lined along the spine and placed several centimeters from the body. The 19-channel Rx array and four-channel Tx array was tested using SNR, g-factor, and B_1^+ maps as well as in high-resolution anatomical images of the cervical spinal cord and compared with a commercial 3T spine array. For a direct 7T coil to 7T coil comparison, a four-channel Rx array, which followed the geometry and strategy of Sigmund et al. (8), was constructed to compare and assess the benefit of adding additional elements.

METHODS

Coil Former

The housing shown in Figure 1 comprises three parts—the neck former and two shelves on the side used for holding the preamplifiers—and was printed in acrylonitrile butadiene styrene plastic using a rapid prototyping three-dimensional printer (Dimension SST 1200es; Dimension, Inc., Eden Prairie, Minnesota, USA). The design of the coil former was based on a digital human body mesh model (Ansys; HFSS, Cannonsburg, Pennsylvania, USA) to obtain a close-fitting surface of the natural morphology of the head and neck. The distance from the center of the spinal cord to the posterior surface of the neck ranged from 5.6 cm (at C1) to 5.9 cm (at C7) in this model (Fig. 1a). The posterior surface was dilated by 1 cm and truncated anteriorly. Square boxes were added above the shoulders to house the preamplifiers. The planar transmit array was positioned at the bottom of the former, a minimum distance of 5.7 cm from the body. The neck former dimensions were 19.4×25.4 cm for the cross-section and 30.3 cm for the length along the neck, covering the entire C-spine and supporting the body (Fig. 1c). A foam layer (3 mm) was added for subject comfort.

Receive Array

The receive array consisted of 19 overlapped loop elements (7 cm diameter) arranged on the curved surface of the coil former. The loops were tiled in a hexagonal tiling pattern (six loops surrounding a central loop) (11) as shown in Figure 2a. The location of each loop was marked on the former with a hexagon pattern embedded in the three-dimensional printed former. The overlap was based on the critical loop center-to-center distance of 0.75 times the loop diameter to minimize mutual inductance (17). On typical adults, the receive array

covers the whole cervical cord with some coverage of the thoracic spine (T1–T3). The comparison four-channel Rx array built on the same coil former is shown in Figure 2b. A 16-awg-thick tin-plated copper wire was used to form the loops with arches in the wire to allow the wires to cross over one another without touching. All of the discrete components in the loops were mounted on separate small FR4 circuit board pads manufactured with a rapid prototyping circuit router (T-Tech-7000; T-Tech Inc., Norcross, Georgia, USA).

The circuit schematic of the 19-channel Rx loop is shown in Figure 2c. We divided each loop symmetrically with six segments separated by the discrete components mounted on a small circuit board. There were three types of discrete component boards: the simplest had a single capacitor (such as C_3 , C_4 , C_5 , or C_6) and pads to attach the wire loop sections; a component board was used for the output matching circuit (matching capacitor C_1 and active PIN diode trap; D and L); a third board was used for the fine-tuning circuit (variable capacitor C_T and fuse F). The variable capacitor was used to fine-tune the loop resonance to 297.2 MHz, and the fuse F was used for passive protection to prevent a large current induced during transmit should the active detuning fail. The output matching circuit board included an active detuning circuit, including a match capacitor C_1 , a hand-wound inductor L, and a PIN diode D (MA4P7446F-1091T; Macom, Lowell, Massachusetts, USA). The L and C_1 together formed a parallel resonant circuit at the Larmor frequency to block current flow during transmit. The resulting trap circuit detunes the receive loop when the PIN diode D is forward biased. The preamplifiers were placed just outside of the field of view (FOV) of both the Tx and Rx arrays to avoid RF losses from induced currents in the preamplifier socket boards and the conductive components of the preamplifiers from degrading Tx or Rx performance. The preamplifiers (Siemens AG, Healthcare Sector, Erlangen, Germany) were mounted on the shelf on each side of the neck with the proper orientation to the B_0 field for optimal performance (18).

To transform the preamplifier input impedance into a high series impedance within the loop (preamplifier decoupling), we first transformed it to a low impedance across the diode, completing the parallel LC₁ circuit tuned to the Larmor frequency. This parallel LC circuit, in turn, introduces a high series impedance in the coil loop. In this case, little current flows in the loop, and the inductive coupling to other coils is minimized. The length of the coaxial cable (UT-85C-form; Micro-Coax, Pottstown, Pennsylvania, USA), which is used to connect the loop and the preamplifier, was carefully controlled to be near a half wavelength (~23 cm) to preserve the low impedance of the preamplifier across the diode. A section of that coaxial cable was incorporated in to a cable trap, which was added immediately in front of the preamplifier to suppress common-mode cable currents. An additional cable trap (not shown in Fig. 2b) was used on the output of the preamplifiers to block cable currents and avoid interaction with the local Tx array.

The circuit schematic of the four-channel Rx loop is shown in Figure 2d. The matching and trap circuitry was altered slightly to accommodate practical component values for this larger coil element.

