HIGH-THROUGHPUT DIFFUSION-TENSOR-IMAGING OF MOUSE BRAINS USING A FOUR-COIL SYSTEM

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In mouse models of neurological diseases, diffusion-tensor imaging (DTI) can reveal subtle injuries to white matter. Statistical methods can detect brain pathologies with remarkable sensitivity, by comparing a sizable number of mouse brains to population-based atlases[1-3]. The total scan time required can be prohibitive. Concurrent imaging of multiple excised mouse brains at 7T has been presented[4-5] before. When the number of specimen imaged in parallel is large, the benefit of decreased imaging time surpasses the detriment of potentially lower SNR. In small-bore magnets dedicated to DTI, the number of specimen that can be imaged in parallel is limited. A multiple-coil system must provide similar SNR as during the imaging of a single mouse brain in order to increase throughput. We present a parallel acquisition, four-coil system designed to image post mortem mouse brains in an 11.5cm gradient bore diameter, 7T scanner without significant SNR penalty.

In a multiple-coil configuration, the coils must be located close to the magnet isocenter to mitigate magnetic field inhomogeneity, gradient warping, and in the case of diffusion tensor imaging, to limit diffusion tensor errors induced by Maxwell gradient cross terms[6]. Also, each coil must be shielded to prevent cross-talk. Because imaging takes place in the coil-dominated noise regime[7], the high quality factor and high SNR performance of each coil will be lost if it couples too closely to an electrically lossy shielding structure. Therefore, in the design of a multiple-coil system, a trade-off exists between the SNR benefit of a greater separation distance between the coils and the detriment of their location further away from the magnet isocenter.

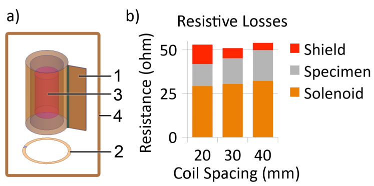

Fig. 1: a) Finite-element model of the coil[1] (24mm long, ∅14mm), coupling loop[2], mouse brain[3], and shield[4]. b) After matching to 50Ω, *transformed* losses show that the 40mm shield causes minimal losses; a larger transformation increases the relative contribution of the specimen.

Fig. 2: Four shielded coil system, with one open shield.

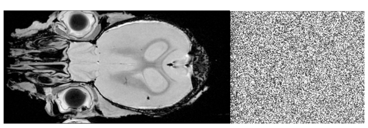

Fig. 4: One coil was loaded with a brain, and the others left empty. No cross-talk can be seen in the images from the empty coil (only one shown, partially cropped).

Numerical Simulations: First, we assessed the amount of electrical noise emitted by the radiofrequency shield in close proximity to our coil. Using a finite-element, three-dimensional electromagnetic simulation software (HFSS 13, Ansys, Canonsburg, PA), we modeled and quantified all the electrical losses in three different coil configurations: where the four coils are 20, 30 and 40mm apart from each other (Fig. 1). In each configuration, the surrounding shield was as distant to the coil as possible to limit its noise contribution.

Validation: When the coil is matched to the characteristic impedance of the radiofrequency chain (50Ω), the 20mm-wide shield dissipates energy similarly to a 11-Ω resistor, or over a third of the energy lost in the coil alone (29Ω). By increasing the distance between coils to 40mm and using a wider shield, the losses within the shield can be lowered by 62% to 4Ω. In that configuration, the losses in the coil were equivalent to 32Ω, and those in the mouse brain model to 18Ω. The 40mm-configuration was selected, its mechanical fixture was 3D-printed (QuickParts, 3D Systems, Rock Hill, SC), the shields and coils were assembled (Fig. 2). Experimentally, quality factor measurements validated the numerical model (Table 1).

