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B_1 Homogenization in MRI by Multi-layer Coupled Coils

Shumin Wang, Joseph Murphy-Boesch, Hellmut Merkle, Alan P. Koretsky, and Jeff H. Duyn

Laboratory of Functional and Molecular Imaging, National Institute of Neurological Disorders and Stroke, National Institutes of Health, 10 Center Dr. 10/B1D728, Bethesda, Maryland 20892, USA.

Abstract

Transmit B_1^+ field homogenization in high-field (> 3.0 T) human MRI is challenging due to radio frequency (RF) wavelength effects. An approach based on appropriately coupling surface coils to a volume coil was investigated. Electro-magnetic simulation results demonstrated the feasibility and effectiveness of this method in proton MRI of the human head at 7.0 T.

Keywords

Magnetic resonance imaging; radio frequency coils

I. Introduction

High field MRI systems (7.0 Tesla and above) are increasingly finding applications for both anatomical and functional MRI of human brain. However, one of the remaining problems slowing the widespread application of high field MRI is the control of the transverse RF magnetic (B_1^+) field, which becomes difficult with the shorter RF wavelengths at high fields [1].

General MRI applications require uniform B_1^+ , which helps to minimize contrast and sensitivity variations over the object. Traditional quadrature-driven volume coils used for B_1^+ excitation provide limited uniformity of the B_1^+ profile [2], which is often not better than about $\pm 50\%$ for proton MRI of human head at 7T (Larmor frequency $f_L \sim 300$ MHz) [1]. Therefore, the ability to homogenize transmit B_1^+ profile is important for full realization of the potential of high-field MRI. Recently, several methods have been proposed in RF coil design to homogenize B_1^+ profiles, which include inserting high-permittivity materials (or passive RF shimming) [3], inverse coils [4][5], active RF shimming [6][7][8], and transmit SENSE [9][10].

It is well-known that surface coils coupled to a volume transmit coil can perturb the B_1^+ profile of the latter. Normally, surface coils are completely decoupled during transmission. However, when the B_1^+ field generated by a traditional volume coil becomes distorted at high-field, such perturbations may become valuable to manipulate the profile. Local B_1^+ correction using one or two coupled surface coils has been demonstrated in Refs. [11] - [14].

In this article, we investigate the feasibility of improving B_1^+ homogeneity by coupling an array of surface coils to a volume coil. This is different from coupling a single surface coil

to a volume coil. Since all surface coils are also mutually coupled through the volume coil, their interaction must be considered as well in global B_1^+ homogenization. Preliminary results of this work have been reported in Refs. [13] and [14].

II. Methodology

A. Multi-layer Coupled Transmit Coil

In this study, the outer-layer is a 25.4-cm (10-in) long 16-rung shielded birdcage coil with a 4-port quadrature excitation. Two types of birdcage coil were considered. The first one has an elliptical cross-section with a major axis of 30-cm (12-in) and a minor axis of 22.86-cm (9-in). The second one has a circular cross-section with a diameter of 30-cm (12-in). The RF shield is 2.54-cm (1-in) away and conformal to each birdcage coil respectively.

The inner layer consists of an eight-element overlapped surface coil array (Fig. 1). The dimension of each inner-layer element is roughly 6-cm (longitudinal) by 9-cm (azimuthal). The overlaps between neighboring loops were adjusted to minimize inductive decoupling. Although the inner-layer elements may couple through the birdcage coil, it was found that this effect depends upon the geometry of the outer-layer coil. With elliptical birdcage coil, this was an insignificant effect.

In addition, capacitive coupling must also be considered. If two overlapping coils are too close (in the radial direction), parasitic capacitance develops between coil conductors that face each other. Thus the overlapping region behaves like another resonant loop and its B_1^+ field dominates in the peripheral region. After trials, each surface coil was set to be 3-mm wide and radially separated from its nearest neighbors by 6-mm.