Transmit Array

Figure 3a shows the geometries of the four-channel transmit array (6×14.8 cm rectangles and 1-cm-wide traces) constructed with an FR4 printed circuit board. The neighboring rectangular loop elements share a common conductor and its series capacitors. All elements were mounted on a Delrin sheet (made with acetal resin) in the bottom of the coil housing. The circuit schematic of the transmit array is shown in Figure 3d. The decoupling, tuning (to 297.2 MHz) and matching (to 50Ω) of all transmit elements was adjusted with high-power trimmer capacitors (60–0716-10016-600; Tronser, Engels-brand, Germany). The structure was tuned to degeneracy by correctly choosing the rung and loop capacitance to achieve decoupling (19,20). C_{trimmer} on the shared rung was used to minimize the coupling between neighboring elements. A series PIN diode D was placed in each loop to block currents in the transmit loop during receive (to detune the Tx array). The DC current used to switch this diode is applied via the RF coax and returned to ground through an RF choke. To drive the four loops, a 180° balun was used after a 50Ω matching circuit (21).

Figure 3b and 3c show the additional pair of individual loops built to assess the geometric coupling expected between the elements of the array. To assess the geometric coupling between next-nearest neighbors and further neighbors, the S_{21} coupling between these loops was measured with a distance of 12 cm and 18 cm between loop centers.

Coil Testing on the Bench

The unloaded-to-loaded ratio of the quality factor Q ratio was assessed with a single receive loop built only for this test. The 3-dB bandwidth of the loaded or unloaded loop was measured using an S_{21} measurement between two loop probes decoupled from one another (double-loop probes), which lightly coupled into the loop with a 5.5cm distance between probes and the loop. The coil Q was given by $Q = f_0 / \Delta f$, where f_0 is the center frequency and Δf is the width of the response at the -3 dB points. After populating all of the receive elements on the former, a custom-made coil plug socket simulator and a human-shape (head/neck/torso) saline phantom (0.9% NaCl and 0.09% gadolinium; The Phantom Laboratory, Salem, New York, USA) were used for bench testing the assembled array. The setup mimicked the coil plug socket configuration in the scanner and provided control of the PIN diode currents (100 mA) for the Rx and Tx elements as well as the 10-V power supply for preamplifiers.

The element tuning, matching and decoupling were tested by S_{11} and S_{21} measurements using a vector network analyzer. The individual elements were tuned to the Larmor frequency and matched to 50Ω by removing the preamplifier from its socket, while monitoring S_{11} at the preamplifier input. Because the S_{21} coupling between two elements also depends on the matching and tuning of the elements under test, coupling measurements were performed after tuning and matching the individual elements on the saline phantom. These coupling measurements were conducted on the nearest neighbor overlapping elements by using a S_{21} measurement with cables directly connected to the preamplifier sockets of the two elements being tested. When measuring the decoupling between an adjacent pair, all other unused elements of the Rx array were actively detuned (PIN diode under forward bias). The wire arches on the loops were manually bent to adjust the geometric overlap and

thus minimize the S_{21} between neighboring pairs. The preamplifier decoupling manifests itself as a high-Q “dip” at the Larmor frequency in the S_{21} plot measured between double-loop probes in proximity to the detector, whereas the detector preamplifier is in its socket and DC powered and all other loops in the array are actively detuned. The preamplifier decoupling was recorded as the change in the S_{21} , when the coil output coaxial cable is terminated in each of two different match conditions (22): in the first case, the coil is terminated with the powered preamplifier (low impedance termination); in the second case, the coil is terminated with $50\ \Omega$. The active detuning also was measured as the difference of the S_{21} between the double-loop probes of detuned coil and the S_{21} of the coil terminated with $50\ \Omega$.

In order to assess the geometrical inductive coupling between elements in the four-channel Tx array, we compared the coupling of two separated loops positioned with the spacing of the four-channel array. The geometric coupling was recorded as the S_{21} between the outputs of the two loop coils (loaded by the human-shape saline phantom) when the rectangular loop elements were placed a distance of 12 cm center to center (corresponding to nearest neighbors in the array) and also placed a distance of 18 cm center-to-center (corresponding to the next-nearest neighbors in the array). The active detuning of the transmit element was measured by recording S_{21} between double-loop probes in proximity to the detector, while the loop was terminated with $50\ \Omega$, the diode was forward biased, and while all loops in the array were detuned. The difference of these two S_{21} values (in dB) was reported as the active detuning.

Simulation and SAR

A three-dimensional full-wave electromagnetic analysis using finite element modeling (Ansys; HFSS, Cannonsburg, Pennsylvania, USA) and an RF circuit cosimulation method (23) was applied to the four-channel Tx coil using a realistic three-dimensional electromagnetic body model to calculate the B_1^+ field and the SAR in the body. The human model is comprised of 300 objects and 33 tissue types, detailed down to 1 mm (Fig. 1). The coil model included all construction details, including precise dimensions and material electrical properties. The simulation results showed that a -45° phase increment in the RF signal to drive each element (from Tx1 to Tx4) produced a relatively uniform B_1^+ field along the C-spine. Therefore, we wired the Tx array using the same -45° phase increment generated by an 8×8 Butler matrix. The array was excited as a single channel in CP mode using a 1 V sinusoidal continuous RF wave for the SAR and B_1^+ field simulations. The resultant B_1^+ field in the human model is shown in Figure 4a. The magnitude map of the electric field from HFSS is shown in Figure 4c. For 10 g averaged local SAR, the electric field matrix and the density matrix were exported to perform the calculation. The transmit power was limited by the scanner based on this 10 g averaged local SAR using the system coil-file parameters supplied for this purpose. Additionally, we compared the simulated B_1^+ map and the measured B_1^+ map and rescaled the transmit power limit.

