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Mitigation of Near-Band Balanced Steady-State Free Precession Through-Plane Flow Artifacts Using Partial Dephasing

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Abstract

Purpose: To mitigate artifacts from through-plane flow at the locations of steady-state stopbands in balanced steady-state free precession (SSFP) using partial dephasing.

Methods: A 60° range in the phase accrual during a TR was created over the voxel by slightly unbalancing the slice-select dephaser. The spectral profiles of SSFP with partial dephasing for various constant flow rates and during pulsatile flow were simulated to determine if partial dephasing decreases through-plane flow artifacts originating near SSFP dark bands while maintaining on-resonant signal. Simulations were then validated in a flow phantom. Lastly, phase-cycled SSFP cardiac cine images were acquired with and without partial dephasing in six subjects.

Results: Partial dephasing decreased the strength and nonlinearity of the dependence of the signal at the stopbands on the through-plane flow rate. It thus mitigated hyper-enhancement from out-of-slice signal contributions and transient-related artifacts caused by variable flow both in the phantom and in vivo. In six volunteers, partial dephasing noticeably decreased artifacts in all of the phase-cycled cardiac cine datasets.

Conclusion: Partial dephasing can mitigate the flow artifacts seen at the stopbands in balanced SSFP while maintaining the sequence's desired signal. By mitigating hyper-enhancement and transient-related artifacts originating from the stopbands, partial dephasing facilitates robust multiple-acquisition phase-cycled SSFP in the heart.

Keywords

flow artifact mitigation; partial dephasing; balanced SSFP (bSSFP); banding

INTRODUCTION

Balanced steady-state free precession (SSFP) is commonly used for cardiac imaging due to its ability to provide high signal-to-noise ratio and desirable myocardium-blood contrast in short acquisition times. In slice-selective acquisitions, inflow-related signal enhancement further increases the blood signal and therefore the contrast. The sequence, however, is

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SUPPORTING INFORMATION

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sensitive to off-resonance, with periodic nulls in its spectral profile leading to dark band artifacts.

At precession frequencies near these steady-state stopbands (henceforth referred to as “near-band”), the fully gradient-refocused nature of balanced SSFP results in unique flow effects not seen in other gradient-echo sequences (1). Spins that flow out of the slice no longer experience radio-frequency (RF) excitation, but still see the gradients. In balanced SSFP, there is no net gradient area to spoil the signal, so these out-of-slice spins may continue contributing to the signal. At the stopbands (which we also refer to as simply “bands”), these spins remain in-phase, potentially causing a manyfold increase in the signal. Furthermore, this out-of-slice signal is a highly nonlinear function of the through-plane flow rate. The implications of near-band through-slice flow in in vivo images include strong enhancement and transient-related artifacts (1-3). In addition, these artifacts make multiple-acquisition balanced SSFP, which is normally used to obtain banding-free balanced SSFP images (4), challenging in the heart. The band falls in the blood pool in at least one of the phase-cycled component images, and standard methods of combining the images do not eliminate the associated artifacts (5-7).

In this study, we investigated using partial dephasing (8) to mitigate near-band flow artifacts. The technique was originally developed to help reduce oscillations during the approach to steady-state (8) and has also been shown to decrease eddy current artifacts (9). In partial dephasing, slightly unbalancing the gradients on the slice-select axis creates a small (up to 120°) range in the phase accrued by spins during a TR within the voxel. Epstein et al. (8) showed that this does not significantly attenuate the steady-state signal of fully balanced SSFP.

We hypothesized that, by causing spins flowing through the slice to see a different amount of phase accrual each TR, partial dephasing would decrease the out-of-slice coherency that flowing spins at the bands experience in standard balanced SSFP. By decreasing the strength and nonlinearity of the dependence of the signal at the bands on the through-plane flow rate, partial dephasing would mitigate the associated artifacts. To test this hypothesis, we used Bloch simulations along with phantom and in vivo experiments. The effects of partial dephasing on artifacts originating from the bands were studied in cardiac images of six human subjects.

