Counteracting RF inhomogeneity on the upper legs at 3T using strongly modulating pulses

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Introduction: Strongly modulating pulses (SMP) [1] have been used to homogenize tissue flip angle (FA) on upper legs at 3 T. The muscle and fat tissues were discriminated in the pulse design algorithm via their resonance frequencies to achieve different flip angles. For a given TR, by implementing the respective nominal Ernst angles of the two different tissues in a spoiled gradient recalled echo (GRE) sequence, a higher signal for both tissues in addition to a better signal uniformity can be in principle obtained. After measuring the B_1 and ΔB_0 maps to generate the corresponding 2D histogram, a strongly modulating pulse with Ernst angles of 19° (for muscle) and 37° (for fat) was designed (TR=80 ms [2]). The pulse was then inserted in a GRE sequence and its performance was compared with the one of a strong hard pulse with nominal flip angle equal to 19°. Measurements of the FA when using the SMP were also performed to confirm the theory.

Theory: Strongly modulating pulses are non-selective pulses [1]. They are designed through the use of a measured $\{B_1,\Delta B_0\}$ 2D histogram in an optimization procedure. The returned pulse is made of a cascade of N hard pulses, each one parameterized by an amplitude, initial phase, duration and frequency. Due to this parameterization, the calculation of the dynamics is fast. In addition it does not rely on the linear approximation of the Bloch equation and is therefore valid for any flip angle. By searching in the parameter space with a genetic and simplex algorithm, a candidate waveform which yields a FA with minimum deviation with respect to the measured distribution is returned in less than 30 seconds on the fly. The theoretical GRE signal is $S(FA) \propto (1-E1)\sin(FA)/(1-E1\cos(FA))$, where $E1=\exp(-TR/T1)$, and achieves its maximum value at the Ernst angle $\theta_E=\arccos(E1)$, which is dependent on T1 and therefore on tissue. At this angle, to first order a deviation of the FA leads to no deviation of the measured signal. As a result, if the respective Ernst angles of the different tissues were achieved, a higher signal and better signal uniformity could be obtained. In the strongly modulating pulses algorithm, this is possible if the different tissues can be sufficiently discriminated via their respective resonance frequencies (not their T1 values) [3].

Methods: We used the actual flip angle imaging (AFI) sequence reported in [4] to measure B_1 and ΔB_0 over the volunteer's legs using a standard 300 µs square pulse (TR2/TR1=167/33 ms, FA=50°, matrix size 64 x 64 x 32, resolution 6 x 6 x 6.5 mm^3, TA=7 min). The same sequence was also used to check the FA obtained with the strongly modulating pulse. We used 9 and 10 bins in the ΔB_0 and B_1 directions respectively for the $\{B_1,\Delta B_0\}$ 2D histogram to design the SMP. The returned waveform had a 2.65 ms duration and a peak B_1 amplitude of 21.5 µT. Also, to compare the SMP (19° for muscle and 37° for fat) with a hard pulse, we performed two spoiled GRE experiments, each with a different pulse scenario, with following parameters: TR/TE=80/2.61 ms , resolution 1.5 x 1.5 x 2 mm^3, matrix size 256 x 256 x 104, IPAT factor = 2 and TA=11 min. We used a 3 T Siemens Trio scanner (Siemens, Erlangen, Germany), a whole body coil for transmission and an array of surface coils for reception. The SMP was designed on the fly during the volunteer's exam. For TR=80ms with the SMP, the SAR was 64% of the limit recommended by the IEC.

