Vessel-wall imaging using water-selective centrically-encoded 3D SSFP-echo with double-inversion recovery

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Introduction: In this work we explore vessel wall imaging of carotid arteries using water-selective 3D SSFP-echo (SSFPe) pulse sequence with intermittent application of double-inversion recovery (DIR) as a means to minimize inflow enhancement. SSFPe signals are formed from superposition of many RF and stimulated-echo pathways[1] that survive several TRs. The flow sensitivity of SSFPe pulse sequence results from the phase-incoherence of echo pathways[2,3] induced by bulk flow, which will be referred to as flow-induced (FI) spoiling. FI spoiling is equivalent to RF spoiling[4] from the perspective of magnetization's reference frame. Preliminary results and simulations indicate that blood signal attenuation via FI spoiling is not sufficient and the SNR of tissue is limited. For more effective blood signal suppression and optimized tissue SNR we propose to implement double inversion for each k_z (outer loop) and centrically-encode the SSFPe pulse sequence along k_y (inner loop).

Methods: The longitudinal magnetization of blood will be attenuated significantly more compared to tissue in SSFPe (**Fig 1**) due to FI spoiling. Low tissue SNR will result because the effective TE of SSFPe is approximately 2×TR (~18 ms) and T2 of tissue is approximately 35 ms at 3T. The DIR module is implemented for each k_z to increase tissue signal and improve blood suppression by minimizing magnetization of the inflowing spins. The double inversion consists of a non-selective hyperbolic-secant (HS) adiabatic inversion (BW=1940 Hz and duration=9.08 ms) immediately followed by a slab-selective (thickness=70mm) HS adiabatic inversion. M_z of out-of-slab blood will reach a steady-state due to periodic application of the non-selective adiabatic inversion pulse. By judiciously choosing the inversion time (TI) M_z of inflowing spins will pass through the null point at the time the SSFPe sequence is launched (in practice dummy loops are required before acquisition begins). In short, a single DIR preparation is used for each k_z. In the imaging slab during the TI, a fraction of the blood will be replaced with "DIR-prepared blood" and tissue M_z will relax. Thus, undesired signal enhancement from inflow is minimized since once the blood enters the imaging slab its magnetization will be further attenuated and/or suppressed from relaxation in response to the FI spoiling effect of the SSFPe pulse sequence. Bloch equation simulations were performed on a hypothetical voxel of 1000 spins with linear velocity distribution (± 0.1 cm/s) about an average value determined by the cardiac phase of the monophasic velocity waveform (**Fig 2**) typically observed in carotid arteries. The inflow effect was simulated for a 2D-slice voxel by resetting a fraction (5%) of the spins to the equilibrium longitudinal magnetization value every TR. The FI spoiling was simulated by letting the phase of each spin evolve based on its velocity, and the longitudinal and transverse magnetization was determined by averaging (weighted by motion- and displacement-induced phase) over all spins in the voxel. 3D in vivo images of the carotid arteries of healthy volunteers were acquired with a water-selective (binomial (11) RF excitation) 3D SSFPe and DIR-SSFPe using custom-built two-element bilateral 4-channel carotid coils. Imaging parameters: flip angle=8°+8°, TE/TR=3.8/8.71ms, FOV=140×128×70 mm^3, voxel size=0.61×0.61×2 mm^3, total acquisition time=68s for SSFPe and approximately 100s for DIR-SSFPe (TI~ 700 ms).

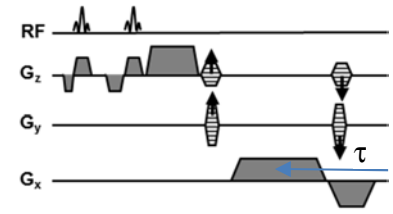

Fig 1 Water-selective 3D SSFP-echo pulse sequence using two RF pulses separated by 1.196 ms at 3T. After the 2nd RF, a crusher dephases the FID and imparts additional phase to moving spins. Effective TE is approximately (2×TR) - τ, i.e. the first echo is formed on the 2nd cycle of the pulse sequence.

Results:

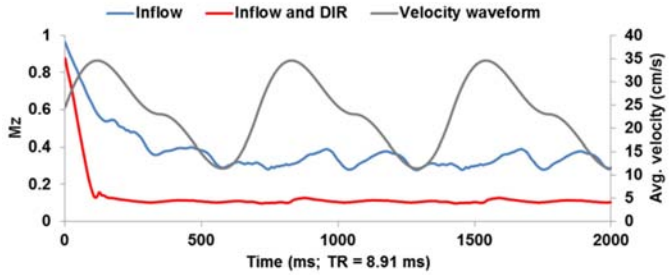

Fig 2 Bloch equation simulation of blood longitudinal magnetization for SSFPe (blue) and DIR-SSFPe (red). For reference, simulated monophasic velocity waveform is also shown (gray).

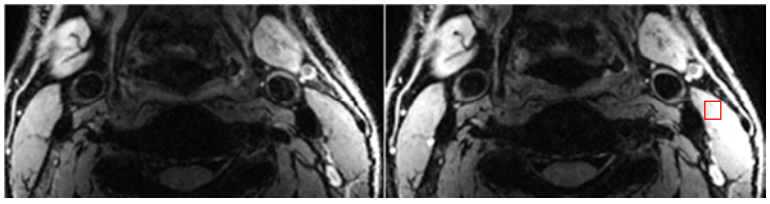

Fig 3 Comparison between images acquired with SSFPe (**a**) and DIR-SSFPe (**b**) at the level of common carotid artery. Images are shown with the same window level. Tissue SNR (red square) increased by about 20% in **Fig 3b**.

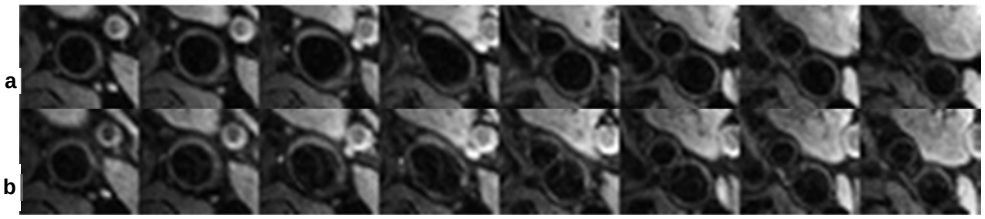

Fig 4 Eight (out of 35) contiguous slices (16 mm) about the bifurcation, acquired using (**a**) DIR-SSFPe and (**b**) SSFPe, are shown to demonstrate that incorporating DIR with centric encoding leads to reduced flow artifacts by minimizing the magnetization of the inflowing spins.

Conclusions: Potential application for vessel-wall imaging using water-selective 3D SSFP-echo pulse sequence with DIR was demonstrated in vivo in healthy young subjects. The efficiency (3s/slice) of the technique makes it suitable for 3D vessel-wall imaging.

References: [1] Wu and Buxton JMR 1990, [2] Buxton MRM 1993, [3] Gudbjartsson and Patz MRM 1995, [4] Zur et al. MRM 1991.

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