Coil Testing in Scanner

The spine coil was developed and tested on a prototype 7T whole-body system (Siemens AG, Healthcare Sector, Erlangen, Germany) equipped with a body gradient coil (SC72, 70

mT/m, slew rate = 200 T/m/s) and 32 receive channels. An 8×8 Butler matrix was used to drive the four Tx channels in the desired B_1^+ shimming phase relationship. To ensure that the coil components would not be heated by absorbing too much RF power from the RF transmitter or gradient system, we ran high duty-cycle high-SAR RF transmit and gradient switching sequences for 15 minutes with a phantom, followed by a temperature inspection with an infrared thermometer to test all the components.

The spine coil was compared with a commercially available 3T 19-channel array (Siemens AG, Healthcare Sector, Erlangen, Germany) comprised of elements from the 12-channel head coil and four-channel neck coil and three elements of the 24-channel spine matrix. This spine coil was used in triple mode, whereby each loop was used independently with a separate receiver channel.

First, we compared the efficiency of the B_1^+ generation of the two systems (body coil at 3T and four-channel local Tx array at 7T) with the head/neck/torso saline phantom ($T_1 = 346$ ms and 360 ms at 3T and 7T, respectively). We measured the B_1^+ transmit field per volt of input RF at the port of the coil (compensated with cable loss and Butler matrix loss) as well as the product of the proton density (ρ) and the array reception profile (Rx) in the sagittal plane. At 3T, the B_1^+ profile of the body coil was obtained by recording the images of a two-dimensional gradient-recalled echo sequence at eight different Tx voltages (10 V to 150 V every 20 V), with 5.1 ms RF pulse duration, 128×128 resolution in plane, FOV 200 mm, 5 mm slice thickness, and $TR/TE/BW = 600$ ms/6.1 ms/890 Hz/pixel and fitting the images to the image intensity equation to obtain a flip-angle map estimate, voxel by voxel, using a standard nonlinear search algorithm in MATLAB (MathWorks Inc., Natick, Massachusetts, USA) in a method similar to that described by Setsompop et al. (24). At 7T, due to the large B_1 field variation and the low efficiency of the transmit coil, we acquired the images at 24 different Tx voltages (10–240 V every 10 V) with 10 ms RF pulse duration and $TR/TE/BW = 600$ ms/7.8 ms/890 Hz/pixel to estimate the flip-angle map for the B_1^+ profile of the four-channel local Tx array. The image intensity equation (25) is:

$$I(V, \vec{r}) = \rho(\vec{r}) Rx(\vec{r}) \frac{(1 - e^{-TR/T_1}) \sin \theta(V, \vec{r})}{1 - e^{-TR/T_1} \cos \theta(V, \vec{r})}. \quad [1]$$

This measurement was used to obtain the flip-angle map $\theta(V, \vec{r})$. B_1^+ profiles can then be estimated from the flip-angle map and the magnitude integral of RF pulse and transmit voltage, based on the following equation:

$$B_1^+(\vec{r}) = \theta(V, \vec{r}) / \gamma \cdot V \int_0^T RF(t) dt. \quad [2]$$

To assess the receive efficiency of the array (without contamination from the transmit efficiency variations of the transmit array), we then calculated the product of $Rx(r)$ and $\rho(r)$. To simplify this calculation, we started with an image obtained with a low flip angle

excitation (30 V applied to the Tx array) and a long TR (1800 ms). A TE of 6.1 ms was used at both 3T and 7T. For this scan, $\cos(\theta) \sim 1$ and $\exp(-TR/T_1) \sim 0$ in Eq. [1]. Then

$$\rho \left(\vec{r} \right) Rx \left(\vec{r} \right) \approx I \left(V, \vec{r} \right) / \sin \theta \left(V, \vec{r} \right). \quad [3]$$

We used this product of the proton density and receive profile in the phantom as a measure of the receive efficiency of the array.

SNR and g-Factor Map

For a fair comparison of the receive coils, we calculated the SNR map that directly represents the properties of receive array, using the B_1^- profile as a sensitivity map in the formula according to Roemer's SNR calculation method (17):

$$SNR = \sqrt{S^H \psi^{-1} S}, \quad [4]$$

where $S = \rho \cdot Rx$ and ψ is the separately measured noise covariance matrix (from data acquired with no RF excitation). For g-factor characterizations relevant to accelerated parallel imaging, the FOV along the phase-encoding direction should be selected to tightly enclose the object to maximize the aliasing induced by the acceleration (26). Given that an L–R phase-encoding direction is usually avoided in spinal cord imaging to prevent aliasing of the shoulders, we evaluated the coil g-factor with an A–P phase-encoding direction in the sagittal plane.

