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Velocity Encoding with the Slice Select Refocusing Gradient for Faster Imaging and Reduced Chemical Shift-Induced Phase Errors

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Purpose: To investigate a novel phase-contrast MRI velocity-encoding technique for faster imaging and reduced chemical shift-induced phase errors.

Methods: Velocity encoding with the slice select refocusing gradient achieves the target gradient moment by time shifting the refocusing gradient, which enables the use of the minimum in-phase echo time (TE) for faster imaging and reduced chemical shift-induced phase errors. Net forward flow was compared in 10 healthy subjects ($N=10$) within the ascending aorta (aAo), main pulmonary artery (PA), and right/left pulmonary arteries (RPA/LPA) using conventional flow compensated and flow encoded (401 Hz/px and TE=3.08 ms) and slice select refocused gradient velocity encoding (814 Hz/px and TE=2.46 ms) at 3 T.

Results: Improved net forward flow agreement was measured across all vessels for slice select refocused gradient compared to flow compensated and flow encoded: aAo vs. PA ($1.7\% \pm 1.9\%$ vs. $5.8\% \pm 2.8\%$, $P=0.002$), aAo vs. RPA+LPA ($2.1\% \pm 1.7\%$ vs. $6.0\% \pm 4.3\%$, $P=0.03$), and PA vs. RPA+LPA ($2.9\% \pm 2.1\%$ vs. $6.1\% \pm 6.3\%$, $P=0.04$), while increasing temporal resolution (35%) and signal-to-noise ratio (33%).

Conclusion: Slice select refocused gradient phase-contrast MRI with a high receiver bandwidth and minimum in-phase TE provides more accurate and less variable flow measurements through the reduction of chemical shift-induced phase errors and a reduced TE/repetition time, which can be used to increase the temporal/spatial resolution and/or reduce breath hold durations. *Magn Reson Med* 71:2014–2023, 2014. © 2013 Wiley Periodicals, Inc.

Key words: phase-contrast MRI; chemical shift; velocity encoding; flow quantification

Measurement of volumetric blood flow rates using phase-contrast MRI (PC-MRI) in the cardiovascular system is a primary clinical application of magnetic resonance imaging. However, quantitative volumetric

imaging of blood flow with PC-MRI is limited by systematic errors in phase (1–4) leading to clinically significant flow errors, which dampen clinical enthusiasm for the technique.

In particular, our previous work has shown that chemically shifted perivascular fat can significantly corrupt PC-MRI flow measurements (5). In that work, we characterized the impact of chemical-shifted lipid signals in PC-MRI and defined a coherent strategy to reduce chemical shift errors. Briefly, the amount of chemically shifted fat pixels that shift into the vessel can be reduced by increasing receiver bandwidth (BW) at the expense of signal-to-noise ratio (SNR). Second, an in-phase echo time (TE; TE_{IN}) will ensure that fat and water resonances are in-phase, which minimizes the resulting errors in the calculated velocity.

Chemical shift errors are more challenging to address at 3 T, when compared with 1.5 T, because of the larger frequency difference between water and fat, which leads to a larger spatial shift of fat for a given BW. Furthermore, the shorter minimum in-phase TE (TE_{IN,MIN}) at 3 T (2.46 ms for $\delta_{3T} \sim 420$ Hz and scanner reported $B_0 = 2.89$ T) is shorter than at 1.5 T (4.76 ms for $\delta_{1.5T} \sim 210$ Hz and scanner reported $B_0 = 1.49$ T), and may not be routinely achievable with standard flow-encoding methods at conventionally targeted velocity-encoding strengths (VENCs).

Herein, we describe and evaluate a method for flow encoding in PC-MRI that uses the slice select gradient and a time-shifted refocusing gradient lobe to generate the target gradient moments, which shortens the TE/repetition time (TR) (increases temporal resolution) and enables the use of TE_{IN,MIN} at 3 T for time-efficient reduction of chemical shift-induced phase errors.

THEORY

For a typical PC-MRI experiment, two velocity-encoded acquisitions with different first moments are necessary to encode flow along a single direction (6–8). All encoding schemes are designed to null the zero gradient moment for both encoding steps, $M_{0,1} = M_{0,2} = 0$, to refocus the position-related phase, but with a net difference in first moments to yield velocity-dependent phase with no dependence on nonvelocity background phase,

$$\Delta\phi = \gamma v_z (M_{1,1} - M_{1,2}) = \gamma v_z M_1. \quad [1]$$

In Eq. [1], $\Delta\phi$ is the phase difference between the two velocity-encoded images, γ is the gyromagnetic ratio, v_z is the velocity in the through-plane direction, and $M_{1,1}$

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FIG. 1. Pulse sequence diagrams for our previous chemical shift-optimized PC-MRI sequence using a conventional FCFE velocity-encoded sequence at HBW and TE_{IN} (a), a conventional bipolar velocity-encoded sequence at HBW and TE_{IN} , which uses a negative and positive velocity-encoded waveform along slice select axis (b), and SSRG velocity-encoded sequence at HBW and $TE_{IN,MIN}$ (c). The SSRG sequence conducts velocity encoding using the slice selection and corresponding refocusing gradients that inherently have a nulled zero gradient moment ($M_0=0$) and a nonzero first moment ($M_1 \neq 0$), which provides velocity sensitivity. For the second measurement, the SSRG is shifted in time to produce the targeted VENC as detailed in Figure 2 and Eq. 8. The SSRG sequence permits the use of $TE_{IN,MIN}$, whereas the other conventional sequences cannot achieve this TE.

and $M_{1,2}$ are the first gradient moments for encoding steps 1 and 2. The VENC, defined as the velocity that produces a phase shift of $\pm\pi$ radians, is determined by ΔM_1 for any encoding strategy:

$$VENC = \frac{\pi}{\gamma|\Delta M_1|}. \quad [2]$$

The target ΔM_1 is conventionally achieved by using either bipolar velocity-encoding gradients or a set of flow compensated and flow encoded (FCFE) velocity-sensitizing gradients. For bipolar encoding, $M_{0,1}=M_{0,2}=0$ and the equal and opposite gradient polarities yield $M_{1,1}=X/2$ and $M_{1,2}=-X/2$ (7,8). For FCFE-encoding strategies, $M_{0,1}=M_{0,2}=0$, while $M_{1,1}=0$ and $M_{1,2}=X$ (6,9). Although not widely appreciated bipolar encoding is more time efficient for low VENCs and FCFE is more time efficient for mid to high VENCs. Furthermore, FCFE provides reduced signal loss owing to intravoxel coherent velocity-related phase dispersion (10,11), such as in the presence of shear or complex flow, and reduced ghosting artifacts in the phase encode direction arising from large M_1 -induced velocity-related phase shifts (8).

