Published in final edited form as:

Magn Reson Med. 2015 April ; 73(4): 1533–1539. doi:10.1002/mrm.25246.

SAR Reduction in 7T C-Spine Imaging Using a “Dark Modes” Transmit Array Strategy

Yigitcan Eryaman^{1,2,3,*}, Bastien Guerin², Boris Keil^{2,10}, Azma Mareyam², Joaquin L. Herraiz^{1,3}, Robert K. Kosior^{3,9}, Adrian Martin^{3,8}, Angel Torrado-Carvajal^{3,7}, Norberto Malpica^{3,7}, Juan A. Hernandez-Tamames^{3,7}, Emanuele Schiavi^{3,8}, Elfar Adalsteinsson^{3,4,5,6}, and Lawrence L. Wald^{2,5}

¹Research Laboratory of Electronics, Massachusetts Institute of Technology, Cambridge, Massachusetts, USA.

²A. A. Martinos Center for Biomedical Imaging, Department of Radiology, Massachusetts General Hospital, Charlestown, Massachusetts, USA.

³Madrid-MIT M+ Vision Consortium, Madrid, Spain.

⁴Department of Electrical Engineering and Computer Science, Massachusetts Institute of Technology, Cambridge, Massachusetts, USA.

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

⁶Institute of Medical Engineering and Science, Massachusetts Institute of Technology, Cambridge, Massachusetts, USA.

⁷Department of Electronic Technology, Rey Juan Carlos University, Móstoles, Madrid, Spain.

⁸Department of Applied Mathematics, Rey Juan Carlos University, Móstoles, Madrid, Spain.

⁹Faculty of Medicine, University of Calgary, Calgary, Canada.

¹⁰Harvard Medical School, Boston, Massachusetts, USA.

Abstract

Purpose—Local specific absorption rate (SAR) limits many applications of parallel transmit (pTx) in ultra high-field imaging. In this Note, we introduce the use of an array element, which is intentionally inefficient at generating spin excitation (a “dark mode”) to attempt a partial cancellation of the electric field from those elements that do generate excitation. We show that adding dipole elements oriented orthogonal to their conventional orientation to a linear array of conventional loop elements can lower the local SAR hotspot in a C-spine array at 7 T.

Methods—We model electromagnetic fields in a head/torso model to calculate SAR and excitation B_1^+ patterns generated by conventional loop arrays and loop arrays with added electric dipole elements. We utilize the dark modes that are generated by the intentional and inefficient

*Correspondence to: Yigitcan Eryaman, Ph.D., Research Laboratory of Electronics, Massachusetts Institute of Technology, Cambridge, MA 02139. yigitcan@mit.edu.

Additional Supporting Information may be found in the online version of this article.

orientation of dipole elements in order to reduce peak 10g local SAR while maintaining excitation fidelity.

Results—For B_1^+ shimming in the spine, the addition of dipole elements did not significantly alter the B_1^+ spatial pattern but reduced local SAR by 36%.

Conclusion—The dipole elements provide a sufficiently complimentary B_1^+ and electric field pattern to the loop array that can be exploited by the radiofrequency shimming algorithm to reduce local SAR.

Keywords

parallel transmit; local SAR; global SAR; radiative dipole; loop–dipole arrays; excitation fidelity

INTRODUCTION

Parallel transmit (pTx) arrays are currently in investigational use to alleviate radio frequency (RF) field inhomogeneity problems in high-field MRI (1–5) and offer the potential to optimize specific absorption rate (SAR) as part of the pulse design process (6–9). The geometry of the individual coils is an important design parameter that can be optimized in order to improve MRI excitation. The coil elements used in the transmit array can have diverse geometries, including microstrip coils (1,5), conventional loop coils (2,3,6–8), and radiative dipole antenna (10). The latter was recently introduced as a transmit element for high-field transmit arrays (10) but was shown to have a higher peak local SAR at 7T compared to a loop element producing identical B_1^+ in the middle of a uniform phantom. In a comparison of these three types of transmit elements, the loop antenna had the smallest local SAR. Nonetheless, the dipole's B_1^+ field penetrated longer distances in a phantom, likely due to the lack of cancellation of currents compared to micro-strip coils or conventional loop elements (10).