Anatomical MRI

The anatomical cervical spinal cord images were acquired on two healthy human subjects, one male and one female, after review of the institution's human research committee, and informed consent was obtained from each subject.

For shim adjustment at 7T, we used a standard double-TE two-dimensional gradient-recalled echo sequence and generated B_0 field maps from the measured phase difference. The region used in the B_0 field optimization was restricted in the A–P direction to improve the shim coil compensation in the area of the spine at the expense of the homogeneity in head. For all FLASH acquisitions, phase stabilization (which uses a short TE navigator to correct the phase of each k-space line) was applied to minimize motion and breathing artifacts.

In one subject, axial gradient echo, multi-echo FLASH images were acquired at 3T and 7T. Slices were centered at the C5/C6 vertebral level and TE-adjusted to match gray/white contrast across field strength. The common imaging parameters were as follows: TR = 512 ms; matrix = 384×384 ; FOV = 192 mm; $R = 2$ (GRAPPA) acceleration; in-plane resolution = 0.5×0.5 mm; 13 slices; mm slice thickness; partial Fourier factor = 6/8; A–P phase-encoding direction; and acquisition time = 1 min 44 sec. For the 3T setting, TE = [17, 25, 32, 40] ms and BW = 260 Hz/pixel. For the 7T setting, TE = [7.8, 15, 22, 29.4] ms and BW = 195 Hz/pixel. At 7T only, axial images with high in-plane resolution were also acquired with the following parameters: TR = 400 ms; TE = [4.9, 16.3, 25.8] ms; in-plane resolution

= 0.3×0.3 mm; matrix = 512×512 ; FOV = 156 mm; $R = 2$ (GRAPPA) acceleration; 10 slices; 3 mm slice thickness; partial Fourier factor = 6/8; BW = 255 Hz/pixel; A–P phase encoding direction; and acquisition time = 3 min 34 sec.

In another subject, sagittal and axial FLASH images were acquired at 7T with the following parameters: TR = 514 ms, TE = [7.8, 15, 22, 29.4] ms (four echoes averaged), 0.35 mm in-plane, 3 mm slice thickness, matrix = 534×480 , $R = 2$ (GRAPPA) acceleration, bandwidth (BW) = 195 Hz/pix, phase stabilization, 2 averages, acquisition time = 4 min 30 sec.

RESULTS

Transmit Array Characteristics

The S_{21} coupling between pairs of Tx elements was less than -16 dB despite a natural geometric coupling of -8.1 dB between the next-nearest neighbors in a loaded case. This is sufficient for driving the coils independently as a parallel transmit array. The active detuning of the Tx loop was approximately -20 dB as calculated by recording the S_{21} between the double-loop probes. Table 1 shows the S-parameter matrix in the four-channel Tx loop array with loading. Table 2 is generated from simulation and shows that the coil model in simulation has nearly identical behavior to the physical coil.

According to the electromagnetic simulation of the electric and magnetic fields in the human model (Fig. 4a and 4c), when the array is driven by a -45° phase increment, the maximum 10 g averaged local SAR is 0.0015 W/kg in the human body with a 1 V sinusoidal continuous RF wave excitation split by the 8×8 Butler matrix to each of the four channels. Figure 4d shows the experimentally determined B_1^+ maps of the sagittal slice in the phantom obtained by the quantitative B_1 mapping method at 7T. The simulated B_1^+ map and measured B_1^+ map (Fig. 4b and 4d) were compared in the spinal cord region-of-interest (ROI) indicated by a curved black box at a depth of 5.8 cm from the model's posterior edge (taken from the anatomy of the simulation model). The average B_1^+ within this ROI in the simulation results was 11.3 nT/V and 14.3 nT/V in the phantom measurement with the power determined at the input port. The maximum B_1^+ was 35.0 nT/V in the simulation and 56.4 nT/V at the input port in the measurement (33.2 nT/V without RF loss compensation). Therefore, a conservative factor of 2 for the SAR limit was included to further ensure patient safety. The average B_1^+ efficiency in the ROI for the 7T Tx array was 14.3 nT/V, which was about half of the average B_1^+ efficiency of 3T body coil (30.5 nT/V) (Fig. 4e).

Receive Array Characteristics

The ratio of unloaded/loaded Q (with phantom) of the 19-channel 7T Rx array elements was ~ 10.9 (240/22) for a 70-cm loop (19-channel Rx array) at 7T, indicating nearly complete body load dominance. The Q ratio of the elements in the four-channel Rx array (size 150 mm $\times$ 70 mm and 105 mm $\times$ 100 mm) was 12.5 (175/14) for the larger element and 12.3 (173/14) for the smaller element. Active detuning provided >25 dB isolation between the tuned and detuned states for all channels in both the four-channel Rx and the 19-channel Rx arrays. Preamplifier decoupling was approximately -20 dB for all elements. The coupling between neighboring elements (S_{21}) ranged from -14 to -18 dB (mean coupling = -15.3

dB). The average of the off-diagonal elements in the noise correlation coefficient matrix was 18.2% from the 19-channel Rx array.