METHODS

Simulations

The spectral profiles for various flow rates with a fully balanced sequence and with 30°, 60°, 90°, and 120° of partial dephasing were determined using Bloch simulations similar to those used by Markl and Pelc (10). $\mathbf{m}_n$, the magnetization during the n th TR at the echo time TE = TR/2, can be determined using the recursive equation

$$\mathbf{m}_n = E_{\text{rel}} \mathcal{R}_{\text{off}} \mathcal{R}_{\text{exc}} \mathcal{R}_{\text{pd}} \mathcal{R}_{\text{off}} [E_{\text{rel}} \mathbf{m}_{n-1} + \mathbf{b}_{\text{rec}}] + \mathbf{b}_{\text{rec}} \quad [1]$$

where $\mathbf{b}_{\text{rec}}$ is a vector representing recovery and E_{rel} , $\mathcal{R}_{\text{off}}$, $\mathcal{R}_{\text{pd}}$, and $\mathcal{R}_{\text{exc}}$ are matrices corresponding to different physical processes. More specifically, $\mathbf{b}_{\text{rec}}$ represents T_1 -recovery and E_{rel} is a diagonal matrix representing T_1 - and T_2 - or T_2^* -relaxation during half of a TR. $\mathcal{R}_{\text{off}} = R_z(\phi / 2)$ represents rotation around the z -axis due to off-resonance, caused for instance by susceptibility interfaces or imperfect shimming. ϕ is the phase accrued by a spin during a TR due to off-resonance; it does not include phase imparted by a non-zero net gradient area. RF excitation is represented by $\mathcal{R}_{\text{exc}} = R_y((-1)^n \alpha(z))$, a rotation around the y -axis by the position-dependent flip angle $\alpha(z)$. The RF pulse is assumed to have alternating sign so that on-resonant isochromats are at the center of the passband.

Finally, $\mathcal{R}_{\text{pd}} = R_y\left(\left(\frac{z}{\Delta z}\right)\Delta\theta\right)$ captures the position-dependent rotation from the unbalanced gradient that results if a partial dephasing amount of $\Delta\theta$ is applied over the slice. It distinguishes Equation [1] from a recursive equation characterizing standard balanced SSFP. The position in the slice-selection direction is represented by z , with the center of the slice at $z = 0$. Δz denotes the full-width at half-maximum (FWHM) slice thickness. For example, with $\Delta\theta = 60^\circ$ of partial dephasing, a spin at $z = \Delta z/2$, one edge of the nominal slice, would rotate -30° due to partial dephasing while a spin at the other edge ($z = -\Delta z/2$) would rotate 30° . Note that $\Delta\theta = 0^\circ$ is equivalent to fully balanced SSFP. (See the Appendix for additional notes.)

As with Markl's simulations, inflow, slice profile, and out-of-slice effects were modeled (see Fig. 1 for an illustration). To model through-plane flow, the spins were moved along z each TR before applying Equation [1]. For flow of velocity v , this was done by increasing the z -position of the magnetization by $v\text{TR}$ and replacing the vacated spots on the in-flow side of the slice with fresh magnetization. Time-varying flow waveforms were modeled by shifting the spins by different amounts each TR. We used a z resolution of 0.125 mm to balance adequate resolution of flow velocities and accuracy of the simulations with reasonable computation time. For a TR of 3.5 ms, the resulting flow-velocity resolution is 3.57 cm/s. The flow rate can also be quantified as the percent of the imaged slice replaced by fresh magnetization each TR. The spin replacement percentage Δs is given by

$$\Delta s = \frac{v\text{TR}}{\Delta z} \times 100. \quad [2]$$

(The use of FWHM in this study results in approximately a factor of 2 discrepancy with Markl's definition of the slice thickness.)

