OPTIMIZATION OF A HIGH SENSITIVITY MRI RECEIVE COIL FOR PARALLEL HUMAN BRAIN IMAGING

Jacco A. de Zwart[1], *Patrick J. Ledden*[3], *Peter van Gelderen*[1], *Peter Kellman*[2], *Yuxi Pang*[1], *Renxin Chu*[1], *Jeff H. Duyn*[1]

[1]Advanced MRI, LFMI, NINDS and [2]LCE, NHLBI, National Institutes of Health, Bethesda, MD, and [3]Nova Medical Inc., Wakefield, MA, USA

ABSTRACT

Two eight-channel MRI receive-only coils were developed to provide whole-brain coverage at 1.5 T and 3.0 T field strength, respectively. Objectives were an image signal-to-noise ratio superior to standard designs throughout the human brain, as well as high parallel imaging performance. Electro-magnetic field simulations were used to determine array diameter and inter-element coil gap. Low mutual inductive coupling was achieved at 1.5 and 3.0 T using high-impedance pre-amplifiers. Coils show an average SNR improvement over commercial birdcage coils of 2.4 and 2.3 for the 1.5 T and 3.0 T design, respectively. The mean of the noise-amplification factor related to reconstruction of under-sampled data (*g*-factor) was 1.03 for 2-fold under-sampled data (rate-2) and 1.22 for rate-3 at 1.5 T. For data acquired with the 3.0 T coil array, these values were respectively 1.06 for rate-2 and 1.37 for rate-3.

1. INTRODUCTION

Well-designed surface coil arrays have the potential to provide image signal-to-noise ratio (SNR) superior to commercial birdcage volume coils throughout the human brain. In addition, availability of individual coil signals allows implementation of parallel imaging techniques such as SMASH (SiMultaneous Acquisition of Spatial Harmonics) [1] or SENSE (SENSitivity Encoding) [2].

For optimal SNR and accelerated imaging performance, most of the head needs to be surrounded by mutually decoupled receive elements. Previous multi-coil designs achieved decoupling using "magic overlap" of neighboring coil elements, combined with the use of high-impedance preamplifiers [3]. This approach poses restrictions on the coil design. Here, negligible coupling is achieved without the need for "magic overlap" of the coil elements by using ultra high-impedance pre-amplifiers [4] with approximately 3 kΩ input impedance, thus providing greater flexibility in positioning the coil elements, ultimately allowing for lower *g*-factors (a measure for SENSE-reconstruction related noise amplification [2]) and higher image signal-to-noise-ratio (SNR).

Some multi-channel coils have been specifically designed for human brain imaging, optimizing sensitivity in specific regions or the entire brain [5,6]. Recently, a 6-channel SENSE-optimized receive coil for cardiac imaging has been presented [7]. Here, design of 8-element brain arrays for 1.5 T and 3.0 T field strength is described, optimized for SNR and SENSE performance.

2. MATERIALS & METHODS

2.1. Simulations

Electro-magnetic field simulations were used to investigate effects of coil spacing (gap) and array diameter on performance. Both image SNR and SENSE *g*-maps were computed. An 8-channel design was chosen since a maximum of 8 receivers was available. The human head was modeled as a cylindrical object with uniform conductivity σ. Coil elements were assumed identical, rectangular, equally spaced and placed on a cylinder parallel to the B_0-field. The coil noise, N_s, and the coil sensitivity profiles, **S**, were derived for 63.8 MHz (the proton resonance frequency at 1.5 T) from the magnetic vector potential, **A**, according to Biot-Savart law.

$$N_S \propto \sqrt{R_S} \propto \sqrt{\sigma \cdot \omega^2 \int_V \mathbf{A} \cdot \mathbf{A} d\mathbf{r}} \tag{1a}$$

$$\mathbf{S} = \left(\vec{\nabla} \times \mathbf{A}\right)_{x,y} \tag{1b}$$

with R_s the equivalent noise resistance, and ω the NMR resonance frequency. In Equation 1a, magnetic and electrostatic noise sources are neglected, and integration is performed over the entire object volume *V*. In Equation

1b, the z-component is ignored. Noise data derived from network analyzer measurements performed at 63.8 MHz show that the assumption that sample noise is the dominant noise source is valid for coils with a surface area greater than 25 cm^2 (results not shown). Inductive coupling was assumed non-existent, as it was negligible in the actual coils due to use of high-impedance pre-amplifiers [3,4].

