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Whole-Brain Arteriography and Venography Using Improved Velocity-Selective Saturation Pulse Trains

Wenbo Li^{1,2,†}, Feng Xu^{1,2,3,†}, Michael Schär¹, Jing Liu^{1,4}, Taehoon Shin^{5,6}, Yansong Zhao⁷, Peter C.M. van Zijl^{1,2}, Bruce A. Wasserman¹, Ye Qiao¹, and Qin Qin^{1,2,*}

¹The Russell H. Morgan Department of Radiology and Radiological Science, Division of MR Research, Johns Hopkins University School of Medicine, Baltimore, MD, USA

²F.M. Kirby Research Center for Functional Brain Imaging, Kennedy Krieger Institute, Baltimore, MD, USA

³Developing Brain Research Lab, Children's National Medical Center, Washington DC, USA

⁴Department of Radiology, Guizhou Medical University Affiliated Hospital, Guiyang, Guizhou Province, China

⁵Division of Mechanical and Biomedical Engineering, Ewha Womans University, Seoul, South Korea

⁶Department of Diagnostic Radiology and Nuclear Medicine, University of Maryland, Baltimore, MD, USA

⁷Philips Healthcare, Cleveland, Ohio, USA

Abstract

Purpose—To develop velocity-selective (VS) MR angiography (MRA) protocols for arteriography and venography with whole-brain coverage.

Methods—Tissue suppression using velocity-selective saturation (VSS) pulse trains is sensitive to B1 inhomogeneity. To reduce its sensitivity we replaced the low-flip-angle hard pulses in the VSS pulse train with optimal composite (OCP) pulses. Additionally, new pulse sequences for arteriography and venography were developed by placing spatially selective inversion pulses with a delay to null signals from either venous or arterial blood. The VS MRA techniques were compared to the time-of-flight (TOF) MRA in six healthy subjects and two patients at 3T.

Results—More uniform suppression of stationary tissue was observed when the hard pulses were replaced by OCP pulses in the VSS pulse trains, which improved contrast ratios between blood vessels and tissue background for both arteries (0.87 vs. 0.77) and veins (0.80 vs 0.59). Both arteriograms and venograms depicted all major cervical and intracranial arteries and veins, respectively. Compared to TOF MRA, VS MRA not only offers larger spatial coverage but also

*Corresponding Author: Qin Qin (Ph.D.), Department of Radiology, Johns Hopkins University School of Medicine, F.M. Kirby Research Center for Functional Brain Imaging, Kennedy Krieger Institute, 707 N. Broadway, Baltimore, MD, 21205, qin@mri.jhu.edu, Tel: 443-923-9516.

[†]These authors contributed equally to this work.

depicts more small vessels. Initial clinical feasibility was shown in two patients with comparisons to TOF protocols.

Conclusion—Non-contrast-enhanced whole-brain arteriography and venography can be obtained without losing sensitivity to small vessel detection.

Keywords

non-contrast-enhanced MRA; cerebral MRA; arteriography; venography; velocity-selective pulse train; optimal control

INTRODUCTION

Non-contrast-enhanced (NCE) MR angiography (MRA) is free of ionizing radiation, does not require an exogenous contrast agent and is completely non-invasive, thus is ideal for first-line screening or frequent monitoring in follow-up exams. Time-of-flight (TOF) MRA (1,2) has been the most established NCE MRA method for evaluating intracranial vessels. Applying a saturation slab below or above the imaging volume, arteriograms and venograms can be obtained, respectively. Based on the effects of through-plane blood flow, TOF-MRA is prone to saturation of slowly flowing blood, which results in poor delineation of distal small arteries or a lumen with severe stenosis (3,4). To reduce this in-plane saturation effect, 3D TOF is typically fused by sequentially acquired multiple thin-slabs (5), thus offering limited spatial coverage.

Phase-contrast (PC) MRA (6) produces velocity measurement by subtracting two scans with paired flow-dependent phase modulation using toggled bipolar gradients. Due to its long acquisition time for 3D angio, PC MRA is commonly applied in 2D to obtain a quick vascular scout image (3,4). Arterial spin labeling (ASL) based techniques have also been developed to obtain cerebral MRA using interleaved labeling and control RF pulses, providing either large anatomical coverage (3D) (7–12), or 4D time-resolved MRA with dynamic filling of arteries (13–19). Similar to ASL perfusion methods, the subtraction of the label and control datasets removes the static tissue background and reveals the labeled arterial anatomy. ASL-MRA is associated with lengthened scan time due to the requirement of two acquisitions and the sensitivity to motion-induced misregistration.