However, in the presence of perivascular fat, the chemical shift difference between water and lipids can give rise to non-velocity related phase, which can be minimized by selection of optimal timings, to keep the water and lipid

signals in-phase (5). The design of the flow-encoding gradients is thus further constrained by their total duration if the criteria for minimum in-phase TE are to be satisfied: $TE_{IN,MIN} = 2.46$ ms at 3 T. Bernstein et al. (12) previously defined strategies for minimizing TE times in PC-MRI sequences for both bipolar and FCFE encoding and our proposed methods make further reductions in TE possible.

Previously, time-efficient velocity encoding without the need for time-consuming bilobe encoding gradients has been achieved by taking advantage of the intrinsic first moment of the slice-selection gradient waveform and refocusing lobe. Markl et al. (13) previously developed a balanced steady-state free precession phase-contrast pulse sequence that used a slice-selection gradient with alternating polarity for each encoding step, thereby combining the velocity encoding with the slice selection and slice select refocusing gradient (SSRG). Thompson and McVeigh (14) used a time shift of the refocusing lobe and bipolar flow-encoding lobe to control the VENC without altering the slice-selection gradient. Herein, we adopt similar principles to significantly reduce the time needed to achieve velocity encoding over a range of clinically useful VENCs by using a time shift of the refocusing lobe and no other flow-encoding lobes. Our proposed technique is referred to throughout as the SSRG sequence.

The most common implementations of flow-encoding gradients are shown for FCFE gradient waveforms in Figure 1a and bipolar gradient waveforms in Figure 1b, whereas our proposed SSRG velocity-encoding technique is shown in Figure 1c.

For our SSRG method, the relationship between the gradient parameters and the resulting zero, M_0 , and first

FIG. 2. Slice select and refocusing gradient (SSRG) lobes for the SSRG PC-MRI sequence. The slice thickness and RF duration determine $M_{0,S}$ of the slice select gradient lobe from $t=0$. A time-efficient triangular refocusing gradient lobe is constructed such that $M_{0,R}=M_{0,S}$. The refocusing gradient lobe for the second measurement is shifted in time by Δt to produce the target VENC. $M_{0,S}$ is the area of the slice select gradient; G_S is slice select gradient amplitude; t_S is the duration of the slice select gradient; $M_{0,R}$ is the area of the refocusing gradient; r_R is the duration of the refocusing gradient ramp from 0 to G_R ; and Δt is the time shift of the refocusing gradient needed to produce the ΔM_1 associated with the target VENC.

moments, M_1 , and thus the VENC, can be determined analytically. Parameters characterizing a typical SSRG waveform for a symmetric radiofrequency (RF) pulse are shown in Figure 2. For any given RF pulse, the amplitude, G_S , is determined by the desired slice thickness, whereas the length of the slice select gradient lobe, t_S , is determined by the duration of the RF pulse. The M_0 of the slice select gradient lobe, $M_{0,S}$, and refocusing gradient lobe, $M_{0,R}$, must be equal ($M_{0,S} = M_{0,R}$) to ensure that all excited spins are refocused over the thickness of the slice after excitation:

$$M_{0,R} = G_S t_S. \quad [3]$$

The amplitude of the refocusing gradient lobe, G_R , can be determined by using the minimum available gradient rise time of the system, r_R , and the result of Eq. 3:

$$G_R = \frac{M_{0,R}}{r_R}. \quad [4]$$

The first moment for a symmetric gradient waveform (i.e., the refocusing lobe) centered at any point in time, t , can be determined by $M_1 = M_0 t$. Thus, for the first encoding step, with the refocusing gradient lobe centered at $t = t_S + r_R$, the first moment of the refocusing gradient lobe can be calculated as $M_{0,R} (t_S + r_R)$. The nonzero first moment of the entire gradient waveform for the first encoding step, $M_{1,1}$, is thus given by

$$M_{1,1} = M_{1,S} + M_{1,R} = M_{1,S} - M_{0,R}(t_S + r_R). \quad [5]$$

For the second encoding step, the refocusing gradient lobe is shifted in time by Δt and is thus centered at $t = t_S + r_R + \Delta t$, resulting in a first moment of $M_{0,R} (t_S + r_R + \Delta t)$. The nonzero first moment of the entire gradient waveform for the second encoding step, $M_{1,2}$, can be determined as

$$M_{1,2} = M_{1,S} + M_{1,R} = M_{1,S} - M_{0,R}(t_S + \Delta t + r_R). \quad [6]$$

The slice select gradient lobe is identical for the two encoding steps and thus its first moment contribution cancels when calculating the net first moment, ΔM_1

$$\Delta M_1 = M_{1,1} - M_{1,2} = M_{0,R} \Delta t. \quad [7]$$

Substituting Eq. 7 in Eq. 2 above, the time shift for the target VENC is

$$\Delta t = \frac{\pi}{\gamma M_{0,R} \text{VENC}}. \quad [8]$$

METHODS

Comparison of Velocity-Encoding Strategies

The minimum available TE as a function of VENC was determined for FCFE, bipolar, and SSRG sequences. The TE and VENC were determined at both low BW (LBW) = 401 Hz/px and high BW (HBW) = 814 Hz/px with all other

parameters held constant (192×120 encoding matrix, $1.6 \text{ mm} \times 1.6 \text{ mm}$ spatial resolution, 5 mm slice thickness, 30° flip angle, four views-per-segment, $500 \mu\text{s}$ RF pulse width, and a maximum slew rate of 157 mT/m/ms).

