Published in final edited form as:

Magn Reson Med. 2015 October ; 74(4): 1189–1197. doi:10.1002/mrm.25817.

A 7T Spine Array Based on Electric Dipole Transmitters

Qi Duan¹, Govind Nair², Natalia Gudino¹, Jacco A. de Zwart¹, Peter van Gelderen¹, Joe Murphy-Boesch¹, Daniel S. Reich², Jeff H. Duyn¹, and Hellmut Merkle¹

¹ Laboratory of Functional and Molecular Imaging, National Institute of Neurological Disorders and Stroke, National Institutes of Health, Bethesda, Maryland, USA

² Division of Neuroimmunology and Neurovirology, National Institute of Neurological Disorders and Stroke, National Institutes of Health, Bethesda, Maryland, USA

Abstract

Purpose—In this work the feasibility of using an array of electric dipole antennas for RF transmission in spine MRI at high field is explored.

Method—A 2-channel transmit array based on an electric dipole design was quantitatively optimized for 7T spine imaging and integrated with a receive array combining 8 loop coils. Using B_1^+ mapping, the transmit efficiency of the dipole array was compared to a design using quadrature loop pairs. The radio-frequency (RF) energy deposition for each array was measured using a home-built dielectric phantom and MR thermometry. The performance of the proposed array was qualitatively demonstrated in human studies.

Results—The results indicate dramatically improved transmit efficiency for the dipole design as compared to the loop excitation. Up to 76% gain was achieved within the spinal region.

Conclusion—For imaging of the spine, electric-dipole based transmitters provided an attractive alternative to the traditional loop-based design. Easy integration with existing receive array technology facilitates practical use at high field.

Keywords

High field MRI; dipole antenna; spinal imaging; transmit array; receive array

INTRODUCTION

MRI of the spine at 3T has been found to offer improvement over 1.5T in terms of signal-to-noise ratio (SNR) and tissue delineation, but with new challenges related to B_1 and B_0 inhomogeneity (1). MRI of the spine at 7T is hampered by a lack of commercial radio-frequency (RF) receive coils and a body transmit coil. Development of 7T spine arrays therefore requires implementation of both transmit and receive functions in the same coil assembly, either via transceiver elements or via separate transmit and receive elements. In the past few years, several groups have proposed various designs (2-7) utilizing loop-based transmit structures. However, given the generally large field-of-view (FOV) required for

spine imaging in comparison with the RF wavelength at 7T (e.g. ~13 cm in muscle), loop-based transmit elements generally require a rather complicated layout and cabling (2-5). Although it is possible to simplify RF excitation by exploiting the traveling wave phenomenon (e.g. by employing a patch antenna at magnet bore entrance (8,9)), this may result in reduced transmit efficiency when compared with transmitters placed close to the area of interest (10).

In recent studies, use of electric dipole antennas has been explored as an alternative mode of RF transmission at high field (11-18), mostly for small FOV imaging. For head or prostate imaging at 7T, the antennas were shortened to ~20 cm to avoid unnecessary RF deposition outside the region-of-interest (ROI). However, when the conventional “half-wave dipole” (19) is constructed only with conductor strips and without additional lumped elements such as inductors, its length is well-suited for the larger superior-inferior (S-I) FOV required for spine imaging. In one application, the electric dipole was used in a so-called “dark mode” (i.e. only generating longitudinal B_I^+ field, which is invisible in MRI) to modify the electric field distribution for a cervical spine array based on loop designs (20). But for spine imaging at 7T, the use of electrical dipoles for B_I transmission has not been reported. In this work, the feasibility of using electrical dipole antennas is explored by evaluating both transmit efficiency and energy deposition, relative to a conventional loop-based design (4).

METHODS

Transmit and receive elements of our spine array were divided between a top layer immediately under the patient surface housing the receive-only loop array and a bottom layer located 4 cm below the patient housing the transmit-only dipole array.