The B_1^+ field pattern generated by any transmit element depends on the orientation of the element with respect to B_0 . For the dipole array, previous uses have sensibly oriented it to produce the maximum B_1^+ field at the center of the body. In this study, we use dipole elements with their conductive element aligned orthogonal to the direction of the static magnetic field (z -axis), as shown in Figure 1. In this arrangement, the dipoles produce RF magnetic fields mostly in the z -direction—and with little B_x or B_y component that can give rise to the desired B_1^+ component. Thus, the B_1^+ component generated by these elements is small compared to that generated by a loop or a conventionally oriented dipole. This property of the dipole elements enables us to energize these elements without generating spin excitation. This situation is similar to that previously shown for the “dark modes” of the degenerately tuned birdcage coil—those modes that generate the wrong circular polarization for spin excitation (11). We have hypothesized that, although not useful for spin excitation, these dark modes or dark elements are potentially useful for generating electric (E -) fields that partially cancel the E -fields associated with spin excitation resulting from conventional elements. In the case of the combined loop and dipole array described here, the loop elements are expected to generate spin excitation (along with unwanted E -fields), and the

dipoles are used to attempt to partially cancel these E-fields and thus reduce local SAR. We validated this concept in a simulation study of a C-spine excitation array at 7 T.

METHODS

We performed electromagnetic (EM) simulations of an array with dimensions similar to a previously constructed four-channel loop array (12) and compared it to an array of identical loops with “dark” dipole elements added, as shown in Figure 1. The width and height of the loop element were chosen as 69 mm and 155 mm, respectively. Four capacitors were distributed around the loops for matching and tuning. The length of the dipoles were chosen as 118 mm. All transmit element models were constructed by using a 10-mm wide copper stripline. The lower part of the array was located approximately 40 mm away from the lower body; and the upper part of the array was located approximately 80 mm away from the head. Simulations were performed with SEMCAD X EM Solver (Speag, Zurich, Switzerland) using the virtual family model “Duke” (IT’IS Foundation, Zurich, Switzerland). The calculation was done on a grid size chosen by the software that varied between 1 mm and 4.5 mm; however, the larger grid size was mainly in areas of free space. After the fields were calculated on the nonuniform grid, we interpolated the results onto a uniform 3-mm grid. Uniaxial perfectly matched layer (UPML) boundaries were placed 500 mm from the field sensor area (400 mm × 220 mm × 340 mm), which was the area over which the fields are calculated (Fig. 1), leading to a total simulation volume of 1400 mm × 1220 mm × 1340 mm. UPML boundaries were placed at a sufficient distance to the RF coil and body model. Loops and dipole antennas were matched and tuned to the Larmor frequency for 7 T (298 MHz). The reflection coefficient seen from the excitation ports of all elements were adjusted to a value less than −17 dB. First, the structure is modeled in SEMCAD (e.g., the loop of the loop coil with no capacitors). Next, using the simulated impedance of this structure—as determined by SEMCAD at the Larmor frequency—we calculated the needed capacitor values to match and tune the loop to 50 Ohms at 297.2 MHz using MATLAB (Mathworks, Inc., Natick, MA). Then we placed these values into the SEMCAD model and repeated the procedure until it converged (after about 2 or 3 iterations). Starting from the drive port and working counterclockwise, the calculated C values (in pF) are as follows: loop 1 = 35, 3.52, 3.52, 3.52; loop 2 = 35, 3.52, 3.52, 3.52; loop 3 = 34, 3.51, 3.51, 3.51; and loop 4 = 30, 3.57, 3.57, 3.57. Similarly for the four dipoles (circuit diagram shown in Fig. 1), simulated L and C were: dipole 1 = 144.4 nH, 34.0 pF; dipole 2 = 144.6 nH, 35.5 pF; dipole 3 = 143.9 nH, 35.1 pF; and dipole 4 = 143.3 nH, 32.5 pF. The copper traces were modeled as perfect conductors. No additional dielectric material was placed between the RF coil and the body. All elements were assumed perfectly decoupled in order to simplify EM simulations. In practice, we previously showed that the loop array could be constructed with a maximum S12 of −14 dB (12).