Recently, velocity-selective (VS) magnetization-prepared MRA has been introduced for visualization of vessels by employing a VS saturation (VSS) pulse train before a single acquisition without subtraction (20,21). A Fourier-transform based VSS pulse train is composed of multiple velocity-encoding steps. In order to reduce off-resonance-induced phase modulation, refocusing pulses are inserted in each velocity-encoding step. A previous implementation for peripheral VS-MRA at 1.5T used a single composite refocusing pulse with tailored RF and gradient waveforms but had only moderate immunity to variations in the B₀ off-resonance and B₁₊ field (21). In the latest studies, we demonstrated both cerebral and peripheral VS-MRA at 3.0T by employing paired and phase-cycled refocusing pulses which have shown to be superior to the single refocusing for maintaining the designated VS profile in the regions of large B₀ inhomogeneities (22,23). When compared to TOF, cerebral VS-MRA showed enhanced delineation of small vessels with slower flow (22).

Current VSS pulse trains are still sensitive to B1+ inhomogeneities, due to the use of hard pulses of low flip angles which are played at the beginning of each velocity-encoding step for weighting of the excitation k-space (22,23). The accumulation of these low flip angles through all the velocity-encoding steps determines the flip angles of the static spins within the saturation band (ideally 90°). Poor B1+ scale degrades the net flip angle of the saturation band, and therefore results in nonuniform background suppression in angiograms.

Successive application of two VSS preparations was proposed to achieve more uniform saturation but at the cost of an additional signal drop for arterial blood and a doubled specific absorption rate (SAR) (22,23). Another limitation of the present cerebral VS-MRA protocol is its inability to distinguish between arteries and veins (22). The velocities and directions of intracranial arterial and venous blood do not differ as significantly as in peripheral vessels (23). No straightforward VS profile design could be used to differentiate arterial and venous blood in the complex cerebral vascular network.

In this work, we aimed to develop a 3D VS-MRA sequence that allows arteriography and venography with whole-brain coverage. To minimize the sensitivity to B0/B1 inhomogeneities, a low-flip-angle optimized composite (OCP) pulse was constructed to replace the hard pulse at the start of each velocity-encoding step of the VSS pulse train. To obtain well-separated cerebral arterio- and venograms, spatially selective inversion (SSI) pulses were applied to invert and null the outgoing venous blood or incoming arterial blood, respectively. The techniques were optimized and evaluated among healthy subjects for separately acquiring 3D arteriograms and venograms with whole-head coverage at 3T.

METHODS

Optimized Composite (OCP) Pulse

The VSS pulse train was constructed with a series of nine excitation pulses (10° each), interleaved with pairs of 180° hard pulses for refocusing and velocity encoding gradient lobes (Fig. 1). Each of the 10° excitations was achieved through an OCP pulse generated using an optimal control method for B0 and B1 insensitive excitation (24). This shallow tip pulse is composed of hundreds of block-shaped subpulses with their amplitudes and phases derived through numerical optimization. Input of the optimal control routine includes the desired flip angle, the maximum amplitude of the RF pulse, the duration of each subpulse and the number of the subpulses. Given the range of B0/B1 offset incurred in the brain at 3T: B0 = ±200 Hz and B1+ scale = ±0.2, we chose a slightly wider desired range of immunity to the B0/B1 inhomogeneities: B0 = ±250 Hz and B1+ scale = ±0.3. Note that individual B0 or B1 maps were not measured and optimization at run time was not required. The optimal control routine in MatPulse (25,26) was utilized to generate a 10° OCP pulse with 300 components. The input parameters set 13.5 μT as the peak amplitude and 6.4 μs dwell time as the duration of each subpulse, both given by the characteristics of the body coil of our 3T scanner.

Numerical simulations of the Bloch equations were performed using Matlab (MathWorks, Inc., Natick, MA, USA) to assess the properties of both the OCP pulse itself and the corresponding VSS pulse train. The velocity field of view (FOV_v) of the VSS pulse train was kept as 45 cm/s with a targeted saturation band within ±4 cm/s. Responses of the

longitudinal magnetizations (M_z) following the VSS pulse trains under various B_0/B_1 conditions were examined for velocities from -60 cm/s to 60 cm/s with intervals of 0.5 cm/s. The effect of T_1 or T_2 relaxation was ignored in the simulations.