Spins were moved through the slice profile of a time-bandwidth-product-2 60° Hamming-windowed sinc with Δz of 6 mm. The imperfect slice profile causes α , the rotation angle due to RF excitation, to be a function of z . Since the RF pulses lead to a spin-echo-like refocusing at TR/2 in balanced SSFP, signal within the slice was assumed to decay with time-constant T_2 . As done in (1) (but not (10)), this was replaced by T_2^* once spins had exited the slice and thus no longer saw the RF excitation.

Because of the very small net gradient area in SSFP with partial dephasing (relative to that in gradient-spoiled gradient-echo sequences), excited spins that flow out of the slice may continue to contribute to the total signal. Therefore, the simulation of a spin should be carried out until the magnetization has decayed sufficiently to be negligible. Since signal exhibits T_2^* -dependence after it has left the slice, we traced spins for $4T_2^*$ after they had flowed out of the slice. This corresponded to a length of $4T_2^*v$ where $\alpha(z) = 0$. If this were greater than 15 cm, 15 cm was used instead to account for the limited extent of cardiac receive coils. The transverse component of the complex sum of the magnetization vectors, both in and out of the slice, was used as an estimate of the signal generated by blood at that through-plane flow rate and precession frequency in a sequence with that amount of partial dephasing. This signal is denoted as $M_{xy}(s, \theta, \phi)$.

To obtain spectral profiles, 144 ϕ values between 0° and 360° were simulated, leading to a spacing of 2.5° . The other parameters used were: TR = 3.5 ms, TE = 1.75 ms, $T_1 = 1000$ ms, $T_2 = 150$ ms (in-slice), and $T_2^* = 75$ ms (out-of-slice). (Relaxation time constants matched those used in (1).) Static spins and flow rates up to 100 cm/s, corresponding to 58.3% spin replacement per TR, were simulated. Simulations were also repeated with TR = 4.6 ms, TE = 2.3 ms, $T_1 = 275$ ms, $T_2 = 260$ ms, and $T_2^* = 130$ ms to mimic our flow phantom experiments, which are used to validate our simulations. Profiles were normalized so that the signal of stationary, on-resonant spins in fully balanced SSFP, $|M_{xy}(0, 0, 0)|$, is 1. Therefore, $M_{xy}(\theta, 0, \phi)$ gives the fraction of the signal of static spins retained by a particular partial dephasing amount.

In addition, we simulated two scalings of a pulsatile flow waveform. For pulsatile flow, the notation $M_{xy}(n, \theta, \phi)$ where n is the repetition number is used instead. For each partial dephasing amount, the level of inconsistency of the spectral profile across different flow rates (for constant flow) or different repetitions (for pulsatile flow) was used as an indication of how much variable flow during image acquisition will cause the signal to vary, thus leading to transient-related artifacts.

The simulation code is provided at https://github.com/adatta92/SSFP_w_Part_Deph_Sim.

Experiments

Since, based on the simulations, the results were not expected to be particularly sensitive to the amount of partial dephasing, all experiments were done only with and without 60° of partial dephasing. To validate the simulations, a constant flow phantom was used. We imaged a slice through the lumen using SSFP sequences with and without 60° of partial dephasing. Partial dephasing was implemented by slightly shortening the slice-select dephaser after readout (see Supporting Fig. S1 for an illustration); the sequences were otherwise identical. To measure the effect of off-resonance, images were acquired with 36 RF-phase-cycling amounts between 0° and 360° . After appropriate phase corrections, the effect of an RF-phase increment of ψ every excitation pulse is equivalent to that of a resonance offset resulting in $\phi = 180^\circ - \psi$ (4). (Note that the basic alternating-RF-sign sequence described by Eq. [1] has $\psi = 180^\circ$.)