Preclinical Evaluation in Normal Volunteers

A preclinical evaluation of SSRG was performed in 10 normal volunteers ($N=10$) to show that SSRG suppresses chemical shift effects while shortening the TE/TR compared to FCFE. All imaging was performed on a Siemens Trio 3 Tesla system (Siemens Medical Solutions, Erlangen, Germany) with 40 mT/m maximum gradient amplitude and 200 T/m/s maximum slew rate. Our university's Institutional Review Board approved the study and informed consent was obtained for each subject before MRI scanning. Subjects were positioned head first in the supine position on the scanner bed and imaged using an anterior six-element body matrix coil and a posterior six-element spine matrix coil for signal reception. Blood flow was measured using PC-MRI in the ascending aorta (aAo), main pulmonary artery (PA), and right/left pulmonary arteries (RPA/LPA) of 10 ($N=10$) volunteers (two females and eight males; age 27.6 ± 4.0 years) with no previous history of cardiovascular disease. High-resolution black blood turbo spin echo images were also acquired with and without fat saturation (15,16) during end systole to define the presence (or absence) of perivascular fat for the vessel of interest at the same slice location used for PC-MRI flow measurements.

The imaging plane for aAo flow was located distal to the aortic valve and coronary ostia. The imaging plane for flow in the PA was located downstream from the pulmonary valve and proximal to the first bifurcation. The imaging planes for the LPA and RPA were located $\sim 1 \text{ cm}$ distal to the pulmonary bifurcation. All imaging planes were prescribed on bSSFP cine images during end systole with end-expiratory breath holds.

PC-MRI flow measurements were obtained using a FCFE cine gradient echo phase-contrast sequence with retrospective ECG gating: mid-phase TE ($\text{TE}_{\text{MID}} = 3.08 \text{ ms}$), which orients the fat vector to be approximately perpendicular to the blood vector, $\text{TR} = 6.04 \text{ ms}$, 192×132 matrix, $1.6 \text{ mm} \times 1.6 \text{ mm} \times 5 \text{ mm}$ acquisition voxel, 30° imaging flip angle, 401 Hz/pixel receiver BW (LBW), four views-per-segment (17), a total scan time of 20 heartbeats, a temporal resolution of 48.3 ms, 20–24 cardiac phases (heart rate dependent) reconstructed during one end-expiratory breath hold with retrospective ECG gating, and GRAPPA (18) parallel imaging with an acceleration factor of 2 and 24 reference (central) k -space lines. Two-dimensional through-plane velocity encoding was performed using VENC of 200 cm/s for all flow territories. Although it is possible to fine-tune the VENC for each subject and each particular vessel, this will introduce changes in the TE and TR, which can be confounding when making comparisons in a controlled study. Therefore, the VENC was kept high enough to reduce the risk of aliasing and was therefore held constant for all subjects and all vessels. Fractional echo was not used for any sequence.

Both $TE_{MID} = 3.08$ ms and $LBW = 401$ Hz/px were found to be close approximations to commonly used values in clinical scans found in current PC-MRI literature (19–21). This combination of TE and BW sensitizes the measurements to the effects of chemical shift. Measurements were also obtained using the SSRG velocity-encoding sequence with identical parameters as described above, except for the following changes: $TE_{IN,MIN}/TR = 2.46/4.46$ ms, 814 Hz/px receiver BW (HBW), a temporal resolution of 35.65 ms, and 27–32 reconstructed cardiac phases (heart rate dependent). Additional data were also acquired using FCFE with our previously defined optimized chemical shift protocol: $TE_{IN}/TR = 4.92/6.91$ ms, 814 Hz/pixel receiver BW (HBW), a temporal resolution of 55.3 ms, and 17–21 reconstructed cardiac phases. Both FCFE at $HBW + TE_{IN}$ and SSRG at $HBW + TE_{IN,MIN}$ theoretically provide the same insensitivity to chemical shift-induced phase errors, but SSRG at $HBW + TE_{IN,MIN}$ provides increased SNR and a shorter TR (i.e., improved temporal resolution or shorter breath-hold durations) owing to the TE/TR reduction.

Image Processing

Data were processed offline using MATLAB (The MathWorks, Natick, MA) and a DICOM viewing tool (Osirix, www.osirix-viewer.com). Eddy current background phase errors were measured and corrected, for all sequences, using a stationary phantom (2,22,23) after each exam. For quantitative flow assessment at each vessel territory, a region-of-interest (ROI) was drawn in Osirix around the contours of each vessel boundary as indicated in the magnitude images. The same ROIs were imported into the eddy current correction images for correction of background phase errors. The resulting ROI information was then exported from Osirix to MATLAB for quantitative velocity and flow analysis. The measured net forward flow of blood was computed by scaling the mean ROI signal intensity (velocity-related phase shift) by $VENC/\pi$ (cm/s). The resulting mean ROI velocities were then multiplied by the area of the ROI (cm²) to calculate the flow rate (mL/s) and finally integrated over the cardiac cycle to yield the net forward flow results (mL).

Data Analysis

Chemical Shift Effects on Net Forward Flow

To analyze the effects of chemical shift on net forward flow, measurements were made in the aAo, PA, RPA, and LPA for FCFE with $LBW + TE_{MID}$ and $HBW + TE_{IN}$ as well as SSRG with $HBW + TE_{IN,MIN}$.

Internal Blood Flow Consistency

An analysis of the internal blood flow consistency was conducted by comparing the measured net forward flow in the aAo and PA, aAo and RPA + LPA, and the PA and RPA + LPA for FCFE with $LBW + TE_{MID}$ and SSRG with $HBW + TE_{IN,MIN}$. In the absence of shunts or regurgitant flow (neither of which is expected within our normal volunteer population), we expect that the blood flow in the $PA = 1.05$ aAo [accounting for coronary flow (24)] = RPA + LPA. A two-sample *t*-test with Holm–Sidak post

hoc correction was used to measure the statistical significance of the differences in the measured net forward flow between the two sequences.

Eddy Current Comparison

Eddy current-induced velocity offsets were measured by copying the manually contoured ROIs from each volunteer and vessel to the eddy current correction images. The eddy current-induced velocity offsets were compared by averaging the mean velocity from each ROI through time for SSRG at $HBW + TE_{IN,MIN}$ and FCFE at $LBW + TE_{MID}$ and $HBW + TE_{IN}$.

The magnitude of chemical shift- and eddy current-induced flow errors was compared. Chemical shift-induced flow errors were obtained by calculating the mean, standard deviation, and minimum/maximum of the flow error (mL) between the aAo and PA, aAo and RPA + LPA, and the PA and RPA + LPA, between FCFE at $LBW + TE_{MID}$ and SSRG at $HBW + TE_{IN,MIN}$. Eddy current-induced flow errors were similarly calculated as the flow difference arising from measurements with and without eddy current correction for both sequences. A two-sample *t*-test was used to measure the statistical significance of the differences in flow errors arising from eddy currents and chemical shift.

