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Velocity-Selective Magnetization-Prepared Non-Contrast-Enhanced Cerebral MR Angiography at 3T: Improved Immunity to B0/B1 Inhomogeneity

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Abstract

Purpose—To develop a Fourier-transform based velocity-selective (VS) pulse train that offers improved robustness to B0/B1 inhomogeneity for non-contrast-enhanced cerebral MR angiography (MRA) at 3T.

Methods—VS pulse train I and II with different saturation bands are proposed to incorporate paired and phase cycled refocusing pulses. Their sensitivity to B0/B1 inhomogeneity was estimated through simulation and compared with a single refocused VS pulse train. The implementation was compared to standard time-of-flight (TOF) among 8 healthy subjects.

Results—In contrast to single refocused VS pulse train, the simulated VS profiles from proposed pulse trains indicate much improved immunity to field inhomogeneity in the brain at 3T. Successive application of two identical VS pulse trains yields a better suppression of static tissue at the cost of 20~30% signal loss within large vessels. Average relative contrast ratios of major cerebral arterial segments applying both pulse train I and II with two preparations are 0.81 ± 0.06 and 0.81 ± 0.05 respectively, significantly higher than 0.67 ± 0.07 of TOF-MRA. VS MRA, in particular, the pulse train II with the narrower saturation band, depicts more small vessels with slower flow.

Conclusion—VS magnetization-prepared cerebral MRA was demonstrated among normal subjects on a 3T scanner.

Keywords

non-contrast-enhanced MRA; cerebral MRA; velocity-selective pulse train

INTRODUCTION

The visualization of the intracranial vasculature is important in the diagnosis of cerebrovascular pathologies, such as stenosis and aneurysms (1). Although catheter-based angiography has been considered the gold standard for cerebral artery assessment, non-contrast-enhanced MR angiography (NCE-MRA) has rapidly evolved owing to the motivation to lower risk and cost as well as the advances of MR hardware and software (2,3).

Expanding from conventional phase-contrast (PC) (4) and time-of-flight (TOF) (5,6) MRA techniques, the current approaches include variant of inflow-based techniques (7,8) and subtraction-based methods: fast-spin-echo acquisition (9–12), or balanced steady-state free precession (bSSFP) acquisition preceded by flow-sensitive preparation pulses (13–15), triggered at systolic and diastolic cardiac phases; spin-tagging based techniques acquired alternatively between labeling and control (16–19). Among them, TOF-MRA has been the most common method used for evaluation of intracranial vasculature but is limited by its long imaging time and limited coverage (2,20). Moreover, TOF-MRA is prone to saturation of slowly flowing blood protons, resulting in signal loss within a severely compromised lumen or distal small arteries (1,20).

Recently, Fourier-transform based velocity-selective (VS) magnetization-prepared MRA has been introduced for visualization of vessels based on the designated flow velocity and allows for a large spatial coverage (21,22). Specifically, the angiographic signal is achieved, by setting the flowing spins in the pass-band and static spins in either the inversion-band (21) or saturation-band (22), therefore it preserves flowing blood signal and avoids motion-induced misregistration associated with subtractive approaches. The combination of non-selective RF pulse trains with embedded velocity-encoding gradients, based on the k -space formalism (23–25), can produce almost arbitrary velocity-selective profiles. However, the original scheme (without refocusing pulses) (25) suffers from off-resonance effect which is manifested as excitation profile shifting along velocity (21). The susceptibility to B0 field inhomogeneity can be alleviated, by incorporating one composite refocusing pulse within each velocity encoding step and modifying the RF and gradient waveforms accordingly, as recently shown for peripheral MRA at 1.5T (22). However, the tolerable B0 offset is limited to ± 80 Hz and the sensitivity to B1 inhomogeneity remains an issue particularly at high field strength. Unfaithful B1+ scale (ratio of actual flip angle to nominal input flip angle) leads to two independent consequences: incorrect RF weighting for the excitation k -space by the hard pulse at the beginning of each velocity encoding step and thus inaccurate flip angle for either the inversion or saturation band; imperfect refocusing during each velocity encoding step and thus degraded velocity selective profile at off-resonance.