The SNR in the i^{th} channel of the coil array is:

$$\mathrm{SNR}_i \propto \frac{\left|\omega\left(\vec{\nabla}\times\mathbf{A}_i\right)_{x,y}\right|}{\sqrt{\sigma\int_V \boldsymbol{A_i}\cdot\boldsymbol{A_i}d\boldsymbol{r}}} \tag{2}$$

Signals from the respective coil channels can be combined as was described by Roemer et al. [3]:

$$\mathrm{SNR}_c = \sqrt{\mathbf{S}^H\mathbf{\Psi}^{-1}\mathbf{S}} \tag{3}$$

with $\mathbf{S}$ a vector containing the sensitivities SNR_i from the individual coil elements and $\boldsymbol{\Psi}$ the noise correlation matrix. Noise amplification in accelerated imaging (g-factor) can be assessed as follows [2]:

$$g_\rho = \sqrt{[(\mathbf{\Sigma}^H\mathbf{\Psi}^{-1}\mathbf{\Sigma})^{-1}]_{\rho,\rho}(\mathbf{\Sigma}^H\mathbf{\Psi}^{-1}\mathbf{\Sigma})_{\rho,\rho}} \tag{4}$$

where $\boldsymbol{\Sigma}$ is the coil sensitivity matrix [2], constituting a reformatted version of $\mathbf{S}$, with number of rows and columns determined by the number of coil elements and acceleration rate, respectively. The parameter g_ρ contains the g-values for the ρ^{th} region in the unaliased image (e.g. for rate-3 SENSE ρ takes on values 0, 1, and 2 and g_ρ is defined over an image section covering $^1/_3$ of the FOV). Assuming that inductive coupling between elements is negligible, $\boldsymbol{\Psi}$ can be calculated from:

$$\Psi_{ij} = \sigma\cdot\int_V \mathbf{A}_i\cdot\mathbf{A}_j d\mathbf{r} \tag{5}$$

2.2. MRI experiments

Experiments were performed on normal volunteers on a 1.5 T Siemens Magnetom scanner and a 3.0 T GE Signa LX platform using the 8-element array for the respective field strength, as well as the standard birdcage coil provided by the manufacturer. Since at the time of the experiment only 4 receive channels were available on the 3.0 T scanner, experiments on that platform were performed in two parts, where only half of the channels was sampled (all channels were connected to a pre-amplifier to maintain a high degree of decoupling). A 16-channel receiver for the 3.0 T GE scanner is currently under construction in-house.

Table 1: *Relative SNR and* g *as function of gap.*

gap	-0.25	0.25	0.50	1.00
SNR in center	0.98	1.00	1.00	0.99
maximum SNR	8.65	10.34	10.37	13.68
average SNR	3.12	3.51	3.64	3.84
maximum g: rate-2	1.16	1.06	1.04	1.03
rate-3	2.68	1.88	1.64	1.46
rate-4	8.80	7.03	6.11	5.13
average g: rate-2	1.03	1.01	1.01	1.01
rate-3	1.43	1.16	1.14	1.12
rate-4	2.74	2.00	1.86	1.74

Table 2: *Relative SNR and* g *as function of coil diameter.*

coil diameter	1.1	1.2	1.3	1.5
SNR in center	1.00	1.01	1.03	1.04
maximum SNR	10.37	7.90	6.86	5.98
average SNR	3.64	3.08	2.98	2.90
maximum g: rate-2	1.04	1.06	1.07	1.09
rate-3	1.64	1.71	1.92	1.98
rate-4	6.11	4.86	5.33	5.78
average g: rate 2	1.01	1.01	1.01	1.02
rate-3	1.14	1.16	1.18	1.21
rate-4	1.86	1.98	2.16	2.48

Common scan parameters were: spoiled gradient echo; 240×240 mm^2 FOV; 20 ms TE. Parameters specific for the 1.5 T experiments: 30º flip angle; 500 ms TR; 256×256 matrix; 17 4-mm thick slices. Parameters specific for the 3.0 T: 90º flip angle; 2000 ms TR; 256×128 matrix; 12 2-mm thick slices.

Average g-factors were computed for the acquired MRI images. Images were first cropped to tightly fit the head in the slice where the head had the largest diameter. These cropped full-FOV images were subsequently used as sensitivity maps for computing of the SENSE reconstruction matrix and corresponding g-maps for 2- and 3-fold accelerated imaging.

