Wire-Wound B_1 Coils for Low Frequency MRI

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INTRODUCTION: Very low field lung imaging using hyperpolarized noble gases [1,2] represents a new frontier in MRI. With operating fields of order 10 mT and resonant frequencies well below 1 MHz, much of the RF coil technology that has been developed for conventional MRI is either not applicable or not practical for use in this new regime. We have begun to research and develop simple wire-wound coils designed solely for the production of B_1 fields at low frequency. Here we outline a number of design considerations and a summary of our experience with the implementation of such coils. Note that the separation of transmit and receive coil functions may let one take advantage of novel approaches to signal detection (*eg.* cryogenic coils, SQUIDS).

SIMPLE DESIGN CONSIDERATIONS: The magnetic field inside a uniformly magnetized object can be described in terms of an equivalent surface current $\boldsymbol{K} = \boldsymbol{M} \times \boldsymbol{n}$, where $\boldsymbol{M}$ is the magnetization and $\boldsymbol{n}$ is a unit vector normal to the surface [3]. For a very long rod, of elliptical cross-section, uniformly magnetized perpendicular to its length (Figure 1) the equivalent surface current is

$$K(\theta) = K_o \sin\phi = K_o \sin(\tan^{-1}(a^2/b^2 \tan\theta)) \qquad (1)$$

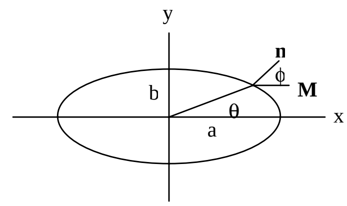

Figure 1: Cross-section - ellipse with axes *a* and *b*, magnetization **M** along *x*.

For a circular cross-section $a = b$, $\boldsymbol{K}(\theta) = K_o \sin(\theta)\, \boldsymbol{k}$, and the field inside the cylinder is $\boldsymbol{B} = \mu_o K_o / 2\, \boldsymbol{i}$. This is the idealized field profile of the birdcage resonator [4], which operates using parallel rungs of current appropriately phased to approximate the $\sin(\theta)$ distribution at any instant in time. For low field MRI, however, the use of such coils may not be practical since it becomes challenging to tune small inductance structures to low enough frequencies [5]. Fortunately, one can resort to the use of simple wire-wound coils to generate very homogeneous oscillating magnetic fields. In this case, the coil is composed of an appropriate angular distribution of single-phase current paths that approximate $\boldsymbol{K}(\theta)$. The RF field produced by such a coil is linearly polarized [6]. In practice, coil lengths are short and current paths must be continuous, and as a result Equation (1) is only useful as a conceptual guide. The coils described below were designed by considering contributions from all current paths: the axial segments along the length of the coil and the necessary return paths at the ends of the coil. The well-known equation for the magnetic field produced by a straight segment of wire [3] is used for this purpose:

$$\mathbf{B} = \mu_o I(\sin\alpha_2 - \sin\alpha_1)/4\pi R\hat{\phi} \qquad (2)$$

DESIGN, CONSTRUCTION AND PERFORMANCE OF A CYLINDRICAL COIL: A length of 90 cm and a diameter of 54 cm were chosen as suitable dimensions for a cylindrical RF coil; a practical number of axial current paths was considered to be 20. A simple Fortran program was employed to find the angular spacing of the axial wires that gives the most homogeneous magnetic field over the volume bounded by the region $\Delta x = \Delta z = \pm 15$ cm and $\Delta y = \pm 10$ cm relative to the midpoint of the coil (*i.e.* the volume to be occupied by the lungs). The circular return paths were approximated by 40 straight-line current segments. The results of the computation gave $\theta_i = \{\pm 26.6^o, \pm 46.8^o, \pm 59.7^o, \pm 76.9^o, \pm 80.1^o\}$ (modulo 180^o) for the angular spacing, and a homogeneity of ± 1.5% over the volume of interest. The coil was wound with a single contiguous wire (1mm O.D. copper) on a fibreglass former, 120 cm long and 1 cm thick, designed as an insert for the 55 cm bore of our low field MR imager. The axial sections were glued (Loctite® 495) into grooves in the former. The winding pattern of the coil is shown in Figure 2, along with a plot of the field homogeneity as measured in 'free space.' These measurements were made by driving the RF coil at 100 kHz and mapping the field with a small pick-up coil; results showed the homogeneity to be close to that which was expected, albeit with a slight asymmetry. Inside the bore of the imager, however, the RF field map was significantly distorted. Similar low frequency RF coils have been used previously inside a conventional MR imager with negligible degradation of the field homogeneity [7]. The most significant difference between the environments in which these two coils were operated was the quantity and proximity of surrounding conducting surfaces in which currents could be induced. Our low field imager is extremely compact; the main magnet windings, 3 gradient coils, 14 shim coils, and metallic tubing for water cooling are all contained within an 8 cm shell. We describe approaches for minimizing induced currents in peripheral conductors, which produce RF field distortions.

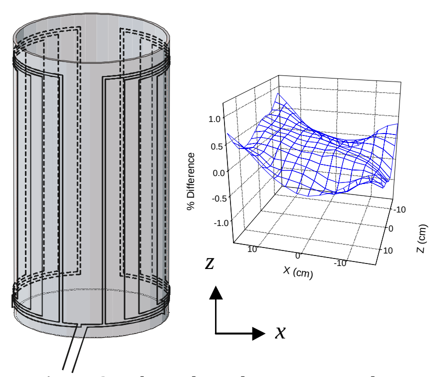

Figure 2: The coil windings. Measured field homogeneity in *xz*-plane.

NOVEL GEOMETRIES AND ACTIVE SHIELDING: In regular high field imaging applications RF fields can be easily confined using passive shields made of thin copper foil. This approach is not feasible in low field (~ 10 mT) applications, as the thickness of copper that would be required to confine the RF fields would also efficiently screen the switched gradient fields needed for imaging. As a first attempt to improve the homogeneity of the coil described above, the design was modified to incorporate an elliptical profile, which increases the distance between the RF coil windings and the imager, while maintaining the working volume that we require for in vivo lung imaging. The 'free space' homogeneity of this elliptical coil was measured to ± 2%. Inside the magnet, the homogeneity was measured to be ± 10% which is tolerable. To improve upon this further, we are exploring the use of active shielding techniques [8,9], which are most commonly used to decouple gradient coils from other structures within an imager.

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