In this study, we aimed to develop the VS saturation pulse trains with further improved immunity to B0/B1 field inhomogeneities for cerebral MRA at 3T. Extended VS pulse trains

were designed with a pair of refocusing pulses within each velocity encoding step and accompanied by phase cycling. The performances of the proposed techniques were compared with single refocused pulse train using numerical simulations. The techniques were further optimized and evaluated with application of one preparation or two preparations in healthy subjects for cerebral MRA at 3T. Preliminary studies were reported in recent ISMRM meetings (26,27).

METHODS

VS Pulse Trains with Paired Refocusing Pulses and Phase Cycling

A single refocused VS saturation pulse train (22) was first constructed with 4 velocity encoding steps with 4 composite refocusing pulses ($90^\circ_x 180^\circ_y 90^\circ_x$) and 5 rectangular excitation pulses through Shinnar-Leroux transform (28) (Fig. 1a). To better combat B0/B1 inhomogeneity, two sets of new VS pulse trains with paired refocusing pulses inserted for each velocity encoding step are investigated: I, 4 velocity encoding steps with 8 composite refocusing pulses ($90^\circ_x 180^\circ_y 90^\circ_x$) and 5 excitation pulses (18° each) through Fourier-transform (Fig. 1b); II, 8 velocity encoding steps with 16 refocusing pulses and 9 excitation pulses (10° each) (Fig. 1c). Further improvement is realized through phase cycling of these refocusing pulses with following schemes: I, MLEV-8: [0-0-180-180-180-0-0-180]; II, MLEV-16: [0-0-180-180-180-0-0-180-180-180-0-0-180-180-0] (29). Note that the paired pulses constitute a full refocusing of the phase of the transverse magnetizations and thus do not require modification of the encoding RF pulses as necessitated when using a single refocusing pulse in each velocity encoding step (22).

The paired refocusing pulses in each velocity encoding step of the pulse train are embedded among a group of four gradient lobes (slice direction) with alternating polarities (Fig. 1b,c), which leads to a flow-sensitizing gradient waveform (30). The shapes of the gradient lobes are chosen as triangular for better self-cancellation of the eddy currents generated from the rising and falling ramps (30). Since the flow velocity of major cerebral arteries are 30~40 cm/s at the diastolic phase for young healthy subjects (31), the velocity field of view (FOV_v) is set to be 45 cm/s for this study. Fig. 1a shows a 15 ms single refocused VS pulse train and Fig. 1b shows a 30 ms VS pulse train I with paired refocusing, both producing a saturation band within ± 8 cm/s (or called cut-off velocity, V_c); Fig. 1c shows a 48 ms VS pulse train II with $V_c = \pm 4$ cm/s. Specific parameters for each pulse are listed in Supporting Table S1.

Numerical Simulation

Numerical simulations of the Bloch equations based on matrix rotation were performed to assess the properties of the proposed VS pulse trains using Matlab (MathWorks, Inc., Natick, MA, USA). Responses of the longitudinal magnetizations (M_z) following the pulse trains under various conditions were examined for velocities from -60 cm/s to 60 cm/s with intervals of 0.5 cm/s. First, the sensitivity to a typical range of B0/B1 offset incurred in the brain at 3T (B0 field : ± 200 Hz; B1+ scale: from 0.8 to 1.2) were evaluated for all three VS pulse trains (Fig. 1a,b,c); Second, the effect of the phase cycling of the refocusing pulses were compared for the pulse trains with paired refocusing (Fig. 1b,c): without any phase-cycling and with different phase-cycling patterns (I, repeat of MLEV-4:

[0-0-180-180-0-0-180-180], and MLEV-8; II, repeat of MLEV-8: [0-0-180-180-180-0-0-180-0-0-180-180-0-0-180], and MLEV-16); Third, the effect of transverse relaxation time (T_2) over the duration of the proposed pulse train I (30 ms) and II (48 ms) were considered for three different T_2 values at 3T: 1500 ms (CSF, (32)), 150 ms (arterial blood, (33,34)), and 70 ms (tissue, (35)), respectively; Lastly, the velocity selective profiles of the proposed pulse train I and II were simulated for velocities with $\pm 5\%$, $\pm 10\%$ and $\pm 20\%$ linear temporal-variation (acceleration) during 50 ms to mimic pulsation or tortuous flow. The effect of T_1 or T_2 relaxation was ignored during these simulations when not explicitly stated.