We constructed a loop and two dipoles and performed S parameter measurements. The loop and the dipoles were matched and tuned at 297.2 MHz. The coupling between the loop and the dipoles at different locations (S12) were measured using a network analyzer. We also acquired B1⁺ maps of the loop and the dipoles (in dark mode and regular operation mode) individually using a uniform phantom, as shown in Supporting Figures S1 and S2. B1⁺ maps were obtained using a uniform phantom with a 3.3 g/L NiCl₂·6H₂O, 2.4 g/L NaCl solution.

The dimensions of the phantom (rectangular box) were $150 \times 150 \times 380$ mm. The conductivity and relative permittivity of the solution were measured as 0.7 S/m and 80, respectively. A standard gradient recalled echo sequence (pulse repetition time = 600 msec) was used to acquire the B_1^+ maps with a resolution of $3.1 \times 3.1 \times 5$ mm. MRI images are obtained using excitation voltages varying from 0V to 200V for each element. The measured image intensities were fitted to sinusoids to calculate the B_1^+ variation of each transmit element in the imaging plane. A Tx/Rx switch was used to receive the MRI signal with the same coil. The EM simulations were also performed to calculate the B_1^+ maps in the same uniform phantom.

Using loop elements alone, we calculated simulated B_1^+ patterns from two different sets of excitation currents. First, we used a simple linear phase increment to the loops (0° , 30° , 60° , 90°) with constant magnitude. Secondly, we used particle swarm optimization (PSO) (13) to optimize the amplitude and phase of each port's excitation (RF shimming) to minimize the magnitude least square (MLS) flip-angle error over the spinal region of interest (ROI). The ROI was drawn manually in only one image plane (thus, it has only 1 pixel depth in the patient L–R direction).

The local SAR distribution of these two methods is calculated for a mean B_1^+ field of $1 \mu\text{T}$ at the spinal cord ROI. A home-written region growing algorithm was used. For each voxel in the body, a neighborhood of voxels that approximates 10g tissue is chosen. Inside the body, growth in this region is roughly isotropic. Near the edge of the body, it stops growing at the surface and only grows inwardly. The 10g SAR is then averaged over this region. After establishing the optimum excitation and resulting SAR pattern from the loop elements alone, we energized the dipole elements to ascertain their effect. The dipole elements are expected to generate small transverse B_1^+ field at the location of the spinal cord due to their orientation. However, the electric fields generated by the dipoles are comparable in magnitude to the loop elements. We optimized the currents on the dipole elements to reduce the overall peak local SAR using PSO. For the optimization, SAR matrices were compressed to a smaller set of 141 virtual observation points (14) to enable fast evaluation of peak 10g local SAR. The optimization was constrained to maximally cancel the local SAR while achieving the same B_1^+ to within 3% at each point in the spinal cord ROI as the loops alone. Supporting Table S1 shows the magnitudes (root means square voltage [Vrms]) and the phases of the voltages on all array elements. In addition to the comparison described above, we calculated optimum RF shimming solutions using an approach that minimizes the difference between the magnitude of the simulated and desired flip-angle profile while explicitly constraining both global and local SAR (7). This was performed for both the loop array and the loop + dipole array. The target pattern was a uniform 25° flip angle excitation on the spinal ROI. The algorithm was free to adjust the phase and amplitude of all elements within the power and global SAR constraints. Each pulse consisted of a sinc profile (3 lobes; 0.8 ms second duration; duty cycle of 10%). Calculating the pulse several times with different local SAR constraints results in an L-curve that shows the trade-off between local SAR and the flip-angle target fidelity [% root-mean-square error (RMSE)] for the magnitude image. The global SAR, maximum peak RF power, and average RF power were explicitly constrained (to 3.2 W/kg, 5 kW and 500 W, respectively).

RESULTS

Figure 2 shows the simulated B_1^+ map of the individual loop and dipole elements. The spinal cord B_1^+ of the loops were significantly larger than the B_1^+ of the dipoles reflecting the choice of orientation of the dipoles, which are arranged to generate magnetic fields mainly in the z -direction (Fig. 2; bottom row). Because of their weak contribution to B_1^+ , the dipole elements can be excited without affecting the B_1^+ of the loop array. Figure 3 shows the simulation results for the B_1^+ distribution of two separate excitation patterns: 1) linear-phased current pattern with loops carrying currents with constant magnitude but with linear phase changing in the z -direction. 2) Currents optimized with PSO to minimize MLS with RF shimming. Figure 3 (3a, 3d) shows B_1^+ due to four loops only. Figure 3 (3b, 3e) shows B_1^+ due to the loop–dipole array (4 loops, plus 4 dipole elements), whereas the dipoles produce the “dark modes,” as explained above. Figure 3 (3c, 3f) shows the B_1^+ on the spinal ROI. As can be seen from both the sagittal B_1^+ maps and the B_1^+ plots, the excitation of the dark modes with the dipole elements did not change the B_1^+ pattern of the loops significantly.