The outer-layer coil is the only actively driven unit to generate a primary B_1^+ field. The inner-layer array passively couples to the outer-layer and produces a modulating B_1^+ field. Control of the intra-layer coupling is achieved by tuning the inner-layer elements away from the Larmor frequency appropriately.

The inner-layer array described above covers a 4-cm thick region of interest (ROI) beneath the elements, which is roughly 5-cm to 8-cm from the top of the human head in the S-I direction. We note that in typical high-field brain imaging, this ROI is of general interest and also seriously affected by B_1^+ inhomogeneity. The B_1^+ field in regions above this ROI is typically more homogeneous (peak-to-peak ratio better than 75%). If B_1^+ modulation is desired in other regions, the same methodology can be applied.

B. B_1 Modulation by Surface Coils

Considering a circular loop coil with radius a and surface normal pointing to $\hat{n}$ direction in a *linearly polarized* uniform time-varying magnetic field $\vec{B}_1 = B_{1n} e^{j\omega t} \hat{n}$, the electromotive force (EMF) $\mathcal{E}$ generated by $\vec{B}$ is given by Faraday's law as

$$\mathcal{E} = -j\omega\mu B_{1n}(\pi a^2) \quad (1)$$

, where $\omega = 2\pi f$. Each surface coil can be modeled by a serial *RLC* circuit with input impedance at the Larmor frequency

$$Z_L = R_s \left[1 + j \frac{\omega L}{R_s} \left(\frac{f_L}{f} - \frac{f}{f_L} \right) \right] \quad (2)$$

where R_s is associated with losses, L is the coil inductance, f_L is the Larmor frequency and f is the actual detuning frequency of a coil element given by $C = 1/\omega^2 L$. Now let $f = f_L + \Delta f$ and assume Δf is relatively small compared to f_L . The impedance obtained after keeping the lowest order terms of the Taylor expansion of Eq. (2) with respect to Δf is

$$Z_L = R_s \left[1 - j2 \frac{\omega L \Delta f}{R_s f} \right] \quad (3)$$

The loss term R_s is typically small, and the current induced on the loop coil can be approximated by

$$I_L \simeq \frac{\mu B_{1n} (\pi a^2)}{2L} \frac{f_L}{\Delta f} \quad (4)$$

Using the expression for the magnetic field generated by a circular loop [15], the modulation field produced by the induced current at distance $k \cdot a$ away from the coil center in the direction of $\hat{n}$ is given by [15]

$$B'_{1n} = \frac{\mu}{L} \frac{\pi a}{4(1+k^2)^{\frac{3}{2}}} \frac{f_L}{\Delta f} B_{1n} \quad (5)$$

where k denotes the normalized distance from the coil center. For wires of radius r_w , the inductance is approximately

$$L = \mu a \left[\ln \left(\frac{8a}{r_w} \right) - 2 \right] \quad (6)$$

Substituting Eq. (6) into Eq. (5), we obtain

$$B'_{1n} = \frac{\pi}{4(1+k^2)^{\frac{3}{2}}} \frac{f_L}{\left[\ln \left(\frac{8a}{r_w} \right) - 2 \right] \Delta f} B_{1n} \quad (7)$$

Since only half of the linearly polarized magnetic field $\vec{B}_1$ contributes to the *circularly polarized* magnetic field B_1^+ , the actual modulation to B_1^+ along the axis of the loop is only half of that given in Eq. (7). Thus the total B_1^+ field at distance $k \cdot a$ from the coil center is

$$B_1^+ = \left\{ 1 + \frac{\pi}{8(1+k^2)^{\frac{3}{2}}} \frac{f_L}{\left[\ln \left(\frac{8a}{r_w} \right) - 2 \right] \Delta f} \right\} B_{1o}^+ \quad (8)$$

where B_{1o}^+ is the original *circularly polarized* magnetic field before any modulation ($\Delta f \rightarrow \infty$). Considering the loop coils in our study, the desired detuning frequency should be about 5% above the Larmor frequency in order to make the minimum B_1^+ field (at distance a from coil surface), which is about 50% of the maximum B_1^+ , equal to the maximum B_1^+ .

