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Quadrature RF Coil for In Vivo Brain MRI of a Macaque Monkey in a Stereotaxic Head Frame

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Abstract

We present a quadrature volume coil designed for brain imaging of a macaque monkey fixed in a sphinx position (facing down the bore) within a stereotactic frame at 3 T, where the position of the monkey and presence of the frame preclude use of existing coils. Requirements include the ability to position and remove the coil without disturbing the position of the monkey in the frame. A saddle coil and a solenoid were combined on a modified cylindrical former and connected in quadrature as to produce a homogeneous circularly polarized field throughout the brain. To allow the loops of the saddle coil to encompass the ear posts, partial disassembly and reassembly were facilitated by embedding pin and socket contacts into separate pieces of the former. Coil design included simulation of the electromagnetic fields for the coil containing a 3D model of a monkey's head. The resulting coil produced adequate homogeneity and signal-to-noise ratio throughout the brain.

Keywords

monkey; MRI; coil; brain; stereotaxy

I. INTRODUCTION

Accuracy of targeting in the brain is critical for the success of cell transplantation and other surgical procedures. When such procedures are to be performed in the brain with minimally invasive methods, it is necessary to have exact knowledge of the position of various nuclei and tracts within the brain and their orientation relative to known spatial references during both the acquisition of 3D whole-head images and performance of the procedure. This can be accomplished with use of an MRI-compatible stereotaxic positioning device (1–3). Use of such a device in the MRI environment, however, can prohibit the use of commonly used coils due to their size and the unusual posture of the subject. In our case, the coil must accommodate the head of a Rhesus or Cynomolgus Macaque monkey in a sphinx position (facing in the direction of B_0 along the bore of the magnet), allowing for ear bars, eye posts, and bite bar to connect the frame to the monkey's head, as shown in Fig. 1. The coil must

also be capable of being positioned and removed without disturbing the monkey or any parts of the frame and provide homogeneous coverage and adequate signal-to-noise ratio (SNR) throughout the monkey's brain. These requirements precluded use of any existing coils or conventional designs. Here, we describe a solution achieving adequate homogeneity and sensitivity over the monkey brain in this situation.

II. METHODS

For an MR system having only one transmit and one receive channel (3 T Bruker Medspec S300), we conceived a design fundamentally consisting of two coils combined in quadrature where the first coil is a saddle coil oriented to produce a horizontal (left-right) field and the second is a solenoid oriented to produce a vertical field through the monkey head.

Simulation

The finite-difference time-domain (FDTD) method (4) was used to evaluate the general coil design. A monkey head model was incorporated in the design to provide reasonably accurate geometric references and loading. The numerical model of the coil loaded with the monkey head is depicted in Fig. 2. The data set for the monkey head model is in the public domain and can be downloaded freely (5). Because the available monkey model appears to be based on a smaller green monkey (*Chlorocebus sabaeus*), it was scaled so the head size was roughly equivalent to that of a Rhesus macaque monkey and remeshed. The resulting model had a mesh resolution of $1.25 \times 1.25 \times 1.25 \text{ mm}^3$. The tissue electrical properties were assigned values from the literature (6) for 125.44 MHz, the frequency of the MRI system. In the first simulation, current sources were placed across gaps spaced less than a quarter wavelength around each coil (7). To assure uniform current distribution in the fairly long paths, we included six sources in the saddle coil and seven sources in the solenoid coil. The first simulation was run with a sinusoidal waveform at 125.44 MHz. Using the impedances of the sources calculated during the first simulation, capacitances across the gaps in each coil required for resonance at the desired frequency were determined (8). The sources in each coil were then replaced with capacitors. Tuning was verified and *S*-parameters were evaluated with a pair of simulations in which each coil was excited with a Gaussian-derivative impulse excitation and Fourier transforms of selected time-domain responses were evaluated. Field homogeneity was then evaluated in results of a subsequent calculation in which the tuned coils were excited simultaneously with sinusoidal voltage sources having equal amplitude and a 90° phase delay between them. All field simulations were accomplished using commercially available software, XFDTD 6.5 (Remcom, PA). To further evaluate design of practical experimental tuning and matching circuits, calculated *S*-parameter values over a range of 10 MHz were then loaded into an RF circuit simulator (Advanced Design System, Agilent Technologies, Santa Clara, CA). Using the circuit simulator, it was possible to ensure that a simple "L" type tuning and matching network (consisting of a variable capacitor for matching in line with the center coaxial conductor before a variable capacitor for tuning between the center and ground conductors and in parallel with the coil circuit) on each channel would provide adequate range for both tuning and matching.