Figure 5a shows the receive profile SNR map of the sagittal image at 3T in the head/neck/torso saline phantom for an image combined with the noise covariance-weighted root sum-of-squares for the commercial head–neck array. Figure 5b and 5c shows the SNR map of the 7T 19-channel Rx array and that of the four-channel Rx array in the same phantom. In the spinal cord ROI marked by the black box, the average SNR of the 7T 19-channel Rx array ($\text{SNR}_{7\text{T},19\text{ch}} = 450$) is approximately 4.2-fold higher than the 3T SNR ($\text{SNR}_{3\text{T}} = 106$), and about 1.5-fold higher than the 7T four-channel Rx array ($\text{SNR}_{7\text{T},4\text{ch}} = 299$). Discounting the expected linear field effect on SNR, whereby we expect a 7/3 gain in SNR between 7T and 3T, leaves an additional factor of 1.8 $\times$, which we attribute to the more highly parallel and closely fitting geometry of the 7T Rx array. Figure 6 shows inverse g-factor maps ($1/g$) for the three coils at various acceleration factors (R), based on the SENSE method (27). The g-factor improvements of the 19-channel Rx array can be summarized by the average $1/g$ value calculated in the phantom 1.1-fold (standard deviation = 0.05) at acceleration $R = 3$ and 1.6-fold (standard deviation = 0.11) at acceleration $R = 4$ compared with the 3T commercial array, 1.6-fold at acceleration $R = 3$ and 5.6-fold at acceleration $R = 4$ compared with the 7T four-channel array.

Conventional Imaging

Figure 7 shows the high-resolution sagittal and axial T_2^* weighted spinal cord images at 7T in both the sagittal and axial planes with 0.35 mm in-plane resolution and 3 mm thick slices. With the 19-channel 7T spine array, the imaging area covered the whole cervical spine, down to T3-T4 levels, depending on the subject's size.

Figure 8 shows the spinal cord image at the C5/C6 vertebral level from the 3T 19-channel commercial coil with 0.5-mm resolution (a) and the 7T 19-channel array with both 0.5 mm (b). The 7T spine coil provided higher SNR than its equal resolution comparison image, likely due to a combination of the four-fold higher Rx sensitivity and improved T_2^* contrast at 7T. Figure 8c shows the 0.3-mm high-resolution image, with the distinct boundary between white matter and gray matter, the apparent ventral and dorsal nerve fibers, and the anterior median fissure.

DISCUSSION

The separate coil design of a four-channel Tx array and a 19-channel Rx array is feasible for imaging the human cervical spinal cord at 7T with high resolution. The tightly fitting 19-channel Rx array improves sensitivity and acceleration compared to the design used at 3T and the design of four-channel Rx array. The odd number of Rx array elements chosen (19 rather than 16, 24 or 32) simply reflects the tiling needed to cover the desired territory with a symmetric arrangement. It is consistent with our design philosophy that the array design should not be constrained by the number of RF channels—the design engineer should be free to choose the best design, not a particular number of channels. There was a frequency shift between the loaded and unloaded cases, both in the receive loop and in the transmit loop. This shift required the use of a loading phantom during element tuning, matching, and

decoupling, which we mention in the section on coil testing on the bench. Although we kept a copper foil cylinder to simulate the presence of the RF shield in the 7T gradient coils, we found that bench measurements were not significantly affected by the shield and preceded with bench measurements, which were not conducted inside this dummy shield. Finally, we observed differences between the simulated and measured B_1^+ maps. These differences most likely stem from the shape, position, and compartment differences between the phantom and the digital human model.

The SNR measured at 7T showed a nearly 4.2-fold increase relative to the 3T images. Although 7/3 of this factor can be attributed to the field strength, the remaining increase can be attributed to the coil and T_2^* difference at 3T and 7T. The TEs used in the SNR measurement and assumptions of a T_2^* of 30 ms at 7T and 40 ms at 3T demonstrate that the T_2^* weighting is only a ~5% effect. Thus, the coil's net SNR contribution factor is 1.8. The in vivo imaging tests qualitatively support the SNR measurement findings. The comparison at fixed (0.5 mm) in-plane resolution shows higher SNR in the 7T array image. Therefore, we showed a 7T high-resolution spinal cord sagittal image and axial images (acquired at level C1–C7) and a 0.3-mm in-plane resolution image to demonstrate the coverage and high resolution offered by this 7T coil design. The g-factors of the 7T array were substantially better than those of the 3T coil; some of this gain is attributable to the B_1^- phase modulations due to the shorter wavelength at 7T (28). Additionally, although the commercial coil also had 19 receive elements, these elements were arranged differently. They were designed to cover the brain as well as the full neck and C-spine. In contrast, the 7T array developed in this work was designed to only image the C-spine.

