Spectrally Sensitive Imaging Using Balanced Steady State Free Precession

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Introduction. Balanced Steady State Free Precession (b-SSFP) MRI[1] allows rapid image formation. Major advantages of b-SSFP are the greatly improved SNR and blood-tissue CNR compared to other rapid gradient echo techniques. After a certain number of RF pulses, the observed MR signal from a sample reaches a steady state (SS). The SS signal depends on underlying T1 and T2 relaxation times, the repetition time TR, the excitation B1 distribution, and the degree of spin dephasing due to spins at off-resonance. Depending on the off-resonance, b-SSFP will demonstrate different SS amplitudes (Fig. 1) and signal phase as a function of the local off-resonance frequency. In conventional b-SSFP with field inhomogeneities across the FOV, this off-resonance dependency of the SS magnetization is apparent as a bothersome resonance-offset-dependent "banding artifact". Recently it was shown that for appropriate pulse sequence parameters, at TE = TR/2 spins from water and fat have exactly opposed phase and hence the spectral features of b-SSFP can be utilized to separate water from fat[2]. Here, we provide a better description of the origin of the SS spectral signal fluctuation and present a completely different approach to interrogate the spectral components within a voxel using an augmented b-SSFP imaging method which utilizes the frequency dependence of the SSFP response.

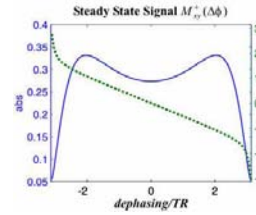

Fig. 1 –Magnitude (blue) and phase (green) of the steady state signal as a function of the dephasing or off-resonance. Notice that this function is periodic with ±π which corresponds to frequency offset of ±1/TR. For example, at 1.5T: +/- 125Hz with TR = 3ms; +/- 94Hz with TR = 4ms.

Materials and Methods. Assuming that the signal from one voxel originates from spins resonating at slightly different Larmor frequencies, the SS signal received is a sum of frequencies weighted by the fractional amount of the species resonating at a particular frequency weighted by the corresponding value of the filter function (Fig.1) at this frequency. By deliberately altering the phase of b-SSFP excitation RF pulses the pass and stop bands of this filter (Fig.1) can be cyclically shifted. The stop/pass band periodicity is 1/TR. Specifically, the RF-phase of the excitation can be varied in linear increments Δf within [0, 2π] in a periodic manner to generate this varying filter which will sweep across a frequency range determined by $f_{SSFP} = 1/TR$. The spectral resolution of the measurement Δf can be freely adjusted by selecting the number of phase increments N_{Inc} in [0, 2π] and is given by $\Delta f = f_{SSFP}/N_{Inc}$. For a given matrix size $N_M = (N_x \times N_y \times N_z)$, the total acquisition time is therefore $N_M N_{Inc} + N_{prep}$, where N_{prep} denotes the dummy cycles that are necessary to reach a SS. Ideally, this filter should be impulse function. In such case, the filter would sweep over the frequency range 1/TR like a spectrum analyzer and would provide direct information about the underlying frequency distribution of spins at off-resonance on a per-pixel basis. However, because the b-SSFP filter has a more complex shape (Fig. 1) and is determined by T1, T2, and excitation B1, the voxel response is actually a convolution of the filter function with the underlying frequency distribution. Here, the voxel response, V, at a certain frequency, η, is the weighted sum of spins S at off-resonance over the frequency range, Ω, covered by the filter kernel, H (Eq. 1). Eq 1 is also known as circular convolution. The sweeping increment, df, determines the frequency resolution of the spectrum and can be approximated numerically (Eq. 2). By deconvolving the b-SSFP signal with the filter kernel, the underlying frequency spectrum in each voxel can be obtained. Using this sweep approach, a good shim is not necessary as long as the relative position of a reference frequency, e.g. water, is known and the field is homogenous over the voxel. Optionally, basis sets for filter kernels can be determined beforehand by measuring calibration phantoms for different species (T1, T2, B1). Notice that to some extent the shape of the filter kernel can be customized by selecting appropriate b-SSFP parameters (i.e. flip angle, TE, TR) for a given range of relaxation times. The deconvolution of the measured SS signal is a classical inverse problem and allows one to solve for an estimate of the underlying resonance frequency distribution in a minimum norm sense (Eq. 3)[4]. Here, **s** is the signal distribution of the range of frequencies in vectorized form.