Coil Construction

A coil based on the design described above was constructed with modifications to accommodate macaque monkeys in the sphinx position. Notably, to allow the neck of the monkey to pass through the back of the cylindrical acrylic former, the path of the lowest turn of the solenoid was modified, bringing a portion of the lowest turn in close proximity to the middle turn. To get adequate coverage of the entire monkey brain it would be necessary that the loops of the saddle coil encompass the ear bars. This combined with the requirement that

the coil be removable without disturbing the monkey or the frame resulted in a design allowing easy disassembly of the coil, especially removal of the lower conductors in the saddle coil. This was accomplished with use of socket contacts (Tyco Electronics, 60619-1) and pin contacts (Tyco Electronics, 60620-1) embedded into the acrylic former, as shown in Fig. 3. The path of the coils was laid with copper strips (5 mm wide, 1 mm width, and 0.3-mm-thick) taped to the acrylic former with polyamide dielectric (Shercon Inc., Santa Fe Springs, CA) tape selected for its ability to withstand high temperatures during soldering. To minimize capacitive coupling between the coils and between different turns of either coil, the copper strips were shaped so that where they crossed each other one would pass over the other with ~1 cm between. A tuning and matching network for each channel and BNC-type coaxial connectors were mounted within and on a plastic electrical connection box shielded on the inside with copper tape. To improve stability of coil tuning and matching, a cable trap was added to the solenoid channel between the BNC connector and the matching capacitor. The cable trap was constructed by wrapping a section of coaxial cable into a tight helical shape with 3.5 turns and a 1.5 cm inner diameter, then exposing a small region of the outer conductor at each end of the helix and attaching a variable capacitor and tuning the resulting resonance to the Larmor frequency of our system, ~125.44 MHz. A copper shield for the coil was created by laying segments of wide copper tape (1.5 inch-wide, 0.003-inch-thick, Chometrics, NH) interleaved with polyester insulating tape (Epsi LLC, Frankville, WI) onto a modified cylindrical former to create a high-pass structure so that the low-frequency gradient coils would not induce eddy currents but RF fields would. Photographs of the coil and shield are shown in Fig. 3.

MRI Experiments

All animal experiments were performed with approval of our Institutional Animal Care and Use Committee (IACUC) and with on-site veterinary care. Briefly, prior to MRI and stereotaxic frame application, after the animal is sedated with glycopyrolate and tiletamine/zolazepam, a catheter was placed into the cephalic vein in the leg of the monkey to administer intravenous anesthesia. In order to induce anesthesia, undiluted propofol is administered to the monkey (9). Vocal cords were sprayed with xylocaine to prevent involuntary muscular contraction of the laryngeal cords and intubated. The animal was kept warm with a combination of clothing, bubble wrap, and pads filled with of warm water. Finally, the monkey was placed in the stereotaxic head frame and fitted with the designed coil. Respiration was monitored throughout the experiment. For demonstration of coil performance and location of stereotaxic frame during imaging, simple proton density-weighted gradient echo images (TE/TR = 6/875ms, FOV = 25×25 cm², matrix size = 256×256 , 2-mm slice thickness, and 50° flip angle) were acquired on coronal and axial planes containing the oil-filled eye and ear posts, as well as on the mid-sagittal plane through the monkey brain. SNR was evaluated in each image by measuring the mean signal in an ovalar region including the approximate center of the brain on the image and approaching (but not including) distinct boundaries at the edge of the brain, and then dividing by the standard deviation of the image intensity in a region of no signal or apparent artifact in a rectangular region at the periphery of each image. This approach is similar to the fourth method in a widely cited document (10). Although this approach is often less preferable to other methods, time constraints while the monkey was under anesthesia precluded the acquisition of additional images required for these methods.