In Vivo Experiments

Experiments were conducted on a 3T Philips Achieva scanner (Philips Medical Systems, Best, The Netherlands) using the body coil for RF transmission (maximum amplitude 575 Hz) and a 32-channel head-only coil for signal reception. The maximum strength and slew rate of our standard gradient coil are 40 mT/m and 200 mT/m/ms, respectively. Eight healthy volunteers (26–55 yrs old, three males and five females) were enrolled after providing informed consent in accordance with the guidelines of Institutional Review Board.

As part of scan planning, PC-MRA survey images were acquired in the sagittal and coronal planes to display the major intracranial and neck arteries (ICA: internal cerebral artery; ECA: external cerebral artery; VA: vertebral artery; BA: basilar artery; ACA: anterior cerebral artery; MCA: middle cerebral artery, PCA: posterior cerebral artery). Both scans used a 50 mm slab, TR / TE = 20 / 6.3 ms, FOV = $250 \times 250 \text{ mm}^2$, and a scan matrix of 256×128 (acquisition time: 20 s for 2 averages).

Most 3D VS magnetization-prepared MRA images were acquired in an axial plane through the circle of Willis with a TOF MRA as the reference. Two proposed VS pulse trains (I and II) were first compared among six volunteers with each applied once or twice consecutively followed by spoiler gradients, and without any VS pulse trains. When not specified, VS saturation applied along the slice direction with two preparations is employed for the remaining experiments to achieve more uniformed saturation effect on background tissues. Then, the gradients of the VS pulse trains were applied either along the anterior-posterior (frequency-encoding), left-right (phase-encoding) or foot-head (slice-selection) direction to examine the effect of velocity-encoding orientations. The VS-MRA was further tested on four subjects for a coronal slab to cover the major cerebral vessels (encoding velocity along the slice direction as well).

Cardiac gating through peripheral-pulse-unit (PPU) triggering was utilized on all subjects and was compared for six subjects without triggering. The PPU triggering was employed with a triggering delay of 350 ms (36) to synchronize the VS pulse trains to the diastolic phase, when there is less variation in the flow velocity for the intracranial arterial blood. A SPIR module (spectral presaturation with inversion recovery) was inserted between the VS pulse trains and the acquisition for fat suppression.

The proposed VS cerebral MRA employed turbo field echo (TFE) as the acquisition module with low-high profile ordering (37) for the acquisition of the center of the k -space right after

VS preparation pulses. A 65 mm-thick slab was acquired with a resolution of $0.7 \times 0.7 \times 1.4 \text{ mm}^3$ and reconstructed to $0.5 \times 0.5 \times 0.7 \text{ mm}^3$ through zero-padding. Other parameters included: readout bandwidth = 193 Hz / pixel, flow-compensation gradients applied in three orthogonal orientations, TR / TE = 11 / 6.5 ms, flip angle = 15° to ensure the optimal signal strength during the transient state before reaching steady state (38), TFE factor = 60, TFE acquisition window = 650 ms, TFE shot interval = 2 heart beats (or 2 sec without PPU triggering, both allowing approximately 1300 ms interval for inflow of fresh blood before VS pulse trains), SENSE factor = 3 along phase-encoding direction, and total scan time = approximately 2.5 min. The exerted SAR was 33% and 60% for VS-MRA using pulse train I and II, respectively.

The standard TOF-MRA was acquired with identical resolution, volume coverage, and acquisition time as VS-MRA. Other parameters included: readout bandwidth = 288 Hz / pixel, flow-compensation gradients applied, 50% of echo in the readout direction acquired, TR / TE = 23 / 3.5 ms, flip angle = 18° , SENSE factor = 3, 3 chunks acquired with 15 slices per chunk to decrease the saturation effects at the end of the volume (39), excitation pulses with linearly-varying flip angle (start: 16.2°) over the chunk applied to reduce the saturation effect of inflowing blood (40). SAR of TOF-MRA was 10%.

Quantitative Analysis

To avoid the complication of spatially variable noise level resulting from the parallel imaging reconstruction with the use of phased array coils (41,42), only relative signal ratio and relative contrast ratio were calculated for comparison here. Relative signal ratios are the ratios of signal intensities of the VS-MRA with one preparation with respect to without preparation, or VS-MRA with two preparations with respect to with one preparation. Relative contrast ratio is defined as $(S_a - S_t) / S_a$ (43), where S_a is the signal intensity of blood and S_t is the signal intensity of cortex tissue (the perfect relative contrast ratio is 1.0). Region of interests were placed over six major cerebral arterial segments (ACA: A1, A2; MCA: M1, M2; PCA: P1, P2) and cortex tissue on maximum-intensity-projection (MIP) images of the axial-oriented angiograms acquired from six subjects. When comparing results from different methods, paired t-test with the two-tailed distribution was used.

