A 4-channel SENSE optimized array coil for rodent brain imaging at 11.7T

S. J. Dodd[1], H. Merkle[1], P. van Gelderen[1], J. H. Duyn[1], A. P. Koretsky[1]

[1]Laboratory of Functional and Molecular Imaging, National Institutes of Health, Washington, DC, United States

Introduction

Parallel imaging has proven to increase the speed and/or signal-to-noise ratio (SNR) in MRI. Advances in this field have primarily been driven by clinical need, and only recently has more attention been paid to animal studies (1,2). The goal of the present work was to design an array for high resolution functional and anatomical imaging of the whole rodent brain. An example design for a 4-element receiver arrays optimized for Sensitivity Encoded (SENSE) imaging of the whole rodent brain at 11.7 T is presented, inspired by earlier work on arrays for human brain (3).

Methods

Position and shape (approximated by a cylinder for calculation purposes) of the rodent brain was considered for the optimization. Although our imaging system operates at 500 MHz the rodent head is small enough that dielectric effects are minimal, therefore estimates of the magnetic field used Biot-Savart summations and the electric field was estimated using only the time derivative of the vector potential for sample noise calculations. The SNR and geometry factor, g, were calculated according to the method of deZwart et al (3). The example presented here was optimized assuming no presence of coil noise. However, the simulated g-factor maps were calculated using additional noise based on the measured quality factor, Q, of the resultant coils, with and without load. With the Q measurements, the ratio of sample noise to coil noise could be determined. The additional term was included in the sample noise matrix with an adjustment to the diagonal elements. Noise contributions from the preamplifier and receiver circuitry were considered negligible.

Numerical optimization was performed using simulated annealing. Wire positions were allowed to move around a defined cylinder, such that the size of the loop could change. Overlapping of coils was allowed. Symmetry was maintained around the y-axis. Coupling between the coils was assumed to be negligible. The array length was set at 28 mm, with a diameter of 32.5 mm. The object was considered to have a diameter of 90% of the coil array. The parameter to be minimized in the optimization was g/SNR. This parameter was calculated for each pixel in the volume of interest with a mean used in the error function. The aliasing direction was always considered in the x-direction for the purposes of SENSE imaging.

Coils were built according to Reykowski et al (4). Each coil was connected with a half-wavelength cable to a commercial preamplifier (Xiaoming Lou) with an input impedance of 2.8 ohms. More than 15 db decoupling was achieved through the preamplifiers, and this was the only method of decoupling used. Cable traps were employed after the amplification stage. No baluns were used. A home-built saddle coil, 69-mm i.d., was used as the transmitter. All equipment could be placed into a 90-mm id gradient set. An 11.7T magnet (Magnex) equipped with a Bruker console with 4 receiving channels was used for the imaging experiments. A generalized SENSE reconstruction (5) was performed offline using IDL (Research Systems Inc).

Results and Discussion

Figure 1 shows a coil design resulting from the optimized for rate-3 SENSE (in the x-direction), assuming only sample noise. An array built using this design gave (loaded Q)/(unloaded Q) of 21.8/31.0 for the smaller coils and for the larger coils, 23/50. Figures 2 A-C show an image of a rat brain acquired as sum of squares, rate-2 and rate-3 SENSE respectively. Simulated and acquired g-factor maps are shown for the rate-3 image in figures 3A-B. These compare quite well when an additional term based on the Q of each coil was added to the noise matrix. For the ROI we are interested in, the maximum g-factor for rate-3 simulations was 3.5, and the measured value was 3.8. Without the inclusion of coil noise the maximum calculated g-factor was 2.0. For the rate-2 image, the maximum calculated g-factor was 1.26, and measured value was 1.3.

When compared with a commercial 25-mm surface coil (Bruker), often used for rat brain studies, the SNR increase was a factor of 1.5 better at the top of the brain, and 1.3 better at the base of the brain. Simulated SNR increases were 2.5 at the top and 1.7 at the bottom, with coil noise added into the calculations. The primary difference here is most likely related to coil construction, and some changes in positioning compared to the simulation (the top coils are not as close to the sample as possible).

In conclusion, it has been demonstrated that numerical optimization methods may be used successfully in the design of array coils for rodents, leading to gains in SNR/speed even for such small coils. It is essential, however, that coil noise considerations are included in addition to sample noise, unless cooling is employed. The effects of coil noise maybe crudely estimated from the length of each coil loop for the purposes of inclusion in the optimization. This results in larger coils in which coupling becomes more of an issue. A mutual inductance term may then be added to the error function to address some of these concerns.

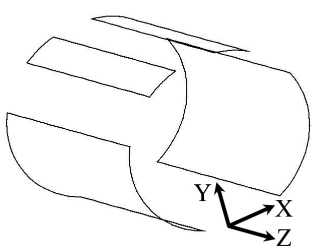

Figure 1. Example of optimized coil. In this case the coil is optimized for rate-3 SENSE (in x-direction). No coil noise is added to produce this result.

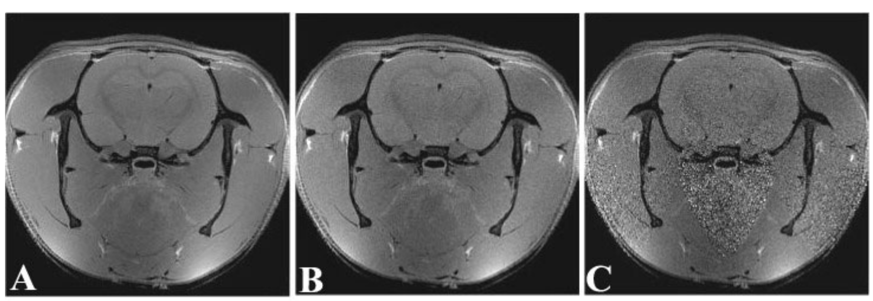

Figure 2. **A** Sum of squares image. **B** Rate-2 SENSE reconstruction. **C** Rate-3 SENSE reconstruction. Image acquisition was spin-echo with 30-mm FOV, 1-mm slice, matrix size of 240 x 240, TR/TE = 4000/12 ms.

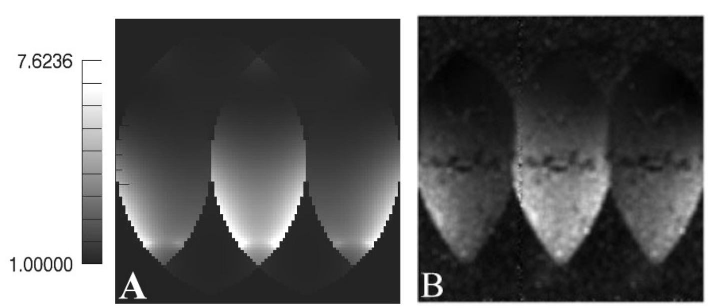

Figure 3. **A** Simulated g-factor map and **B** calculated from rate-3 image in Figure 2C. The maximum g-factor over the rat-brain is 3.5 for simulated data, and 3.8 for the measured data.

References

1. Beck BL et al. Rev. Sci. Instrum. 72(11) :4292 -4294, 2001 2. Zhang X. Et al Proc. 12th ISMRM, Kyoto, 2004, p39
3. DeZwart J. et al MRM 47:1218-1227, 2002 4. Reykowski A. et al MRM 33:848-52, 1995 5. Pruessman K. et al MRM 42:952-962, 1999