The absolute velocity of the flowing liquid was kept constant. Instead, the slice thickness was varied to generate different spin replacement percentages. This mimics different flow velocities through a single slice thickness. The velocity was measured five times over the scan session using a commercial phase-contrast sequence with the velocity encoding set to 100 cm/s. A circular region of interest (ROI) of radius 2.9 mm was selected in the center of the tube. The average velocity in this ROI ranged between 28.8 and 35.1 cm/s during the scan session, with a mean of 31.8 cm/s. Based on this range of velocities, the nominal slice thicknesses of 8, 6, 4, and 2 mm corresponded to spin replacement percentages of 16.6–20.2%, 22.1–26.9%, 33.1–40.4%, and 66.2–80.8%, respectively. To obtain empirical spectral profiles, the mean signal in the ROI, normalized by the slice thickness, was plotted versus RF-phase-cycling amount. Thus, we could compare the measured spectral profiles for these spin replacement percentages with the profiles from simulations with similar values.

All experiments were performed on a 1.5 T scanner (Signa Excite, GE, Milwaukee, WI). The parameters of the SSFP sequence used to image the phantom were 4.6 ms TR, 2.0 ms TE, 12 cm $\times$ 12 cm field of view, and 0.71 mm $\times$ 0.71 mm resolution. Using a TE not equal to TR/2 is not expected to have a significant effect as far as flow artifacts. With the assumption of a single ϕ in each voxel, a shift in the TE should only cause a bulk phase shift due to off resonance and a small change in the amount of relaxation. Note that movement during this time in the z direction due to through-plane flow does not have any effect because the signal is summed over the z direction anyway. The water was doped with copper sulphate to 2.0 mM concentration, resulting in $T_1 \approx 275$ ms and $T_2 \approx 260$ ms (11).

In addition, six healthy volunteers were imaged to assess if partial dephasing mitigates near-band flow artifacts. Cardiac cine images of axial slices were acquired with and without 60° of partial dephasing. Images with 180° and 0° RF phase-cycling were acquired during a single breath-hold. When imaging slices that were not at isocenter, the RF phase-cycling amount was automatically offset based on the slice position to keep the passband in the same place regardless of the partial dephasing amount. If the center of the slice is at z_s , an RF phase-cycling offset of $\left(\frac{z_s}{\Delta z}\right)\Delta\theta$ accounts for the bulk precession from the unbalanced gradient. The linear gradient in the y direction was deliberately offset from the optimal shim to create increased off-resonance. (In healthy volunteers at 1.5 T, bands in the blood pool can often be avoided by good shimming. Here, we add a “deshim” to evaluate the effectiveness of the proposed method.) The scan parameters were 3.4 ms TR, 1.4 ms TE, 24 cm $\times$ 24 cm field of view, and 1.25 mm $\times$ 1.43 mm resolution.

RESULTS

Bloch simulations indicate that the strength and nonlinearity of the dependence of the near-band signal on the through-plane flow rate are substantially reduced by partial dephasing. In fully balanced SSFP, the simulated signal at $\phi = 180^\circ$ (i.e., the center of the stopband) has a highly irregular dependence on the spin replacement percentage (Fig. 2a, c). In comparison, with 60° of partial dephasing, the spectral profiles morph smoothly as a function of the flow rate (Fig. 2b, d). In addition, the maximum amount of hyper-enhancement is decreased by partial dephasing. Without partial dephasing, due to out-of-slice coherency, off-resonant

signal with $s = 58\%$ (for example, blood flowing at 100 cm/s through a 6 mm thick slice if the TR is 3.5 ms) is simulated to be nearly 40 times as bright as stationary on-resonant signal and 20 times as bright as flowing on-resonant signal. With $\theta = 60^\circ$, the intensity of the brightest signal is less than four times that of flowing on-resonant signal. (The results of the constant flow simulations with other amounts of partial dephasing are included in Supporting Fig. S2.) Thus, partial dephasing decreases both the amount of enhancement off-resonance and the complexity of its flow-rate dependence.