3. RESULTS & DISCUSSION

3.1. Simulations

Results of simulations are summarized in Table 1 and Table 2. SNR is expressed relative to the SNR of a birdcage coil, coil gap as a fraction of element diameter and coil array diameter relative to the object diameter. Table 1 shows that SNR in the center of the sample decreases slightly with increasing coil gap, while maximum and average SNR increase. The g-factors improve when gap increases, leveling off over gap 0.5.

Table 2 shows that SNR and g improve with decreasing coil array diameter. Note that a decrease in

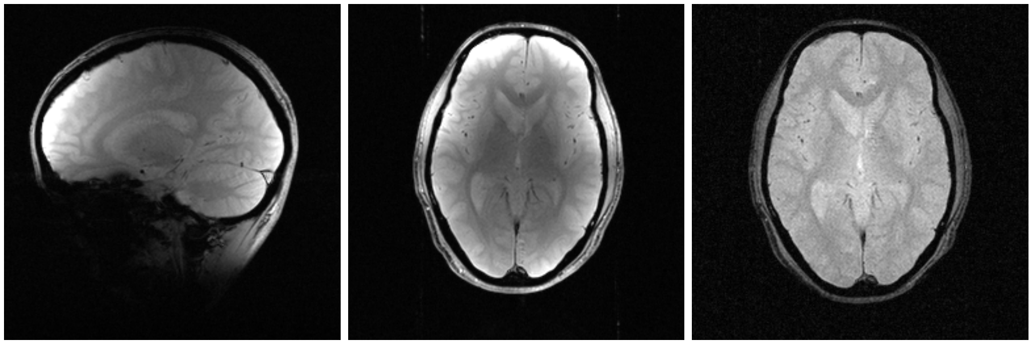

Figure 1: *SNR maps obtained with the 8-element head array (left: sagittal, middle: axial) and the standard birdcage head coil (right: axial) on the 1.5 T Siemens scanner. Coil array images are scaled 0-105, the birdcage image is scaled 0-35.*

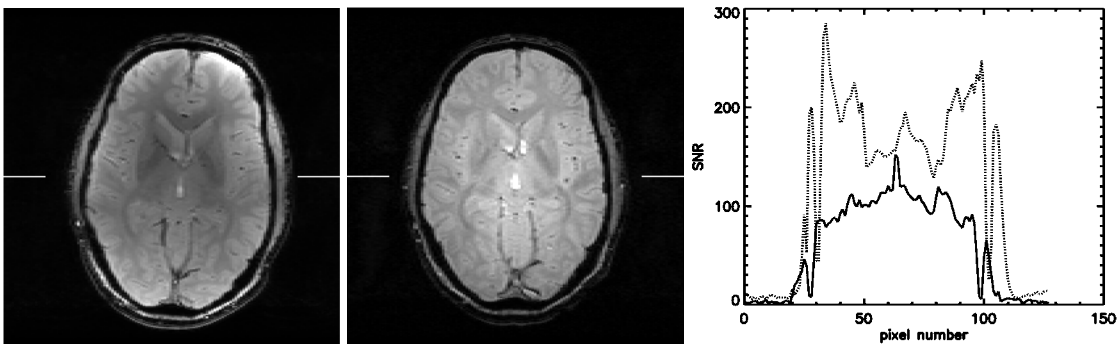

Figure 2: *SNR maps from axial data acquired at 3.0 T on a GE Signa scanner. The 8-element coil data (left) are scaled 0-450, the GE birdcage image (center) is scaled 0-150. As an illustration, profiles from these two images at the location of the tick-marks are plotted on the right (broken line: 8-element array, solid line: birdcage).*

array diameter leads to smaller elements closer to the sample.

3.2. Construction of the coils

Guided by these results, prototype coils were built to tightly fit the head, with an inter-element gap of 0.5 times the element width. Spacing between coil and head is approximately 1 cm. Anatomically shaped formers were constructed, closely fitting the average head size (largest anterior-posterior dimension: 22 cm; largest left-right dimension: 18 cm). The former was divided in two segments, on each of which 4 coil elements were made out of 12.7 mm wide, 50 μm thick copper tape. Posterior elements were made slightly smaller and longer than anterior elements to obtain full brain coverage, which was not accounted for in the simulations. Surface area and inductance of the elements were very similar, 75±10 cm^2 and 250±25 nH, respectively.