Design and Optimization of the Electric Dipole Array

Our transmit-only dipole array (Figure 1a) employed two transmit elements, each consisting of a conventional half-wave dipole. The dipoles were constructed with straight 25 mm wide copper tape and fed at their centers with RF. For each of the two segments of each dipole element, the copper strips mounted on a 5 mm thick acrylic board. The dipoles were positioned 4 cm below the patient surface to allow space for the receive array. As expected, each dipole was found to have a resonance with a wavelength corresponding to twice the total dipole length. The resonant frequency of the dipole shifted lower when it was placed near a highly dielectric and conductive load such as human body. For tuning, a phantom that presented loading characteristics similar to a human body at 297.2 MHz was placed 4 cm above the dipoles. The lengths of the dipole segments were adjusted to bring the resonant frequencies close to the NMR Larmor frequency (297.2 MHz at 7T). Each dipole was fed via a Lorenz (lattice) balun (21) (Figure 1b) matched to ~90 Ω (i.e. two 35 nH inductors and two 8 pF capacitors, consistent with theoretical prediction). Two diodes were added in series with the feeding points of each dipole so that it could be detuned during signal reception (Figure 1b). The RF power of the transmit port of the 7T scanner (Siemens Magnetom 7T, Erlangen, Germany) was split equally between each dipole using a home-built lumped-element quadrature hybrid (22), using a minimal amount of metal to avoid eddy currents in

close proximity to the dipoles. Reflected power from the isolation port of the hybrid was terminated with a high-power 50- Ω load (Figure 1a).

A body-mimicking phantom was constructed using a plastic container of 425 mm width, 600 mm length, and 170 mm depth. It was used to optimize coil tuning and perform initial imaging experiments. To mimic the bulk electrical properties of muscle of the back at 7T ($\sigma = 0.79$ S/m, $\epsilon_r = 59$) (23), the phantom was filled to a depth of 110 mm with a sugar-gel recipe developed earlier (24) (available at <http://www.amri.ninds.nih.gov/phantomrecipe.html>).

Two additional optimization steps were performed on the dipole array before the detuning circuits were added. First, with the elements connected in quadrature (90° inter-channel phase delay), the B_1^+ fields of the elements were mapped in the central axial plane using a modified Bloch-Siegert approach (25). The separation between the dipoles was empirically minimized until coupling between the elements significantly changed their B_1^+ distributions. In the next step, the transmit efficiency (defined as B_1^+ in μ T per unit input voltage in V) was measured at the center axial plane by varying the phase delay between the hybrid and individual dipoles. Phase delays were adjusted by inserting additional cable of different lengths between the dipole feeding points and either hybrid output ports. The phase delay was varied in $\sim 30^\circ$ steps, and the average transmit efficiencies were fit using 6th order polynomials as a function of inter-channel phase delay. Beside phase delay, the offset in L-R direction between the center of the array and the center of the ROI could also affect the transmit efficiency. Thus similar to (4), the transmit efficiency within a 20 mm diameter circular ROI at 50 mm depth was then numerically maximized over the two variables. After this optimization, the dipole array was converted to a transmit-only configuration by implementing the detuning circuits as mentioned above.

Receive Array Design

Since the main goal of this work was to improve transmit efficiency, a conventional array of overlapping loops was used for reception. Starting with the array described in (4), two columns of 10cm square coils, four per column, were configured in receive-only mode. A third column of transmit-only coil as well as the transmit feature in the center column was removed. Each coil was tuned and matched using the loading phantom. For reception, high reflection, low noise GaAs-FET preamplifiers (type Siemens 10185702 + 10185751; Stark Contrast, Erlangen, Germany) were used, which included active detuning circuits on their circuit boards. For additional protection from RF transmit interference, passive detuning (including an inductor and a pair of cross-diodes) was added to one capacitor of each element. To prevent inter-channel coupling through the RF receive cables, floating cable traps (26) were added on each cable bundle that contains all four coaxial cables connecting the preamplifiers on receive coils to the coil system plug.

Use of the Loop-based Transmit-Receive Spine Array as Reference

In order to compare the transmit efficiency and the RF heating of the dipole-based transmit array to the loop-based design (4), the latter was slightly modified: 1) inline traps were added on the coax cables connecting the four receive-only coils; 2) the bundle traps were

replaced with custom-built floating traps (26); and 3) the transmit cables were cut to the determined optimal lengths.