Pulsatile flow simulations corroborate that the level of inconsistency of the spectral profile across different flow rates is indicative of how much the signal changes with variable flow. For two scalings of a pulsatile flow waveform, without partial dephasing, the near-band signal changes unpredictably from TR to TR and can be greatly enhanced (Fig. 3b, e). With partial dephasing, although some enhancement remains, the spectral profile remains fairly consistent between consecutive repetitions (Fig. 3c, f). (Note that we plot the spectral profiles during the second cycle of the pulsatile flow waveform to minimize the effects of SSFP start-up transients.) Since changes during the acquisition of an image, such as within a cardiac phase for segmented cardiac imaging, result in transient-related artifacts, our simulations indicate that SSFP with partial dephasing may be less sensitive to near-band flow than fully balanced SSFP. (The evolution of the spectral profile with amounts of partial dephasing other than 60° are included in Supporting Fig. S3.)

On-resonant signal, however, is largely unaffected by partial dephasing. With 60° of partial dephasing, the signal from on-resonant static spins is 98.0% of that in standard balanced SSFP. The amplitude of flowing on-resonant blood in SSFP with partial dephasing is also very similar to that in fully balanced SSFP. (The 2-fold enhancement due to the inflow of fresh magnetization may be desirable (12).) Therefore, simulations suggest that partial dephasing may mitigate near-band flow artifacts while maintaining balanced SSFP's high SNR.

As hypothesized, the decreased hyper-enhancement of near-band signal appears to be because partial dephasing disrupts the coherence of signal from out-of-slice spins. As illustrated in Figure 4 for 58.3% spin replacement, while the magnitude plots of the transverse components of the simulated magnetizations are largely unchanged by partial dephasing, the phase plots look very different between the two conditions shown. Without partial dephasing, spins at the stopband remain in-phase for an extensive distance out of the slice, so all of the signal adds constructively. In contrast, with partial dephasing, at all frequencies, there are phase wraps outside the slice since flowing spins see a different amount of phase accrual each TR. Some out-of-slice contributions may still occur at resonance frequency offsets where, for a limited distance past the slice, ϕ partially cancels the precession from partial dephasing. (For both θ values, the resonance offset that results in the highest total signal is indicated by a dotted red line.) However, since the rotation due to the net gradient area, $\left(\frac{z}{\Delta z}\right)\Delta\theta$, changes linearly with position while the precession from off-resonance is assumed to be a constant, this undoing of the partial dephasing can only occur over a short distance. Therefore, the peak enhancement is substantially reduced

relative to fully balanced SSFP, although it may occur at a range of frequencies surrounding the band. (See the Appendix for more details about the asymmetry of the spectral profile.)

Constant-flow simulations without partial dephasing replicated Markl's results when we matched their parameters (and accounted for the factor of 2 discrepancy in spin replacement percentage from using the FWHM slice thickness). In addition, with and without 60° of partial dephasing, phantom measurements at various spin replacement percentages validate the constant flow simulations—the shapes and relative magnitudes of the empirical spectral profiles are similar to the simulated ones (Fig. 5). Therefore, the model used appears to capture the effects of steady through-plane flow at any constant resonance offset.

The phantom and in vivo images acquired indicate that partial dephasing mitigates the near-band flow artifacts to which balanced SSFP is sensitive. Although the flow phantom had close to constant flow, without partial dephasing, the images of the lumen at frequencies surrounding the stopband¹ have other artifacts in addition to enhancement (Fig. 6a). In comparison, the images with 60° of partial dephasing appear more uniform and do not have signal in the background around the tube (Fig. 6b). In vivo, partial dephasing removed many of the artifacts seen in the standard balanced SSFP images. With the considerable amount of off-resonance applied over the heart, the balanced SSFP images acquired during cardiac phases with significant flow have regions of very hyper-intense signal and transient-related artifacts propagating in the phase-encode direction from the band (Fig. 7a). Partial dephasing substantially mitigated these flow artifacts, although the images still have banding artifacts (Fig. 7b). Dark bands, however, can be addressed by combining multiple RF-phase-cycled SSFP images (13,14). The root-sum-of-squares-combined multiple-acquisition images with partial dephasing (Fig. 8b) do not have any obvious artifacts, in stark contrast with the images without partial dephasing (Fig. 8a). (Images from other cardiac phases and subjects are included in Supporting Fig. S4.) While standard combination methods remove signal nulls, they do not address bright artifacts, which therefore corrupt the final combined balanced SSFP images. Therefore, in our study, partial dephasing enabled multiple-acquisition, banding-free SSFP imaging in the heart by mitigating near-band SSFP flow artifacts.