The primary benefit of the four-channel Tx array is the good coverage of whole C-spine with a relatively uniform B_1^+ field and the flexibility to be able to drive the transmit as an array for parallel transmit applications. On the other hand, the B_1^+ efficiency of the four-channel Tx array in the spinal cord at 7T is lower than that of the body coil at 3T. This likely arises from the large coil–patient gap we imposed on the coil geometry to reduce the potential for local E field hot spots. Further work will be required to fully optimize the gap between transmit array and the human body to fully maximize the performance given the strict safety constraints (29). The best insight into the practical use of the transmit coil probably comes from the comparison with the 3T body coil, which is the gold standard familiar to most spine imagers. The fact that our transmit coil is about half of the efficiency of the 3T body coil suggests that it will be a limiting factor for some excitations and that a relatively high transmit voltage will be needed to achieve a given flip angle. To the extent that 7T might demand even higher peak-power pulses to overcome off-resonance effects exacerbates this problem. Improving the transmit coil for the spine array is therefore an area for future work.

CONCLUSIONS

A highly parallel Rx array with parallel local Tx capabilities was designed, bench tested, and validated in phantom and human imaging at 7T. We compared the coil's transmit efficiency to that of the 3T body coil and assessed its uniformity across the spinal cord with B_1^+ simulations and phantom mapping. The SNR maps were corrected to remove the effect of

spatially varying flip angles and compared with those obtained at 3T with a commercial head–neck C-spine coil and a four-channel 7T Rx array based on a previously published 7T spine array. The 7T 19-channel spine array was found to have 4.2 times the SNR than the 3T setup, a gain consistent with the high-resolution imaging comparisons. The 19-channel Rx array showed a 1.5-fold improved SNR at the spinal cord compared with the four-channel Rx array. This comparison, as well as the 0.3-mm resolution spinal cord image, shows the benefits of 7T MRI for delineating lesions in the spinal cord (30,31) and that the ultra–high-resolution T_2^* -weighted brain imaging that has become a mainstay of 7T applications can be effectively translated to the C-spine.

ACKNOWLEDGMENTS

We thank Eli Siskind for help with coil former design.

Grant sponsor: National Institutes of Health; Grant numbers: P41EB015896, K24NS59892; Grant sponsor: National Multiple Sclerosis Society; Grant number: FG1892A1/1.

REFERENCES

1. Gilmore CP, Geurts JGG, Evangelou N, Bot JCJ, van Schijndel RA, Pouwels PJW, Barkhof F, Bü L. Spinal cord grey matter lesions in multiple sclerosis detected by post-mortem high field MR imaging. *Mult Scler*. 2009; 15:180–188. [PubMed: 18845658]
2. Bü L, Geurts J. Magnetic resonance imaging as a tool to examine the neuropathology of multiple sclerosis. *Neuropathol Appl Neurobiol*. 2004;106–117. [PubMed: 15043708]
3. Bachmann R, Reilmann R, Schwindt W, Kugel H, Heindel W, Krämer S. FLAIR imaging for multiple sclerosis: a comparative MR study at 1.5 and 3.0 Tesla. *Eur Radiol*. 2006; 16:915–921. [PubMed: 16365731]
4. Sicotte NL, Voskuhl RR, Bouvier S, Klutch R, Cohen MS, Mazziotta JC. Comparison of multiple sclerosis lesions at 1.5 and 3.0 Tesla. *Invest Radiol*. 2003; 38:423–427. [PubMed: 12821856]
5. Wu B, Wang C, Krug R, et al. 7 T human spine imaging arrays with adjustable inductive decoupling. *IEEE Trans Biomed Eng*. 2010; 57:397–403. [PubMed: 19709956]
6. Zhang X, Ugurbil K, Chen W. Microstrip RF surface coil design for extremely high-field MRI and spectroscopy. *Magn Reson Med*. 2001; 46:443–450. [PubMed: 11550234]
7. Kraff O, Bitz AK, Kruszona S, Orzada S, Schaefer LC, Theysohn JM, Maderwald S, Ladd ME, Quick HH. An eight-channel phased array RF coil for spine MR imaging at 7 T. *Invest Radiol*. 2009; 44:734–740. [PubMed: 19809342]
8. Sigmund EE, Suero GA, Hu C, McGorty K, Sodickson DK, Wiggins GC, Helpert JA. High-resolution human cervical spinal cord imaging at 7 T. *NMR Biomed*. 2012; 25:891–899. [PubMed: 22183956]
9. Keltner J, Carlson J. Electromagnetic fields of surface coil in vivo NMR at high frequencies. *Mag Reson Med*. 1991; 22:467–480.
10. Tropp J. Reciprocity and gyrotropism in magnetic resonance transduction. *Phys Rev A*. 2006; 74:1–13.
11. Wiggins GC, Triantafyllou C, Potthast A, Reykowski A, Nittka M, Wald LL. 32-channel 3 Tesla receive-only phased-array head coil with soccer-ball element geometry. *Magn Reson Med*. 2006; 56:216–223. [PubMed: 16767762]
12. Cohen-Adad J, Mareyam A, Keil B, Polimeni JR, Wald LL. 32-channel RF coil optimized for brain and cervical spinal cord at 3 T. *Mag Reson Med*. 2011; 66:1198–208.
13. Wiggins GC, Potthast A, Triantafyllou C, Wiggins CJ, Wald LL. Eight-channel phased array coil and detunable TEM volume coil for 7 T brain imaging. *Mag Reson Med*. 2005; 54:235–240.