Pulmonary to Systemic Blood Flow Ratios

Pulmonary to systemic blood flow ratios (Qp/Qs , PA flow divided by aAo flow) were calculated for each volunteer. A two-sample *t*-test with Holm–Sidak post hoc correction was used to measure the statistical significance of the differences in the measured Qp/Qs ratios for FCFE with $LBW + TE_{MID}$ and SSRG with $HBW + TE_{IN,MIN}$.

SNR Comparison

A quantitative analysis of the measured SNR for FCFE with $HBW + TE_{IN}$ and $LBW + TE_{MID}$ and SSRG with $HBW + TE_{IN,MIN}$ was conducted by placing two ROIs within the magnitude images to produce an estimate of the signal and the noise. One set of ROIs was contoured to the aAo, PA, RPA, or LPA to obtain a measurement of the mean signal for each cardiac phase. Another set of ROIs was placed outside the subject's body to obtain an estimate of the background signal standard deviation (noise) for each cardiac phase. The noise ROIs were placed near the top left portion of the images in a region that was devoid of any artifacts as observed under window/level extremes. The mean $\pm$ standard deviation SNR for each vessel and each subject was computed as the mean of the mean signal intensity divided by the standard deviation for each cardiac phase for each sequence. The mean $\pm$ standard deviation of the SNR for each vessel territory was compared between sequences.

RESULTS

Comparison of Velocity-Encoding Strategies

Figure 3 shows the minimum available TE (ms) plotted as a function of the available VENC (cm/s) for FCFE, bipolar, and SSRG velocity-encoding strategies at both LBW (Fig. 3a) and HBW (Fig. 3b). SSRG is the only

FIG. 3. Minimum achievable TE (ms) plotted as a function of the target VENC (cm/s) for conventional bipolar velocity encoding (red), conventional FCFE velocity encoding (green), and SSRG velocity encoding (blue) shown for (a) LBW = 401 Hz/px and (b) HBW = 814 Hz/px. The SSRG sequence is the only encoding strategy that permits the use of $TE_{IN,MIN}$ = 2.46 ms, which occurs for VENCs $\geq$ 190 cm/s. The SSRG sequence also permits the shortest TE for all VENCs $\geq$ 165 cm/s. The FCFE sequence yields the shortest TE for 35 cm/s < VENC < 165 cm/s. The bipolar sequence permits the shortest TE for VENCs $\leq$ 35 cm/s. As the VENC increases, a theoretical minimum TE is reached because of limitations associated with the plateau of the readout gradient overlapping with the refocusing gradient. The SSRG sequence can generate a larger M_1 (lower VENC) within this time compared with the other encoding strategies, thus permitting the use of a shorter TE. All other parameters were held constant (192×120 encoding matrix, $1.6 \text{ mm} \times 1.6 \text{ mm} \times 5 \text{ mm}$ acquisition voxel, 30° flip angle, four views-per-segment, $500 \mu\text{s}$ RF pulse width, and a maximum slew rate of 157 mT/m/ms). [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

encoding strategy that permits the use of $TE_{IN,MIN}$ = 2.46 ms at HBW, which occurs for VENCs $\geq$ 190 cm/s. The minimum achievable TE at LBW is 2.84 ms, which can only be reached with SSRG for VENCs $\geq$ 190 cm/s. SSRG also permits the shortest TE for all VENCs $\geq$ 165 cm/s. FCFE yields the shortest TE for 35 cm/s < VENC < 165 cm/s. The bipolar sequence provides the use of the shortest TE for VENCs $\leq$ 35 cm/s. FCFE allows shorter mini-

um TEs compared to the bipolar sequence for the VENCs used in this study; therefore, our new SSRG velocity-encoding sequence was only compared to FCFE.

In Vivo Studies

Chemical Shift Effects on Net Forward Flow

Figure 4 shows the difference in the measured net forward flow data from the aAo, PA, RPA, and LPA for FCFE with LBW + TE_{MID} and SSRG with HBW + $TE_{IN,MIN}$, which qualitatively illustrates chemical shift-induced flow errors. The turbo spin echo images with and without fat saturation were used to detail the presence or absence of perivascular fat for each vessel in each volunteer (Fig. 5). In the presence of perivascular fat, FCFE with LBW + TE_{MID} led to an overestimation of the net forward flow compared to SSRG with HBW + $TE_{IN,MIN}$ (aAo: $7.2\% \pm 1.6\%$ vs. $1.6\% \pm 0.3\%$; PA: $8.5\% \pm 2.7\%$ vs. 3.8% ; RPA: $11.4\% \pm 2.3\%$ vs. 3.3% ; and LPA: $8.8\% \pm 2.8\%$ vs. $2.3\% \pm 2.8\%$). The mean difference, across all volunteers and vessels, between the measured net flow for FCFE with HBW + TE_{IN} and SSRG with HBW + $TE_{IN,MIN}$ was 1.1 mL (2.1%).

Internal Blood Flow Consistency

Table 1 shows an analysis of the internal consistency for net forward flow measurements made in the aAo, PA, RPA, and LPA from our preclinical evaluation of 10 normal volunteers ($N = 10$). Included in the analysis are the mean percent agreement and the minimum and maximum percent agreement (shown in brackets). In all three territory comparisons (aAo vs. PA, aAo vs. RPA + LPA, and PA vs. RPA + LPA) a statistically significant difference ($P < 0.05$) between the measured net forward flow for FCFE with LBW + TE_{MID} and SSRG with HBW + $TE_{IN,MIN}$ is observed. This indicates significant improvement in the internal consistency of flow measures through the reduction of chemical shift-induced phase errors using SSRG with HBW + $TE_{IN,MIN}$. A comparison of the measured net flow between FCFE with HBW + TE_{IN} and SSRG with HBW + $TE_{IN,MIN}$ showed no significant difference.