DISCUSSION AND CONCLUSIONS

Flow phantom and in vivo images indicate that partial dephasing alleviates the near-band through-plane flow artifacts to which slice-selective balanced SSFP is sensitive, and Bloch simulations and phantom results support the proposed mechanisms. (These artifacts are less of a concern in 3-D balanced SSFP because, at physiological flow velocities, spin replacement percentages are low due to the large excited thickness.)

The simulation results were not particularly sensitive to the amount of partial dephasing and suggest that a wide range of θ s (for example, anywhere from just 30° up to 120° of dephasing) may mitigate near-band flow artifacts. Based on the simulations, larger amounts

¹More precisely, these are the images with RF phase-cycling amounts such that the flow tube's resonance frequency falls near the stopband.

of partial dephasing may suppress flow artifacts more effectively than smaller amounts. However, some out-of-slice coherency and therefore hyper-intense signal remains in SSFP with partial dephasing, and, as θ increases, this may be spread over a wider range of frequencies. Partial dephasing also broadens the stopband for static spins (see Supporting Fig. S1(b)). Sixty degrees seems to provide a good balance—it achieves most of the improvement in increased spectral profile consistency and decreased enhancement possible from using partial dephasing (see Supporting Fig. S3) while having minimal effect on the steady-state signal. However, in the flow phantom, the artifacts in standard balanced SSFP images were not confined to a narrow band, as would be expected from the simulations. In addition, in the cine images, partial dephasing did not appear to significantly broaden the enhanced regions in the blood pool. Together, these findings suggest that the tradeoffs of partial dephasing may be less significant in practice.

The simulations and phantom experiments had some limitations. The off-resonance in the slice direction was assumed to be constant to keep the parameter space manageable. In addition, we solely examined the impact of partial dephasing on through-plane flow. The effects in SSFP with partial dephasing of in-plane and oblique flow, which also lead to artifacts in standard balanced SSFP (2,15,16), may merit investigation.

Overall, the flow phantom signal with partial dephasing closely followed the simulated curves. Note that we did not measure the field in the phantom—the location of the band (labeled 180° in Fig. 5) was inferred from the shape of the curves. Without partial dephasing, there were some slight discrepancies. The flow was likely not entirely constant and laminar, as is also suggested by the transient-related artifacts in the images in Figure 6. In addition, the empirical profiles shift to the right relative to the simulated ones as θ increases. This may be because it was challenging to make the field homogeneous and the flow tube perfectly normal to the slice. Given the experimental non-idealities, the phantom results seem to validate the simulations. The flow phantom images also suggest that, even with well-behaved flow, partial dephasing may be noticeably helpful.

In our experience, the artifact mitigation in vivo was robust, with a marked improvement in several cardiac phases of all of the cine datasets we acquired. The first few cardiac phases after the plethysmograph trigger and, in some cases, the last cardiac phase tended to show the most improvement. These cardiac phases, which corresponded to late systole, likely had faster flow than the other cardiac phases. The slices acquired in the different subjects were in different parts of their hearts, but more data should be acquired to systematically analyze if the effectiveness of partial dephasing depends on the slice location.