MR Experiments

Phantom measurement of RF transmit efficiency—The experimental setup is illustrated in Figure 1c. A ROI representing the expected location of the spine is marked as a white circle in the center axial plane of the phantom (marked by yellow/black dashed rectangle), centered 5 cm above the bottom of the phantom in anterior-posterior (A-P) direction. Unless specified otherwise, experiments were conducted with the center planes of the phantom ($x'=0$ plane and $z'=0$) coinciding with $x=0$ plane and $z=0$ plane, as indicated in Figure 1c. For simplicity, we define this position as the “center position”.

Transmit efficiency was measured using Bloch-Siegert B_1^+ mapping (25) as follows. First, for calibration, the RF envelope voltage required to generate an 11.7 μT (500 Hz) B_1^+ field was estimated at the center of the ROI ($x'=0, y'=0, z'=0$) at the center position. After transmitter calibration, axial B_1^+ maps were acquired. Given the length of the dipole and the increasing gradient nonlinearity in the S-I direction, mapping was performed in sequential axial scans by moving the patient table to cover sections of the FOV. Each section covered a 10 cm region along the long-axis (S-I) of the FOV. Five sections centered on the long-axis of the phantom (S-I) were selected and imaged. The acquired B_1^+ maps were converted to B_1^+ efficiency by dividing the measured B_1^+ amplitudes (in μT) by the RF envelop voltages (in V) used to acquire the data.

Phantom measurement of RF induced heating—To evaluate the RF energy deposition, MR thermometry based on the water proton resonant frequency (PRF) shift was performed for each array. Multi-echo gradient echo (GRE) images were acquired while depositing RF energy with a 32 kHz off-resonance pre-pulse that deposited substantial RF energy deposition without affecting MRI signal. In order to avoid impractically long experiments (due to the required cool-down time between the acquisitions), thermometry was only performed at a single section that covered the area of largest expected temperature increase, which was close to the feeding point (i.e. $z'=0$) based on the theory (19) and simulation. To minimize the effects of gradient non-linearities, the experiment was performed at the center position, as indicated in Figure 1c. Coronal rather than axial slices were chosen to maximize coverage parallel to the bottom part of the phantom. Duration of the heating was 10 minutes with an average power deposition of approximately 33 W, as measured using the power monitor on the scanner. Temperature changes could be derived from changes in image phase (induced by PRF shift) before and after the heating. As mentioned in (24), the multiple resonances of sucrose, whose resonance frequencies do not change notably with temperature, can confound PRF-based MR thermometry (similar to what is observed in water-lipid environments (27)). Therefore, a nonlinear least squares fitting approach was used to determine changes in water proton resonance frequency in the presence of the known sucrose signal (28), rather than a linear scaling of phase changes as is done in conventional PRF-based MR thermometry (29).

Anatomical MRI—To demonstrate practical use of the array, anatomical images of the vertebral column and the spinal cord were acquired on two healthy volunteers; all imaging was approved by the Institutional Review Board. On the first volunteer, single slice images were acquired with breathholding and without fat saturation or any saturation band. First, an axial B_1^+ map with a matching anatomical GRE image was acquired at $z=0$ to confirm the B_1^+ efficiency *in vivo*. Then two additional single slice GRE images in axial and sagittal planes were acquired with imaging parameters similar to what was presented in the literature, e.g. (2,4,5). On the second volunteer, a high-resolution image of the spinal cord was acquired without breathhold using a 2D GRE sequence. Imaging artifacts caused by breathing were corrected based on the built-in navigator echoes, in combination with information about the TE dependence of the artifact, derived from three otherwise identical pre-scan repetitions without phase encoding.

RESULTS

Array Configurations

The prototype transmit dipole array, receive array, and transmit/receive (T/R) loop array are shown in Figure 2a, 2b, and 2c respectively. For the transmit array, a length of 370 mm was found to provide the best tuning for the 7T resonance frequency (297.2 MHz in this case), which is shorter than the half wavelength in vacuum (~505 mm). This shortening is attributed to the interaction of the dipole with the high dielectric constant of the body (or the phantom with similar properties). A minimum separation of 6 cm between the dipoles was required to prevent significant inter-element coupling. In this setup, the reflection coefficient S_{11} for each dipole was < -30 dB, the coupling coefficient S_{21} between the elements was ~ -12 dB, and the active detuning circuit typically provided better than 20 dB of detuning using an S_{21} measurement with a pick-up probe. The S-parameters for the loop-based arrays (receive-only and T/R arrays) were typically, $S_{11} = -35$ dB for matching, $S_{21} = -24$ dB between elements (measured without the present of other electronics), and the detuning circuit typically provided 33 dB of detuning. The unloaded/loaded Q-ratio for each channel was around 7.3. Finally, the preamplifiers typically provided about 17 dB of preamp decoupling on each receive-only element.