Protocols for Arteriography and Venography

Variations of the VS MRA protocols were performed with the pulse sequence diagram shown in Fig. 2a. When no SSI pulse (Fig. 2a, green dashed box) is employed, the obtained angiogram depicts both arteries and veins with a whole-brain coverage (Fig. 2b, orange box). In order to remove the venous blood for obtaining an arteriogram, a SSI pulse is applied to the entire intracranial region (right above the petrous segment of internal carotid arteries) (Fig. 2c, green dashed box) with a delay time (T_{inv}) inserted before the VSS pulse train. The SSI pulse was a 20 ms-long HS n adiabatic pulse ($n = 4$, $\beta = 4$, bandwidth = 2500 Hz) (27,28) with a slab thickness of 130 mm. Since the T_1 of blood at 3T is about 1.8 sec (29,30) and the transit times from the carotid to intracranial vessels range from 0.5–1.2 sec (14), the inversion delay T_{inv} was set as 0.9 sec for a total TR per shot (TR_{shot}) of 2 sec. Thus the inverted outgoing venous blood magnetization recovers to the nulling point at acquisition while incoming fresh arterial blood fills the major head and neck arteries.

Similarly, for acquiring a venogram, the SSI pulse was applied with a 320 mm slab from the cervical region (right below the superior bulb) to the chest (Fig. 2d, green dashed box) where the incoming arterial blood within carotid and vertebral arteries are inverted. A 50 ms HS n pulse ($n = 4$, $\beta = 4$, bandwidth = 3000 Hz) (27,28) was used as the SSI pulse to better invert the blood in the chest. T_{inv} was set as 0.8 sec, which is slightly shorter than T_{inv} for the arteriography. This is mainly due to the observed lower inversion efficiency of SSI pulse at the chest where larger B_0 and B_1 inhomogeneities are present (30). This could also be partially caused by the additional T_2 decay during the 50 ms adiabatic pulse, during which the magnetization will have a transverse component (31).

Experiments

Experiments were performed on a 3T Philips Achieva scanner (Philips Medical Systems, Best, The Netherlands) using the body coil for RF transmission (maximum B_1 amplitude 13.5 μ T) and a 32-channel head-only coil for signal reception. The maximum strength and slew rate of the gradient coil were 40 mT/m and 200 mT/m/ms, respectively. Six healthy volunteers (age: 36 ± 9 years, three males and three females) and two patients (age: 46-year-old male, 48-year-old female) were enrolled after providing informed consent in accordance with the Institutional Review Board guidelines.

As part of scan planning, 2D PC-MRA survey images were acquired in the sagittal and coronal planes to display the major intracranial and neck arteries. Both scans used a 50 mm slab, repetition time/echo time (TR/TE), 20/6.4 ms; field of view (FOV), 250×250 mm²; scan matrix, 256×128 ; acquisition time, 20 s for each.

For each VS MRA protocol, the VSS pulse train was followed immediately by a SPIR module (spectral presaturation with inversion recovery) for fat suppression and the multi-shot turbo field echo (TFE) acquisition module with low-high profile ordering (32). A 220 mm-thick axial slab with a FOV of 180×180 mm² 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: TR / TE = 7.5 / 2.4 ms; flip angle = 7° to ensure the optimal signal strength during the transient state before reaching steady state (33); flow-compensation gradients applied in three orthogonal orientations; 62.5% of echo in the readout direction acquired; TFE factor = 70; TFE acquisition window = 520 ms; readout bandwidth = 134 Hz / pixel; SENSE factor = 2×2 (phase-encoding direction and slice direction). The time between the end of acquisition and the next SSI pulse ($T_{\text{pre_inv}}$) was set to be around 0.6 sec for $\text{TR}_{\text{shot}} = 2$ sec, and total scan duration was 5 min.

Five different VS MRA protocols were used for each volunteer: three without employing SSI pulses, and two with SSI pulses for arteriography and venography respectively. When SSI pulses were not applied, the performance of the VS MRA using the VSS pulse trains with 10° OCP pulses were compared with the ones using 10° hard pulses with one VSS preparation and the hard pulse-based ones with two consecutive VSS preparations, respectively. The exerted SAR of the VSS pulse trains using all hard pulses was 26% of the SAR limit for one VSS preparation and 52% for two VSS preparations, vs. 38% for the ones using OCP pulses. When SSI pulses were applied with 23% of the SAR limit for arteriography (20 ms HS_n pulse) and 57% of the SAR limit for venography (50 ms HS_n pulse), only VSS pulse trains with OCP pulses were performed. All SAR percentages are relative to 3.2 W/kg, the head-averaged SAR limit of the scanner. All the velocity-encoding gradients in VSS pulse trains were applied along the foot-head direction. No PPU triggering was utilized in these protocols.

