Reference-less flow measurements using refocused SSFP

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Introduction: Phase-contrast (PC) MRI is an established flow quantitation technique, but suffers from limited spatio-temporal resolution, and inferior SNR compared to balanced SSFP sequences. Recently, several groups [1-4] have proposed PC-MRI methods based on SSFP, which achieve accurate flow quantitation with high SNR efficiency. However, the total acquisition time for any phase-contrast-based method is intrinsically high, since one or more separate phase reference scans must be acquired. We propose a flow quantitation approach that exploits the intrinsic refocusing property of SSFP, to achieve 50% reductions in scan time.

Methods: Experiments were performed on a GE Signa 3T EXCITE HD system (gradients capable of 4 G/cm amplitude and 15 G/cm/ms slew rate), using a 4-channel receive coil array. A time-resolved 3DFT SSFP image volume was acquired in the common carotid artery in a healthy volunteer (1x1x2 mm^3 voxel size; FOV 16 cm; TR=6.4 ms), using the readout waveform and sequence timing illustrated in Fig. 1 (solid lines). The echo time was set to one-half the sequence repetition time (TE=TR/2), which ensures that the image phase is independent of local resonance offsets [5]. This is illustrated in Fig. 2, which shows the simulated spin phase for blood at rest (simulation parameters T1/T2=1000/200 ms; TR=5 ms; flip angle 50^o; RF phase incremented by 180^o every TR). Fig. 2 demonstrates the anticipated "spin-echo-like" refocusing property of balanced SSFP. As a consequence, the image phase in stationary image regions is largely object-independent, and reflects system-related sources of phase such as gradient/acquisition time delays and complex coil sensitivities, which are generally smoothly varying across the field-of-view. In this work, this smooth, object-independent background phase was estimated from stationary regions adjacent to the carotid artery, fitted to a 2nd order 2D polynomial, and subtracted from the image. The remaining image phase depends only on the flow-encoding properties of the imaging gradients themselves, and hence only on spin velocity. With knowledge of the gradient first moment at the echo, the velocity can therefore be calculated directly, without the need for a separate phase-reference acquisition. Note that flow along the phase-encode direction does not affect the image phase, but rather manifests as a spatial shift along y. Also note that in this work we neglect the small flow-encoding moment produced by the slice-select gradient. For validation, we performed a second acquisition with the readout gradient inverted (dotted line in Fig. 1), and calculated the velocity along the readout direction from the two images using conventional PC processing [3].

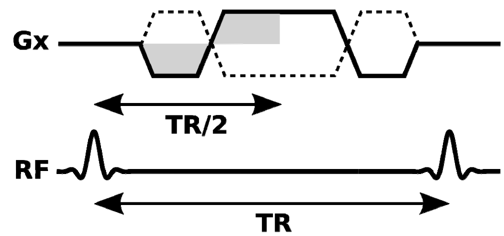

Figure 1: SSFP pulse sequence for reference-less flow (solid lines). At the echo, spin phase is independent of local resonance offset, and is determined by the velocity and the gradient first moments at the echo. The shaded area indicates the gradient lobe responsible for flow-encoding. The dotted line shows the readout gradient used for the second acquisition, which was used to perform conventional phase-contrast processing (for validation purposes only).

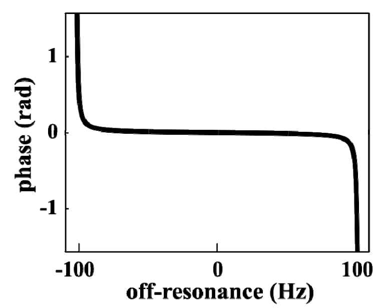

Figure 2: Simulated blood phase for a range of resonance offsets, assuming zero velocity. The band edges are located at -100 and 100 Hz. The phase is near zero across most of the SSFP band, indicating "spin-echo-like" refocusing.

Results: Fig. 3 shows the velocity along the readout direction (vertical in the image) in the common carotid artery of a healthy volunteer in peak systole obtained with (left) conventional SSFP PC-MRI using two SSFP acquisitions, and (center) the proposed reference-less (or "non-PC") method using only one SSFP acquisition. Note that the PC map shown in Fig. 2 was obtained after correcting for a significant linear phase "roll" (or "shading") along the readout direction, which is due primarily to a small gradient/acquisition time delay. Velocity profiles along the blue and read dotted lines are plotted on the right, and are in good agreement.

Discussion: A drawback of the proposed technique compared to the gradient inversion method is the reduced effective VENC, which reduces the velocity SNR. In addition, the proposed method cannot be used in applications containing little or no stationary tissue within the imaging field-of-view. Furthermore, depending on the choice of receive coil, it may be necessary to perform a higher-order (>2) polynomial fit to the background phase. Also note that if the local resonance offset is larger than 1/(2TR) (or smaller than -1/(2TR)), spin phase may be pi out of phase, and the calculated velocity will be the negative of the true spin velocity.

Conclusion: The intrinsic refocusing property of SSFP can be used to shorten the scan-time of SSFP phase contrast methods by a factor of two.

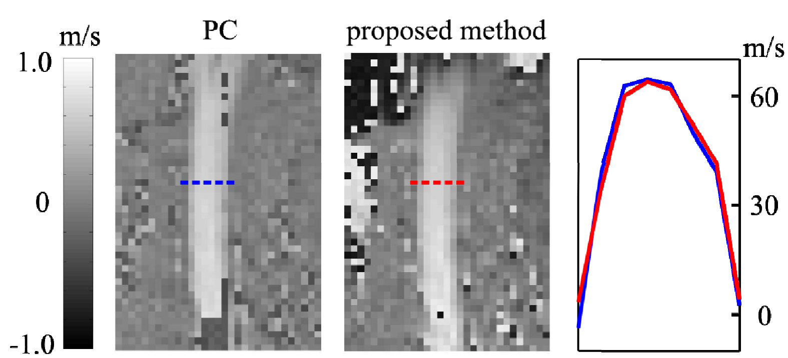

Figure 3: Velocity maps in the common carotid artery in a healthy volunteer, obtained using (left) SSFP phase-contrast MRI using data from two acquisitions, and (center) the proposed method, calculated from only one SSFP acquisition. Velocity profiles along the blue and dotted lines are plotted on the right.

[1] Overall et al, MRM 2002; 48: 890
[2] Markl et al, MRM 2003; 49:945
[3] Grinstead et al, MRM 2005; 54: 138
[4] Nielsen et al, ISMRM 2006; p. 879
[5] Scheffler et al, MRM 2003; 49: 395