On the SSFP Signal

O. Bieri[1], and K. Scheffler[1]

[1]MR-Physics, Department of Medical Radiology, University of Basel / University Hospital, Basel, Switzerland

Introduction. SSFP has been successfully applied to many areas, such as neuroradiology, musculoskeletal, abdominal, cardiac, and cartilage imaging, because of its notably improved signal-to-noise ratio (SNR) over gradient-echo techniques, its fast performance and excellent contrast-to-noise ratio (CNR). Consequently, its signal property has been investigated over several years by many groups and it is generally accepted that the SSFP signal is a function of relaxation times, excitation angles, and spin densities only (Fig. 1). However, it was shown that in tissues balanced SSFP suffers from a considerable signal loss by up to a factor of two as compared to theory (1). This signal loss is due magnetization transfer (MT). A modification of the SSFP sequence scheme is presented that modulates signal attenuations in SSFP to either strongly attenuate or recover the near full steady state amplitude. Implications to SNR (CNR) for sequence design for clinical applications are analyzed.

Methods. Measurements were performed on a Siemens Avanto 1.5 T system. The SNR of a generic pulse sequence is given by

$$SNR \propto S \cdot \Delta V \cdot \sqrt{\eta} \cdot \sqrt{T_S} \qquad [1]$$

where S is the signal amplitude, ΔV is the voxel volume, η is the sequence efficiency (duration of readout, RO, per TR) and T_S is the time for image acquisition. For SSFP, the steady state signal might be strongly affected by MT and Eq. [1] modifies to

$$SNR_{SSFP} \propto \xi \cdot S \cdot \Delta V \cdot \sqrt{\eta} \cdot \sqrt{T_S} \qquad [2]$$

where S again represents the unperturbed signal (Fig. 1, right) and ξ is a factor that captures MT-related dependencies. From this, CNR simply relates to the SNR difference (Eq. [2]). Prior findings (1) suggest up to 2-fold signal reduction (ξ=1/2) in tissues exhibiting strong MT from saturation. The average rate of saturation (W) in the longitudinal magnetization of protons associated with macro-molecules is given by (see (2))

$$\langle W(\Delta) \rangle \propto \tau_{RF}^{-1} \int dt \cdot \omega_1^2(t) \propto BW_{RF} \int dt \cdot \omega_1^2(t) \qquad [3]$$

where $\omega_1(t) = \gamma |\mathbf{B}_1(t)|$ describes the RF excitation field strength. Elongation of the RF pulse duration by a factor of β, i.e. $\tau_{RF} \rightarrow \beta \cdot \tau_{RF}$, as shown in Fig. 1, reduces the pulse amplitude (and bandwidth, BW) by the same amount, i.e. $\omega_1 \rightarrow \omega_1/\beta$, for identical flip angles. As a result, saturation effects should scale like $\langle W(\Delta) \rangle \rightarrow \langle W(\Delta) \rangle / \beta^2$ and therefore affect ξ. Thus optimal SNR (CNR) for SSFP may substantially depend on the target, especially in tissues where strong MT effects can be expected.

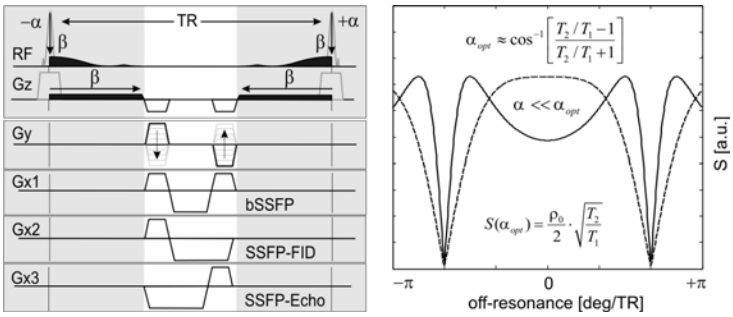

Fig. 1: Sequence (left) for saturation and thus signal modulation in SSFP. Theoretically, SSFP depends on relaxation times, flip angle, and ρ_0 only (right).