Although the dipoles have minimal effect on the B_1^+ pattern, the peak 10g local SAR was reduced when the dipoles were energized. SAR reduction is demonstrated for both the linear phased current pattern (Fig. 4) and the uniform B_1^+ pattern obtained with RF shimming (Fig. 5). Figure 4 and Figure 5 show the 10g local SAR distribution of two arrays in three planes chosen so that the axial plane always contains the peak local SAR hotspot. Energizing the dipole elements reduced the peak local SAR by 16% and 36%, respectively, for the linear phase and full RF shimming cases.

The global SAR values for the loop-only and loop + dipole array were 1.21 W/kg and 1.29 W/kg for the linear phase excitation. For the MLS RF shimming, the global SAR were calculated as 1.40 W/kg and 1.44 W/kg for the loop-only and loop + dipole array, respectively.

Figure 6 shows the local SAR tradeoff of the loop array and the loop + dipole array as L curves. The addition of the dipoles-reduced local SAR for all of the excitations studied. SAR reduction was 28% for the approximate optimal operating point on the L-curve (an RMSE excitation fidelity of 5% over the spinal cord ROI).

One loop and two dipole elements were constructed with the same dimensions discussed in the methods section. Supporting Figure S1 shows the matching-tuning (S11) and the decoupling (S12) behavior of the loop and the dipoles at different locations. Data in row 1 of the table in Supporting Figure S1 was obtained using a single loop and a single dipole. The position of the dipole was changed whereas the position of the loop was kept constant. When the dipole was located symmetrically in the middle of the loop, decoupling value (S12) of -22.6 dB was obtained. It is also shown that the loop couples strongly with the neighboring dipole ($S12 = -5.4$ dB.) The coupling between the loop and the next nearest dipole and the farthest dipole was weaker (-14.1 dB and -18.2 dB, respectively). Row 2 of the table in Supporting Figure S1 shows the coupling between the two dipole elements: in the nearest-neighbor positions and in the next two most distant positions. Supporting Figure

S2 shows the simulated and the measured B_1^+ of the loop and the dipole in the two orthogonal orientations of the dipole. In general, we have a good qualitative agreement between the simulated and measured B_1^+ for both elements. Note that the transmit efficiency (B_1^+) of the dipole is reduced in the dark element orientation when compared to the regular orientation.

DISCUSSION

For spine imaging, the local SAR reduction is achieved by energizing “dark” elements—the dipoles which do not change significant B_1^+ profiles, but which do provide additional degrees of freedom to cancel E-fields (and thus reduce SAR). For RF shimming excitation, the maximal local SAR hot spot occurred in the inferior C-spine where the loops are closest to the body. For the linear phased current excitation, the maximal local SAR hot spot occurred in the neck. In both cases, adjuvant excitation with dipoles enabled overall peak SAR reduction. The position of the hot spot did not change for these examples. However, the addition of the dipoles increased the local SAR hot spot at the neck by 3.1% for the RF shimming excitation pattern. Global SAR did not increase significantly when the dipoles are energized. The peak 10g local SAR remained as the limiting RF safety factor in these examples. Hard constraints can be added to the optimization problem in order control global SAR properly while reducing local SAR.

We refer to the dipoles as “dark” in the sense that they produced little effect on B_1^+ due to their orientation. However, they do not generate the ideal null B_1^+ fields in space. In order to ensure that no or little B_1^+ distortion was caused by the dipoles, the dipole currents were constrained in the optimization process. By constraining B_1^+ perturbation levels in the optimization, a prior B_1^+ pattern obtained by loops on the whole spinal ROI was preserved with no significant change due to the excitation of the dipoles. An alternative approach would be to simply include the dipole fields in a general pulse-optimization process that explicitly constrains local SAR, such as that demonstrated by Guérin et al. (7). Although inclusion of the dipole B_1^+ fields would have little effect on the B_1^+ distribution of the studied configuration, the extra computational time would likely be acceptable for an array with elements that were not truly “dark.”

