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7T Human Spine Imaging Arrays With Adjustable Inductive Decoupling

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Abstract

Ultrahigh-field human spine RF transceiver coil arrays face daunting technical challenges in achieving large imaging coverage with sufficient B_1 penetration and sensitivity, and in attaining robust decoupling among coil elements. In this paper, human spine coil arrays for ultrahigh field were built and studied. Transceiver arrays with loop-shaped microstrip transmission line were designed, fabricated, and tested for 7-tesla (7T)MRI. With the proposed adjustable inductive decoupling technique, the isolation between adjacent coil elements is easily addressed. Preliminary results of human spine images acquired using the transceiver arrays demonstrate the feasibility of the design for ultrahigh-field MR applications and its robust performance for parallel imaging.

Index Terms

Array; decoupling; high-field MRI; human spine; microstrip; parallel imaging; RF coil

I. INTRODUCTION

High magnetic fields and parallel imaging are synergistic and complementary, providing both high spatial resolution and high temporal resolution in MR imaging and spectroscopy. Parallel imaging with multicoil arrays provides a solution to many of the problems encountered with fast acquisitions at high magnetic fields [1]–[5]. In turn, high fields are expected to improve parallel imaging performance due to the more complex sensitivity profile of each coil element and increased SNR [3], [6]–[8]. A challenge at ultrahigh fields, however, is the increased difficulties encountered when designing arrays, which are optimized for parallel imaging, especially for samples requiring a large field of view (FOV) such as the human spine. The conventional method utilizing a large body volume transmit coil for spin excitation and a separate array of multiple local surface coils for signal reception at lower field strengths is not readily extendable to ultrahigh fields, because the required large transmit volume coil is usually unavailable. In addition, the transmit B_1 (i.e., B_1^+) profile generated by the large volume coil at ultrahigh fields is highly nonuniform in the human body owing to the body's high permittivity and large size. Alternatively, a promising approach when working with ultrahigh fields is to build transceiver coil arrays that allow for independent phase and amplitude control for each coil element. This method would enable B_1 shimming commonly required at high fields and parallel excitation. These transceiver arrays are usually built with the structure of a microstrip transmission line (MTL) [3], [9]–[16] which, in its simplest form, consists of a thin strip conductor and a ground plane separated by a low-loss dielectric substrate [9]. Due to its unique semiopen structure, substantial electromagnetic energy is stored in the dielectric substrate. As a consequence, this structure combines the advantages of phase-adjustment on each coil element and intrinsic RF shielding of the transmission-line, enabling low radiation losses, reduced mutual inductive coupling and improved sensitivity distribution. In contrast to conventional lumped-element RF coils, the MTL has demonstrated superior advantages in ultrahigh-field transceiver coil array designs for human parallel MR imaging applications.

A number of technical challenges and design considerations in the design of transceiver arrays for the imaging human spine at the ultrahigh field of 7T need to be addressed. Sufficient imaging coverage and B_1 penetration with efficient MR signal excitation and reception are required in order to detect the signals from the entire length of the spine. The strong coupling of coil elements to the sample mediates interactions among the coil elements, making it more difficult to electromagnetically decouple them [17]. Although the interconnecting L/C

components [18], [19] would be employed for decoupling coil elements at 7T or higher fields [3], [13], [15], and [16], the method is not always efficient for coil arrays with heavier loading. In addition, the required decoupling capacitance is often impractically small and extremely sensitive to different loads, resulting in the difficulties of on-site tuning and matching [15], [16]. Inductive decoupling is an improved scheme for high fields [15], but in practice it is not convenient to achieve optimized decoupling by tuning a multiturn inductor.

In this work, we explore the feasibility of designing transceiver arrays for human spine parallel MR imaging at 7T, using loop-type microstrip elements with proposed adjustable inductive decoupling. Preliminary 7T spine images acquired from healthy volunteers are presented. Accelerated images with Generalized autocalibrating partial parallel acquisition (GRAPPA) are also obtained with MATLAB to demonstrate its parallel imaging performance at the ultrahigh field of 7T.

