# Homogeneity and shimming requirements for a field-cycled MRI/PET scanner

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**Introduction:** Field-cycled MRI employs two concentric, actively controlled resistive magnets to replace the conventional superconducting magnet in clinical MRI [1,2]. The inner magnet is of high-field strength (~1 T), yet not necessarily uniform, and is used to polarize the sample. This magnet is activated for approximately 1 s, after which an outer, low-field (~0.1 T) homogenous (readout) magnet is turned on during k-space acquisition. A field-cycled MRI system, intended for small animal imaging, was designed and built by the authors (figure 1) [2,3]. A 9 cm gap was placed in the center of the scanner to allow for the insertion of a PET ring. The restriction placed on system geometry due to the PET ring reduces the magnet homogeneity. However, the homogeneity requirements of a field-cycled MRI scanner are less stringent than that of a clinical scanner due to the low readout-field-strength. The following abstract describes the shimming of a field-cycled MRI scanner to meet these requirements, and presents the first NMR spectrum obtained from the home-built scanner.

**Theory:** For low-field imaging, the phase accrued due to the applied gradient must be greater than the phase accrued due to field inhomogeneity [2,4]. For a 10 cm FOV and a typical 2.3 mT/m gradient strength, the maximum tolerable inhomogeneity for a 5 MHz readout field is 20 ppm $cm^{-1}$ [2]; this sets the necessary requirements for magnet shimming. Field-cycled MRI therefore requires a less homogeneous readout field than clinical MRI. Bandwidth and gradient strength can be adjusted during scanning to compensate for residual inhomogeneity.

In field-cycled MRI, the low magnetic field present during the application of gradient pulses reduces the Lorentz forces on the gradient coil. This, in turn, relaxes the constraints on the mechanical stability and construction tolerances of the gradient coil.

**Methods:** A gradient coil was constructed with a 12.7 cm inner diameter, 5.5 kg mass, and 61.0 cm length (figure 2). All gradient axes had efficiencies of 1-2 mT/m/A. The inner *z*-axis was mounted on a PVC tube, while the *x*- and *y*-axes were pressed into teflon sheets, which were epoxied onto the PVC tube.

Solenoid RF coils were constructed of 12 AWG round copper wire mounted on acrylic formers. The *Q* values of the RF coils ranged from 10 – 300. High *Q* mica capacitors were used for tuning. In addition to the magnet, the gradient coil, RF coils, and receive circuit were built entirely in-house.

A 2-mm-long sample was placed inside a 1-cm-diameter RF coil and positioned at various locations inside the scanner. The resonant frequency was measured as a function of position to map the magnetic field profile of the readout magnet. The scanner was allowed to reach thermal equilibrium before field measurements were acquired. The inner and outer pairs of coils in the readout magnet were iteratively repositioned to maximize the homogeneity. Coils were located on threaded rods (figure 1) to facilitate axial movement. The coils were positioned to an accuracy of 0.1 mm. The gradient coil described above was then used to apply first order shimming fields. The signal length was maximized for the 1-cm-diameter RF coil by adjusting the gradient strengths. The magnet was shimmed electromagnetically by shimming on increasingly larger sample sizes, up to an 8 cm diameter-sphere-volume (DSV).

**Results and discussion:** Thermal heating of the readout magnet can cause the center frequency to drift. To reduce temporal instability, the magnet was allowed to reach equilibrium before the acquisition of data. Inhomogeneity caused by the thermal expansion of the magnet wire is negligible over a typical 5 min scan time.

Figure 3 shows the homogeneity of the readout magnet along the bore of the magnet, as a function of the *y*-coordinate. The homogeneity of the readout magnet after mechanical shimming was approximately 350 ppm along the *z*-axis and 750 ppm in the *xy*-plane, for a 10 cm DSV. The field measurements had a 10 ppm precision. A typical spectrum, post mechanical shimming, is shown in figure 4. A higher homogeneity and SNR was achieved with the use of first order shimming. The homogeneity of the magnet, after first order shims are applied, is adequate for imaging at low field based on the requirements specified above. Higher-order electromagnetic shims are being incorporated into the scanner to further improve homogeneity and SNR.

The anticipated additional inhomogeneity due to the insertion of a PET ring is minimal. This dual-modality field-cycled MRI/PET system combines the required homogeneity for low-field MRI with the necessary geometry for the inclusion of PET.

**References:**

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[3] Gilbert K M et al. 2006 *Magn. Reson. Eng.* **29**B(4) 168-75.
[4] Morgan P N et al. 1996 *Magn. Reson. Med.* **36** 527-36.

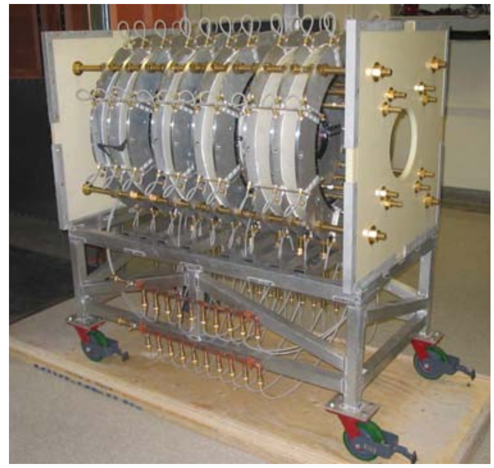

**Figure 1:** The field-cycled MRI scanner.

**Figure 2:** The gradient coil constructed for use in the field-cycled MRI scanner.

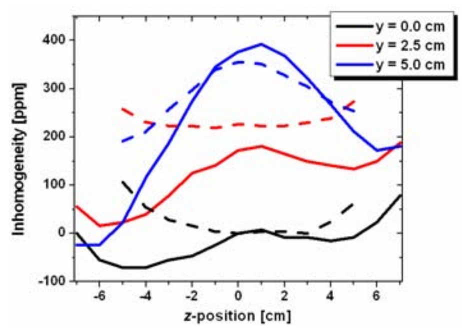


**Figure 3:** The inhomogeneity, as a function of position, after mechanical shimming. The readout magnet was operated at 1 MHz.

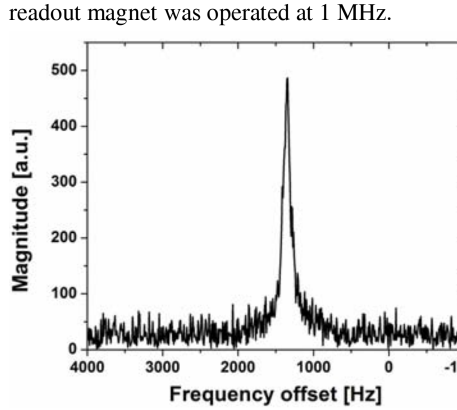


**Figure 4:** A typical spectrum acquired post mechanical shimming.