RESULTS

Numerical Simulation

For the single refocused pulse train and our proposed pulse trains with paired and phase cycled refocusing, Fig. 2 displays the Mz responses of VS pulse trains over the plane of velocity (x-axis) vs. B0 off-resonance frequency (y-axis) at three different B1+ scales (0.8 (left column), 1.0 (middle column) and 1.2 (right column)) respectively: for the single refocused pulse train (first row), significant pass-band distortion is apparent at high off-resonance, even with correct B1+ setting; for pulse trains I (second row) and II (third row), the simulated VS profiles are well maintained at different B0/B1 conditions. The Mz signal intensity within the saturation band still suffers from B1 inhomogeneity due to the hard pulses used at the beginning of each velocity encoding step. This can be further mitigated by a repeat application of the VS pulse train as will be shown in the *in-vivo* experiments.

The results of pulse trains I and II without phase cycling for the refocusing pulses are shown as Fig. 3a and 3d, respectively, which are considerably worse than those with phase cycling schemes of MLEV-4 (Fig. 3b), MLEV-8 (Fig. 3c and 3e) and MLEV-16 (Fig. 3f) applied. It is evident that more complete phase cycling for the refocusing pulses shows more improved robustness to B0 field inhomogeneity.

When taking into account the T_2 relaxation, Mz responses over the velocity (x-axis) are displayed with $T_2 = 1500$ ms (CSF, Fig. 4a and 4d), 150 ms (arterial blood, Fig. 4b and 4e), and 70 ms (tissue, Fig. 4c and 4f) for prolonged pulse train I (30 ms, first row) and II (48 ms, second row) respectively. For arterial blood, the Mz intensity at the passband is less than 10% lower (Fig. 4b and 4e) compared to the cases with much longer T_2 (Fig. 4a and 4d); for static tissue, the remaining Mz intensity at the stopband is about 5% (I, Fig. 4c) and 14% (II, Fig. 4f) respectively. This weakened background suppression caused by short T_2 values of tissue can also be mitigated when using VS pulse trains with two preparations.

The Mz responses for a range of temporal changes of velocities (0%, red; $\pm 5\%$, green; $\pm 10\%$, blue; and $\pm 20\%$, magenta) during the pulse train I and II are visualized in Fig. 5a and 5b, respectively. Longer pulse train and larger velocity changes give rise to more signal loss at the fast-velocity side of the passband. Conversely, the velocity selective profiles between the passband at the slow-velocity side and the saturation band are not sensitive to this range of velocity changes. This is similar to the finding in Fig. 9 of (21).

In Vivo Experiments

The utility of the VS pulse train can be much appreciated, when first compared with the acquisition without the VS module (Fig. 6a). The MIP images of VS-MRA acquired with applications of one or two preparations for VS pulse train I (Fig. 6b–c) and VS pulse train II (Fig. 6d–e) are exhibited, respectively, with the same intensity scales. The measured relative signal ratios of six different major cerebral arterial segments between the applications with two VS preparations (Fig. 6c,e) and the corresponding ones with only one preparation (Fig. 6b,d) across six subjects are listed in Supporting Table S2 for both pulse train I and II, with averages of 0.73 ± 0.06 and 0.73 ± 0.05 , respectively. Note that these ratios are found much higher than the averaged relative signal ratios between the results with one VS preparation (Fig. 6b,d) and without any preparation (Fig. 6a), 0.43 ± 0.10 and 0.50 ± 0.09 , respectively, which is likely due to the partial volume effect of unsaturated tissue in the images without VS preparation. The relative signal ratios of cortex tissue between the results with two VS preparations (Fig. 6c,e) and without any preparation (Fig. 6a) are 0.07 ± 0.01 and 0.08 ± 0.01 , respectively. By assuming 0.73 as the remaining fraction of blood signal after each VS preparation, the VS-MRA with two VS preparations lead to about half of the signal loss of the major vessels, in contrast to more than 10 times reduction of the static tissues.

