

Design and Implementation of a Quadrature RF Volume Coil for *In-Vivo* MR Brain Imaging of Rhesus Macaque Monkey in a Stereotaxic Head Frame

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Abstract -- We describe the design and implementation of a unique coil for *in-vivo* rhesus macaque brain imaging in a stereotaxic device. The RF volume coil consists of a 2 turn solenoid and a saddle coil configured and fed in quadrature. Finite difference time domain method was used to design the coil. Images acquired show excellent homogeneity and SNR throughout the monkey's brain.

Index Terms -- Brain imaging, FDTD simulation, magnetic resonance imaging, quadrature coil.

I. INTRODUCTION

In magnetic resonance imaging (MRI), the magnetic fields of interest are identified by B_0 , B_1 and B_{1+} . B_0 is the main magnetic field of the MRI scanner used to align the spins of the hydrogen nuclei [1]. B_1 is the field introduced by the RF imaging coil to excite the aligned spins of the hydrogen nuclei. If the coil is driven in quadrature (orthogonal channels are out of phase by 90°) the channels produce a circularly polarized magnetic field, referred to as B_{1+} , to excite the aligned spins of the hydrogen nuclei.

Radio frequency (RF) coils used in MRI are resonators that do not include any magnetic material in their design. RF coils are tuned to Larmor frequency, at which high signal amplitudes are applied to the coils [1]. RF coils commonly used in MRI are transmit, receive, and transmit-receive coils. Quality of MR images depends on the signal-to-noise ratio (SNR), which is affected by the proximity of the coil to the subject and signal intensity at resonance in both transmit and receive modes. In transmit mode, the coil is a transmitter and produces a homogeneous B_1 or B_{1+} field within the region of interest (ROI). In receive mode, the coil acts like a receiving antenna to detect the signal from the subject's head as the disturbed spins return to their equilibrium [1].

The accuracy of targeting brain tissue is critical for the success of cell transplantation and other procedures. It is essential to perform brain procedures with minimal invasiveness. Furthermore, exact knowledge of the position and orientation of the head and brain is required for 3D whole head image acquisition prior to surgery [2]. A stereotaxic device [2] (as seen in Fig. 1) enables exact positioning of the monkey head and brain. Utilizing such a device prevents the use of conventional coils due to its size and the animal posture.

In order to facilitate the use of a stereotaxic frame, the coil must accommodate the head of a Rhesus Macaque monkey in a sphinx position, facing the direction of the B_0 field. Additionally, the coil must allow for ear bars (Fig. 1), eye posts, and a bite bar that connect the frame to the monkey's head (Fig. 1). The coil must also be capable of being positioned and removed without disturbing the monkey or any parts of the frame. Furthermore, it is necessary that there is a homogeneous field distribution and an adequate SNR level throughout the monkey's brain. These requirements prevent the use of existing coils or conventional designs.

Previously, MRI of rhesus macaque monkey brain was performed at the Department of Radiology at Penn State College of Medicine using a single channel birdcage coil in our 3T system. This birdcage coil housed the monkey in the stereotaxic head frame. MR images of the rhesus macaque monkey brain using the birdcage coil showed poor SNR.

In this paper, we present the design of a new transmit-receive quadrature RF coil for *in-vivo* MRI of Rhesus Macaque monkey brain in a stereotaxic head frame. In Section II, we discuss the design issues of the quadrature coil and introduce its topology. The simulation method for evaluating the field distributions for the introduced design is also presented in Section II. Section III presents the results of this study including the field simulation, measured scattering parameters, and MR images.

II. METHOD

Finite difference time domain (FDTD) method [3] was used for coil simulation. The FDTD method is quite popular in biological applications, due to its simplicity of implementation for inhomogeneous media and wide range of frequencies that could be evaluated with one simulation.

The design of our RF coil consisted of a 2-turn solenoid oriented to produce a vertical B field and another coil consisted of a saddle coil oriented to produce a horizontal B field through the monkey's head (Fig. 2). The coil was designed using XFDTD, a commercial FDTD solver (Remcom Inc., State College, PA), for attaining optimal homogeneity within the ROI. The coil model had a radius of 7 cm and height of 14 cm.