Optimal Positioning and Phasing of the Dipole Array

With the dipole elements connected in T/R configuration, the transmit efficiency of the dipole pair as a function of inter-element phase delay was measured using B_1^+ mapping. The maps of transmit efficiency (in $\mu\text{T/V}$) are shown in Figure 3a, with regions of low MRI signal masked out. The capability of “steering” the transmit field by incrementing the phase shift between the elements is clearly demonstrated. Figure 3b shows the dependence of transmit efficiency in the target ROI (at the expected depth of the spine) on the phase delay and on the ROI position relative to the center of the array (i.e. spatial offset in L-R direction). Given that the width of the phantom was much larger than the separation of dipole pair (425 mm v.s. 60 mm), the effect of the spatial offset could be simulated by moving the ROI in the data analysis, as opposed to repeating the B_1^+ mapping while physically shifting the phantom over the array. The optimum phase delay and spatial offset were found to be 142° and 9 mm, respectively. However, it was found that performance

remained within 97% of this optimum value (Figure 3c) if the dipole pair were simply driven in quadrature (90°) and no L-R offset was used. Based on this, *in vivo* experiments were performed with quadrature drive and with center line of the array in S-I direction approximately aligned with the spinal cord. To further simplify the setup, the centerlines of the receive array (in L-R and S-I) were aligned with those of the transmit array. While this may not be optimal for receiving in the presence of twisting effects in the receive B_I profile (4), the deterioration in receive sensitivity was expected to be minimal.

Performance Comparison between Dipole and Loop Array

Estimated calibration voltages for the RF amplitude were 347 V for the dipole array versus 548 V for the T/R loop array, corresponding to transmit efficiencies as of 0.0338 $\mu\text{T/V}$ and 0.0214 $\mu\text{T/V}$ respectively. The latter is comparable with that reported in (4). Axial B_I^+ efficiency maps (in $\mu\text{T/V}$) calculated from B_I^+ mapping experiments for the two arrays are summarized in Figure 4. It is clear that the dipole-based transmit array has much higher transmit efficiency along the longitudinal centerline (S-I) as compared with the loop-based array. We attribute this to the difference in B_I^+ field distributions between the two arrays, given that the average B_I^+ efficiencies over the entire phantom were nearly identical (0.026 $\mu\text{T/V}$ for both arrays).

Maps of temperature change after normalization for input power for the two arrays are shown in Figure 5. The dipole transmitter (Figure 5a) showed ~10% higher maximum RF induced heating as compared to the loop transmitter (Figure 5b) for the 10-minute heating test. Heating for the dipole array is more concentrated on the left side, toward the dipole with 90° phase delay, whereas the heating generated by the loop transmitters is more evenly distributed.

These results suggest that, for a given amount of input power, the dipole array provides a higher B_I^+ in the region of the spine than the loop-based array, but it also generates higher peak local SAR. Therefore, to properly compare transmit efficiency under SAR-limited conditions, we normalized the B_I^+ by the square-root of associated maximum local heating for a 10 minute session (from now on referred to as the heating-corrected transmit efficiency, in $\mu\text{T}/\sqrt{K}$ as temperature change in Kelvin). The square root is needed for the temperature change because B_I^+ is proportional to the input RF envelope voltage, whereas the local temperature change is proportional to the power (the square of voltage). Figure 6 shows line profiles through the center of the spinal ROI in three orthogonal directions: L-R, A-P, and S-I. These results confirm that the gain in B_I^+ efficiency by the dipole array is higher than the associated slightly increased peak RF heating. In fact, the gain in these three profiles within 10 cm around the center of the spinal ROI ($x'=0, y'=0, z'=0$) (highlighted in grey in Figure 6) can be as high as 76%.