14. Adriany G, Auerbach EJ, Snyder CJ, Gözübüyük A, Moeller S, Ritter J, Van de Moortele P-F, Vaughan T, Ugurbil K. A 32-channel lattice transmission line array for parallel transmit and receive MRI at 7 tesla. *Magn Reson Med*. 2010; 63:1478–1485. [PubMed: 20512850]
15. Duan, Q.; Sodickson, DK.; Lattanzi, R.; Zhang, B.; Wiggins, GC. Optimizing 7 T spine array design through offsetting of transmit and receive elements and quadrature excitation.. *Proceedings of the Joint Annual Meeting ISMRM-ESMRMB 2010*; Stockholm, Sweden. 2010. p. 51
16. Vossen M, Teeuwisse W, Reijnierse M, Collins CM, Smith NB, Webb AG. A radiofrequency coil configuration for imaging the human vertebral column at 7 T. *J Magn Reson*. 2011; 208:291–297. [PubMed: 21134773]
17. Roemer PB, Edelstein WA, Hayes CE, Souza SP, Mueller OM. The NMR phased array. *Magn Reson Med*. 1990; 16:192–225. [PubMed: 2266841]
18. Possanzini, C.; Boutelje, M. Influence of magnetic field on preamplifiers using GaAs FET technology.. *Proceedings of the 16th Annual Meeting of ISMRM*; Toronto, Canada. 2008. p. 1123
19. Jevtic, J. Ladder networks for capacitive decoupling in phased-array coils.. *Proceedings of the 9th Annual Meeting of ISMRM*; Glasgow, Scotland, UK. 2001. p. 17
20. Lee RF, Giaquinto RO, Hardy CJ. Coupling and decoupling theory and its application to the MRI phased array. *Mag Reson Med*. 2002; 48:203–213.
21. Peterson, DM.; Wolverson, BL. Simulation and Analysis of balanced matching circuits at 3 Tesla.. *Proceedings of the 10th Annual Meeting of ISMRM*; Honolulu, Hawaii, USA. 2002. p. 885
22. Reykowski A, Wright SM, Porter JR. Design of matching networks for low noise preamplifiers. *Magn Reson Med*. 1995; 33:848–852. [PubMed: 7651124]
23. Kozlov M, Turner R. Fast MRI coil analysis based on 3-D electromagnetic and RF circuit co-simulation. *J Magn Reson*. 2009; 200:147–152. [PubMed: 19570700]
24. Setsompop K, Alagappan V, Gagoski B, et al. Slice-selective RF pulses for in vivo B_1^+ inhomogeneity mitigation at 7 tesla using parallel RF excitation with a 16-element coil. *Mag Reson Med*. 2008; 60:1422–1432.
25. Nishimura, DG. *Principles of Magnetic Resonance Imaging*. Stanford University: Self-Published; 1996. p. 155
26. Adriany G, Van de Moortele P-F, Wiesinger F, et al. Transmit and receive transmission line arrays for 7 Tesla parallel imaging. *Mag Reson Med*. 2005; 53:434–445.
27. Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI. *Mag Reson Med*. 1999; 42:952–962.
28. Wiesinger F, Van de Moortele P-F, Adriany G, De Zanche N, Ugurbil K, Pruessmann KP. Parallel imaging performance as a function of field strength—an experimental investigation using electrodynamic scaling. *Magn Reson Med*. 2004; 52:953–964. [PubMed: 15508167]
29. Kraff O, Wrede KH, Schoenberg T, Dammann P, Noureddine Y, Orzada S, Ladd ME, Bitz AK. MR safety assessment of potential RF heating from cranial fixation plates at 7 T. *Med Phys*. 2013; 40:042302. [PubMed: 23556915]
30. Cohen-Adad J, Zhao W, Keil B, Ratai E-M, Lawson R, Dheel C, Wald LL, Rosen BR, Cudkowicz M, Atassi N. 7 T MRI of the spinal cord can detect corticospinal tract abnormality in ALS. *Muscle Nerve*. 2013; 47:760–762. [PubMed: 23553571]
31. Cohen-Adad J, Zhao W, Wald LL, Oaklander AL. 7 T MRI of spinal cord injury. *Neurology*. 2012; 79:2217. [PubMed: 23183281]

FIG. 1.

Design of separate transmit and receive arrays for spinal cord imaging relative to the body model. **a:** The anatomically contoured receive array was placed on the neck former close to the body. The C-spine vertebrae (C1–C7) are shown in blue. The depth of the spinal cord (magenta) ranges from 5.6 to 5.9 cm in the cervical spine. The planar transmit array is positioned further from the body at a minimum distance of 5.7 cm and a maximum distance of 9.7 cm. **b:** Top view of the assembly. **c:** The coil housing pictured from above. The housing consists of three parts (the neck former in the center and the shelves on either sides) with a 3-mm foam padding for subject comfort (dark gray). [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

FIG. 2.