As listed in Table 1, average relative contrast ratios of major cerebral arterial segments of VS-MRA with both pulse train I and II increased from 0.63 ± 0.11 and 0.63 ± 0.10 (one preparation) to 0.81 ± 0.06 and 0.81 ± 0.05 (two preparations), respectively, which are also significantly higher than 0.67 ± 0.07 of TOF-MRA ($p < 0.01$). Compared to TOF-MRA (Fig. 6f), VS-MRA depicts more small distal MCA and PCA branches with slow flow at directions parallel to the axial slab. As expected, with the same velocity FOV (45 cm/s) but

half the saturation band ($V_c = 4$ vs. 8 cm/s), VS pulse train II (Fig. 6d–e, Table 1) generates similar appearance for large vessels (A1 and A2 of ACA, M1 and M2 of MCA, P1 and P2 of PCA) and delineates more noticeable small vessels than VS pulse train I (Fig. 6b–c). With only one VS saturation preparation, static tissue background is well suppressed in the center of the brain but degrades at the periphery cortical area (Fig 6b,d), which complies with the parabolic distribution of B1 transmission field in human brain at 3T (44,45); With two preparations, better background suppression is appreciable especially at the edge of the cranium, although at the cost of 20~30% signal loss for large vessels (Fig. 6c,e, Supporting Table S2).

The effect of the velocity-encoding orientation (anterior-posterior, left-right, foot-head) in VS pulse train II is illustrated in Fig. 7a–c, respectively. The majority of vessels are depicted conspicuously in all three images (Table 1: 0.81 ± 0.05 (foot-head), 0.77 ± 0.08 (anterior-posterior), 0.78 ± 0.07 (left-right)), presumably as a result of the curvy and oblique nature of most brain vessels. The dependence on velocity-encoding-direction is revealed for some blood vessels: e.g., the distal ACA along the middle line of the brain is better delineated in the anterior-posterior encoded image (Fig. 7a, red arrowhead) and the sagittal sinus at the posterior part of the brain is pale in the left-right encoded image (Fig. 7b, blue arrowhead). Note that Fig. 7a is tainted by some horizontally oriented stripe artifact in static tissues, which might be related to the stronger eddy currents found in this direction on our scanner (data not shown). Fig. 7c is also compared with Fig. 7d generated without PPU triggering, showing some minor signal loss at the M1 sections of MCA at both sides (red arrow) (Table 1: 0.81 ± 0.05 (with PPU triggering) vs. 0.80 ± 0.05 (without PPU triggering)), where pulsation effect is strongest compared to other major branches (31). Fig. 7e is the TOF-MRA as the reference.

The MIPs of the two VS pulse trains (Fig. 8a: I; 8b: II) acquired with coronal orientation are exhibited. Most major arteries (ICA, ECA, VA, BA, ACA, MCA) and their small branches, especially those with slower blood flow velocity (such as VA and ACA), are better illustrated by VS pulse train II (Fig. 8b). Signal intensities are higher in ICA than in ACA and MCA (Fig. 8), largely due to the partial volume effect from the small size of intracranial arteries (1~3 mm in diameter) relative to the acquired slice thickness (1.4 mm).

Hypointensity or even flow void is sometimes observed at the Petrous portion of ICA (Fig. 8) of our healthy subjects. This is believed to be the result of inadequate flow compensation in the readout in the presence of complex flow within the curved vessels, which can be mitigated by shortening TE (e.g. partial echo, data not shown).

DISCUSSION

We have developed a new Fourier-transform-based VS pulse train technique for NCE cerebral MRA at 3T. By replacing single refocusing pulse in each velocity encoding step (Fig. 1a) with dual refocusing pulses (I and II, Fig. 1b,c) and complete phase cycling concurrently, the robustness to B0/B1 field is significantly improved (Fig. 2). Although composite pulses or even adiabatic pulses can afford more superior performance, pulse II using 16 hard pulses phase-cycled for refocusing achieved adequate immunity to B0/B1 field inhomogeneity for cerebral application at 3T, which has similar RF power as pulse I

with 8 composite refocusing pulses and is thus less constrained by the SAR limit for the sequence employed. Compared to pulse I, pulse II exhibits better blood vessel illustration (Fig. 6 and 8), mainly due to the generation of twice of the velocity encoding resolution and half of the saturation band.