Figure 7a shows an *in vivo* B_I^+ efficiency map overlaid on an anatomical image. For comparison, the B_I^+ efficiency map acquired from the same location on the phantom (center slice in Figure 4a) is shown in Figure 7b, with the ROI corresponding to the cord region marked with white square (about 4cm deep). The average B_I^+ efficiency within the ROI was 0.0405 $\mu\text{T/V}$ for the *in vivo* data and 0.0399 $\mu\text{T/V}$ for the phantom. Examples of *in vivo*

images obtained with the dipole array are shown in Figure 7c-e and demonstrate the ability to obtain excellent sensitivity in the region of the spine at high resolution.

DISCUSSION

In this study, the feasibility of using electric dipole antennae as transmit elements in 7T MR spinal imaging was explored with careful positioning of the dipole elements. Phantom experiments showed that significant gains in transmit efficiency can be achieved without significantly increasing tissue heating. Given the excellent match between the length of the “half-wave dipole” and the expected FOV for spine imaging at 7T, use of electric dipole elements can simplify transmit array design for human spinal applications by reducing total number of channels required to cover a large area (in this example, two dipoles instead of eight loops). Minimizing the number of cables that are used can further reduce potential cable coupling and minimize the parts count for supporting components, such as RF traps and power splitters. One potential drawback with this approach is that the large size of the dipoles used here limits the maximum number of the dipoles that can be used simultaneously without detrimental inter-element coupling. For spine imaging at 7 T, 6 cm separation was found to be necessary to prevent excessive inter-element coupling. Using more half-wave dipoles in parallel with similar separations in L-R direction would place them too far away from the spine ROI. Other dipole configurations would be required to fully take advantage of multi-channel transmission, which usually comes with eight parallel transmit channels. For clinical MRI of the spine, it may be necessary to shorten dipoles (14,18), or combine them with loop transmitters.

The gain in the transmit efficiency at the spinal cord region for the dipole seems mostly due to a different distribution of the B_1^+ field in comparison with the loop-based design, given the average B_1^+ efficiency was almost identical. This is probably due to the removal of the return path that is always 180-degrees out-of-phase, i.e. less field cancellation for positions far from the coil. However, the actual mechanism needs further investigation.

In this work, array configurations were optimized experimentally with the help of a modified Bloch-Siegert B_1^+ mapping method (25). Because this is a laborious approach, only a relatively small portion of the parameter space was explored. Further optimization of the dipole transmitters may be possible by using RF simulations, which readily allow coverage of a larger parameter space. In addition, RF simulations would facilitate the optimization of the dipole configuration using heating-corrected transmit efficiency (or equivalent) rather than RF amplitude normalized transmit efficiency, as was done here, and this may result in further improvement in performance.

CONCLUSION

A coil that employed an efficient transmit array based on electric dipoles and designed for use with a loop-based receive array was constructed for 7T spine imaging. Steering of the B_1^+ field in the axial plane was accomplished by varying the phase delay between the dipoles, which allowed RF shimming and transmit efficiency optimization for a pre-selected ROI. The transmit efficiency of the dipole array was compared with a previously described

design based on quadrature-loop pairs. The dipole antenna provides a much simpler and more efficient transmitter design, and it can be readily combined with standard receive arrays.

Acknowledgements

This research was supported by the Intramural Research Program of the National Institute of Neurological Disorders and Stroke.