The RF coil model shown in Fig. 2(a) was implemented with a mesh resolution of $1.25 \times 1.25 \times 1.25 \text{ mm}^3$. First, for field homogeneity study, the simulation was done with a sinusoidal source at a single frequency of 125.44 MHz, i.e., the Larmor frequency of our 3T system. Fifteen cycles of the sinusoidal input was used to reach proper convergence. The problem size was $210 \times 195 \times 214 \text{ mm}^3$. A low-loss dielectric phantom was assumed inside of the coil to approximate the monkey's head. The dielectric parameters of the phantom modeled were $\sigma = 0.785 \text{ S/m}$ and $\epsilon_r = 43.8$, i.e., average complex permittivity values for head tissues. The modeled phantom was spherical with a radius of 4 cm. The time step was 2.407 ps and the total simulation time was 0.12 μs . The actual PC run time, including the post processing time, was 2 hours. For field homogeneity, it was necessary to have 4 current sources in the saddle coil and 7 current sources in the solenoid coil for generating homogenous B_1 field [4]. Each of the source port impedances was then used to calculate the capacitor (marked as C in Fig. 2(a)) needed for resonance as explained in [5]. The two orthogonal (saddle and solenoid) coils were each eventually fed at one input location (marked as F in Fig. 2(a)), and the sources at other locations were replaced by the proper capacitor (C) [5].

In the second simulation, the model was excited by a Gaussian derivative pulse with a pulse width of 77 ps, to get frequency response of the coil. A Gaussian derivative pulse was chosen in order to prevent DC offsets, which can be problematic for the simulation in reaching its steady state conditions [6]. At the center of the coil, the time domain response of the H field was recorded. XFDTD version 3.3, does not directly provide the B field for a model excited with the Gaussian pulse. Therefore, the H field response was obtained in this work. Matlab was then used to plot the H field versus frequency at the center of the coil, using FFT (Fig. 5).

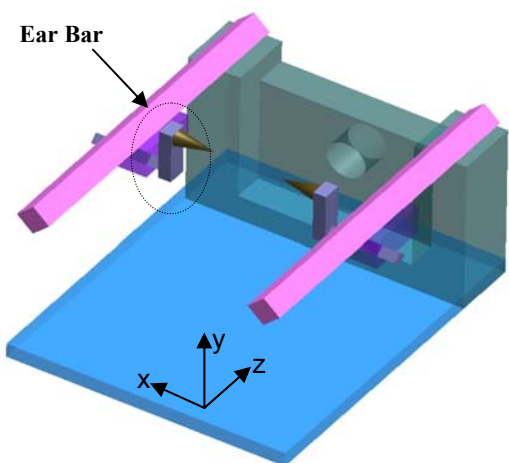


Fig. 1. Computer model of stereotaxic frame showing ear bars. Eye posts and bite bar not shown.

During construction, some deviations from this basic geometry were necessary to accommodate the monkey's neck, ear bars, eye posts, and bite bar. The fabricated quadrature RF coil is shown in Fig. 2(b). The RF coil is installed around the monkey's head rather than surrounding the stereotaxic frame and monkey as seen in Fig. 3. The coil was fabricated by placing copper tapes on the surface of a non-magnetic material with low RF absorption, i.e., plexiglass. The coil was then tuned and matched at the Larmor frequency, while measuring the scattering parameters, for both loaded and unloaded cases, using a network analyzer. The actual phantom was a saline sphere approximately the size of the monkey's head (radius = 4 cm). Tuning circuit consisted of several fixed valued capacitors placed around the body of the coil. The Matching circuit consisted of an L-section impedance matching circuit of one variable and one fixed value capacitor for each feed.

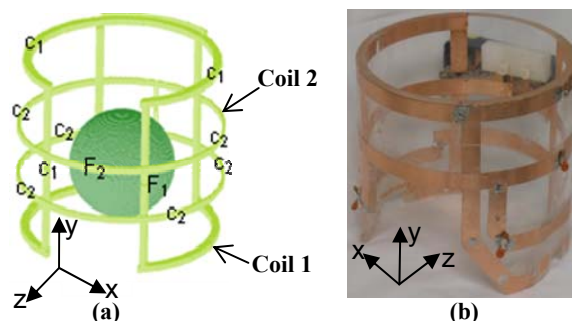


Fig. 2. a) Basic structure for the coil is a 2-coil array consisting of a saddle coil (Coil 1) and a 2-turn solenoid (Coil 2) fed in quadrature. Note that feed (F) and capacitors (C) locations are shown. Figure. b) The fabricated coil with allowances for monkey's neck, ear posts, eye posts, and bite bar.