III. RESULTS

The distribution of the modulus of the simulated B_1 field throughout the model of the monkey head is shown in Fig. 4. Quantitative evaluation of the field homogeneity along lines through the center of the brain in the left-right, anterior-posterior, and inferior-superior directions (as defined by monkey anatomy) showed relative standard deviations (standard

deviation divided by center value times 100%) of 0.51% in the left-right direction, 5.25% in the inferior-superior direction, and 3.41% in the anterior-posterior direction.

Plots of the simulated and experimentally measured S -parameters are shown in Fig. 5. Both indicate that the simple L-type tuning and matching circuit provides adequate range to tune and match each channel when loaded with the monkey head. At 125.44 MHz, input return losses for the saddle coil and solenoid coil (S_{11} and S_{22} , respectively) in both simulation and experiment are all near -20 dB. Coupling (S_{21}) is less than -10 dB in each case. S -parameter plots between experiment and simulation are similar but not exactly the same. Differences are easily attributable to differences in coil geometry, sample geometry, capacitor tolerances, and resulting differences in capacitor values needed to achieve tuning and matching.

Experimental images through the monkey brain are shown in Fig. 6. External references consisting of oil-filled 2-mm channels in the ear and eye posts are visible on the sagittal and coronal images. Good image homogeneity throughout the cerebrum is seen. SNR in the brain was measured to be ~ 59.8 , 47.2 , and 68.6 on the coronal, axial, and sagittal images, respectively.

IV. DISCUSSION

To allow for MRI of the brain of a monkey in a stereotaxic frame with a number of design constraints, we devised and constructed an array of two separate coils connected in quadrature that met all practical and imaging requirements. Simulations of the fundamental design indicated good homogeneity over the monkey brain and deviation from the fundamental design to allow for specific geometric considerations did not have notable adverse effect on homogeneity or SNR in experiment. Although decoupling between the two channels relied only on coil geometry and was adequate for our purposes, it could be further improved with dedicated decoupling methods.

A quadrature volume coil fitted to the anatomy of interest is a good design for systems having a single transmit and single receive channel and not having a body transmit coil available. As this describes the system for imaging of large animals at our site, we pursued the design described here. While this should, in principle, be capable of producing better homogeneity and SNR and use less RF power than a linear volume coil in reception with a system body coil used in transmission (11), if transmit power is not a concern use of a system body transmit coil could show better homogeneity of the excitation field. In cases where multiple receive channels are available and transmit power is not a concern, excitation with the system body coil and reception with a dedicated array of receive coils (12) should be capable of producing superior SNR than the coil used here, at least in the periphery of the head.

The coil presented here will facilitate research at our site, and its simple and unusual design may be used to overcome challenges in future research in other areas as well. It may be interesting to note, for example, that the original basic design could meet many of the requirements for MR of the breast, wrist with the forearm oriented orthogonal to the main field, or the human brain with the subject in the sphinx position.

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Figure 1.
Rhesus Macaque monkey positioned in stereotaxic frame.

Figure 2.

Numerical model of coil design without (a) and with (b) models of monkey head, stereotaxic frame, and RF shield. In (a), F indicates feed location, C indicates capacitor location, subscript 1 and red color indicate saddle coil, subscript 2 and yellow color indicate solenoid. In (c) and (d), views of model from front and side are shown. Note that numerical model has no body and is scaled from a smaller green monkey to roughly match brain size of a rhesus monkey. The coil was designed to be large enough to accommodate a larger cynomolgous monkey, as well as a rhesus.

Figure 3. Experimental coil and shield geometry. Views of (a) back and (b) front of coil as well as (c) detail of connectors at detachable portion and (d) the front of the shield. “T&M” indicates location of tuning and matching circuit.

Figure 4. Simulated B_1 field magnitude distribution in monkey head. Color scale is normalized relative to value near center of monkey's brain.

Figure 5.
Simulated (left) and experimentally measured (right) S -parameters for the coil.

Figure 6. Gradient echo images on coronal, axial, and sagittal (left to right) images through a monkey's head showing the coverage and homogeneity achievable with experimental coil.