II. MATERIALS AND METHODS

A. Size of Coil Elements and B_1 Penetration

Loop-shaped MTL coils were selected to build the spine array. These loops were linearly placed along superior–inferior direction without coil overlap to optimize its parallel imaging performance. Sufficient B_1 penetration is critical for such planar spine arrays to study spinal cords and discs. Unlike the conventional low-field spine arrays in which the homogeneous B_1^+ is easily achieved with a body volume coil, this spine array may produce nonuniform B_1^+ field, which is highly confined around coil elements. For this reason, the size of each MTL loop and number of loops were investigated for a sufficient imaging coverage.

Three coil geometries were considered with different coil size and coil number. The element dimensions are 12 cm $\times$ 12 cm with four channels, 12 cm $\times$ 8 cm with six channels, and 12 cm $\times$ 6 cm with eight channels. Gaps between nearest neighbors for all the versions were 1 cm and low loss substrate with 1.2 cm thickness was selected. The total lengths of those arrays were all approximately 50 cm to fit the human spine scan. Coil elements with a size larger than 12 cm $\times$ 12 cm are difficult to build at 7T ($\sim$ 300 MHz), and those, which have a size smaller than 12 cm $\times$ 6 cm may not have enough B_1 depth. To investigate the B_1 penetration of those three arrays, three MTL coils with the three sizes were built with a resonant frequency of 300 MHz and tested with a shielded 1.5-cm single-loop inductive pickup probe. S_{21} was taken on an Agilent Model E5070B network analyzer by connecting the measured coil and the sniffer into two ports of the network analyzer. The penetration measurements for each coil were taken along the main axis of the loop and based on the depth that the picked voltage decayed to 5% of the maximum voltage. The comparison results are shown in Table I.

Based on our measured results, the MTL loop with the size of 12 cm $\times$ 6 cm has limited penetration less than 9 cm, which is insufficient for spine detection. Other two versions with larger coil elements were finally selected to build the 7T transceiver spin array.

B. Decoupling Scheme for Transceiver Coil Array

Although the RF shield of MTL reduces the mutual coupling between coil elements, in most cases when the substrate thickness is increased to pursue deeper B_1 field penetration, or the coil array is built with narrow gaps between coil elements, the mutual coupling among coils may be strong enough to cause the resonance peak to split. Hence, additional decoupling circuitry is necessary to isolate individual coil elements. This becomes particularly important in transceiver arrays where low-input impedance preamplifier decoupling is not easily implemented. Decoupling methods such as placing capacitors across nearest coil elements are

currently used for microstrip arrays [3], [14], [16]. As shown in Fig. 1(a), the required decoupling capacitance can be expressed as follows [13]:

$$C_d = \frac{1}{L\omega^2} \frac{2(1+C_t/C_1)^2 k}{(C_t/C_1)^2(1-k^2) - 2(C_t/C_1)k(1+k)}, \quad \text{for } \frac{C_t}{C_1} > \frac{2k}{(1-k)}, \quad (1)$$

where L is the selfinductance of each coil element; k is the coupling coefficient defined as $k = M / \sqrt{L_1 L_2}$, in which M is mutual inductance; L_1 and L_2 are the selfinductances of each loop. For the special case when $L_1 = L_2 = L$, k can be defined as $k = M/L$. ω is the desired frequency. C_t and C_1 are tuning capacitors on the MTL loop. If there is no mutual inductance involved between coil loops, the relationship between C_t and C_1 and resonance frequency ω can be simply expressed as

$$\frac{1}{\omega} \left(\frac{1}{C_t} + \frac{1}{C_1} \right) = \omega L.$$

Based on (1), the decoupling capacitance is inversely proportional to square of the resonance frequency. At a resonance frequency higher than 300 MHz, the required decoupling capacitance is usually very tiny and quite sensitive to different loadings. As a result, the coil adjustments for each patient are complicated and time consuming. The alternative method is applying inductors across the nearest neighbors for decoupling, which is shown in Fig. 1(b). The required decoupling inductance is expressed in (2) [13]

$$L_d = L \frac{(C_t/C_1)^2(1-k^2) - 2(C_t/C_1)k(1+k)}{2k(1+C_t/C_1)^2}, \quad \text{for } \frac{C_t}{C_1} > \frac{2k}{(1-k)}. \quad (2)$$

It is revealed that implementing inductors is particularly advantageous at ultrahigh fields since the decoupling inductance is independent of resonance frequencies. After choosing the appropriate decoupling inductance, the coil array will be more stable to different loads. In this case, the loss from inductors is negligible since the sample loss is dominated at ultrahigh fields. A disadvantage to this method, however, is that it requires inconvenient decoupling adjustments.