Receive array. **a:** Photograph of the 19-channel receive array. **b:** Photograph of the four-channel receive array reconstructed as closely as possible following the strategy and geometry of Sigmund et al. (8). The four-channel receive array was built on the same coil former as our 19-channel receive coil's former. **c:** Circuit schematic for an element of the 19-channel Rx array. C₃, C₄, C₅, and C₆ are single capacitors; F is the fuse of the loop; C_T is the tuning capacitor; C₁ and L consist of an LC circuit; and D is the PIN diode to provide a parallel resonant circuit to detune the receive loop. **d:** Circuit schematic of the four-channel receive array. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

FIG. 3.

Transmit array. **a:** Photograph of the four-channel C-spine transmit array. Tx1, Tx2, Tx3, and Tx4 refer to the independent loops, which can be excited in parallel. **b:** Photograph of two separate loops with a distance of 12 cm (the distance between the next-nearest neighbors in the transmit array). **c:** Photograph of two separate loops with a distance of 18 cm (the distance between Tx1 and Tx4). The S_{21} coupling between the loops in panels b and c were measured to provide a reference for the values measured in the full array. **d:** Circuit schematic of the four-channel transmit array. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

FIG. 4.

Electromagnetic simulation and measured B_1^+ map for the transmit coils used. The curved black box in each map indicates the location of the spinal cord, which was determined from the human digital model and transferred to the uniform saline phantom by assuming the cord is 5.8 cm from the posterior surface for the phantom. **a:** B_1^+ map of CP mode of the four-channel transmit array driven with -45° phase increment for each channel (CP mode). **b:** Zoomed version of simulated B_1^+ for the four-channel transmit array. **c:** Magnitude map of the simulated E field for the four-channel transmit array. **d:** Measured B_1^+ map for the four-channel transmit array in uniform saline phantom. **e:** Measured B_1^+ profile for the 3T body coil in the same phantom. **f:** Rescaled version (to match the scale of panel e) of the measured B_1^+ map for the four-channel transmit array.

FIG. 5.

Comparison of SNR maps from the 7T spine coils and the 3T spine coil measured in the saline phantom. The spinal cord ROI locations are marked by curved black boxes. **a:** 3T SNR map from the commercial head–neck array (average SNR in the ROI: $\text{SNR}_{3\text{T}} = 106$). **b:** 7T SNR map from the 19-channel spine array (average SNR in the ROI: $\text{SNR}_{7\text{T}, 19\text{ch}} = 450$). **c:** 7T SNR map from the four-channel spine array (average SNR in the ROI: $\text{SNR}_{7\text{T}, 4\text{ch}} = 299$).

FIG. 6.

Inverse g-factor maps (plotted as $1/g$) for a closely fitted sagittal FOV prescribed for the cervical region ($\text{FOV} = 200 \times 200 \text{ mm}^2$) with the A–P phase-encoding direction calculated from the measured receive maps and noise correlation matrix. **a:** Results for the 3T head–neck array. **b:** Results for the 7T 19-channel spine array. **c:** Results for the 7T four-channel spine array.

FIG. 7.

Sagittal and axial T_2^* -weighted 7T spinal cord images from the 19-channel 7T spine coil. Left: Sagittal image showing the region of coverage (C1–C7). Right: Axial spinal cord images of C1–C7 using gradient-recalled echo multi-echo FLASH (RMS combination from the four echoes), 0.35 mm in-plane (no interpolation), and 3-mm slice thickness in 4 min 30 sec.

FIG. 8.

Comparison of 3T and 7T C-spine images (C5/C6). **a:** 3T image using the 19-channel commercial coil with $0.5 \times 0.5 \times 3$ mm resolution (acquisition time = 1 min 44 sec). **b:** 7T image using the 19-channel receive array and four-channel transmit array with $0.5 \times 0.5 \times 3$ mm resolution (acquisition time = 1 min 44 sec). **c:** 7T image using the 19-channel receive array and four-channel transmit array with $0.3 \text{ mm} \times 0.3 \text{ mm} \times 3 \text{ mm}$ resolution (no interpolation) (acquisition time = 3 min 43 sec).

Table 1

S-Parameters in Four-Channel Tx Loop Array and S_{21} of Two Separate Loops with Distances 12 cm and 18 cm Measured on Bench with the Human-Shaped Saline Phantom

	Tx1	Tx2	Tx3	Tx4
Tx1	-28.2	-16.6	-16.2	-19.1
Tx2		-32	-16.1	-17.4
Tx3			-29.8	-22.5
Tx4				-30.2
Two separate loops			-8.1 (12 cm)	-14.2 (18 cm)

Values are presented in dB.

Table 2

S-Parameters of Electromagnetic Simulation Model in the Four-Channel Transmit Array with Digital HFSS Digital Human Model

	Tx1	Tx2	Tx3	Tx4
Tx1	-30.0	-18.2	-20.0	-19.4
Tx2		-29.9	-15.8	-19.8
Tx3			-29.9	-20.0
Tx4				-30.0

Values are presented in dB.