By applying the VS pulse train twice, B1+ sensitivity is further reduced (Fig. 6) with improved relative contrast ratios (Table 1) at the cost of 20–30% signal reduction for large arteries around circle of Willis when one and two VS modules were compared (Fig. 6b vs. 6c and 6d vs. 6e, respectively, Supporting Table S2). Signal drop caused by T₂ relaxation is less than 10% (Fig. 4) during one VS pulse train for arterial blood. Another factor for signal loss can be attributed to the variation in blood velocity due to pulsation or tortuous flow orientation during the pulse (22): 10~20% signal loss for spins with fast velocities experiencing $\pm 5\sim 10\%$ changes during the pulses are found in our simulations (Fig. 5). Cardiac triggering at diastolic phase was found to only slightly improve the angiographic quality at these large vessels (Fig. 7c vs. 7e). Note that the velocity selective profiles are affected for all the spins by their T₂ relaxations, whereas distortions of the passband occur primarily for the fast flowing spins with large velocity changes (high acceleration values). Further improvement of hardware (e.g. higher RF amplitude and gradient strength) will enable shortening durations of VS pulse trains, hence attenuating the blood signal loss from both factors.

Stripe artifacts are observed in VS-MRA with encoding directions other than slice direction. We tentatively speculate that it is caused by eddy currents or other gradient imperfection. Specifically, the incomplete cancelation of the unipolar gradient lobes closely surrounding the refocusing pulses might effectively introduce a spatial modulation of longitudinal signal of background tissues, similar to the tagging technique (46) and the artifacts in the MRA using flow-sensitive preparation pulses with unipolar gradients (13).

The TFE acquisition scheme following our VS magnetization-preparation also deserves further optimization. Current protocol has an acquisition resolution of $0.7 \times 0.7 \times 1.4 \text{ mm}^3$ with a SENSE factor of 3, resulting in a total scan time of 2.5 min. For clinical applications, acquisition with isotropic resolution over a large spatial coverage within 5 min is desired. bSSFP has been employed in many NCE-MRA studies, mostly at 1.5T (8,13–15,18,21,22,47,48) and few at 3T (19,49), owing to its high SNR efficiency and inherent flow-compensation capability,. The challenges of 3T, including off-resonance artifacts and high SAR, have made bSSFP more preferable at 1.5T (50). Recently, an accelerated 3D TFE technique with radial acquisition was proposed for ASL-based MRA (51,52). The benefit of advanced acquisition techniques can potentially be combined with the VS magnetization-preparation.

One limitation of our study is its lack of separation between arteries and veins. The velocities of intracranial arterial blood and venous blood do not differ as significantly as in peripheral counterparts. The complex cerebral vascular network also makes it difficult to differentiate arteries and veins based on their flowing directions. A potential solution for this problem maybe depends on the nature of acceleration of arterial blood during the systolic phases (15).

CONCLUSION

A NCE cerebral MRA method has been developed at 3T, based on VS magnetization preparation pulse trains which improve the robustness to B0/B1 field inhomogeneities via paired and phase cycled refocusing. Compared with TOF MRA, VS-MRA enhances the delineation of more distal branches of cerebral arteries and allows for flexible scan orientation without requiring the section to be positioned orthogonal to the direction of flow. Although technical feasibility was shown by the excellent depiction of the cerebral arteries in healthy subjects, the clinical value of the proposed technique needs to be investigated through comparisons with more established methods such as contrast-enhanced MRA in patients with cerebral vascular disorders.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.

Diagram of (a) VS pulse train with a single composite refocusing pulse in each velocity encoding step, (b) VS pulse train I with paired composite refocusing pulses and MLEV-8 phase cycling scheme and (c) VS pulse train II with paired hard pulses for refocusing and MLEV-16 phase cycling scheme.

Figure 2.

The simulated Mz-velocity responses at different B0 conditions after applying various VS pulse trains (pulse train with single refocusing pulses (a–c); pulse trains with paired and phase cycled refocusing pulses: I (d–f) and II (g–i)) at representative B1+ scales of 0.8 (a, d, g), of 1.0 (b, e, h), and of 1.2 (c, f, i).

Figure 3.

The simulated Mz-velocity responses with different phase cycling schemes. Pulse train I: (a) without any phase cycling; (b) MLEV-4; (c) MLEV-8; Pulse train II: (d) without any phase cycling; (e) MLEV-8; (f) MLEV-16. Here B1+ scale is at correct setting.

Figure 4.

