A Field-Cycling MRI System with Detection at 0.45T

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Introduction:

In recent years there has been increased interest in the use of magnetic field cycling with MRI. In field-cycling the magnetic field is switched between two levels: the evolution field B_0^E (often a low value) at which the spins evolve, and the detection field B_0^D at which RF pulses and signal detection occur. B_0^D should be as high as possible in order to improve SNR. The technique is used in field-cycled proton-electron double-resonance imaging (FC-PEDRI), to image the distribution of free radicals in biological samples [1]. Here the magnetic field is reduced to B_0^E (typically ~4 mT) at which the EPR of the free radical is irradiated for ~$3\times T_1$, transferring polarization from electron spins to water protons via the Overhauser effect. After ramping the field up to B_0^D in a time $<T_1$, signal detection occurs. A second use for field-cycled MRI is in relaxometric imaging, where field-cycling is employed to allow the acquisition of images showing the behaviour of T_1 with field strength, obtained on a single instrument. Such T_1-dispersion measurements can highlight so-called quadrupole dips in the T_1 *versus* B_0^E plots, caused by ^1H-^{14}N interactions in immobilized proteins [2-4]. In this work we have built a field-cycled MRI system with a detection magnetic field of 0.45 T, suitable for both of the above applications.

Apparatus:

***Field-Cycling Magnet Specification*:** The system is designed to cycle to any field between zero and 0.45 T, with a field switching time of under 40 ms. The maximum sample diameter is 60 mm To avoid image artefacts the detection magnetic field must be homogeneous (shimmed to ±1 ppm over a 60 mm diameter spherical volume (DSV)), stable (±1 ppm within 10 ms of a field switch) and repeatable (±1 part in 10^8 from one acquisition to another). Relaxometric imaging measurements can tolerate relatively large spatial and temporal variations (~5%) in B_0^E , but FC-PEDRI is less forgiving, so the specification was tailored to FC-PEDRI studies using narrow EPR-line (~50 μT) free radical contrast agents. In order to achieve uniform and efficient irradiation of the free radical's EPR, B_0^E (typically ~4 mT) should be homogeneous (shimmed to ±10 μT over a 60 mm DSV), stable (±5 μT during the evolution period) and reproducible from cycle to cycle.

***Magnet Implementation*:** A dual, coaxial magnet system is employed. The detection magnetic field is provided by a whole-body sized superconducting magnet with inner bore 830 mm (Oxford Magnet Technology, UK) operated at a constant 0.45 T. Within its bore, and coaxial with the superconducting magnet, is situated a resistive, actively-shielded field-offset ("bucking") coil, the net field from which is in opposition to the superconducting magnet's 0.45 T at the coils' centre. Active shielding of the offset coil is necessary to avoid eddy currents in the superconducting magnet's cryostat, and to minimize inductive coupling with that magnet. The actively-shielded field-offset coil assembly, which also incorporates gradient and shim coils, was constructed to the above specification by a commercial source (Tesla Engineering Ltd., UK). The free bore inside the shield/offset/gradient/shim coil assembly is 120 mm in diameter. In order to fully cancel the field at the sample, the field-offset and active-shield coils (connected in series) require a drive current of 880 A. This is supplied by a unit comprising four power-supply amplifier modules in parallel (Copley Controls Inc., USA). The power supply incorporates a high-precision zero-flux current transducer (Danfysik A/S, Denmark) which provides automatic correction for any drift in gain of the power-supply amplifier.

***Console and RF*:** The system is controlled by a commercial NMR/MRI console (Tecmag Inc., USA), via a standard PC. The magnetic field is set by the console via a home-built, 16-bit, high-precision DAC module whose output is fed to the field-offset magnet's power-supply amplifier. The console also provides RF at the NMR frequency (19.14 MHz) and at the EPR frequency (typically around 100 MHz) for FC-PEDRI experiments. These studies made use of two RF coil assemblies: one was a double-resonance birdcage resonator, tuned to both the NMR and the EPR frequency, with all modes being circularly polarized (i.e. NMR Tx and Rx, plus EPR Tx). The other used a saddle coil for NMR (Tx and Rx) and an Alderman-Grant resonator for EPR Tx. For relaxometric measurements another birdcage resonator (NMR Tx and Rx, without an EPR channel) was used.

Results and Conclusions:

The system has been used successfully in both FC-PEDRI and relaxometric measurements. *In vivo* FC-PEDRI studies were conducted on anesthetized adult Sprague-Dawley rats injected with a triaryl methyl free radical contrast agent. High resolution images of the free radical distribution were obtained (5 mm slice thickness, 0.8 mm in-plane resolution) that showed, for example, the animal's kidneys and ureters. Relaxometric studies on a boiled hen's egg test object demonstrated the expected marked reduction in T_1 of egg white, measured on a quadrupole dip (65 mT) relative to the T_1 measured either side of the dip (56 mT and 75 mT). The relatively high detection field (for a field-cycling system) of 0.45 T increases the SNR, and hence improves the quality of images and dispersion plots.

References:

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