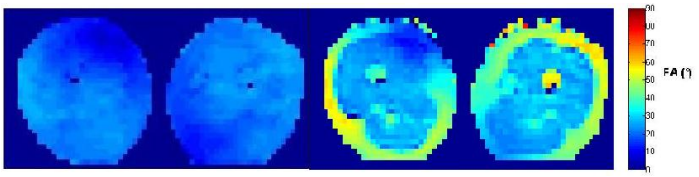
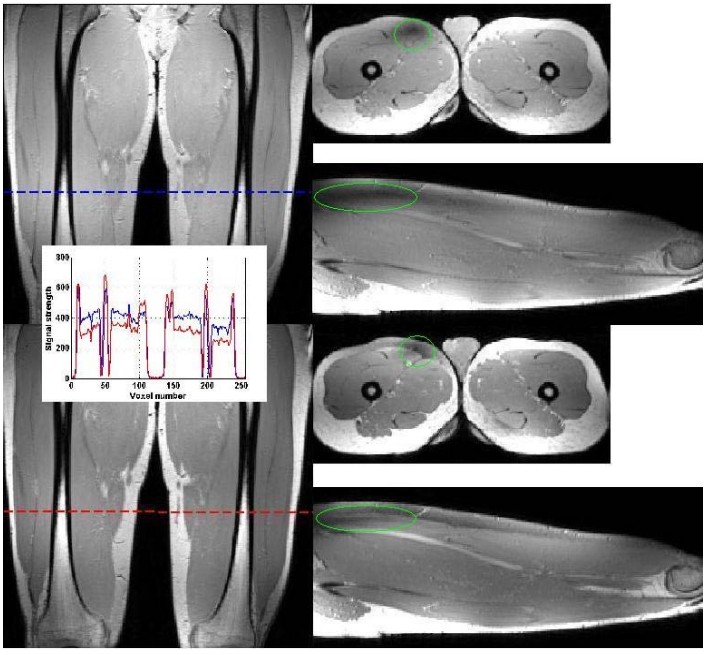


Fig. 1: Axial slice of the measured FA for a square pulse calibrated by the scanner (left), and a strongly modulating pulse (right) where one can see a clear difference between the FA achieved in the fat and in the muscle.



Results and Discussion: The FAs obtained when using the hard pulse and the SMP are shown in figure 1. One can clearly see that a higher flip angle was in general achieved in the fat tissue. In the muscle, the average FA was measured to be 26° because the FA formula in [4] does not take into account magnetization transfer between the free and the macromolecular protons [5, 6]. An analysis shows that any loss of magnetization in fact makes the FA calculation in [4] overestimated. The regions of very low B_1 intensities are also visible on these images. The results of the GRE experiments are shown in figure 2. The upper and lower three slices correspond to the hard pulse and SMP results respectively. On the left, one can also see the voxels'signal along the dashed lines. As expected, the signal in the fat tissue is higher (and hence also the contrast). The low B_1 regions have been circled in green. It is clear that the SMP also has been able to compensate to a good extent for the low B_1 values, with the signal being up to a factor of two higher. The signal drop in the muscle (20-25%) increases the contrast and is due to magnetization transfer. This effect could eventually be reduced either by taking a longer TR or by using an RF pulse requiring lower power.

Conclusion: We have demonstrated a new application of the strongly modulating pulses in MRI, namely the implementation of different flip angles in different tissues, while mitigating RF inhomogeneity effects. The key requirement is the sufficient discrimination of the two tissues via their respective resonance frequencies. Achieving different flip angles depending on the resonance frequency could be very useful for instance in water/fat suppression applications.

References: [1] N. Boulant et al. MRM 60:701-708 (2008). [2] E. Han et al. ISMRM 2003. [3] E. Fortunato et al. J. Chem. Phys. 116:7599-7606 (2002). [4] V. L. Yarnykh. MRM 57:192-200 (2007). [5] G. J. Stanisz et al. MRM 54:507-512 (2005). [6] X. Ou et al. MRM 59:835-845 (2008).

Fig. 2: GRE images using a 19° hard pulse (three upper images) and a 19°/37° (muscle/fat) SMP (three lower images). On the right the green circles show the locations of the B_1 holes which have been partially compensated for. On the left also is shown the signal along the dashed lines. As expected, the signal in the fat tissue has been increased while the one in the muscle has been reduced due to magnetization transfer.