Actual Flip Angle Imaging in the Pulsed Steady State

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Introduction: Knowledge of spatial distribution of flip angles (FA) produced by an imaging sequence is needed for many MRI engineering applications such as quality control of radio frequency (RF) coils, specific absorption rate (SAR) estimations, and validation of theoretical models for electromagnetic field calculation. Information about actual FA is important for correction of results in various quantitative methods (T_1, T_2, and magnetization transfer measurements, quantitative spectroscopy), especially in high magnetic fields. All existing methods for FA (or B_1 field) mapping rely on series of measurements with different FA (1,2) or application of sequences with non-identical RF pulses, i.e. spin echo (3-5) or stimulated echo (6), for acquisition at two or one nominal FA values. The assumption about the exact ratio of flip angles in such sequences (usually 2) may not hold due to slice profile and transmitter nonlinearity effects. Existing methods also require long TR (~ seconds), which preclude their 3D implementation for mapping the entire object. We propose a new single-measurement FA mapping method (Actual Flip Angle Imaging, AFI) based on a fast gradient-echo sequence with identical RF pulses.

Theory: Consider a pulse sequence consisting of two identical RF pulses with FA α followed by two delays TR_1 and TR_2 (Fig. 1). For clarity, we assume $TR_1 < TR_2 < T_1$. The FID signals S_1 and S_2 are observed after the corresponding pulses in a gradient-echo form. We further assume that the sequence is ideally spoiled, i.e.: 1) all transverse coherencies are dephased at the end of the delay TR_1, and 2) all transverse coherencies are irreversibly destroyed at the end of the delay TR_2. We refer to this sequence as the double-delayed spoiled gradient-echo, DD-SPGR. The ratio of signals generated by this sequence in the pulsed steady state is independent of T_1, if the first-order approximation can be applied for exponential terms:

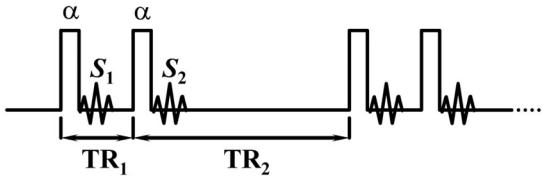

Fig. 1. DD-SPGR pulse sequence.

$$r = S_2 / S_1 = \frac{1 - E_1 + (1 - E_2) E_1 \cos\alpha}{1 - E_2 + (1 - E_1) E_2 \cos\alpha} \approx \frac{1 + n\cos\alpha}{n + \cos\alpha}$$

where $E_{1,2} = \exp(-TR_{1,2}/T_1)$, and $n = TR_2/TR_1$. Thus, for short TR_1 and TR_2 the ratio r can be used as a measure of the flip angle: $\alpha = \arccos((rn-1)/(n-r))$.

Methods: DD-SPGR pulse sequence was implemented on a 1.5 T whole-body MR scanner (GE Signa). A standard RF spoiling scheme with a 117° phase increment was applied throughout the pulse train. Additionally, strong spoiling gradients (23 mT/m, 50 ms duration) were applied along z- and x-axes at the end of the period TR_2 in order to facilitate diffusion dephasing of any residual transverse coherencies. This allowed avoiding incomplete spoiling in the presence of static field inhomogeneities. Three experiments were conducted in order to validate the concept of AFI:

1. A series of DD-SPGR scans was obtained from a small phantom containing media with different T_1. The images were acquired at variable nominal flip angles. The phantom was placed at the center of a head transmit-receive (T/R) coil. Transmitter power was calibrated manually as accurately as possible. Experimental signal behavior for different T_1 and FA was compared with theoretical predictions (Fig. 2).
2. An experiment on T_1 mapping by a variable FA method was performed with a surface single-turn T/R coil and a uniform phantom. Such a coil provides extremely non-uniform RF excitation (Fig. 3). AFI was used to generate a B_1 map, which allowed determining flip angles for reconstruction of a T_1 map.
3. In vivo whole-body 3D B_1 map was obtained from a human subject (Fig. 4) with following parameters of the DD-SPGR sequence: $TR_1/TR_2/TE$= 20/100/2.8 ms, nominal FA=60°, resolution 2x2x5 mm, scan time ~ 7 min. No cardiac and respiratory triggering was applied.

