Signal-to-Noise Considerations for Parallel Imaging with Hyperpolarized Gases

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Introduction: For conventional proton MRI, the signal-to-noise ratio (SNR) for a parallel-imaging acquisition is decreased compared to a full, gradient-encoded acquisition by two factors: the square root of the reduction in imaging time, which for proton MRI is unavoidable because less time is spent collecting data, and a spatially-dependent noise amplification that depends on coil-array geometry and the relationships between coil sensitivities [1]. The latter factor can be minimized through optimization of the parallel-imaging strategy [2] and the coil array, and in a well-designed parallel acquisition the imaging-time factor dominates the SNR loss. In contrast, for imaging with hyperpolarized gases, the non-equilibrium nature of the magnetization leads to a fundamentally different relationship for the SNR in parallel imaging. The purpose of this work was to explore the SNR behavior for parallel imaging with hyperpolarized gases.

Theory: Let us consider imaging by using a conventional low-flip-angle GRE pulse sequence, which is by far the most widely used technique for ^{3}He MRI of the lung. It is reasonable to assume that the data-sampling period per excitation (or equivalently, the receive bandwidth per pixel) would be chosen based on some criterion such as the minimum achievable echo time or the resulting sensitivity of the pulse sequence to field inhomogeneities. For the following discussion, we will therefore assume that the data-sampling period is held constant as other pulse-sequence parameters are varied.

Of particular relevance to parallel imaging is how the SNR changes as the number of excitations is reduced. For conventional, multi-shot pulse sequences (e.g., GRE) applied to biological tissues and used in the typical steady-state mode (i.e., steady state of either the longitudinal magnetization alone as in a spoiled pulse sequence or of the complete magnetization vector), reducing the number of excitations, and thus the imaging time, by a given factor decreases the SNR by the square root of this factor because less time is spent collecting data. In a well-designed parallel-imaging experiment, this imaging-time factor should dominate the SNR loss. To answer the same question for imaging of hyperpolarized gases, we must choose a scheme for selecting the flip angles for the excitation RF pulses as the number of pulses is reduced. There are several considerations that influence the selection of the optimum excitation flip-angle scheme and a full discussion of these is beyond the scope of this work. Nonetheless, the results from two specific schemes illustrate the essential point.

First, consider the ideal variable-flip-angle scheme that provides a uniform signal level throughout the GRE acquisition and uses all of the magnetization [3,4]. The requisite flip angles are:

$$\theta_n = \operatorname{atan}\left(1/\sqrt{N-n}\right), \qquad [1]$$

where N is the number of excitations and θ_n is the flip angle for the n^{th} excitation. (T1 decay is neglected because the typical acquisition time (~1 s) is much less than the T1 for hyperpolarized gases in the lung (~20 s).) For this flip angle series, the SNR is:

$$\mathrm{SNR} \propto (N \sin\theta_1)/\sqrt{N}\ , \qquad [2]$$

where N is the number of excitations and θ_1 is the flip angle for the first excitation pulse from Eq. 1. Combining Eqs. 1 and 2, we find that the SNR dependence on the number of excitations (N) vanishes. As the second example, consider the case of a constant flip angle and sequential phase encoding. The SNR is:

$$\mathrm{SNR} \propto \sqrt{N}\cos^{(n_o-1)}\theta_c \sin\theta_c\ , \qquad [3]$$

where n_o is the phase-encoding step corresponding to the center of *k* space and θ_c is the constant flip angle. The flip angle that maximizes Eq. 3 is:

$$\theta_{c,max} = \operatorname{atan}\left(1/\sqrt{n_o-1}\right). \qquad [4]$$

As a function of the number of excitations, the maximum SNR for constant flip angle is initially (i.e., for N = 1) equal to the SNR for variable flip angles, and then, as N increases, oscillates toward an asymptotic SNR value that is 86% of that for variable flip angles. For 10 phase-encoding steps or greater, the maximum SNR for constant flip angle is within 5% of its asymptotic value. So, for values of N that are likely to be used in practice, the SNR for constant flip angle is also independent of the number of excitations as long as the flip angle is calculated from Eq. 4.

For the flip-angle schemes discussed in the preceding paragraph we find that, in contrast to steady-state proton MRI, the SNR is independent of the number of excitation RF pulses. As the number of RF pulses is decreased, the signal level can be increased by using larger flip angles, and this increased signal exactly balances the concomitant increase in the noise level secondary to the decrease in the total time spent sampling data. *Thus, when the number of phase-encoding views is reduced, for example to perform parallel imaging, there is no SNR penalty from the decrease in acquisition time.*

Experimental confirmation: For the case of constant flip angle and sequential phase encoding, we acquired GRE images (TR/TE 6.7/2.9) of 1-liter Tedlar bags filled with 100 ml of hyperpolarized ^{3}He and 900 ml of N_2. These studies were performed on a commercial 1.5-T scanner (Sonata, Siemens Medical Solutions). ^{3}He gas was polarized by collisional spin exchange with an optically-pumped rubidium vapor by using a commercial system (Model 9600 Helium Polarizer, Magnetic Imaging Technologies, Inc.). Measurements of the total magnetization in each bag were used to compensate for any differences in the polarization or volume of gas between bags. While holding the voxel volume constant, images with 32, 64 and 128 phase-encoding views were collected. For each case, the flip angle for the excitation RF pulse was calculated from Eq. 4. The resulting normalized image SNRs for 32, 64 and 128 phase-encoding views were 0.94 ± 0.05, 1.01 ± 0.05 and 1.05 ± 0.05, respectively. For conventional steady-state proton imaging, the corresponding SNRs would be 0.5, 0.7 and 1.0, respectively. Potential sources of error include an inaccurate transmitter calibration, or slight differences between measurements in the shape or orientation of the Tedlar bag. At this time we do not have an estimate for the magnitude of these potential errors, although we believe that they would each be on the order of a few percent. All factors considered, these preliminary measurements show very good agreement with theory and confirm that our theoretical formalism is valid.

Conclusions: Theoretical and preliminary experimental results indicate that the non-equilibrium nature of the hyperpolarized magnetization leads to the interesting circumstance that the SNR for an appropriately-designed GRE acquisition can be maintained independent of the number of excitation RF pulses (and thus the imaging time), and therefore the primary source of SNR reduction for parallel imaging with proton MRI can be eliminated for parallel GRE-based MRI using hyperpolarized gases. Specifically, parallel imaging applied to hyperpolarized ^{3}He holds the promise of lung MRI with a several-fold decrease in acquisition time without a substantial SNR penalty.

References:
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