REFERENCES

1. Shapiro MD. MR Imaging of the Spine at 3T. *Magnetic Resonance Imaging Clinics of North America*. 2006; 14(1):97–108. [PubMed: 16530638]
2. 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(11):734–740. [PubMed: 19809342]
3. Wu B, Wang C, Krug R, Kelley DA, Xu D, Pang Y, Banerjee S, Vigneron DB, Nelson SJ, Majumdar S, Zhang X. 7T Human Spine Imaging Arrays With Adjustable Inductive Decoupling. *Biomedical Engineering, IEEE Transactions on*. 2010; 57(2):397–403.
4. Duan, Q.; Sodickson, DK.; Lattanzi, R.; Zhang, B.; Wiggins, G. Optimizing 7 T spine array design through offsetting of transmit and receive elements and quadrature excitation. *International Society for Magnetic Resonance in Medicine; 18th Annual Meeting & Exhibition of ISMRM; Stockholm, Sweden*. 2010. p. 324
5. 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(2):291–297. [PubMed: 21134773]
6. 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(7):891–899. [PubMed: 22183956]
7. Zhao W, Cohen-Adad J, Polimeni JR, Keil B, Guerin B, Setsompop K, Serano P, Mareyam A, Hoeft P, Wald LL. Nineteen-channel receive array and four-channel transmit array coil for cervical spinal cord imaging at 7T. *Magn Reson Med*. 2014; 72(1):291–300. [PubMed: 23963998]
8. Brunner DO, De Zanche N, Frohlich J, Paska J, Pruessmann KP. Travelling-wave nuclear magnetic resonance. *Nature*. 2009; 457(7232):994–998. [PubMed: 19225521]
9. Andreychenko, A.; Voogt, I.; Kroeze, H.; Klomp, D.; Lagendijk, J.; Luijten, P.; van den Berg, C. An Advantageous Combination of Travelling Wave and Local Receive for Spine MR Imaging at 7T: Local SAR Reduction and SENSE Reconstruction; 18th Annual Meeting of ISMRM; Stockholm, Sweden. 2010. p. 430
10. Zhang B, Sodickson DK, Lattanzi R, Duan Q, Stoeckel B, Wiggins GC. Whole body traveling wave magnetic resonance imaging at high field strength: Homogeneity, efficiency, and energy deposition as compared with traditional excitation mechanisms. *Magnetic Resonance in Medicine*. 2012; 67(4):1183–1193. [PubMed: 21842501]
11. Raaijmakers AJ, Ipek O, Klomp DW, Possanzini C, Harvey PR, Lagendijk JJ, van den Berg CA. Design of a radiative surface coil array element at 7 T: the single-side adapted dipole antenna. *Magn Reson Med*. 2011; 66(5):1488–1497. [PubMed: 21630342]
12. Ipek O, Raaijmakers AJE, Klomp DWJ, Lagendijk JJW, Luijten PR, Berg CATvd. Characterization of transceive surface element designs for 7 tesla magnetic resonance imaging of the prostate: radiative antenna and microstrip. *PHYSICS IN MEDICINE AND BIOLOGY*. 2012; 57(2):343. [PubMed: 22170777]
13. Ipek O, Raaijmakers A, Lagendijk J, Luijten P, van den Berg C. Optimization of the radiative antenna for 7-T magnetic resonance body imaging. *Concepts in Magnetic Resonance Part B: Magnetic Resonance Engineering*. 2013; 43B(1):1–10.