We notice from Eq.(8) that field cancelation can occur if a surface coil is tuned below the Larmor frequency, i.e. $\Delta f < 0$. Furthermore, Eq.(8) estimates the worst scenario where the original B_{1o}^+ field needs to be doubled. In general, peripheral B_1^+ field can be higher and less compensation may be required. Thus Eq. (8) provides a lower bound for the detuning frequency. The actual magnetic field distribution is more complicated than assumed. Since

the B_1^+ profile of a surface coil depends on its relative position to a phantom, the optimal detuning frequency is expected to be phantom-dependent. To this end, we applied full-wave simulations for more accurate results.

III. Numerical Results

A Method of Moments (MoM) program based on the PMCHWT (Poggio, Miller, Chang, Harrington, Wu, Tsai) formulation was developed and applied [16][17]. The human head phantom (Fig.1) has relative permittivity of $\epsilon_r = 36$ and conductivity of $\sigma = 0.657 S/m$. The conductivity corresponds to 55 mM NaCl solution. The permittivity is derived in such a way that the electrical size of the cross-section of the human head model (in terms of wavelength) is similar to that of the white matter region ($\epsilon_r = 43$ at 298 MHz). Two parameters were evaluated on four axial slices in the ROI. The first is the peak-to-peak ratio of B_1^+ intensity. The second is the ratio of the standard deviation of B_1^+ to the average B_1^+ , or the inhomogeneity coefficient.

The loaded elliptical and circular birdcage coils were numerically tuned to 298 MHz by thirty-two 5.86 pF capacitors. The frequency response of the unloaded birdcage coil (tuned with a difference capacitance) is shown in Fig. 2. Also shown in Fig. 2 is the frequency response of the loaded birdcage coil with and without the inner surface elements. With the inner elements, the loaded birdcage coil is slightly detuned and the Q factor drops. Both can be explained by the fact that the intra-layer coupling effectively brings the birdcage coil closer to the human head.

It was found that four-port quadrature driving is sufficient and necessary. No obvious improvement was observed by driving more than four ports and much deteriorated results were observed by driving only two ports. Figure 3 compares the B_1^+ profiles of the elliptical and the circular birdcage coils. The simulation result of the circular birdcage coil corresponds to our previous 7T experimental results [18]. It can be seen that the elliptical birdcage coil has a different symmetric pattern than the circular birdcage coil, and its homogeneity is slightly better. However, the main reason for choosing the elliptical coil is its more efficient coupling to the inner-layer array.

We started from improving the homogeneity in the nearest neighborhood of each inner-layer element individually by adjusting the detuning frequency from 310 MHz to 340 MHz. The tuning capacitance of the birdcage coil remains the same in all simulations. The detuning frequency of each element that yields an average (on four axial slices) minimum-to-center B_1^+ ratio of 0.8 (in its nearest neighborhood) was chosen. If the average minimum-to-center B_1^+ ratio is above 0.8 when a coil was detuned to 340 MHz, 340 MHz was chosen.

After each individual optimization, all inner-layer elements were simulated simultaneously inside the birdcage coil to achieve the maximum global minimum-to-center B_1^+ ratio. In this procedure, the detuning frequency of one element was adjusted, while all the others were fixed. Thus one cycle of global optimization consisted of eight individual adjustments. This procedure was continued until no more improvement could be achieved. It was found that global optimization is *not* mandatory for the elliptical birdcage coil, while four cycles of computation were required for the circular birdcage coil. Each surface coil could be optimized individually to achieve global optimization with the elliptical birdcage coil. The detuning frequencies are listed in Table I.

The B_1^+ profiles of the elliptical birdcage coil on two axial slices, which are 5-cm and 8-cm from the top of the head, are shown in Figs. 4(a) and (c) respectively. The peak-to-peak

ratios, as listed in Table II, range from 0.53 to 0.74. The homogenized B_1^+ profiles on corresponding slices are shown in Figs. 4(b) and (d) respectively, where homogeneity improvements can be clearly identified. The decrease of inhomogeneity coefficients (as listed in Table III), ranges from 30% on the top slice to 38% on the bottom slice. While the average improvement was about 1/3, the peak-to-peak ratio increases the most on the bottom slice, which was about 20%.