C. Adjustable Inductive Decoupling

If the tuning capacitors C_1 and interconnecting inductors L_d in Fig. 1(b) are treated as a decoupling network, the decoupling condition can be achieved by varying C_1 instead of L_d . This new method is depicted in Fig. 1(c), in which C_1 is marked as C_d for decoupling adjustment. For achieving the best decoupling performance among the elements, the overall inductance of the decoupling circuitry is adjusted by tuning the decoupling capacitor while the interconnecting inductance keeps unchanged. Although a capacitor is introduced in the decoupling circuitry, this scheme is still inductive decoupling, thus it is particularly useful and convenient in practice at ultrahigh fields. The relationship between inductor L_d and the ratio C_t/C_d was plotted in Fig. 2 according to (2), assuming the selfinductance of each coil is 1 μH . It is clear that C_t/C_d is proportional to interconnecting inductance. As indicated in Fig. 2, coils with weak mutual coupling ($k < 0.2$) should use small interconnecting inductance values (less than 1 μH), otherwise, adjusting C_t and C_d will be difficult both for decoupling and frequency tunings. Very small inductance values (less than 0.4 μH) may be sensitive to different loadings and should also be avoided.

D. Design of 7T Spine Array

Two spine arrays with different coil element sizes were designed and built for proton MR imaging at 7T, corresponding to the resonance frequency of 298.3 MHz. The structures of two spine arrays are depicted in Fig. 3. One array had six coil elements [Fig. 3(a)]. The size of the coil elements in the square shape was $8\text{ cm} \times 12\text{ cm}$. The second array had four elements [Fig. 3(b)], each element with a bigger loop size of $12\text{ cm} \times 12\text{ cm}$. MTL loops in the two designs were all linearly placed with only 1 cm innergaps. All the substrates (dielectric material) used were 1.2-cm-thick polytetrafluoroethylene (PTFE), which has a low loss tangent ($\tan\delta < 0.00015$) and a permittivity of 2.1. Other kinds of low-loss dielectric materials with different permittivities can certainly be used. Because corners of the coil tend to radiate surface waves, and potentially creating hot spots in images and degrading the Q values of the coils, the corners had been chamfered to reduce the radiation loss and improve B_1 distribution. The strip conductors and ground planes of MTL loop arrays were made from 36- μm -thick adhesive-backed copper tape (3M, St. Paul, MN). The width of the strip conductor for all coils was 1.2 cm. Three trimmer capacitors (Voltronics, Denville, NJ) ranged within 19 pF were mounted on each loop, for frequency tuning, matching, and decoupling. Two homemade nonmagnetic inductors were connected between the nearest coils as indicated in Fig. 1(c). These inductors were made of 15-turns copper wires with 5 mm diameter. Inductances were approximately 0.8 μH at 100 kHz. By slightly changing the decoupling capacitance and the tuning capacitance on each loop, the decoupling condition was finally achieved. On-site tuning and matching of each coil should be conducted with the array inside the magnet bore and loaded with the human spine. It is unnecessary to adjust the decoupling capacitors again at the loading case, which indicates an advantage of the inductive decoupling scheme.

Each coil element was connected to the T/R switch of the MR scanner via a coaxial cable. For each coil array the resonance frequency and decoupling were confirmed using the reflection coefficient S_{11} and the transmission coefficient S_{21} measurements taken on a network analyzer. The reflection coefficient S_{11} measurement was also used to measure the coils' Q values for the loaded (with the human spine) case.