14. Raaijmakers AJE, Italiaander M, Voogt IJ, Luijten PR, Hoogduin JM, Klomp DWJ, van den Berg CAT. The fractionated dipole antenna: A new antenna for body imaging at 7 Tesla. *Magnetic Resonance in Medicine*. 2015 10.1002/mrm.25596.
15. Lakshmanan, K.; Cloos, M.; Lattanzi, R.; Sodickson, D.; Novikov, D.; Wiggins, G. The Circular Dipole. *International Society for Magnetic Resonance in Medicine; Joint Annual Meeting ISMRM-ESMRMB; Milan, Italy*. 2014. p. 315
16. Duan, Q.; Gudino, N.; Zwart, JAd; Gelderen, P; Murphy-Boesch, J.; Duyn, JH.; Merkle, H. Array RF Transmitter for 7T MRI of the Spine Based on Dipole Antennas. *International Society for Magnetic Resonance in Medicine; Joint Annual Meeting ISMRM-ESMRMB; Milan, Italy*. 2014. p. 316
17. Chen, G.; Cloos, M.; Lattanzi, R.; Sodickson, DK.; Wiggins, G. Bent Electric Dipoles: A Novel Coil Design Inspired by the Ideal Current Pattern for Central SNR at 7 Tesla. *International Society for Magnetic Resonance in Medicine; Joint Annual Meeting ISMRM-ESMRMB; Milan, Italy*. 2014. p. 402
18. Chen, G.; Cloos, M.; Sodickson, DK.; Wiggins, G. A 7T 8 Channel Transmit-Receive Dipole Array for Head Imaging: Dipole Element and Coil Evaluation. *International Society for Magnetic Resonance in Medicine; Joint Annual Meeting ISMRM-ESMRMB; Milan, Italy*. 2014. p. 621
19. Silver, S. The "Half-wave Dipole". In: Silver, S., editor. *Microwave Antenna Theory and Design*. McGraw-Hill Book Company, Inc.; New York: 1949. p. 98-99.
20. Eryaman Y, Guerin B, Keil B, Mareyam A, Herraiz JL, Kosior RK, Martin A, Torrado-Carvajal A, Malpica N, Hernandez-Tamames JA, Schiavi E, Adalsteinsson E, Wald LL. SAR reduction in 7T C-spine imaging using a "dark modes" transmit array strategy. *Magn Reson Med*. 2014 DOI: 10.1002/mrm.25246.
21. Lorenz C. Schaltungsanordnung zum Übergang von einer symmetrischen, elektrischen Anordnung zu einer unsymmetrischen, insbesondere bei Hochfrequenzanwendungen. Germany patent. 1932 603816.
22. Mispelter, J.; Lupu, M.; Briguet, A. Quadrature Driving. In: Wei, TK., editor. *NMR Probeheads for Biophysical and Biomedical Experiments*. Mainland Press; Singapore: 2006. p. 131-186.
23. Gabriel S, Lau RW, Gabriel C. The dielectric properties of biological tissues: II. Measurements in the frequency range 10 Hz to 20 GHz. *Phys Med Biol*. 1996; 41(11):2251–2269. [PubMed: 8938025]
24. Duan Q, Duyn JH, Gudino N, de Zwart JA, van Gelderen P, Sodickson DK, Brown R. Characterization of a dielectric phantom for high-field magnetic resonance imaging applications. *Medical Physics*. 2014; 41(10):102303. [PubMed: 25281973]
25. Duan Q, van Gelderen P, Duyn J. Improved Bloch-Siegert based B(1) mapping by reducing off-resonance shift. *NMR Biomed*. 2013; 26(9):1070–1078. [PubMed: 23355474]
26. Seeber DA, Jevtic J, Menon A. Floating shield current suppression trap. *Concepts in Magnetic Resonance Part B: Magnetic Resonance Engineering*. 2004; 21B(1):26–31.
27. de Zwart JA, Vimeux FC, Delalande C, Canioni P, Moonen CT. Fast lipid-suppressed MR temperature mapping with echo-shifted gradient-echo imaging and spectral-spatial excitation. *Magn Reson Med*. 1999; 42(1):53–59. [PubMed: 10398950]
28. Deniz, CM.; Brown, R.; de Zwart, JA.; Collins, CM.; Sodickson, DK. ISMRM scientific workshop "Safety in MRI: Guidelines, Rationale & Challenges". *International Society for Magnetic Resonance in Medicine; Washington, DC, USA*: 2014. Multi-Channel Array Safety Using Least Squares Fitting based MR Thermometry.
29. Ishihara Y, Calderon A, Watanabe H, Okamoto K, Suzuki Y, Kuroda K, Suzuki Y. A precise and fast temperature mapping using water proton chemical shift. *Magnetic Resonance in Medicine*. 1995; 34(6):814–823. [PubMed: 8598808]

Figure 1.

(a) Schematic of the electric dipole array for transmission, (b) circuit diagram for bias T for the PIN control signal, lattice balun for matching, and the detuning circuits, and (c) illustration of the layout of transmit and receive arrays v.s. the loading phantom, and associated coordinates. On the right, the scanner system (x, y, z) is marked in black. Another coordinate system parallel with (x, y, z) system was defined as (x', y', z') (shown in purple) with its origin residing on the center of the ROI at the L-R and S-I center of the phantom, and 5 cm from the bottom on the A-P centerline of the phantom. The (x', y', z') coordinate system, referred as the phantom coordinate system, was always centered at the center of the ROI at the axial center plane of the phantom and was moving with the whole setup, i.e. the phantom and the combined T/R array, whereas the scanner coordinate system (x, y, z) system stayed with the scanner, with its origin always at the scanner iso-center.

Figure 2.