3.3. MRI experiments

Figure 1 shows 1.5 T SNR data acquired on a normal volunteer using the prototype array and birdcage coil. Note that scaling of the coil array images is different from

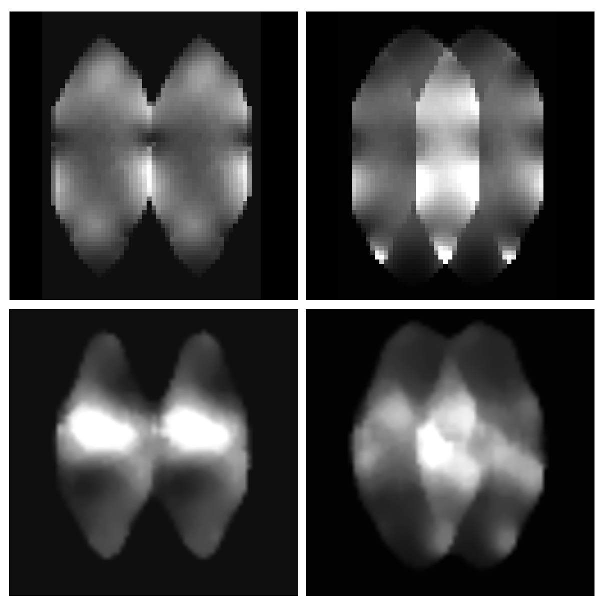

Figure 3: *Comparison of SENSE* g*-factors in the human brain for 8-channel coil arrays at 1.5 T (top) and 3.0 T (bottom). Rate-2* g*-maps (left) were scaled 1.00-1.15, rate-3* g*-maps (right) were scaled 1.0-2.0.*

the birdcage image to preserve image contrast. The array yielded whole brain coverage with an average SNR-gain, computed over all slices, of 2.4 times birdcage SNR (SNR-gain in the center was approximately a factor of 1.4). Figure 2 shows similar data obtained at 3.0 T. SNR-gains were 1.5 (center) and 2.3 (average). Both coils showed a 3-4 fold SNR gain in brain periphery.

Maps of g were computed using Equation 4, where the relative coil sensitivity, estimated from the experimental data, was used for $\mathbf{\Sigma}$. The resulting g-maps are shown in Figure 3. The obtained average g-factors for 1.5 T were 1.03 for rate-2 and 1.22 for rate-3 SENSE, which is somewhat higher than was predicted by simulations (see Table 1 and Table 2). This is probably due to model oversimplification. At 3.0 T, the average g-factors that were obtained were 1.06 (rate-2) and 1.37 (rate-3), respectively.

4. CONCLUSION

With the aid of electro-magnetic field simulations, prototype 1.5 T and 3.0 T 8-element brain coils with high SENSE performance and high SNR throughout the brain were developed using a gapped-element design, combined with pre-amplifier decoupling.

5. REFERENCES

[1] D.K. Sodickson, W.J. Manning, "Simultaneous acquisition of spatial harmonics (SMASH): Fast imaging with radiofrequency coil arrays.", *Magn. Reson. Med.* **38**, pp. 591-603, 1997

[2] K.P. Pruessmann, M. Weiger, M.B. Scheidegger, P. Boesinger, "SENSE: Sensitivity encoding for fast MRI.", *Magn. Reson. Med.* **42**, pp. 952-962, 1999

[3] P.B. Roemer, W.A. Edelstein, C.E. Hayes, S.P. Souza, O.M. Mueller, "The NMR phased array.", *Magn. Reson. Med.* **16**, pp. 192-225, 1990

[4] P.J. Ledden, S. Inati, "Four channel preamplifier decoupled phased array for brain imaging at 1.5 T.", *Proceedings of the 9th scientific meeting and exhibition, ISMRM, Glasgow, UK*, pp. 1117, 2001

[5] J.R. Porter, S.M. Wright, A. Reykowski, "A 16-element phased array head coil.", *Magn. Reson. Med.* **40**, pp. 272-279, 1998

[6] T. Schäffter, P. Börnert, C. Leussler, I.C. Carlsen, D. Leibfritz, "Fast ^{1}H spectroscopic imaging using a multi-element head-coil array.", *Magn. Reson. Med.* **40**, pp. 185-193, 1998

[7] M. Weiger, K.P. Pruessmann, C. Leussler, P. Röschmann, P. Boesinger, "Specific coil design for SENSE: A six-element cardiac array.", *Magn. Reson. Med.* **45**, pp. 495-504, 2001