Figure 6 shows an analysis of the internal consistency for net forward flow (aAo vs. PA, aAo vs. RPA + LPA, and PA vs. RPA + LPA) measured by FCFE with LBW + TE_{MID} and SSRG with HBW + $TE_{IN,MIN}$ with and without eddy current correction. SSRG with HBW + $TE_{IN,MIN}$ improves the intrasubject forward flow agreement for every subject. Eddy current-corrected SSRG with HBW + $TE_{IN,MIN}$ leads to better intrasubject flow agreement on average compared to FCFE with LBW + TE_{MID} .

Table 2 shows the importance of comparing relative and absolute blood flow comparisons for both SSRG with HBW + $TE_{IN,MIN}$ and FCFE with LBW + TE_{MID} from a single volunteer with observed perivascular fat.

Eddy Current Comparison

Eddy current-induced velocity offsets for SSRG with HBW + $TE_{IN,MIN}$, FCFE with LBW + TE_{MID} , and FCFE with HBW + TE_{IN} were 0.6 ± 0.37 , 0.4 ± 0.19 , and 0.52 ± 0.27 cm/s, respectively. The average magnitude of chemical shift-induced flow errors was 2.9 ± 1.6 mL [0.0,

FIG. 4. Differences in measured net forward flow between the conventional FCFE velocity-encoding sequence at $\text{LBW} + \text{TE}_{\text{MID}}$ and the SSRG sequence at $\text{HBW} + \text{TE}_{\text{IN,MIN}}$ expressed as a percentage. Vessels with observed perivascular fat on turbo spin echo black blood images are shown to the left of the vertical dashed line, whereas vessels with minimal perivascular fat are shown to the right of the vertical dashed line. The data are presented based on the magnitude of the net forward flow difference (from greatest to least). Results were obtained in the (a) aAo, (b) main PA, and the (c) RPA and (d) LPA. In the presence of perivascular fat, FCFE at $\text{LBW} + \text{TE}_{\text{MID}}$ leads to an overestimation in net forward flow (theoretically and empirically) compared with SSRG at $\text{HBW} + \text{TE}_{\text{IN,MIN}}$ sequence for all vessel territories except volunteer number 10 in the LPA. Chemical shift-induced phase errors are largest for low-pressure vessels (PA, RPA, and LPA) compared with the aAo that has a thicker vessel wall, thus limiting the perivascular fat signal from spatially shifting into the vessel lumen. In the presence of minimal perivascular fat, the difference between the two sequences is much lower.

6.4 mL] (mean $\pm$ SD [min, max]). The average magnitude of eddy current-induced flow errors was 1.56 ± 1.16 mL [0.03, 4.6 mL].

Pulmonary to Systemic Blood Flow Ratios

The Qp/Qs ratios for SSRG with $\text{HBW} + \text{TE}_{\text{IN,MIN}}$ were lower and significantly different than FCFE with $\text{LBW} + \text{TE}_{\text{MID}}$ (1.00 ± 0.02 vs. 1.05 ± 0.04 , $P < 0.005$).

SNR Comparison

The average measured SNR for SSRG with $\text{HBW} + \text{TE}_{\text{IN,MIN}}$, FCFE with $\text{HBW} + \text{TE}_{\text{IN}}$, and FCFE with $\text{LBW} + \text{TE}_{\text{MID}}$ was 58.5 ± 24.8 , 87.7 ± 40.8 , and 116.8 ± 54.4 , respectively.

Sequence Timing Comparison

Table 3 shows a comparison of the timing parameters for FCFE with $\text{LBW} + \text{TE}_{\text{MID}}$ and $\text{HBW} + \text{TE}_{\text{IN}}$ as well as

SSRG with $\text{LBW} + \text{TE}_{\text{MIN}}$ and $\text{HBW} + \text{TE}_{\text{IN,MIN}}$ for a VENC of 200 cm/s.

DISCUSSION

Comparison of Velocity-Encoding Strategies

In Figure 3, as the VENC increases, a theoretical minimum TE is reached because of limitations associated with the plateau of the readout gradient overlapping with the refocusing gradient. SSRG can encode larger velocities within this time, compared to the other encoding strategies, thus permitting the use of a shorter TE. For a fixed slice select gradient and RF pulse, smaller VENC values require a Δt that causes the refocusing gradient lobe to overlap with the readout gradient plateau for $\text{TE}_{\text{IN,MIN}}$, thus causing an increase in the minimum achievable TE.

In Vivo Studies

Chemical Shift Effects on Net Forward Flow

In Figure 4, it can be seen that chemical shift-induced phase errors are largest in the PA, RPA, and LPA

FIG. 5. Black blood turbo spin-echo images highlight the presence of perivascular fat around the aAo (a), main PA (b), RPA (c), and LPA (d). TV, tricuspid valve; MV, mitral valve; dAo, descending aorta; LA, left atrium; LV, left ventricle.

compared to the aAo. For low-pressure vessels (PA, RPA, and LPA) the vessel wall is thicker, thus limiting the complex signal from perivascular fat from spatially shifting into the vessel lumen. In the presence of minimal perivascular fat, the difference between the two sequences is much lower. Only one volunteer lacked perivascular fat in the PA and another volunteer lacked perivascular fat in the RPA. Only once did FCFE with LBW + TE_{MID} lead to an underestimation of the net forward flow compared to SSRG with HBW + TE_{IN,MIN}.

Internal Blood Flow Consistency

Table 1 shows a statistically significant difference ($P < 0.05$) between the measured net forward flow for FCFE with LBW + TE_{MID} and SSRG with HBW + TE_{IN,MIN} in all three territory comparisons (aAo vs. PA, aAo vs. RPA + LPA, and PA vs. RPA + LPA). This indicates significant improvement in the internal consistency of flow

measures through the reduction of chemical shift-induced phase errors using SSRG with HBW + TE_{IN,MIN}.