For comparison, the standard TOF-arteriography was conducted through the circle of Willis with identical resolution and in-plane coverage as VS MRA. The through-plane coverage was 80 mm as typically prescribed for TOF in our center. Other parameters included: TR/TE = 26/5.8 ms; flip angle = 20° ; flow-compensation gradients applied; 62.5% of echo in the readout direction acquired; readout bandwidth = 72 Hz/pixel; 4 chunks acquired with 29 slices per chunk (5); the excitation pulse with linearly varying flip angles (start: 17° , end 23°) over the chunk (34); SENSE factor = 2 along the phase encoding direction; SAR = 100% of the limit. Total scan duration for TOF was 3.5 min.

VSS prepared arteriography was performed on two patients. To match the TOF protocol with the higher spatial resolution ($0.55 \times 0.55 \times 1.10 \text{ mm}^3$, 5 chunks) and relatively long acquisition time (5.5 min) used in these clinical exams, the VS MRA acquisition protocol was modified by increasing TR_{shot} to 2.5 sec, choosing SENSE factor of 2, and reducing spatial coverage to 77 mm. Given that the reduced imaging slab is covered by the SSI pulse, the signal of the static tissue was partly suppressed at the prescribed T_{inv} of 1.1 sec, in addition to the saturation effect of the VSS pulse. Thus 10° hard pulses, not the 10° OCP pulses, were employed in the VSS pulse train for the patient studies to have shorter velocity-encoding steps (5 ms).

Quantitative Assessment

In order to assess the performance of our various MRA protocols on healthy volunteers, relative contrast ratios between blood vessels and tissue background were analyzed as in our recent studies (22,23). Relative contrast ratio is defined as $(S_{a,v} - S_t) / S_{a,v}$, where $S_{a,v}$ is the

signal intensity of arterial blood or venous blood and S_t is the signal intensity of stationary tissue (the perfect relative contrast ratio is 1.0) (35). Regions of interest (ROIs) were placed on maximum-intensity-projection (MIP) images at major segments of both cervical and intracranial arteries (ICA: internal carotid artery; VA: vertebral artery; BA: basilar artery; ACA: anterior cerebral artery; MCA: middle cerebral artery, PCA: posterior cerebral artery), cerebral veins (IJV: internal jugular vein; TS: transverse sinus; SS: straight sinus; ISS: inferior sagittal sinus; ASSS: anterior superior sagittal sinus; MSSS: middle superior sagittal sinus; PSSS: posterior superior sagittal sinus), and stationary tissue.

RESULTS

Unlike the simple short hard pulse (Fig. 3a), the waveform of the OCP pulse is complexed with rapid phase modulation (Fig. 3c). The pulse duration of a 10° OCP pulse was 1.92 ms (Fig. 3c) vs. 48 μ s for the 10° hard pulse (Fig. 3a). The simulation results show that the acting OCP pulse offers sufficient bandwidth and much less sensitivity to B1 inhomogeneity than the hard pulse (Fig. 3b vs. 3d). The color scale is the Mz magnetization after experiencing the pulses. The blue lines indicate the typical range of B0/B1 offset incurred in the brain at 3T. The OCP pulse returned by the optimal control routine delivered uniform tipping within the specified range of B0/B1 offsets.

The duration of each velocity-encoding step increased from 5 ms with the hard pulse to 8 ms with the OCP pulse. The total duration of the VS pulse train (T_{VS}) with eight steps expanded from 40 ms to 64 ms (Fig. 1). The Mz responses of VSS pulse trains employing nine 10° hard pulses and 10° OCP pulses are displayed over the plane of velocity vs. B0 off-resonance frequency (Figs. 4a,b) and B1+ scales (Fig. 4c,d), respectively. When the 10° hard pulses were used for excitation, the Mz signal intensity in the saturation band suffers from B1+ sensitivity (Fig. 4c). In contrast, the substitution with 10° OCP pulses significantly mitigates the dependence on the B1+ scale (Fig. 4d). For B1+ scale = 0.8 or 1.2, Mz of the static spins ($v = 0$ cm/s) are 0.06 and -0.01 for the VSS pulse train with OCP pulses (Fig. 4d), compared to 0.3 and -0.3 with hard pulses (Fig. 4c).