$$V(\eta) = \int_{\Omega} S(f)\, H(f-\eta)\, df \qquad [1]$$

$$\begin{pmatrix} V_0 \\ V_{\Delta f} \\ \vdots \\ V_{(n-1)\Delta f} \end{pmatrix} = \begin{pmatrix} H_0 & H_{(n-1)\Delta f} & \cdots & H_{\Delta f} \\ H_{\Delta f} & H_0 & \cdots & H_{2\Delta f} \\ \vdots & \vdots & \ddots & \vdots \\ H_{(n-1)\Delta f} & H_{(n-2)\Delta f} & \cdots & H_0 \end{pmatrix} \begin{pmatrix} S_0 \\ S_{\Delta f} \\ \vdots \\ S_{(n-1)\Delta f} \end{pmatrix} \rightarrow \mathbf{v} = \mathbf{H}\,\mathbf{s} \qquad [2]$$

$$\mathbf{s} = \left(\mathbf{H}^{T}\mathbf{H}\right)^{-1} \mathbf{H}^{T}\mathbf{v} \qquad [3]$$

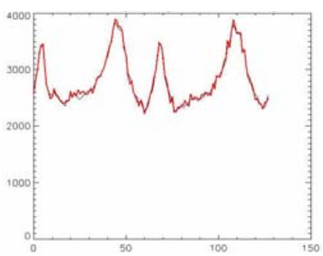

Fig. 2 - b-SSFP Voxel response, V, (black line) and after noise was added (red) (4% of maximum signal).

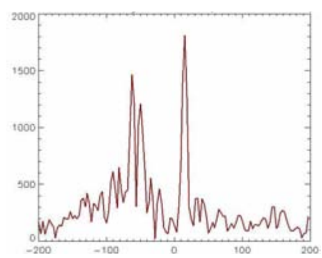

*Fig. 3 – Original MR signal spectrum (black line) and deconvolved spectrum, **s**, (red line) using $SV_{thresh} = 0.05 \cdot SV_{max}$. Almost no differences are apparent.*

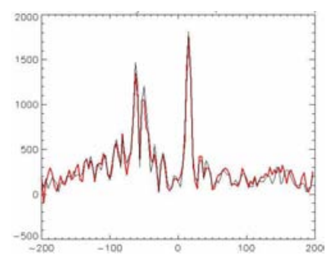

Fig. 4 – Estimated spectrum deconvolved from the noisy b-SSFP voxel response (Fig.2) using TCSVD (5% SV_{max} threshold). The original spectrum is shown in black. Differences are due to regularization.

Results. Fig. 2 demonstrates a typical voxel response, V, from a sweeping b-SSFP acquisition for the underlying frequency distribution shown in Fig. 3. A truncated ($SV_{thresh} = 0.05 \cdot SV_{max}$) circular singular value decomposition (TCSVD) was used to regularize the solution of the spectrum. For measured voxel responses in the high SNR range there is almost no difference between the underlying and the deconvolved spectrum (Fig. 3). With increasing noise levels, however, the spectral distribution gets increasingly degraded but can be corrected for by regularization (Fig. 4).

Conclusion. The contribution of relaxation times, excitation B1 field, and off-resonant spins to banding artifacts in b-SSFP has been characterized. Moreover, a new variant of b-SSFP in concert with a postprocessing method has been introduced to map the underlying spectral distribution of spins within a frequency range given by 1/TR. The b-SSFP banding can be characterized as a result of a circular convolution between the underlying spectrum and a filter function determined by relaxation times and B1. Potential applications of this new approach include improved frequency selective imaging, rapid determination of spectral distribution of spins, and better characterization of molecular tracers (e.g. SPIOS). The proposed spectral decomposition might be advantageous for spectroscopic imaging where high spatial resolution is required without the need for high spectral selectivity. Since water and fat peaks can be clearly identified an advanced Water/Fat separation approach is feasible. In this study the regularization was adjusted manually to find a tradeoff between regularization and signal oscillations. To improve the robustness of deconvolution, more sophisticated regularization approaches, such as L-curve criteria, generalize cross-validation or even total variation could be implemented[4]. Since the filter function originates from (noisy) measurements the utilization of the Total Least Squares method is advisable[4].

References. *[1]Oppelt A, et al. Electromedica 1986; 54:15-18. [2]Vasanawala S, et al. MRM 43: 82-90, 2000; [3]Eberhardt KW, et al. Proc. 13th ISMRM, Miami, FL, 2005, p. 2510; [4]Golub & Van Loan, Matrix Computations, 1983;* ***Acknowledgements.*** *This work was supported in part by the NIH (1R01EB002771), the Center of Advanced MR Technology at Stanford (P41RR09784), Lucas Foundation, and Oak Foundation. The authors are grateful to Drs. D. Clayton, D. Mayer, B. Hargreaves for helpful and stimulating discussions.*