Photos of interior of prototype (a) dipole transmit array (with dipole fine-tuned by copper tape (top) or copper wire (bottom)), (b) loop-based receive array, and (c) an existing loop-based transmit/receive design (4) that was used as a reference, with labeling of major functional component types.

Figure 3.

Optimization of the transmit efficiency: (a) Axial B_1^+ efficiency map (unit $\mu\text{T/V}$) at the center of the dipole array as a function of inter-channel phase delay. White circles represent the ROI targeting the approximate location of spinal cord; (b) average B_1^+ efficiency within the ROI as a function of phase delay and spatial offset; (c) measured (red dot) and fitted B_1^+ efficiency (blue line) with 6th order polynomial for quadrature drive, assuming no additional spatial offset (i.e. spatial offset = 0).

Figure 4.

Axial B_1^+ efficiency maps ($\mu\text{T/V}$) acquired on a phantom in a multi-station fashion, covering 50 cm FOV in the longitudinal (S-I) direction for (a) the detunable dipole-based and (b) the loop-based spine arrays. Imaging parameters for the B_1^+ mapping were: FOV $200 \times 400 \text{ mm}^2$, image matrix 64×128 , yielding $3.1 \times 3.1 \text{ mm}^2$ resolution, slice thickness 5 mm, slice spacing 5 mm, repetition time (TR) = 800 ms, echo time (TE) = [13.30, 14.63] ms. The white scale bars in the image represent 5 cm. White circles represent the approximate locations of spinal cord in a human subject.

Figure 5.

Coronal normalized temperature change maps (K/W, i.e. temperature change after 10 minute RF heating normalized by the input power) calculated based on MR thermometry data for (a) dipole-based and (b) loop-based spine arrays. The image resolution was $4 \times 4 \times 3 \text{ mm}^3$, $\text{FOV} = 480 \times 480 \text{ mm}^2$, with 13 coronal slices and 1 mm slice spacing.

Figure 6.

Heating-corrected transmit efficiency profiles (B_I^+ divided by the square-root of associated maximum local heating in 10 minutes, $\mu T / \sqrt{K}$) through the center of the spine ROI in (a) L-R, (b) A-P, and (c) S-I directions. Shaded regions represent the approximate locations of spinal cord.

Figure 7.

(a) An axial B_1^+ efficiency map (FOV $112.5 \times 300 \text{ mm}^2$, image matrix 48×128 , yielding $2.3 \times 2.3 \text{ mm}^2$ resolution, slice thickness 5 mm, TR = 150 ms, TE = [13.28, 14.51] ms) overlaid on a corresponding anatomical slice (TR = 54 ms, TE = 6.9 ms, image matrix 96×256 , yielding $1.17 \times 1.17 \text{ mm}^2$ resolution) acquired on a healthy volunteer, compared with (b) the one acquired on the phantom (Figure 4a) at the corresponding location. The white square on the *in vivo* data in (a) represented the cord ROI, which is about 4cm deep. A match ROI (white square) was then draw on the phantom data in (b). Example anatomical spine images acquired in a healthy volunteer using the proposed array at 7T: (c) 2D sagittal GRE image (FOV $234 \times 300 \text{ mm}^2$, image matrix 300×384 , TR = 20 ms, TE = 3.6 ms, flip angle = 20° , slice thickness 3 mm). (d) 2D axial GRE image (FOV $156 \times 200 \text{ mm}^2$, image matrix 300×384 , yielding $0.52 \times 0.52 \text{ mm}^2$ resolution, TR = 54 ms, TE = 6.9 ms, flip angle = 20° , slice thickness 3 mm). (e) High resolution GRE cord image acquired on a healthy volunteer (FOV $160 \times 320 \text{ mm}^2$, image matrix 512×1024 , yielding $0.31 \times 0.31 \text{ mm}^2$ resolution, TR = 33 ms, TE = 9.3 ms, flip angle = 35° , slice thickness 3 mm, and 40 repetitions) after navigator-based artifact correction. The main image was scaled from 0 to 100 a.u. The bottom-left insert is a crop-out magnitude image of the cord region after receiver profile correction then rescaled to [35 65] a.u. for better contrast within the cord.

The bottom-right insert is the corresponding phase map (scaled as $[-5\ 5]\text{Hz}$), showing excellent gray/white matter contrast.