Internal consistency measures can only reflect a relative agreement in net forward flow, whereas for PC-MRI to provide the most clinically useful results, absolute flow measurements are preferred. In Table 2, FCFE with LBW + TE_{MID} shows a small intrasequence difference of ~ 1 mL (1.4%). Importantly, SSRG with HBW + TE_{IN,MIN} shows a similar result (Δ intraflow < 1 mL). However, when the intersequence flow differences are compared FCFE with LBW + TE_{MID} shows an overestimation of blood flow by ~ 8 mL (9.3%). In this work, FCFE with LBW + TE_{MID} led to an overestimation of blood flow in all but one vessel territory (39 of 40) in 10 volunteers, which is also consistent with our previous results (5). Although this study reports a consistent overestimation of flow when using LBW + TE_{MID}, the magnitude and sign of the flow discrepancy depend on the superposition of the complex fat and water vectors. For this work, the use of LBW causes more fat pixels to partial volume

Table 1

Inpatient Percent Flow difference from the Preclinical Evaluation of 10 Normal Volunteers ($N = 10$) Expressed as a Mean $\pm$ Standard Deviation [Minimum, Maximum]

	FCFE LBW + TE _{MID}	SSRG HBW + TE _{IN,MIN}	FCFE HBW + TE _{IN}
aAo vs. PA	5.8% $\pm$ 2.8% ^a [0.98, 8.9%]	1.7% $\pm$ 1.9% [0.16, 2.8%]	1.5% $\pm$ 0.9% [0.20, 2.1%]
aAo vs. RPA + LPA	6.0% $\pm$ 4.3% ^a [0.85, 9.8%]	2.1% $\pm$ 1.7% [0.60, 2.5%]	2.1% $\pm$ 1.1% [0.85, 2.4%]
PA vs. RPA + LPA	6.1% $\pm$ 6.3% ^a [0.11, 7.6%]	2.9% $\pm$ 2.1% [0.57, 2.2%]	2.8% $\pm$ 2.1% [0.77, 2.4%]

^a $P < 0.05$ showing a statistical significant difference compared with SSRG HBW + TE_{IN,MIN}. No significant difference was measured between SSRG HBW + TE_{IN,MIN} and FCFE HBW + TE_{IN}.

FIG. 6. Net forward flow measurement error between the conventional FCFE velocity-encoding sequence at $LBW + TE_{MID}$ without (orange) and with (blue) eddy current correction compared with the SSRG sequence at $HBW + TE_{IN,MIN}$ without (green) and with (red) eddy current correction. Data from individual subjects ($N = 10$) are connected to show that eddy current-corrected SSRG at $HBW + TE_{IN,MIN}$ improves the intrasubject forward flow agreement for every subject compared with eddy current-corrected FCFE with $LBW + TE_{MID}$. Correcting for eddy currents further improves the agreement. The box plot shows the median and 25th and 75th percentiles and the error bars show the 95% confidence intervals. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

with blood pixels within the vessel, whereas the use of TE_{MID} causes the fat vector to be oriented approximately perpendicular to the blood vector. Together, $LBW + TE_{MID}$ leads to an increase in the measured phase (flow), whereas our chemical shift-optimized sequence reduces the extent of fat pixels that partial volume with blood pixels by using a HBW and orients the fat vector more closely with that of slow-flowing blood near the vessel wall by using TE_{IN} . Table 2 shows that FCFE with $LBW + TE_{MID}$ demonstrates excellent relative agreement in blood flow (Δ intraflow) between the PA and the RPA + LPA, but a poor agreement in absolute blood flow (Δ interflow) because of chemical shift effects arising from the presence of perivascular fat surrounding these vessels for this volunteer. Similar results are observed in flow phantom experiments, where the measured flow is known (data not shown). These examples highlight why improvements need to be made to make PC-MRI an absolute method for quantifying blood flow and also underscore an important source of variability in clinical PC-MRI measurements.

Eddy Current Comparison

Eddy current-induced velocity offsets for SSRG with $HBW + TE_{IN,MIN}$ were greater than both FCFE with $LBW + TE_{MID}$ and $HBW + TE_{IN}$, but importantly just within the 0.6 cm/s threshold of acceptable eddy current-induced velocity offsets (19). Eddy current background phase error corrections significantly improve the flow accuracy of PC-MRI (2,25,26). In this study, we have shown that the average magnitude of chemical shift-induced flow errors are almost two times larger and

statistically different ($P < 5e-5$) than eddy current effects. Therefore, for studies in which flow accuracy is important both eddy currents and chemical shift-induced phases errors need to be minimized.

Pulmonary to Systemic Blood Flow Ratios

Relative to SSRG, FCFE measurements of Q_p/Q_s are higher on average by 0.05. Chemical shift effects primarily account for the 0.05 difference between SSRG and FCFE. Our Q_p/Q_s results for SSRG at $HBW + TE_{IN,MIN}$, however, do not match the expected result of 1.05 when a 5% coronary flow contribution is assumed (24). Uncorrected phase errors could be influencing the measurements made with either SSRG or FCFE. Uncorrected phase errors that differentially contribute ~ 0.3 cm/s, for example, to the aAo and PA measures could give rise to an error of ± 0.05 in Q_p/Q_s (19), for either SSRG or FCFE, and could account for the deviation of the reported results from the expected value. The source of this error is incompletely understood.

SNR Comparison

A quantitative analysis of the measured SNR shows that the SNR for SSRG with $HBW + TE_{IN,MIN}$ is reduced by 50% compared to FCFE with $LBW + TE_{MID}$ and 33% higher than in our previous chemical shift-optimized FCFE velocity-encoding sequence with $HBW + TE_{IN}$. Both of these values are in good agreement with our theoretical expectation of 46 and 35%, respectively, determined based on the steady-state gradient echo signal equation and the applied flip angle, TE/TR , receiver BW, and estimates of T_1/T_2^* (1400/50 ms).

Sequence Timing Comparison

Table 3 shows that with chemical shift-optimized sequences, SSRG offers a 35% increase in temporal resolution compared with FCFE (35.65 vs. 55.3 ms). When chemical shift effects are not of a concern, the minimum available TE/TR for SSRG offers a 12% increase in temporal resolution compared with FCFE (41.95 vs. 47.85 ms). Regardless of chemical shift-induced errors, SSRG always provides the shortest TE/TR (highest temporal resolution) compared with FCFE for the previously described VENC range.

Table 2

Internal Consistency from a Single Volunteer with Observed Perivascular Fat Showing the Importance Between Absolute and Relative Blood Flow Quantification

	FCFE LBW + TE_{MID} (mL)	SSRG HBW + $TE_{IN,MIN}$ (mL)	Δ Interflow (mL)
PA	99.3	90.7	8.6
RPA + LPA	98.0	89.9	8.1
Δ Intraflow	1.3	0.8	

The SSRG technique is known theoretically and empirically to produce more accurate flow measurements. Δ Interflow = FCFE $LBW + TE_{MID}$ - SSRG $HBW + TE_{IN,MIN}$ for each flow territory. Δ Intraflow = PA - (RPA + LPA) for each sequence. All vessels contained perivascular fat.

