## A 0.2 T Homogeneous Resistive Knee Magnet for Remotely Polarized MRI

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**Introduction:** Unlike conventional MRI, remotely polarized MRI methods (e.g., hyperpolarized gas, hyperpolarized $^{13}$C, PHIP, prepolarized MRI) show insignificant image improvement with field strength above a threshold NMR center frequency, called the body noise dominance frequency [1, 2]. This is because the polarization does not vary with the imaging field. For example, the Orsay group [3] has demonstrated excellent hyperpolarized $^3$He lung ventilation images at 0.1 T (~4 MHz center frequency). Here we detail construction methods for a 32-cm free bore, water-cooled resistive magnet that can operate up to 8 MHz proton frequency. We believe this is the most cost effective method to perform low field remotely polarized MRI.

**Methods:** A classical approach in homogeneous magnet design is to expand the magnetic field on a sphere into a sum of Legendre functions and then minimize the higher order terms by using a nonlinear solver. This technique is shown in [4] for circular current loops. Here, we extended this method for thick cylindrical coils. Our optimization was programmed in Matlab (Mathworks, Cambridge, MA). Our routine solves for the coil diameters ($d_k$), and the ratio of the current density $J_1/J_2$ (see Figure 1). The desired ratio $J_1/J_2$ can be determined by the ratio of wire gauges for the individual coils. The overall resistance can be determined by wire gauge choice. For shimming, additional single turns can be added to the inner an outer coil by the minimum-power magnet design [5]. For efficient cooling the magnet was wound in "pancake coil pairs" [6], having cooling plates between each pancake, an efficient cooling method [6]. The cooling plates were split every 22.5 degrees (16 times) to minimize eddy currents (see Figure 2).

**Results:** We designed a cylindrical shaped 4-coil homogeneous 0.2 T magnet, wound on a 13" diameter G10 former (see Figure 1). The field inhomogeneity was designed to be less than 50 ppm within a 14 cm DSV. The total magnet length is 22"; the total conductor mass is 200 kg. To match the power supply (210 V, 90 A) HAPTZ AWG 7 square copper wire was chosen, which leads to a total resistance of 2.4 Ω. The power dissipation is approximately 17.5 kW at 0.2 T. A total flow of 4 gal/min chilled water guarantees no more than 40°C temperature rise at the hottest spot inside the magnet.

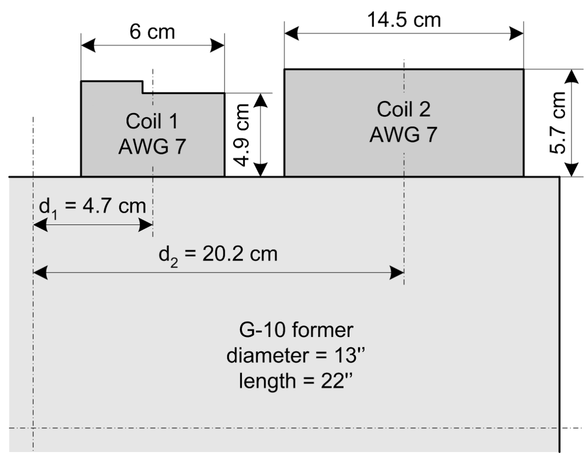


Figure 1: Cross section through symmetric 4-coil homogeneous 0.2 T knee magnet. 14 cm DSV designed for 50 ppm. Total cost $6000.

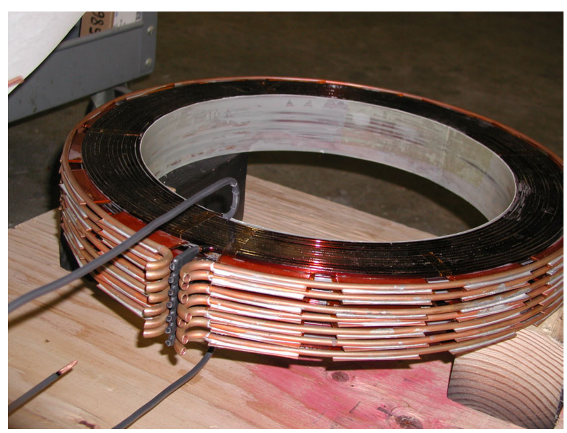

Figure 2: Photo of one coil from 4-coil knee magnet. Bore is 32 cm. Cooling loops for each pancake shown. This magnet should improve SNR of our current MRI wrist scanner by a factor of 4.

**Discussion:** This magnet design method offers excellent thermal stability, low cost construction ($6000), and good mechanical tolerances (~1 mm). While a 6-coil magnet may theoretically improve homogeneity, machining tolerance may render this impossible. Hollow wire cooling is another viable construction method, but the cooling and electrical layout are more coupled. The magnet is now being constructed by Stangenes Industries for less than $6000. This should improve the SNR of human wrist images by a factor of 4 relative to our current 1.1 MHz scanner, assuming $f^{3/4}$ SNR variation [1]. Other applications would be low-field hyperpolarized gas or PHIP.

**References:**

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