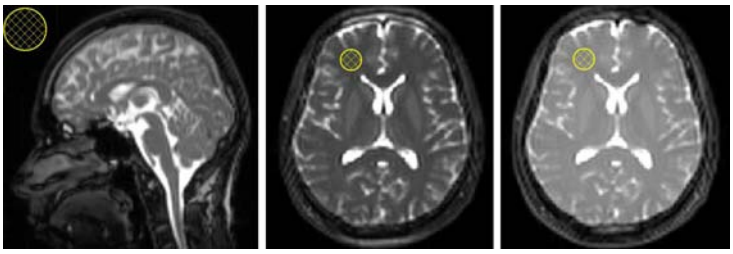

Fig. 2: 3D balanced SSFP images (2×2×2mm, α=36°). A noise of 0.5 was equal to both image acquisitions (left), where the reduced acquisition time (1/2) is not yet accounted for. WM signals of 69 (middle) and 139 (right) are found for maximized and minimized saturation (Eq. [3]), respectively. ROIs indicated in yellow.

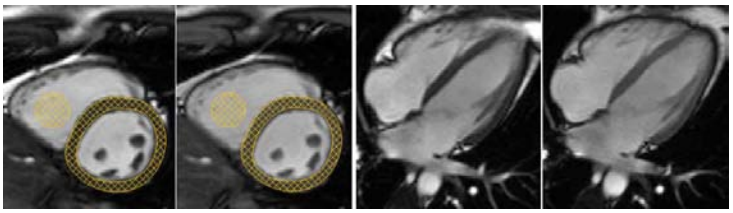

Fig. 3: Short- and long axis view of a cardiac phase using balanced SSFP (α=36°). Shown are identical phases with either maximized (TR=2.9ms, τ_{RF}=270μs) or minimized (TR=4.3ms, τ_{RF}=1700μs) saturation. Measurements yielded a SNR_{SAT} (SNR_{noSAT}) of 121, 31 (105,48) for blood and myocard, respectively. Noise was less in the saturated acquisition (2.6 vs. 2.9). From this, CNR_{SAT}=90 and CNR_{noSAT}=57.

Results & Discussion. The new scheme for saturation-related SSFP signal modulation, as presented in Fig. 1, may entail a revision of optimal SSFP sequence design for clinical applications. An introductory example to this issue is presented in Fig. 2. Here, signal in white matter (WM) is strongly affected by saturation (left: TR=2.5ms, RO=1.6ms, τ_{RF}=200μs), whereas near full signal can be attained by RF pulse modification (right: TR=5.0ms, readout RO=2.5ms, τ_{RF}=2000μs). From Eq. [3], $SNR_{noSAT} \propto S\sqrt{\eta}\sqrt{T_S}$ whereas $SNR_{SAT} \propto \frac{1}{2}S\sqrt{\eta}\sqrt{2T_S}$, where $\sqrt{2}$ corrects the halved acquisition time. Thus SNR is increased by about 25% (experimental: 35%) if saturation can be avoided. In cardiac imaging, SSFP offers an excellent contrast between blood and myocardium. Saturation affects contrast by almost a factor of two, as can be appreciated in Fig. 3, since blood, in contrast to muscle, is hardly affected by MT. The elucidatory examples presented here base on balanced SSFP type of acquisitions. However, it should be clear, that SNR and CNR considerations for non-balanced SSFP type of sequences (i.e. SSFP-FID and SSFP-echo or echo combinations) are in complete analogy.

Conclusion. We have shown that the signal from SSFP sequences depends on MT and thus on the tissue. The gain (or loss) in SNR or CNR may be substantial and an application specific sequence design is indicated. Whereas for optimal CNR, MT effects may be helpful, for simple signal considerations, sequence misdesign can result in a loss of SNR by up to about 30% for tissues exhibiting strong saturation effects.

References. 1. Bieri & Scheffler, *MRM* **56** (2006) 2. Graham & Henkelman, *JMRI* **7** (1997).