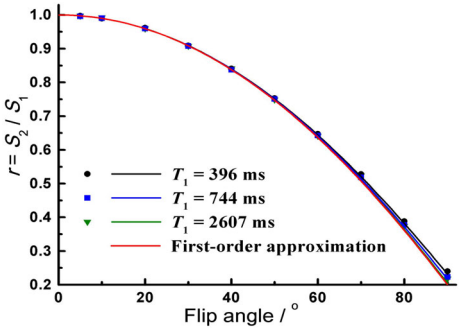

Fig. 2. Experimental (points) and simulated (lines) plots of the ratio r vs. FA obtained using DD-SPGR sequence with $TR_1/TR_2/TE$ = 30/150/5.6 ms for media with T_1 = 396, 744, and 2607 ms. Red line shows the first-order (T_1-independent) solution.

Results: Experimental and simulated plots of r as a function of the FA (Fig.2) demonstrated excellent agreement. To obtain optimal sensitivity of the ratio r to variations of FA, the parameter n should be in range 4-6, while TR_1 is typically 15-30 ms. At larger n and FA>80-90°, the signal behavior deviates from the first-order model, i.e. T_1-sensitivity becomes significant. Fig. 3 compares uncorrected and corrected T_1 maps obtained with a surface T/R coil from a homogeneous phantom. The use of actual flip angles derived from the B_1 map results in complete correction of nonuniformity on the T_1 map. In vivo whole-body B_1 maps (Fig.4) demonstrated typical patterns of RF field variations and did not show tissue-dependent features. Of note, the method is quite insensitive to flow and motion artifacts, because the signals S_1 and S_2 have identical phase properties.

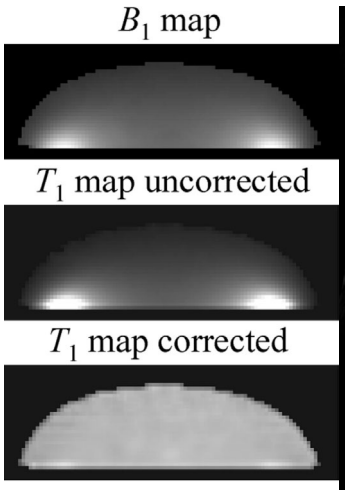

Fig. 3. Application of AFI for RF correction in T_1 mapping with a surface T/R coil. Maps were processed with an appropriate intensity threshold.

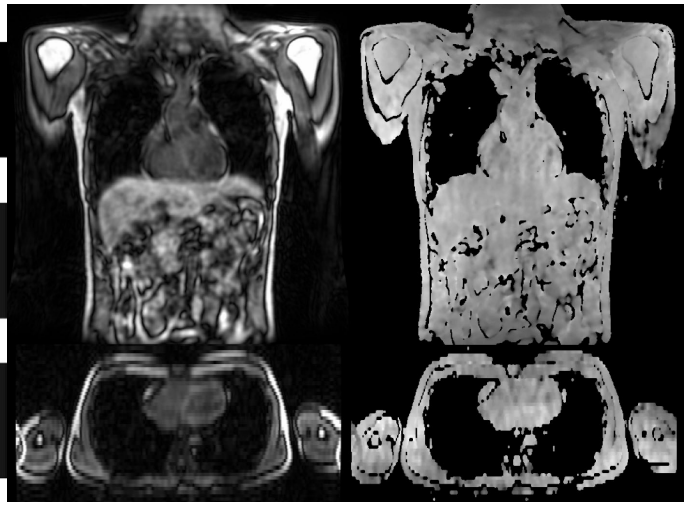

Fig. 4. Whole-body 3D B_1 mapping using an AFI method. Images show coronal and axial cross-sections corresponding to the signal S_1 (left) and calculated B_1 map (right). Maps were processed with an appropriate intensity threshold.

Conclusions: The uniqueness of the proposed method is determined by the fact that it requires a single measurement with the sequence containing identical RF pulses. To date, AFI is the most time-efficient method for RF field measurements, as it works with an effective TR (i.e. TR_1+TR_2) about 100-150 ms. AFI can be used for any applications related to B_1 mapping or RF correction in quantitative methods.

References: 1) Hornak JP et al. *MRM* 1988;6:158. 2) Barker GJ et al. *Br J Radiol* 1998;71:59. 3) Insko EK, Bolinger L. *JMR* 1993;A103:82. 4) Stollberger R, Wach P. *MRM* 1996;35:246. 5) Oh CH et al. *MRI* 1990;8:21. 6) Akoka S. et al. *MRI* 1993;11:437.