In conclusion, SSFP with partial dephasing shows promise in being less sensitive to through-plane flow artifacts while maintaining the high SNR efficiency of balanced SSFP. Although it does not address the signal nulls in balanced SSFP's spectral profile, partial dephasing decreases the out-of-slice phase coherency that occurs due to through-plane flow at the stopbands in balanced SSFP. It achieves this by causing spins moving in the through-slice direction to rotate by a different amount each TR. By thus decreasing the strength and complexity of the band signal's dependence on the flow rate, partial dephasing mitigates hyper-enhancement and transient-related artifacts originating at the bands. Partial dephasing

therefore helps to prevent flow artifacts from dark bands from corrupting other parts of the SSFP image, making the banding less disruptive. In addition, we showed that partial dephasing facilitates multiple-acquisition balanced SSFP in the heart, which is banding-free.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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APPENDIX

In this study, we implemented partial dephasing by shortening the gradient lobe after readout and before excitation. Note that partial dephasing can also be applied immediately after excitation and before readout (Supporting Fig. S1(c)). This makes the recursive relation

$$\mathbf{m}_n = E_{\text{rel}} \mathcal{R}_{\text{off}} \mathcal{R}_{\text{pd}} \mathcal{R}_{\text{exc}} \mathcal{R}_{\text{off}} [E_{\text{rel}} \mathbf{m}_{n-1} + \mathbf{b}_{\text{rec}}] + \mathbf{b}_{\text{rec}} \quad [\text{A1}]$$

instead of that in Equation [1]. Analogous to the differences between gradient-spoiled gradient-echo (FFE, FISP, GRASS, FAST) and reversed gradient-spoiled gradient-echo (T_2 -FFE, PSIF, CE-FAST) sequences, these results in slightly different signal profiles (Fig. 9). The effects in vivo, however, were very slight, so we chose to apply the partial dephasing after readout to be consistent with Epstein et al. (8).

Unlike balanced SSFP, SSFP with partial dephasing has an asymmetric spectral profile for flowing spins. This is because they experience either an increasing or a decreasing amount of precession each TR due to the unbalanced gradient, $\gamma \theta / \omega$, depending on the direction of the flow relative to the polarity of the slice-select gradient. The spectral profiles would be reflected across the stopband frequency if the gradient had opposite polarity, the dephaser were slightly lengthened instead of shortened, or the flow were in the opposite direction.

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FIG. 1.

Illustration of the modeling of inflow, slice profile, and out-of-slice effects for an ensemble of spins with different resonance frequencies, similar to that in (10). Inflow was modeled by shifting the spins along each TR by an amount that was proportional to the flow rate and an integer multiple of the z -resolution. The spins on the in-flow side of the slice were replaced with fresh magnetization. The spins within each sub-slice were flipped by different amounts to model the imperfect slice profile, in this case that of a time-bandwidth-product-2 60° Hamming-windowed sinc pulse. (Note that, although the excitation angle may happen to equal the amount of partial dephasing applied across the slice, these two quantities are unrelated.) The nominal slice thickness Δz was defined as the FWHM, and the range within which the excitation angle was nonzero was assumed to not extend past $\pm 1.25 \Delta z$. Spins that had flowed out of the slice experienced no further excitation, but were simulated until their signal had significantly decayed. The total signal was calculated as the complex sum of the transverse magnetization over all z and thus included both in-slice and out-of-slice contributions. Due to the slightly nonzero gradient area in SSFP with partial dephasing, flowing spins were rotated by a different amount at the end of each TR.

FIG. 2.

Simulated spectral profiles of spins with various velocities without partial dephasing (**a** and **c**) and with 60° of partial dephasing (**b** and **d**). No flow to 58.3% spin replacement is shown in (a) and (b) while (c) and (d) zoom in to show 4% to 44% spin replacement. Signals are normalized by the signal of static, on-resonant spins in fully balanced SSFP. The plots are centered on the stopband. The lines trace the signal with the same resonance frequency but varying Δs , highlighting the level of inconsistency of the spectral profile across different flow rates. Without partial dephasing, the stopband signal is a highly non-linear function of the flow rate, while with partial dephasing the signal at all frequencies changes more smoothly as a function of flow velocity.