SNR Performance: Because the MR acquisition takes place in the coil-dominated noise regime, we designed a copper sulfate phantom loading the coil slightly, in the exact same fashion as a mouse brain. We adjusted the volume of water to obtain the same frequency shift as a mouse brain (0.04ml), and controlled the concentration of copper sulfate to obtain the same electrical load as a mouse brain (100mM $CuSO_4$). For the mouse brain and the phantom respectively, the frequency shifts were 0.85 MHz and 0.80 MHz, while the loaded quality factor in the same coil were 330 and 360, indicating that our phantom loads the coil similarly to a mouse brain.

In the four shielded coil configuration, an SNR penalty occurs because of shimming over a larger volume. We optimized shimming over an 8-cm sphere located around the magnet isocenter. We acquired images of our phantom in the single unshielded coil located at four different locations: on magnet isocenter, 20-mm off center along z (position "B"), and 20-mm along both z and y (position "C"). The SNR penalty, respectively, was 0%, 17% and 22%. The four-coils are located within a 6-cm sphere around magnet isocenter; after re-shimming on that smaller volume, the SNR penalty incurred by the four-coil system was reduced to 19% (Fig. 3).

Quality Factor	
Unloaded: 540	Loaded: 340

Modelled Resistance (Ω)	
Coil: 32	Specimen: 18
Shield: 4	

Table 1: Experimentally, the ratio of unloaded over loaded quality factor (1.58) is in good agreement with the ratio of resistances calculated by the finite-element model (1.44).

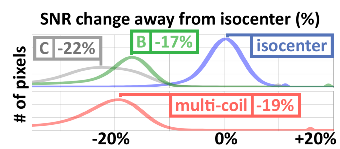

Fig. 3: histograms from a homogeneous phantom, normalized by the mean SNR at isocenter, showing the SNR penalty due to shimming (see main text for position offsets).

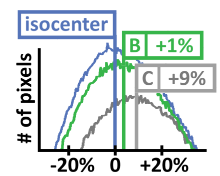

Fig. 5: Fractional anisotropy measurment bias in the anterior commissure of a mouse brain due to off-center imaging (see main text for position offsets)

The efficiency of the shield at limiting cross-talk was quantified by acquiring four image sets simultaneously from four shielded coils: one coil loaded with a mouse brain, and the other coils empty. No ghosting was apparent in any of the three noise images (Fig. 4). Finally, a mouse brain was used to assess the bias on fractional anisotropy measurements due to off-center imaging. A full DTI dataset (baseline + 6 b values) was acquired at the same locations described above. Histograms of the fractional anisotropy in the anterior commissure showed that the mean fractional anisotropy was overestimated by 9% at the location of the four coils (Fig. 5).

Discussion and Conclusion: During the concurrent imaging of excised mouse brains by four coils in a small bore magnet, we limited the SNR degradation due to the radiofrequency shield and shimming to 19%, as compared to a single, unshielded coil on isocenter. The same SNR per brain can be acquired by averaging 50% more; the four shielded coils provide a decrease in imaging time per brain specimen of 63%. If some SNR penalty is acceptable, the full decrease of 75% in imaging time per specimen can be achieved. A further improvement of the four coils is possible by bringing each coil slightly closer to the magnet isocenter. Each coil will be closer to the radiofrequency shield and incur more SNR loss due to shielding; however the SNR penalty due to shimming over a smaller volume will be decreased.

References: **[1]** Dorr AE, ..., Henkelman RM. (2008), Neuroimage, 42(1), 60–69. **[2]** Bock NA, ..., Henkelman, RM. (2006), Journal of Neuroscience, 26(17), 4455–4459. **[3]** Badea A, ..., Johnson, GA. (2012), Neuroimage, 63(3), 1633–1645. **[4]** Nieman BJ, ..., Henkelman, RM. (2007), Nmr in Biomedicine, 20(3), 291–303. **[5]** Dazai J, ..., Henkelman RM. (2011), Journal of visualized experiments, 48, 2497. **[6]** Bammer R, ..., Moseley ME. (2003), Magnetic Resonance in Medicine 50(3), 560–569. **[7]** Glover P, Mansfield P. (2002), Reports on Progress in Physics, 65(10), 1489–1511.