Strengths and Limitations of Pulsing Coils in an Array Sequentially to Avoid RF Interference in High Field MRI

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INTRODUCTION: RF wavelength and interference effects present significant problems in attempts to achieve uniform flip angle distributions in high field MRI. Varying magnitude and phase of individual elements in an array during excitation can yield some improvement. Other possible methods may involve pulsing different coils sequentially in time to avoid interference effects. Here we examine the potential for such methods at frequencies up to 470 MHz using full vector representation of the interaction between the spin magnetization $\mathbf{M}$ and the $\mathbf{B}_1$ fields.

METHODS: The finite-difference time-domain method was used to model a human brain within an 8-element coil array at 470 MHz. The field produced by each element was calculated and recorded. Then the results were loaded into home-built Matlab codes and the available signal intensity distribution was calculated in three different ways:

A) Assuming all coils are pulsed simultaneously, available signal intensity is calculated as

$$|\sin(\gamma\tau\Sigma_n \mathbf{B}_n^+)|$$

where γ is the gyromagnetic ratio, τ is the pulse duration, and the summation of the circularly-polarized vector components $\mathbf{B}^+$ is performed for n=1 to 8 coils in the array. Coils were driven with equal magnitudes and phase equal to locational azimuth measured from the center of the array.

B) Assuming coils are pulsed sequentially, but during the same TR and with no relaxation between pulses, we can use the vector equation $d\mathbf{M}/dt=\gamma\mathbf{M}\mathrm{x}\mathbf{B}$ to find $\mathbf{M}$ after the nth pulse $\mathbf{M}_n$ as

$$\mathbf{M}_n=\mathbf{M}_{n-1}+\int_\tau \mathbf{M}_{n-1}\mathrm{x}\mathbf{B}_n \, dt$$

where $\mathbf{M}_0$ is a unit vector in the z-direction and $\mathbf{B}_n$ is the magnetic field produced by the nth coil. This equation was implemented numerically to calculate $\mathbf{M}_8$ after sequential pulsation of all 8 coils timed and phased as if to be equivalent to the case in A.

C) Assuming each coil was pulsed alone during a different TR, with full relaxation between, and combing the available signal intensities afterwards, the final signal intensity can be calculated as

$$\Sigma_n|\sin(\gamma\tau\mathbf{B}_n^+)|$$

In the simplest implementation, this third method could be seen as the result of summing images acquired by exciting different coils – a method which has shown remarkable promise in a human head image at 470 MHz (1).

In all cases it is assumed that effective parallel imaging reconstruction algorithms are performed. When this is the case, available signal intensity is calculated at each location considering the sensitivity distribution of each coil and thus the image intensity distribution is ideally not a function of receptivity distribution, though SNR distribution is.

RESULTS: A representation of the brain in the 8-element array is given in Figure 1. Figure 2 shows the available signal intensity distribution calculated using each of the methods described. Although calculated with very different methods, the signal intensity distributions in cases A and B are indistinguishable. The distribution in case C is much more homogeneous than that in case A or B.

DISCUSSION: Pulsing different coils during different TRs can be an effective way to avoid RF interference and improve image homogeneity. If the coils are pulsed at different times but within the same TR it is necessary to consider the vector nature of the spin magnetization, rather than merely summing the $\mathbf{B}_1$ field magnitudes, as has been done previously (2). In effect, the vector nature of each $\mathbf{B}_1$ pulse is recorded in the spin magnetization vector and an interference-like effect is seen even if the RF pulses from different coils do not overlap in time unless the magnetization vector is allowed to relax between pulses. Unfortunately, pulsing different coils during different TRs will likely require either new reconstruction techniques or longer total imaging times. Further calculations, not shown here, indicate that very homogeneous images can be created with as few as 2 excitations when using carefully selected magnitudes and phases in the coil elements. It may also be possible that using these techniques with new reconstruction algorithms that take into account the dependence of signal intensity distribution on excitation distribution of the different coils will facilitate calculation of spin density with less dependence on excitation distribution.

REFERENCES:
1) Smith MB *et al*., 13th ISMRM, *submitted*, 2005
2) Ledden P, 12th ISMRM p. 38, 2004

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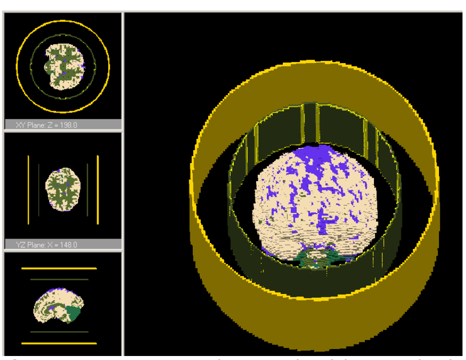

Figure 1: Geometry of an excised human brain in 8-element array.

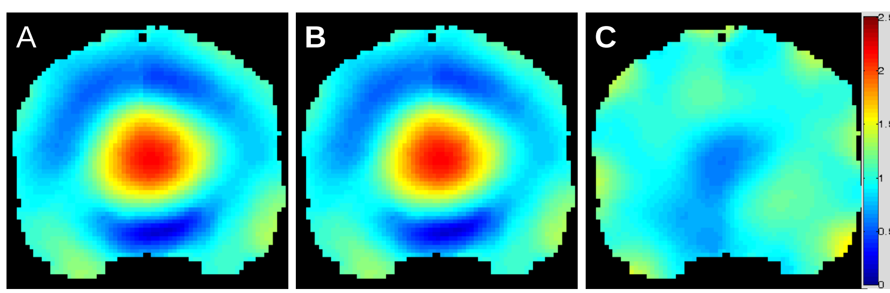

Figure 2: Signal intensity distribution assuming effective parallel image reconstruction for A) all coils driven simultaneously with equal magnitudes and with phases equal to coil positional azimuth, B) all coils pulsed sequentially during each TR with no overlap in time and with delays and phases such that the effective phases are as in A (and neglecting relaxation), and C) only one coil pulsed during each TR, requiring either new reconstruction algorithms or a longer total imaging time. Scale gives fraction of mean intensity value on the plane shown.