Mz responses over the velocity (x-axis) for prolonged pulse train I (30 ms, a–c) and II (48 ms, d–f) with T₂ of 1500 ms (CSF, (a) and (d)), 150 ms (arterial blood, (b) and (e)), and 70 ms (tissue, (c) and (f)) respectively. No offset of B₀ or B₁+ scale is present in this simulation.

Figure 5.

The performance of the proposed pulse train I (a) and II (b) encountering various velocity changes: constant velocities (red), from 95% to 105% (green), from 90% to 110% (blue), and from 80% to 120% (magenta) within 50 ms. No offset of B_0 or B_1 scale or T_2 effect is present in this simulation.

Figure 6.

Representative MIP images of (a) results without any VS pulse trains and magnetization-prepared MRA using VS pulse train I with (b) one preparation and (c) two preparations; using VS pulse train II with (d) one preparation and (e) two preparations; (f) TOF as reference. Major cerebral arterial segments for quantitative analysis are labeled with color at (a): A1 (yellow); A2 (magenta); M1 (cyan); M2 (red); P1 (green); P2 (blue); cortex tissue (black). Note the difference of depiction of small distal MCA branches in the red dashed box with the zoomed-in view between VS-MRA with one preparation (b,d), two preparations (c,e), and TOF (f).

Figure 7.

MIPs of VS-MRA from another subject: VS pulse train II with two preparations encoded along (a) anterior-posterior, (b) left-right and (c) foot-head directions; (d) acquisition without cardiac-triggering using the same VS pulse train for (c); (e) TOF. The poor visualization of the distal anterior cerebral artery (red arrowhead) and the sagittal sinus (blue arrowhead) in (b) indicates the dependence of the signal sensitivity of VS-MRA on the velocity-encoding direction. Comparison between images acquired with (c) and without (d) PPU triggering reveals the pulsation effect on VS-MRA which leads to some minor signal loss at the M1 sections of MCA at both sides (red arrow).

pulse train I with two prep.

pulse train II with two prep.

Figure 8.

The MIPs of VS-MRA acquired from a coronal orientation with slice-direction encoding and two preparations of VS pulse train (a) I and (b) II. Both head and neck arteries are labeled on (b).

Table 1

Quantitative measurement of relative contrast ratios of major cerebral arterial segments for different pulse train configurations.

pulse	VS-MRA										TOF
	I	I	II	II	II	II	II	II	II	II	
	one	two	one	two	two	two	two	two	two	two	
prep.	F-H	F-H	F-H	F-H	F-H	A-P	L-R	F-H	F-H	F-H	
dir.	Y	Y	Y	Y	Y	Y	Y	Y	Y	N	
trigger	Y	Y	Y	Y	Y	Y	Y	Y	Y	N	
A1	0.62±0.08	0.81±0.04	0.62±0.07	0.80±0.04	0.77±0.07	0.77±0.06	0.79±0.05	0.83±0.04	0.79±0.05	0.64±0.08	
A2	0.64±0.07	0.80±0.05	0.64±0.07	0.81±0.04	0.78±0.05	0.79±0.04	0.80±0.05	0.83±0.04	0.80±0.05	0.72±0.04	
M1	0.75±0.05	0.86±0.03	0.73±0.04	0.85±0.03	0.82±0.07	0.83±0.04	0.83±0.04	0.83±0.04	0.77±0.05	0.71±0.03	
M2	0.58±0.10	0.77±0.06	0.59±0.08	0.78±0.04	0.75±0.07	0.78±0.03	0.78±0.03	0.77±0.05	0.77±0.05	0.59±0.08	
P1	0.68±0.07	0.84±0.05	0.66±0.07	0.83±0.05	0.78±0.09	0.76±0.12	0.83±0.03	0.83±0.03	0.77±0.04	0.64±0.04	
P2	0.52±0.13	0.76±0.07	0.53±0.12	0.77±0.06	0.72±0.09	0.73±0.09	0.73±0.09	0.77±0.04	0.77±0.04	0.64±0.04	
Ave.	0.63±0.11	0.81±0.06	0.63±0.10	0.81±0.05	0.77±0.08	0.78±0.07	0.80±0.05	0.80±0.05	0.80±0.05	0.67±0.07	

Note: prep.: preparation; dir.: direction; F-H: foot-head; A-P: anterior-posterior; L-R: left-right; trig.: triggering; A1, A2: ACA; M1, M2: MCA; P1, P2: PCA;