Table 3
Comparison of Parameters for Different Velocity-Encoding Sequences for a VENC of 200 cm/s

	Reduced chemical shift-induced phase errors		Fastest sequence at low bandwidth	
	FCFE HBW + TE _{IN} ^a	SSRG HBW + TE _{IN,MIN}	FCFE LBW + TE _{MIN}	SSRG LBW + TE _{MIN}
TE (ms)	4.92	2.46	3.29	2.84
TR (ms)	6.91	4.46	5.98	5.24
Temporal resolution (ms)	55.3	35.65	47.85	41.95

^aTE_{IN,MIN} is unachievable with the FCFE velocity-encoding sequence at 3 T without the use of higher BW, fractional echo, or decreasing the resolution.

Flow compensated images from a FCFE sequence have reduced flow-related ghosting artifacts in the phase-encode direction. However, FCFE always has one encoding step that is not flow compensated; therefore, flow-related ghosting artifacts cannot be avoided within this encoding step. The SSRG sequence has no flow compensated waveforms, but the moment distribution compared with FCFE is similar and consequently flow-related ghosting artifacts were nearly identical to FCFE. For example, when using SSRG with a VENC of 200 cm/s, $M_{1,1} = 2.3$ (mT ms²)/m and $M_{1,2} = 8.17$ (mT ms²)/m, whereas for FCFE $M_{1,1} = 0$ and $M_{1,2} = 5.87$ (mT ms²)/m.

SSRG can also be combined with the shared velocity-encoding (SVE) technique proposed by Lin et al. (27) to offer an even greater increase in the effective temporal resolution. The SVE concept shares velocity-encoding k-space data between adjacent frames. As a result, the SVE technique cannot be combined with FCFE-encoding strategies because adjacent frames sharing the same velocity-encoded k-space data would not provide any new information. Thus, an SSRG sequence combined with SVE image reconstruction would represent an important and significant increase in temporal resolution compared with conventional FCFE approaches.

Conventional wisdom for PC-MRI parameter selection has always been to pick a LBW for improved SNR and the corresponding minimum TE to further improve SNR and reduce the TR. Although this approach has advantages, a judicious choice of BW and TE can lead to increased measurement accuracy by reducing the effects of chemical shift. Additionally, the 50% decrease in SNR with SSRG at HBW + TE_{IN,MIN} compared with FCFE at LBW + TE_{MID} should not negatively impact measurement accuracy. For example, Westenberg et al. previously reported that a 55% reduction in SNR (187 ± 116 vs. 84 ± 60) did not lead to flow inaccuracies in measurements across the mitral valve (28).

SSRG enables the use of TE_{IN,MIN}, but under the conditions evaluated the use of FCFE at HBW + TE_{MIN} results in a TE that is close to TE_{IN,MIN} ($\Delta = 0.14$ ms), but this TE would orient the fat vector at $\sim 21^\circ$ relative to stationary blood (water) for the 3-T scanner (scanner reported $B_0 = 2.89$ T) used in this study. Using our previous numerical simulations at 3 T (5) indicates that FCFE at HBW + TE_{MIN} (2.6 ms) can result in a forward flow error from chemical shift as large as 5% in the RPA/LPA (data not shown); a 2% reduction in SNR; and a 3% decrease in temporal resolution compared with SSRG at HBW + TE_{IN,MIN}. As a result, SSRG offers reduced chemical shift effects, faster scanning, and increased SNR for VENCs ≥ 190 cm/s.

Limitations

SSRG permits the use of a range of VENCs that are within the range of typical blood flow velocities for imaging the aAo, PA, RPA, and LPA. Smaller VENCs lead to a large time shift, Δt , for the refocusing gradient lobe during the second encoding step. Under these low VENC regimes, SSRG loses its velocity-encoding time efficiency compared with the FCFE or bipolar sequences, which are more time efficient at encoding lower velocities. Therefore, SSRG is not well suited for measuring myocardial motion, CSF flow measurements, and other studies that require low VENCs (see Fig. 3). Furthermore, in this study, the VENC was prospectively chosen to accommodate easily the maximum expected velocity in these subjects. Reducing the VENC requires increasing ΔM_1 , which necessitates an increased M_1 for the time-shifted SSRG in SSRG. Consequently, this leads to increased intravoxel spin dephasing and potential ghosting artifacts (29). Furthermore, if the VENC had been more closely matched to the peak velocity, the impact of the higher M_1 values in SSRG compared with FCFE may have been more apparent (8,29).

The SSRG velocity-encoding sequence allows the use of a shorter, previously unachievable TE_{IN,MIN} of 2.46 ms at 3 T. At 1.5 T, TE_{IN,MIN} = 4.76 ms (assuming ~ 210 Hz/px) and there is no shorter available TE_{IN} at this field strength. As a result, if chemical shift reduction strategies are to be used at 1.5 T, SSRG does not hold an advantage over FCFE. However, even if reducing the effects from chemical shift is not of interest, SSRG can still afford the use of a shorter TE/TR combination (Table 3), resulting in increased temporal resolution, breath hold duration, and/or spatial resolution compared with FCFE.

We have assumed that the use of TE_{IN} aligns the complex fat vector with slow-flowing blood at the vessel periphery, but this assumption is not always valid. For example, if the flow velocity is high as may occur for complex flow patterns, then the phase of fat may no longer closely align with the phase accorded by the local velocity. This highlights the importance of using HBW to minimize the contribution of perivascular fat.

When comparing our measured total flow results between the SSRG and FCFE techniques, the temporal resolution was not held constant. Instead, the minimum available TR was chosen for each technique to provide sequence efficiency. Despite the nonmatched temporal resolutions, improvements in total flow agreement are considered to be a result of chemical shift effects and not temporal resolution as shown in our

previous work (5) where the temporal resolution was held constant.

CONCLUSION

PC-MRI with SSRG velocity encoding is a more time-efficient velocity-encoding strategy for medium to large VENCs when used in conjunction with HBW. SSRG leads to more accurate and less variable flow measurements through the reduction of chemical shift-induced phase errors in addition to shortening the TE/TR, thus permitting the use of better temporal/spatial resolution and/or reduced breath hold duration.