The length of the dipoles used in this study was smaller than the quarter wavelength. In contrast to Raaij-makers et al. (10), the purpose of our study was not to maximize transmit efficiency. Our goal was to provide an E-field for cancelling local SAR without perturbing the excitation ($B_1^+ = 0$ ideally). Thus, although perhaps the shortened structure is less than optimal for transmit applications, it was successful for the purpose for which it was employed.

In this Note, an array of four loop and four dipole elements is compared to four loop elements in terms of excitation fidelity and peak local SAR tradeoff. Alternatively, the loop–dipole array could have been compared to an array of equal number of elements, for example, an eight-loop array. We chose not to pursue this comparison because the linear nature of the spine precludes simply adding more channels in this array geometry in a way that preserves the B_1^+ excitation efficiency in the C-spine cord. The two possibilities for an

eight-loop array are 1) preserving the loop size and extending the array well beyond the C-spine (thus, additional elements have little excitation efficiency at the C-spine) and 2) decreasing the loop size (thus reducing the $B1^+$ efficiency at the depth of the cord).

The loops are located in a coronal plane below the subject's back, as shown in Figure 1D. The dipole elements are in this same plane, centered on the spine, and are oriented so that the dipole's conductor points left-right in the patient's coordinate system, as shown in Figure 1c. The high SAR areas of a loop tends to be under and just beyond the copper strip, with a low SAR region under the center of the loop (i.e., directly under a dipole's conductors). Since we have four loops stacked, the high SAR regions are likely to be under neighboring loops/dipoles. The SAR pattern for a dipole is mainly under the dipole, providing a chance to cancel the field of neighboring loops. It should be noted, however, that it is difficult to generalize this description to a body model because the SAR is shaped by the geometry of the conductive regions.

All elements are assumed to be decoupled in order to simplify EM simulations. Measurements for coupling parameters can be found in Supporting Figures S1 and S2. In practice, the addition of dipole elements could affect the decoupling performance. Decoupling strategies that have been previously proposed in the literature (15) can be utilized to overcome these problems.

In order to apply the dark mode concept, the E-field maps of the dipoles must be known. For this we rely on simulations. We note that, for all regions, we will have the simulated B-field and E-field maps, and if we are proceeding with the calculation, then we trust that the simulated and measured values are in good agreement. However, $B1^+$ of the dark dipole is difficult to measure because it is small. Therefore, we could also use the simulated $B1^+$ maps in these problem areas. We would then proceed with a pulse calculation providing the best spatial excitation fidelity subject to regulatory SAR levels and the amplifiers peak and average power constraints.

We mapped the $B1^+$ of the loop and the dipole in a uniform phantom using a 7 T MR scanner and validated our EM simulations. SAR reduction using dark elements was demonstrated with EM simulations only. For an in-vivo implementation of the loop-dipole array, additional safety validation experiments should be performed. For this purpose, E-field/SAR measurements can be made locally in phantoms by using E-field probes (16) or temperature measurement techniques.

The dark elements actively reduce the local SAR without significantly changing the flip angle distribution. Thus, transmit failure of any of these elements may lead to elevated SAR levels in the patient. In practice, such a failure must be detected by circuitry that either monitors the reflected power (thus, detected an alteration in the dipole tuning or matching) or detects the RF level with a loop coil immediately under the dipole coil. Many pTx systems incorporate automatic shutdown using these sorts of detectors (17). Automatic tuning and matching networks can be used to both properly set the tune/match and actively detect and correct any loss in the transmit performance (18). In addition, techniques that

broaden the bandwidth of the dipoles can also be used to reduce to RF-loading sensitivities (19).