Figure 5 compares the electrical field distributions of an isolated birdcage coil with those of the coupled transmit coil. The electrical field becomes more concentrated in the immediate neighborhood of the inner-layer coil array and the Specific Absorption Rate (SAR) is expected to increase in those regions. However, the peak SAR of the head can be a different issue. For instance, Figures 5(c) and (d) show that the peak SAR of the head occurs around the eyes. Since the eyes are away from the inner-layer array, the electric field actually decreases at this location and the peak SAR is reduced by about 15%. Indeed, the SAR is patient-dependent and it should be better estimated by using a detailed head model. This is one of our ongoing research topics.

IV. Conclusions

We investigated a B_1^+ homogenization approach in high-field MRI by coupling an inner-layer surface coil array to an outer-layer volume transmit coil. Theoretical analysis shows that field compensation occurs when an inner-layer element is appropriately tuned above the Larmor frequency. From numerical simulations of an eight-element surface coil array, it was estimated that on average, the B_1^+ homogeneity can be improved by about 1/3 for 7T brain imaging. Compared to other B_1^+ homogenization methods, this approach has the distinct advantage that it can be implemented without the need for independent RF channels, thereby reducing MRI system complexity. Furthermore, it can be readily adapted to existing coil systems by detuning surface coils rather than decoupling them, during the transmit phase.

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References

- [1]. Jin J, Shen G, Perkins T. On the field inhomogeneity of a birdcage coil. *Magn Reson Med*. 1994; vol 32:418–422. [PubMed: 7984076]
- [2]. Hayes CE, Edelstein WA, Schenck JF, Mueller OM, Eash M. An efficient, highly homogeneous radiofrequency coil for whole-body NMR imaging at 1.5 T. *J. Magn. Reson*. 1985; vol 63:622–628.
- [3]. Yang QX, Mao W, Wang J, Smith MB, Lei H, Zhang X, Ugurbil K, Chen W. Manipulation of image intensity distribution at 7.0 T: Passive RF shimming and focusing with dielectric materials. *J. Magn. Reson. Imaging*. 2006; vol. 24:197–202. [PubMed: 16755543]
- [4]. Fujita H, Petropoulos LS, Morich MLA, Shvartsman SM, Brown RW. A hybrid inverse approach applied to the design of lumped-element RF coils. *IEEE Trans. Biomed. Eng*. 1999; vol. 46:353–361. [PubMed: 10097470]
- [5]. Lawrence BG, Crozier S, Yau DD, Doddrell DA. A time-harmonic inverse methodology for the design of RF coils in MRI. *IEEE Trans. Biomed. Eng*. 2002; vol. 49:64–71. [PubMed: 11794773]
- [6]. Ibrahim TS, Lee R, Robitaille PML. Effect of RF coil excitation on field inhomogeneity at ultra high fields: a field optimized TEM resonator. *Magn Reson Med*. 2001; vol 19:1339–1347.