E. MRI Experiments

The MR imaging experiments with these coils were performed on a 7T/90 cm magnet (GE healthcare, Milwaukee, WI). This scanner is equipped with two quadrature transmit channels and two T/R switches. To test the transceiver spine arrays on this system, scans were conducted by connecting two coil elements into the transmit channels each time and combined offline. To avoid the signal cancellation from the phase difference of the channels, each scan was performed on two nonadjacent coil elements. All other coil elements without scanning were terminated with 50 Ω terminators. Ten healthy volunteers were tested with these arrays by using gradient echo (GRE) images with different echo time (TE) and repetition time (TR) at 20° flip angle. Other MRI parameters were matrix size: 256×256 , FOV: $35\text{ cm} \times 35\text{ cm}$ (the maximum achievable FOV) at sagittal plane and $20\text{ cm} \times 20\text{ cm}$ at axial plane.

The individual channel images are shown to demonstrate the isolation of coil elements. Combined images at axial and sagittal planes are also shown. To test its parallel imaging performance, GRAPPA [20] with the subset selection strategy [21] was applied at reduction factors of 2, 3, and 4 (autocalibration lines, ACS line = 60).

III. RESULTS

Each element of the transceiver array was tuned to 298.2 MHz (the proton Larmor frequency of the 7T magnet that we used in this paper) and matched to system's 50 Ω . At ultrahigh fields where the sample loss is dominated, the losses from the coil and decoupling circuits are

negligible. Decoupling between the nearest neighbors was approximately -23 dB without loading and approximately -17 dB after loading of the human spine. The loaded isolation decreased by several decibels due to the coil's strong coupling to the sample, but it is still in the acceptable range for such transceivers. Isolations among all nonadjacent elements were better than -25 dB both with and without loading. Fig. 4 shows the measurement results on scattering parameters S_{11} , S_{22} , and S_{21} of the prototype transceiver array loaded with the human spine.

Spine images acquired from the same healthy volunteer by using the four-element array and six-element array are shown in Fig. 5. The thoracic-spine disc images were acquired with a GRE sequence within one breathhold (flip angle: 20° , matrix size: 256×256 , slice thickness: 5 mm, TE/TR: 4.2ms/117 ms). Power used for transmission was approximately 250 W per channel. The measurements showed that the four-element array had a deeper penetration of ~ 12 cm while the six-element array had ~ 9 cm, which may show limitations in intervertebral disc imaging unless higher transmit power (>250 W for each channel) is applied. This conclusion is in accordance with our results from bench tests. Thus, the four-element array was finally selected for the following spine imaging.

Individual channel images in Fig. 6(a)–(d) illustrate the good coil isolation between the four elements. After combination with the sum-of-squares (SoS) method, the whole spine image [see Fig. 6(e)] has an acceptable homogeneity at the spinal disc area. Almost all the discs are clearly visible except the ones between the coil element gaps. Fig. 7(a) shows the sagittal image of lumbar spine (GRE flip angle: 30° , matrix size: 256×256 , FOV: $35 \text{ cm} \times 26 \text{ cm}$, slice thickness: 4 mm, TE/TR: 3.7 ms/150 ms, NEX 10). An axial image is shown in Fig. 7(b) (GRE flip angle: 20° , matrix size: 256×256 , FOV: $20 \text{ cm} \times 20 \text{ cm}$, slice thickness: 5 mm, TE/TR: 4.2 ms/117 ms), in which the spinal cord is clearly visible.

The parallel imaging capability of this array was also tested. Datasets were reconstructed offline with a customized GRAPPA-based parallel reconstruction algorithm developed in our laboratory in MATLAB (The Math Works, Natick, MA). Parallel imaging with reduction factors 2, 3, and 4 was successfully employed with 60 autocalibration signals lines and therefore the real acceleration rates are 1.62 [see Fig. 8(b)], 2.04 [see Fig. 8(c)], and 2.35 [see Fig. 8(d)]. The curvature distortion of the images was caused by gradient nonlinearity and imperfect B_0 shimming. This effect was not corrected in the images, since our research goal at this point was to demonstrate parallel imaging performance.