CONCLUSION

In this Note, we demonstrated an extension of the conventional array optimization process in which all of the elements are designed with transmit efficiency in mind. In the conventional design process, either SAR concerns are ignored or efficient spin excitation is equated with efficient $B1^+$ field generation. Although it is certainly true that electric fields must accompany oscillating magnetic fields, this generalization ignores the potential for cancellations of electric fields at discrete locations—and the fact that $B1^+$ is only one component out of the three ortho-normal components ($B1^+$, $B1^-$, $B1_z$) necessary to completely specify the magnetic field vector $B1$ in three-dimensional space. Our goal was to show that generalizing the design goals of transmit arrays to include local SAR and utilizing elements with a wider range of B-field components can be useful for lowering SAR. Although only shown only for a specific case, we hope the lessons learned can be broadly applied to a range of pTx array designs.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

Grant sponsor: National Institutes of Health NIBIB; grant numbers: R01EB006847; R01EB007942; Grant sponsor: Comunidad de Madrid, Madrid-MIT MIVision Consortium.

REFERENCES

1. Metzger GJ, Snyder C, Akgun C, Vaughan T, Ugurbil K, de Moortele V. Local $B1^+$ shimming for prostate imaging with transceiver arrays at 7T based on subject-dependent transmit phase measurements. *Magn Reson Med*. 2008; 59:396–409. [PubMed: 18228604]
2. Setsompop K, Alagappan V, Gagoski B, et al. Slice-selective RF pulses for in vivo $B1$ inhomogeneity mitigation at 7 Tesla using parallel RF excitation with a 16-element coil. *Magn Reson Med*. 2008; 60:1422–1432. [PubMed: 19025908]
3. Thalhammer C, Renz W, Winter L, et al. Two-Dimensional sixteen channel transmit/receive coil array for cardiac MRI at 7.0 T: design, evaluation, and application. *J Magn Reson Imaging*. 2012; 36:847–857. [PubMed: 22706727]
4. Gräßl A, Winter L, Thalhammer C, et al. Design, evaluation and application of an eight channel transmit/receive coil array for cardiac MRI at 7.0 T. *Eur J Radiol*. 2011; 82:752–759. [PubMed: 21920683]
5. Adriany G, Auerbach EJ, Snyder CJ, Gözübüyük A, Moeller S, Ritter J, Van de Moortele PF, 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]
6. Zhu Y. Parallel Excitation With an Array of Transmit Coils. *Magn Reson Med*. 2004; 51:775–784. [PubMed: 15065251]
7. Guérin B, Gebhardt M, Cauley S, Adalsteinsson E, Wald LL. Local specific absorption rate (SAR), global SAR, transmitter power, and excitation accuracy trade-offs in low flip-angle parallel transmit pulse design. *Magn Reson Med*. 2014; 71:1446–1457. [PubMed: 23776100]
8. Lee J, Gebhardt M, Wald LL, Adalsteinsson E. Local SAR in parallel transmission pulse design. *Magn Reson Med*. 2011; 67:1566–1578. [PubMed: 22083594]