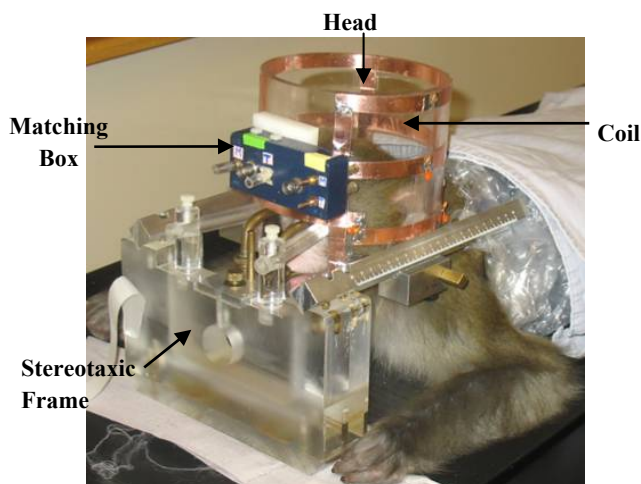


Fig. 3. Monkey contained in stereotaxic frame and head coil.

III. RESULTS

XFDTD plot of the B_1 field is shown in Fig. 4. It illustrates homogenous field distribution within the ROI. The calculated H fields, associated with B_1 , showed a natural resonance frequency of 115 MHz (Fig. 5). This was corrected in actual design by tuning to Larmor frequency.

The fabricated RF coil was initially tuned with no load, and its resonance was set higher than the Larmor frequency. Fig. 6 shows the measured unloaded (Fig. 6(a)) and loaded (Fig. 6(b)) S_{11} and S_{22} of the coil. S_{11} and S_{22} are the return losses of the saddle and solenoid coils, respectively. The addition of the phantom, i.e., saline solution, loaded the coil and lowered the resonant frequency. Also, it is seen that the loaded S_{11} and S_{22} are good at the Larmor frequency.

S_{21} was also measured and illustrated a lowest coupling value of -27 dB, between the orthogonal channels, at 128 MHz, slightly higher than the Larmor frequency of 125.44 MHz.

A Rhesus Macaques monkey was anesthetized for the MRI. The monkey was supplied with both anesthesia and oxygen through the MRI process. The animal's breathing was also monitored during the scan.

MR images of the monkey's brain are shown in Fig. 7. Thin channels of vitamin E oil in the ear bars and eye posts (designed to ensure exact alignment of imaging planes to the frame) are seen in the axial image (Fig. 7(c)).

SNR is evaluated qualitatively by observing the MR images. The MR images show excellent SNR throughout the entire brain.

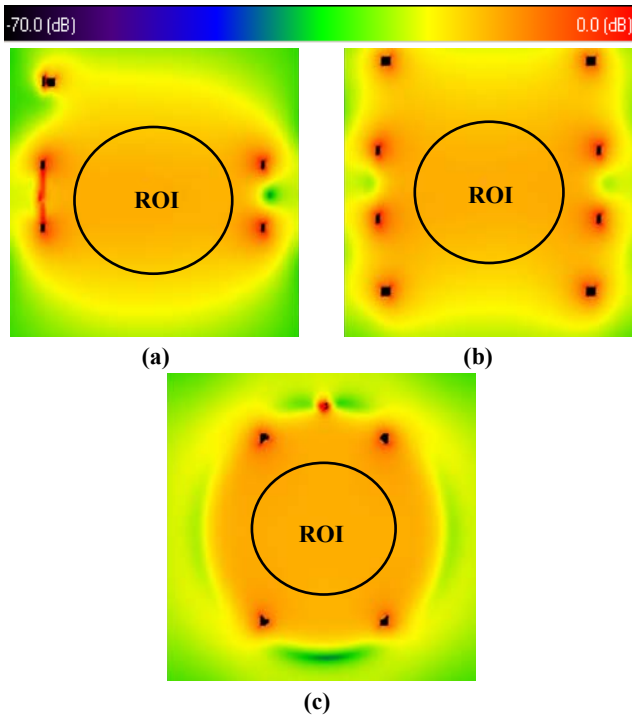


Fig. 4. Simulated B_1 distributions on three orthogonal planes; a) sagittal (yz), b) coronal (xy), and c) axial (xz).

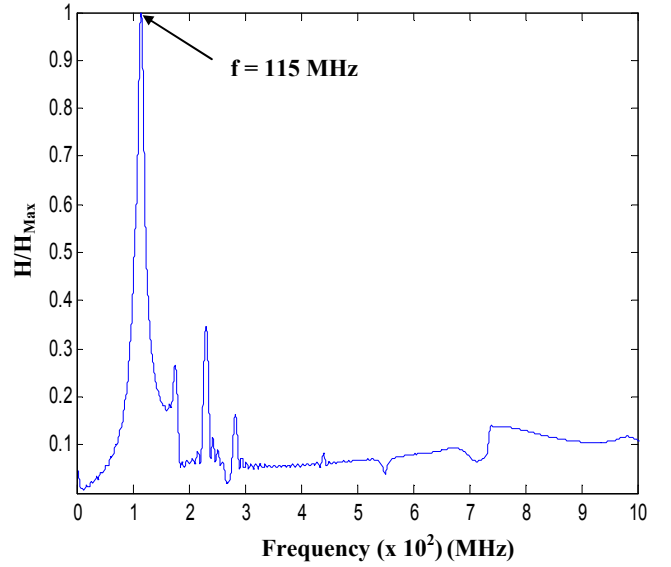


Fig. 5. Normalized H field associated with B_1 versus frequency.