FIG. 3.

Simulated evolution of the spectral profile of balanced SSFP (**b** and **e**) and of SSFP with 60° partial dephasing (**c** and **f**) during pulsatile flow. The two scalings of a hypothetical flow waveform that were used are shown in (**a**) and (**d**), with red indicating the TRs during which the spectral profile is plotted. We start plotting the spectral profile during the second cycle to avoid SSFP start-up transients. Once again, signals are normalized by the signal of static, on-resonant spins in balanced SSFP, and the plots are centered on the stopband. The lines trace the signal at the same resonance frequency over time, highlighting the level of inconsistency of the spectral profile from TR to TR due to pulsatile flow. (Some lines are made more transparent to decrease occlusion of the near-band signal.) Without partial dephasing, the near-band signal varies very unpredictably and can be greatly hyper-intense, while, with partial dephasing, the signal at all frequencies changes more smoothly.

FIG. 4.

Simulated transverse magnetization of spins flowing through the 6 mm-FWHM slice (in the positive z direction) at 100 cm/s (58.3% spin replacement) without partial dephasing (**a**, **c**, and **e**) and with 60° of partial dephasing (**b**, **d**, and **f**). All plots are centered on the stopband. The net signal (**a** and **b**), found by complex-summing over the slice-select direction, illustrates that SSFP with partial dephasing decreases the amount of near-band enhancement but that it causes the off-resonance frequency corresponding to maximum total signal, indicated by the dotted red line, to shift. While the magnitude plot is largely unchanged by θ (**c** and **d**), the phase plot with partial dephasing (**f**) has phase wraps at all frequencies. This indicates that partial dephasing disrupts the out-of-slice phase coherency at the stopband in balanced SSFP (**e**).

FIG. 5.

Measured spectral profiles without partial dephasing **(a)** and with 60° of partial dephasing **(b)**. The lines are plotted at the spin replacement percentage given by flow of the mean velocity of measurement during scan session through a slice of the specified thickness. The semi-transparent regions show the range of simulated values for the spin replacement percentages corresponding to the range of measured flow velocities. The measured profiles look similar to the corresponding simulated profiles, validating the Bloch simulations.

FIG. 6.

Phase-cycled SSFP images of the flow phantom lumen, with approximately 37% spin replacement per TR, at frequencies surrounding the stopband. From left to right and then top to bottom, the RF phase-cycling of each of the 20 images is offset by 10° from the previous. In the dataset without partial dephasing (**a**), the central images have transient-related flow artifacts that are not present in the images with 60° of partial dephasing (**b**).

FIG. 7.

Systolic SSFP images of two subjects (top and bottom rows) with 0° and 180° RF phase-cycling (right and left images in each pair). The fully balanced SSFP images (a) all contain regions of very hyper-intense signal (solid arrows) and transient-related artifacts propagating in the phase-encode direction (dotted arrows) that are mitigated in the analogous images with 60° of partial dephasing (b). The standard SSFP dark bands (triangles), however, remain. The results from the other four subjects are similar.

FIG. 8.

Root-sum-of-squares combinations of the images in Figure 7. The top and bottom rows correspond to the two subjects in Figure 7. In the images without partial dephasing (**a**), the flow artifacts from the component images persist in the combined images. In contrast, the combined images with 60° of partial dephasing (**b**) are free of these artifacts, as well as not having the signal nulls of the single-acquisition component images in Figure 7. Similar results were observed in the other four volunteers.

FIG. 9.

Simulated spectral profiles of spins with various velocities with partial dephasing applied after readout (a), as done in the rest of this work, and before readout (b). Applying 60° of partial dephasing before readout slightly decreases the peak signal relative to partial dephasing after readout. However, the differences between applying the partial dephasing after and before readout are slight.