The MIP images of angiograms (without applying SSI pulses), arteriograms and venograms of one subject are displayed in Fig. 5 with three orthogonal orientations. Note that these MIP images were produced directly from raw data without applying any manual enhancement and are exhibited with the same intensity scales. Compared to running the VSS preparation with hard pulses once (Fig. 5a), much more uniform suppression of background tissues throughout the FOV is evident when either repeating the VSS preparation with hard pulses twice (Fig. 5b) or replacing with the OCP pulses (Fig. 5c). The arteriograms depict all the major intracranial and cervical arteries and their small branches (Fig. 5d). The venograms delineate all the major intracranial and cervical veins with some arterial contamination (mainly in carotid and vertebral arteries) (Fig. 5e). Combination of the arteriograms (Fig. 5d) and venograms (Fig. 5e) resembles the full angiograms (Fig. 5c).

The MIP images of the VS MRA using the improved VSS pulse trains of all six subjects are arrayed in Fig. 6 with the top row for the coronal view of arteriograms and the bottom row for the sagittal view of venograms, respectively. Blood vessels from the bifurcation of

carotid arteries up to the superior top of the brain are delineated in the same scan with this method (Fig. 6). MIP images from a fraction of the whole-brain arteriogram of VS MRA are illustrated in Fig. 7, which has the same spatial coverage as TOF MRA. In addition to offering almost 2 times the volume coverage per unit time (220 mm in 5 min vs. 80 mm in 3.5 min), the extracted slabs from the VS MRA depict more small distal branches with slow flow at directions parallel to the axial slab.

ROIs placed on MIP images are exhibited in Supporting figure S1. Table 1 shows the mean relative contrast ratios of various arterial and venous ROIs of 5 different VS MRA protocols across 6 subjects. The averaged relative contrast ratios of the full angiograms using OCP pulses are higher than those using one preparation of VSS pulse trains with hard pulses for both arteries (0.87 ± 0.03 vs. 0.77 ± 0.04) and veins (0.80 ± 0.07 vs. 0.59 ± 0.13). Compared with employing two consecutive hard pulse-based VSS preparations, using one preparation of OCP-based VSS have similar relative contrast ratios for the arteries (0.87 ± 0.03 vs. 0.86 ± 0.03), and 14% higher for the veins (0.80 ± 0.07 vs. 0.70 ± 0.12). This is most likely a result of the shortening of the overall VS pulse duration by eliminating the second preparation (64 ms vs. $40 \times 2 = 80$ ms), which has more effects on the venous blood with much shorter T_2 relaxation times relative to arterial blood (~ 60 ms vs. ~ 150 ms, (36,37)). In addition to the increase of the mean of the relative contrast ratios, also noticeable is the decrease of the standard deviation (STD) of these numbers of veins (Table 1), which reflects a more uniform vessel contrast to background. At the same time, both the relative contrast ratios of arteries in the arteriograms and veins in the venograms are comparable to those in the angiograms without separating them (0.90 ± 0.03 vs. 0.87 ± 0.03 and 0.78 ± 0.08 vs. 0.80 ± 0.07), respectively, which indicates that the inserted SSI pulses did not interfere with the desired vessel signal.

Figs. 8 and 9 show two clinical examples of VS MRA in patients with intracranial stenosis, along with the corresponding TOF images. While the two approaches show great overall correlation in visualizing the arteries, VS MRA shows the potential to delineate distal arteries which likely involve very slow flow. More uniform suppression of tissue background by VS MRA is also observed for both the intracranial and cervical regions.