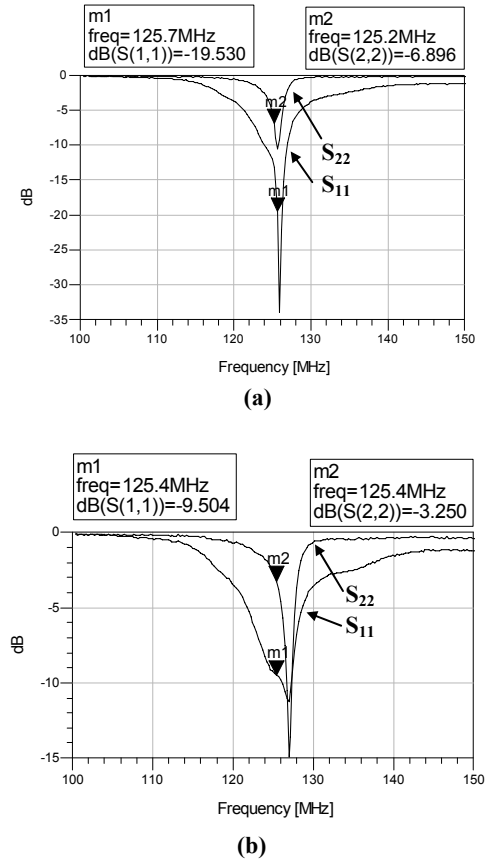


Fig. 6. a) S_{11} and S_{22} unloaded and b) S_{11} and S_{22} loaded with phantom

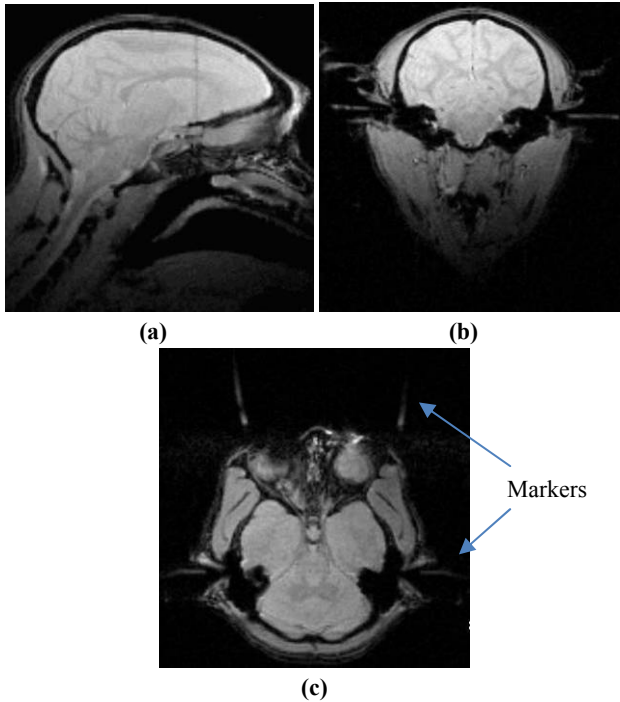


Fig. 7. Monkey brain MR images; a) sagittal (yz), b) coronal (xy), and c) axial (xz) center slices respectively. Note that ear and eye markers are present in the axial slice.

IV. CONCLUSION

To allow for MRI of the brain of a monkey in a stereotaxic frame with a number of design constraints, we devised a fairly simple and unique 2-coil array arranged and fed in quadrature that met all design requirements. Deviation from the basic design to allow for our specific geometric considerations did not have any notable adverse affect on the homogeneity or SNR.

Excellent MR images with high SNR obtained in Fig. 7 illustrate the proper design of the coil. However, our measurements illustrated that the two channels are best orthogonal at 128 MHz, slightly greater than the Larmor frequency of 125.44 MHz for our 3T system. Future activity involves quantifying the SNR, slight modification of the coil to achieve the best orthogonal configuration, evaluated in terms of S_{21} , at 125.44 MHz, and more precise tuning of the loaded coil.

This coil will facilitate important research at the Center for NMR Research at the Penn State College of Medicine, and its simple and unconventional design may be used to overcome challenges in similar future research.

ACKNOWLEDGEMENT

This work was supported in part by the NIH through R01 EB000454 and by the Pennsylvania Department of Health.

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