Signal optimization of magnetic field pulses in field-cycled MRI

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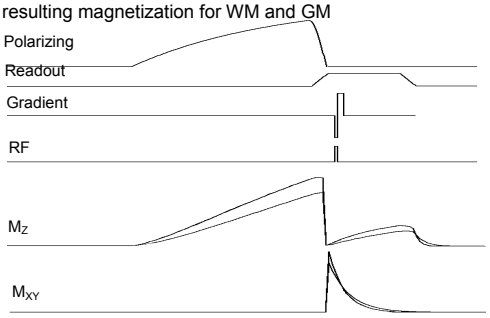
Introduction: In field-cycled MRI [1], the homogeneous and static magnetic field produced by the superconducting magnet is replaced by two dynamically controlled magnetic fields. One field, the polarizing field, is responsible for magnetizing the sample. This field needs to be strong, but not particularly uniform or stable. The other field, the readout field, is required during the detection of the precessing magnetization. This field needs to be both stable and uniform, but not particularly strong. A typical field-cycled pulse sequence is shown in Figure 1. Contrast and SNR in field-cycled MR images are determined by a combination of the polarizing field waveform, the readout field waveform, and of course the usual RF and gradient characteristics. The particular nature of field-cycled MRI (FCMRI) allows additional tissue contrasts to be generated by tailoring the pulse sequence to take advantage of intrinsic T1 dispersion of the tissue. Early work by Bottomly *et al.* [2] produced a general mathematical description modeling T1 in the form $T1 = \alpha\nu^{\beta}$ where α and β characterize the tissue and ν is the frequency of precession. The T1 dispersion curve determines how well various tissues respond to different magnetization waveforms. In superconducting MRI this is less of a concern since there is no control over the field strength, but in field-cycled MRI the strength of the polarizing field can be tailored to the particular type of imaging being performed. This research investigates the field-cycled pulse required to produce maximum magnetization in a given time and compared these waveforms to the more traditional exponential and sinusoidal waveforms produced by constant-voltage and capacitive driven power-supplies.

Figure 1. Field-Cycled MRI Pulse Sequence and the resulting magnetization for WM and GM



Methods and Results: A computer program was written in Matlab based upon Euler's Method [3] to solve the classical non-linear differential Bloch equations. The first simulation considered the simplest power-supply: application of a constant voltage across the polarizing magnet. This resulted in a simple exponential ramping described by the maximum voltage and the magnet time-constant τ. Over time, the sample's magnetization grows according to the Bloch equations and the tissue's T1 dispersion function. After 1s the magnetization was noted. This experiment was repeated for several different tissue types and τ values ranging from 0 (a perfect square pulse) to 0.9 (nearly linear). To force an inductor to reach maximum current instantly would require infinite voltage and is unrealistic, however it allowed determination of the maximum theoretical amount of tissue magnetization. Figure 2 shows how the sample magnetization varies for different τ values. Tissues with the longest T1 value experienced a greater relative reduction in magnetization compared to tissues with short T1 such as adipose experienced the least. What is less obvious from these graphs is the heating of the magnet. Heating is equal to power deposited and therefore proportional to field *squared*. While waveforms with the longest τ performed the worst, losing ~ 25% of their magnetization, they generated much more magnetization for a given amount of magnet heating. The maximum voltages of the waveforms were adjusted so that each waveform generated the same amount of coil heating. The resulting heat-normalized waveforms produced more tissue magnetizations than the square pulse. In Figure 3 shows that maximum magnetization occurs when τ is 0.4-0.5s for white and grey matter.

Figure 2.
Magnetization of Tissues in Exponential Field

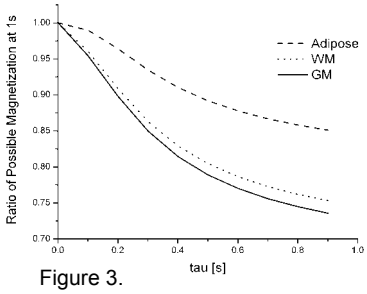
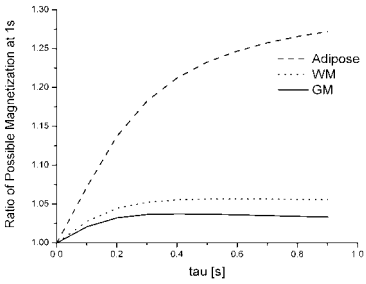


Figure 3.
Tissue Magnetization due to Exponential Waveforms



The simulation was repeated using a model of a capacitive power-supply capable of rapidly injecting current into the magnet. Again it was discovered that a slow changing ramping waveform with a high final current produced maximum magnetization. The difference in optimum magnetization per joule of deposited heat between capacitive and constant-voltage supplies was found to be only 1-2% depending on tissue type. This demonstrates that there is little difference in maximum magnetization obtained using these techniques.

To find the theoretical waveform resulting in maximum magnetization for a given amount of magnet heating, the simulation was modified to include an iterative search method similar to that of simulated annealing. The resulting waveforms are shown in Figure 4. The increase in magnetization over the capacitive and constant-voltage methods was less than 3%.

Conclusions: There are many different ways to generate magnetization using resistive magnets. Some are exotic, requiring specialized waveforms while others are the simple application of a constant-voltage. This study has shown that using simple methods to produce waveforms results in very little loss of magnetization compared to more complicated waveforms.

[1] Macovski et al., Magn Reson Med 30, 221-230 (1993)
[2] P.A. Bottomley, et al, Med. Phys. 11 (4), Jul/Aug 1984
[3] Numerical Recipes in C++, Cambridge University Press

Figure 4.
Optimum Magnetization Waveforms for GM, Muscle and WM