9. van den Bergen B, van den Berg CAT, Klomp DW, Lagendijk JJ. SAR and power implications of different RF shimming strategies in the pelvis for 7T MRI. *J Magn Reson Imaging*. 2009; 30:194–202. [PubMed: 19557737]
10. Raaijmakers AJE, Ipek O, Klomp DWJ, Possanzini C, Harvey PR, Lagendijk JJW, van den Berg CAT. Design of a radiative surface coil array element at 7 T: The single-side adapted dipole antenna. *Magn Reson Med*. 2011; 66:1488–1497. [PubMed: 21630342]
11. Setsompop, K.; Wald, LL. SAR reduction through dark modes excitation. *Proceedings of the 19th Annual Meeting of ISMRM; Montreal, Canada*. 2011. p. 3890
12. Zhao W, Cohen-Adad J, Polimeni JR, Keil B, Guerin B, Setsompop K, Serano P, Mareyam A, Hoeht P, Wald LL. Nineteen-channel receive array and four-channel transmit array coil for cervical spinal cord imaging at 7T. *Magn Reson Med*. 2014; 72:291–300. [PubMed: 23963998]
13. Robinson J, Rahmat-Samii Y. Particle swarm optimization in electromagnetics. *IEEE Trans Ant Prop*. 2004; 52:397–407.
14. Eichfelder G, Gebhardt M. Local specific absorption rate control for parallel transmission by virtual observation points. *Magn Reson Med*. 2011; 66:1468–1476. [PubMed: 21604294]
15. Mahmood, Z.; Guerin, B.; Adalsteinsson, E.; Wald, LL.; Daniel, L. An automated framework to decouple pTx arrays with many channels. *Proceedings of 21st Annual Meeting of ISMRM; Salt Lake City, Utah, USA*. 2013. p. 2722
16. Seifert, F.; Klepsch, T.; Lindel, TD.; Hoffmann, W.; Ittermann, B. TEM cell for calibration of an electro-optic E-field sensor in a clinical scanner. *Proceedings of 19th Annual Meeting of ISMRM; Montreal, Canada*. 2011. p. 3777
17. Gagoski, B.; Bhat, H.; Hoeht, P., et al. Threshold criteria for real time RF monitoring in 7T parallel transmit system. *Proceedings of 19th Annual Meeting of ISMRM; Montreal, Canada*. 2011. p. 3848
18. Keith, G.; Rodgers, C.; Hess, A.; Snyder, C.; Vaughan, JT.; Robson, M. Computerised tuning of an 8-channel cardiac TEM array at 7T: an integrated system using piezoelectric actuators and power monitors. *Proceedings of 21st Annual Meeting of ISMRM; Salt Lake City, UT*. 2013. p. 2743
19. Winter L, Özerdem, Hoffmann W, Santoro D, Müller A, Waiczies H, Seemann R, Graessl A, Wust P, Niendorf T. Design and evaluation of a hybrid radiofrequency applicator for magnetic resonance imaging and RF induced hyperthermia: electromagnetic field simulations up to 14.0 Tesla and proof-of-concept at 7.0 Tesla. *PLoS One*. 2013; 8:e61661. [PubMed: 23613896]

FIG. 1.

B_1^+ field of the loop was larger than the z component of the loop's magnetic field. B_1^+ of the dipole in the above orientation was smaller than the z component of the dipole's field. Alignment of the dipoles in this orientation enables generating "dark modes" (a, b). Loop array and the loop-dipole array (c) were designed for spine imaging. Red and blue cones show the voltage sources and lump elements, respectively. Relative distance of the arrays to the body is shown from side view (d).

FIG. 2.

B_1^+ distribution due to 1 root means square voltage excitation of individual loop and dipole elements are shown. B_1^+ of loop elements were larger than B_1^+ of dipole elements (oriented as in Fig. 1) in the spine ROI. B_z distribution due to 1 V excitation of dipole elements is also shown. In comparison, dipoles (oriented as in Fig. 1) generate magnetic fields mostly in z -direction, which makes them dark elements for MR excitation. Rectangles show the location of the coil elements.

FIG. 3.

$B1^+$ distribution (sagittal plane) due to linear-phased current pattern (top row) and MLS RF shimming (bottom row) for loop array and loop-dipole arrays are shown. Supporting Table S1 shows the magnitudes (Vrms) and the phases of the voltages on all array elements.

Position of the spinal cord is marked with “white dots” in the map. **(a,d)** Four loops only.

(b,e) Loop–dipole array (4 loops plus 4 dipole elements). **(c,f)** The $B1^+$ on the spinal ROI.

The addition of the dipole elements does not change the $B1^+$ pattern of the loops significantly.

FIG. 4.

Local SAR (10 g) distribution in axial, coronal, and sagittal planes for the loop array and loop + dipole arrays. The four loops were excited with a linear-phased current pattern (as in Fig. 3 a, 3b, 3c) and an optimized MLS RF shimming pattern (as in Fig. 3d, 3e, 3f). Row 1 and row 3 show the SAR distribution due to loops alone. In the loop + dipole case, the dipole currents were chosen with PSO to minimize peak local SAR. Row 2 and row 4 show the SAR distribution due to loops + dipoles. The axial plane through the largest hot spot is shown. Adding the dipoles allowed the same $B1^+$ pattern to be achieved with 36% reduction in 10g SAR.

FIG. 5.

L curves showing the trade-off between excitation fidelity (% RMS error) and peak 10g local SAR for the four loops alone and the loop + dipole array. An axial cut through the peak SAR hotspot is shown for each array. The MLS RF shim was optimized using the internal point constrained optimization method. The addition of dipoles reduced the peak 10 g SAR by 28% at a given excitation fidelity.