- [7]. Li BK, Liu F, Crozier S. Focused, eight-element transceive phased array coil for parallel magnetic resonance imaging of the chest - Theoretical considerations. *Magn. Reson. Med.* 2005; vol. 53:1251–1257. [PubMed: 15906277]
- [8]. Vaughan T, DelaBarre L, Snyder C, Tian J, Akgun C, Shrivastava D, Liu W, Olson C, Adriany G, Strupp J, Andersen P, Gopinath A, van de Moortele P, Garwood M, Ugurbil K. 9.4T human MRI: Preliminary results. *Magn Reson Med.* 2006; vol. 56:1274–1282. [PubMed: 17075852]
- [9]. Katscher U, Bornert P, Leussler C, van den Brink JS. Transmit SENSE. *Magn Reson Med.* 2003; vol. 49:144–150. [PubMed: 12509830]
- [10]. Zhang Z, Yip CY, Grissom W, Noll DC, Boada FE, Stenger VA. Reduction of transmitter B1 inhomogeneity with transmit SENSE slice-select pulses. *Magn Reson Med.* 2007; vol. 57:842–847. [PubMed: 17457863]
- [11]. Schmitt, M.; Feiweier, T.; Voellmecke, E.; Lazar, R.; Krueger, G.; Reykowski, A. B1-Homogenization in abdominal imaging at 3T by means of coupling coils. 13th Annual Meeting ISMRM; Miami, Florida. May 2005; p. 331
- [12]. Wiggins, GC.; Zakszewski, E.; Wiggins, CJ.; Wald, LL. B1 Transmit Field Manipulation at 7 Tesla Using Controlled Decoupling of Array Coil Elements (CODACE). 15th Annual Meeting ISMRM; Berlin, Germany. May 2007; p. 1054
- [13]. Wang, S.; Merkle, H.; Koretsky, AP.; Duyn, JH. Floating Elements for Controlling High-Field Transmit B1 Profile; 48th ENC Conference; Daytona Beach, FL. April 2007; p. 109
- [14]. Wang, S.; Merkle, H.; Koretsky, AP.; Duyn, JH. Improving high-field transmit B1 field homogeneity using coupled inner elements. 15th Annual Meeting ISMRM; Berlin, Germany. May 2007; p. 3275
- [15]. Ramo, S.; Whinnery, JH.; van Duzer, T. *Fields and Waves in communication electronics.* Reading, John Wiley & Sons; New York: 1965.
- [16]. Harrington RF. Boundary integral formulations for homogeneous material bodies. *Journal of Electromagnetic Waves and Applications.* 1989; vol. 3:1–15.
- [17]. Wang S, Duyn JH. Analyzing radio-frequency coil arrays in high field magnetic resonance imaging by the combined field integral equation method. *Physics in Medicine and Biology.* 2006; vol. 51:3211–3229. [PubMed: 16757872]
- [18]. Ikonomidou, VN.; de Zwart, JA.; Duyn, JH.; van Gelderen, P. B1 mapping based on signal modeling with magnetization prepared 3D EPI. 13th Annual Meeting ISMRM; Miami, Florida. May 2005; p. 899

Fig. 1. The high-pass birdcage coil and the inner-layer array. Gray patches on the birdcage coil represent tuning capacitors and red rectangles denote excitations. The outmost ellipse in (b) denotes the shield.

Fig. 2. Frequency response of the unloaded birdcage coil and the loaded one with and without the inner array (insertion).

Fig. 3. B_1^+ profiles of the elliptical birdcage coil (a) and of the circular birdcage coil (b) on axial slice 8-cm from the top of the head.

Fig. 4.

The B_1^+ profiles of the birdcage coil along on axial slices 5-cm (a) and 8-cm (c) from the top of the head and the homogenized profiles on corresponding slices (b) and (d). Note that each B_1^+ map has been normalized to its own peak, so that the same color represents the same normalized value.

Fig. 5. The electric field distributions of the birdcage coil along (a,c) and the coupled transmission coil (b,d) on the axial slice 8-cm from the top of the head (a,b) and the sagittal slice (c,d). Results were normalized to the maximum value of each individual plot.

TABLE I

Detuning frequency of each inner-layer element.

Coil	0	1	2	3	4	5	6	7
f (MHz)	340	317	317	325	340	315	315	340

TABLE II

Peak-to-peak B_1^+ ratios on different axial slices.

Distance (cm)	Birdcage coil alone	Homogenized
5	0.744	0.757
6	0.652	0.680
7	0.583	0.625
8	0.530	0.639

TABLE III

Inhomogeneity coefficients on different axial slices.

Distance (cm)	Birdcage coil alone	Homogenized
5	0.081	0.056
6	0.109	0.076
7	0.137	0.090
8	0.159	0.098