IV. DISCUSSION AND CONCLUSION

The feasibility of 7T human spine arrays was studied. Four-channel spine transceiver arrays with MTL structure were designed and constructed. Preliminary *in vivo* images of the human spine were also acquired and demonstrated with the coil array. The microstrip loop design associated with the extra adjustable inductive decoupling is suitable for human spine imaging, providing large image coverage, appropriate B_1 penetration, and decent parallel imaging performance.

As preliminary work for ultrahigh-field human spine imaging, several issues need to be addressed. First of all, multiple transmit channels are necessary to be utilized for such coil arrays. Due to the lack of transmit channels, we tested this spine array through multiple scans and combined the individual channel images offline. Thus, the independent phase control of each channel was not investigated. With the capability of phase control, better homogeneous and B_1 depth may be achieved. Second, the pulse sequences applied to the spine arrays were all GRE sequences with small flip angles for subject safety and MR system reliability considerations. To avoid a large specific absorption rate, we have not tested pulse sequences

that require high-RF power for imaging such as fast spin echo. We believe that higher RF power (bigger flip angles) and better B_0 shimming may improve the image quality with deeper B_1 penetration for imaging deeper structures such as the intervertebral discs. Third, the motion artifacts caused by B_0 distortion from respiration and heartbeat may significantly degenerate the image quality especially for thoracic spine imaging. As a result, the thoracic spine images were acquired within one breathhold to minimize the motion artifacts. The spinal imaging SNR, therefore, was not optimized. The future work will be focused on parallel excitation and investigation of the transmit efficiency of the proposed design.

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Fig. 1.

(a) Schematic of decoupling circuits with interconnecting capacitors, (b) interconnecting inductors, and (c) adjustable inductive decoupling.

Fig. 2.

Explanation of adjustable inductive decoupling. Decoupling condition is achieved by varying C_t/C_d , when interconnecting inductors have constant value.

Fig. 3.
(a) Photos of the six-channel and (b) four-channel human spine array at 7T.

Fig. 4. Scattering parameters S_{11} , S_{22} , and S_{21} measurements between the nearest neighbors of the four-element array after loading with the human spine. The isolation is -17 dB. Frequency span is 30 MHz.

Fig. 5. Images acquired with (a) six-channel array and (b) four-channel array. Spinal disc images indicate that four-channel array with 12 cm $\times$ 12 cm coil size has deeper B1 penetration and better SNR for discs. The same transmit power $\sim$ 250 W per channel was applied. Images were acquired with GRE sequences (flip angle: 20°, matrix size: 256 $\times$ 256, slice thickness: 5 mm, TE/TR: 4.2 ms/117 ms).

Fig. 6.

(a)–(d): Images acquired from each of the loops of the spine array. (e): Composite image produced using the root of the SoS method. Images were acquired with a GRE pulse sequence (flip angle: 20° , matrix size: 256×256 , slice thickness: 5 mm, TE/TR: 4.2 ms/117 ms, FOV: $35 \text{ cm} \times 35 \text{ cm}$). A portion of the images covering the human spine is shown.

Fig. 7.

GRE images (flip angle: 30° , matrix size: 256×256 , FOV: $35 \text{ cm} \times 26 \text{ cm}$, slice thickness 4 mm, TE/TR: 3.7 ms/150 ms, NEX: 10) were acquired from a healthy volunteer in (a) sagittal and (b) axial orientations (flip angle: 20° , matrix size: 256×256 , FOV: $20 \text{ cm} \times 20 \text{ cm}$, slice thickness: 5 mm, TE/TR: 4.2 ms/117 ms).

Fig. 8. Spine images reconstructed using the root of the SoS method (a), and using GRAPPA of reduction factors 2, 3, and 4 (ACS line = 60), so the real reduction factors were (b) 1.62, (c) 2.04, and (d) 2.35. The distortion of the images is mainly caused by gradient nonlinearity and imperfect B_0 shimming.

TABLE I**B₁ Penetration of Three Coil Sizes**

Coil size (cm)	Channel number	<i>B₁</i> field penetration (cm)
12 × 12	4	11.6
12 × 8	6	10.2
12 × 6	8	8.5