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REFERENCES

- Markl M, Bammer R, Alley MT, Elkins CJ, Draney MT, Barnett A, Moseley ME, Glover GH, Pelc NJ. Generalized reconstruction of phase contrast MRI: analysis and correction of the effect of gradient field distortions. *Magn Reson Med* 2003;50:791–801.
- Chernobelsky A, Shubayev O, Comeau CR, Wolff SD. Baseline correction of phase contrast images improves quantification of blood flow in the great vessels. *J Cardiovasc Magn Reson* 2007;9:681–685.
- Bernstein MA, Ikezaki Y. Comparison of phase-difference and complex-difference processing in phase-contrast MR angiography. *J Magn Reson Imaging* 1991;1:725–729.
- Bernstein MA, Zhou XJ, Polzin JA, King KF, Ganin A, Pelc NJ, Glover GH. Concomitant gradient terms in phase contrast MR: analysis and correction. *Magn Reson Med* 1998;39:300–308.
- Middione MJ, Ennis DB. The effects of chemically shifted perivascular fat in quantitative phase contrast MRI. *Magn Reson Med* 2013;69:391–401.
- Axel L, Morton D. MR flow imaging by velocity-compensated/uncompensated difference images. *J Comput Assist Tomogr* 1987;11:31–34.
- Bryant DJ, Payne JA, Firmin DN, Longmore DB. Measurement of flow with NMR imaging using a gradient pulse and phase difference technique. *J Comput Assist Tomogr* 1984;8:588–593.
- Nayler GL, Firmin DN, Longmore DB. Blood flow imaging by cine magnetic resonance. *J Comput Assist Tomogr* 1986;10:715–722.
- van Dijk P. Direct cardiac NMR imaging of heart wall and blood flow velocity. *J Comput Assist Tomogr* 1984;8:429–436.
- Keller PJ, Wehrli FW. Gradient moment nulling through the Nth moment: application of binomial expansion coefficients to gradient amplitudes. *J Magn Reson* 1988;78:145–149.
- Axel L. Blood flow effects in magnetic resonance imaging. *AJR Am J Roentgenol* 1984;143:1157–1166.
- Bernstein MA, Shimakawa A, Pelc NJ. Minimizing TE in moment-nulled or flow-encoded two- and three-dimensional gradient-echo imaging. *J Magn Reson Imaging* 1992;2:583–588.
- Markl M, Alley MT, Pelc NJ. Balanced phase-contrast steady-state free precession (PC-SSFP): a novel technique for velocity encoding by gradient inversion. *Magn Reson Med* 2003;49:945–952.
- Thompson RB, McVeigh ER. Real-time volumetric flow measurements with complex-difference MRI. *Magn Reson Med* 2003;50:1248–1255.
- Tanaka S, Yoshiyama M, Imanishi Y, Nakahira K, Hanaki T, Naito Y, Imai M, Tanaka M. MR measurement of visceral fat: assessment of metabolic syndrome. *Magn Reson Med Sci* 2006;5:207–210.
- Tanaka S, Yoshiyama M, Imanishi Y, et al. Measuring visceral fat with water-selective suppression methods (SPIR, SPAIR) in patients with metabolic syndrome. *Magn Reson Med Sci* 2007;6:171–175.
- Atkinson DJ, Edelman RR. Cineangiography of the heart in a single breath hold with a segmented turboFLASH sequence. *Radiology* 1991;178:357–360.
- Griswold MA, Jakob PM, Heidemann RM, Nittka M, Jellus V, Wang J, Kiefer B, Haase A. Generalized autocalibrating partially parallel acquisitions (GRAPPA). *Magn Reson Med* 2002;47:1202–1210.
- Gatehouse PD, Rolf MP, Graves MJ, et al. Flow measurement by cardiovascular magnetic resonance: a multi-centre multi-vendor study of background phase offset errors that can compromise the accuracy of derived regurgitant or shunt flow measurements. *J Cardiovasc Magn Reson* 2010;12:5.
- Rolf MP, Hofman MB, Gatehouse PD, et al. Sequence optimization to reduce velocity offsets in cardiovascular magnetic resonance volume flow quantification—a multi-vendor study. *J Cardiovasc Magn Reson* 2011;13:18.
- Stalder AF, Frydrychowicz A, Russe MF, Korvink JG, Hennig J, Li K, Markl M. Assessment of flow instabilities in the healthy aorta using flow-sensitive MRI. *J Magn Reson Imaging* 2011;33:839–846.
- Holland BJ, Printz BF, Lai WW. Baseline correction of phase-contrast images in congenital cardiovascular magnetic resonance. *J Cardiovasc Magn Reson* 2010;12:11.
- Miller TA, Landes AB, Moran AM. Improved accuracy in flow mapping of congenital heart disease using stationary phantom technique. *J Cardiovasc Magn Reson* 2009;11:52.
- Ganong WF. Review of medical physiology, 22nd ed. New York: Lange Medical Books/McGraw-Hill; 2005.
- Walker PG, Cranney GB, Scheidegger MB, Waseleski G, Pohost GM, Yoganathan AP. Semiautomated method for noise reduction and background phase error correction in MR phase velocity data. *J Magn Reson Imaging* 1993;3:521–530.
- Lankhaar JW, Hofman MB, Marcus JT, Zwanenburg JJ, Faes TJ, Vonk-Noordegraaf A. Correction of phase offset errors in main pulmonary artery flow quantification. *J Magn Reson Imaging* 2005;22:73–79.
- Lin HY, Bender JA, Ding Y, Chung YC, Hinton AM, Pennell ML, Whitehead KK, Raman SV, Simonetti OP. Shared velocity encoding: a method to improve the temporal resolution of phase-contrast velocity measurements. *Magn Reson Med* 2012;68:703–710.
- Westenberg JJ, Roes SD, Ajmone Marsan N, Binnendijk NM, Doornbos J, Bax JJ, Reiber JH, de Roos A, van der Geest RJ. Mitral valve and tricuspid valve blood flow: accurate quantification with 3D velocity-encoded MR imaging with retrospective valve tracking. *Radiology* 2008;249:792–800.
- Wedeen VJ, Weisskoff RM, Poncelet BP. MRI signal void due to in-plane motion is all-or-none. *Magn Reson Med* 1994;32:116–120.