DISCUSSION

In this work, we proposed two additional modifications for the emerging VS MRA technique: 1) replacing the low-flip-angle hard pulses with B1+-robust OCP pulses for excitation at the start of velocity-encoding steps of VSS pulse trains; 2) adding a SSI pulse with a delay before applying the VSS pulse train such that the inverted magnetization of downstream venous blood or upstream arterial blood recovers to the nulling point at acquisitions, resulting in an arteriogram or venogram, respectively. The large spatial coverage demonstrated with VS MRA, although requiring more k-space points to be acquired, does not necessarily increase the scan time proportionally since it can benefit from an additional parallel-imaging acceleration along the slice-encoding direction. Our VS MRA protocols were acquired within 5 min using SENSE factor = 2×2 for a 220 mm slab (Fig. 6), while TOF MRA was obtained in 3.5 min using SENSE factor = 2×1 for a 80 mm slab. In addition to the doubled volume coverage per unit time, small vessel delineation is more

conspicuous with VS MRA than with TOF for the volunteers (Fig. 7). Initial clinical feasibility was shown in two patients with comparison to TOF protocols with matched resolution, spatial coverage and acquisition time (Figs. 8 and 9).

The utilization of nine 10° OCP pulses (~ 2 ms each) in the VSS pulse trains is needed to yield uniform 90° flip angles for static tissue counteracting the presence of large B_0/B_1+ inhomogeneities. With the same RF hardware limits (peak amplitude and dwell time), the well-known BIR-4 pulse (38) can generate arbitrary flip angles but requires a much longer pulse duration (at least 8 ms) and much higher SAR value. Numerically optimized composite pulses for low flip angles have been investigated more recently (24,39–41). The computation scheme developed by Boulant et al. (39,40) requires the calculation of pulses on the fly, after obtaining B_0/B_1+ maps for each subject as the input for the optimization algorithm (dubbed SMP: strongly modulating pulse). Moore et al. (41) has generated pulses using a non-linear constrained minimization algorithm (dubbed OPT: optimized composite pulse). The design of the OCP pulses in our work adopted the optimization routine described in Liu and Matson's paper (24) (dubbed NSS: non-slice-selective pulse), which was based on earlier studies in the NMR field on broadband excitation pulses using the optimal control theory by Skinner et al. (42,43). OCP pulses based on an optimal control algorithm were shown to offer similar performance but with shorter pulse duration, due to the composition of a much larger number of subpulses (24).

For the new pulse train, it is noticed that M_z of the static spins is -0.1 when the B_1+ scale is at the correct setting (Fig. 4b) rather than 0 with hard pulses (Fig. 4a). This is speculated to be related to the fact that the optimization criterion is targeted for the individual OCP pulse with small tip angles, not for concatenation of a series of these pulses towards a large flip angle (24). Despite this deviation of M_z within the saturation band when $B_1+ = 1.0$, it still provides a much more uniform tipping in the full range of B_1+ scales under investigation (Fig. 4d), compared to its counterparts with hard pulses (Fig. 4c). This issue may be mitigated by slightly reducing the targeted flip angles for each velocity-encoding step.

The proposed approach to differentiate arteries and veins is through the employment of a SSI pulse with an inversion delay set to null the outgoing venous blood or incoming arterial blood. This strategy has been applied successfully in NCE abdominal MRA by inverting and nulling the static tissue while allowing the fresh blood to flow in (44–48). Since tissues often have different structures with different T_1 values (such as white matter, gray matter, and cerebrospinal fluid, whose T_1 values span from 0.7 sec to 4.5 sec at 3 T (49,50)), a single SSI pulse with a preset delay would not diminish the entire signal from the stationary background. In contrast, our arteriography and venography techniques utilize a preceding SSI pulse to invert and null the undesired blood signal and a sophisticated VSS pulse train to suppress static tissue right before image acquisition.

Note that blood T_1 values depend on both hematocrit and oxygenation (30,51–56) and range from 1.6 sec to 2.0 sec at 3T for healthy subjects (29,30). When blood T_1 values deviate ± 0.2 sec from the assumed 1.8 sec, their recovered absolute signal at acquisition would increase less than 6% of the equilibrium magnetization, based on our sequence timing for arteriography ($T_{\text{inv}} = 0.9$ sec, $TR_{\text{shot}} = 2$ sec). For patients with abnormal blood

compositions (such as anemia), blood T_1 can be quickly (1 min) and noninvasively (no blood draw) estimated using recently developed techniques (29,30,57–59).

One limitation of the current method to suppress either upstream arterial or downstream venous blood signal by exerting a SSI pulse with a delay is its sensitivity to slow inflow of arterial blood. Although the 0.9 sec T_{inv} used for six healthy subjects (Fig. 6 and Fig. 7) and the 1.1 sec T_{inv} for the two clinical examples (Fig. 8 and Fig. 9) worked well, they may not be sufficient for patients with significantly reduced blood flow. With slow arterial inflow, distal arterial branches would not be replenished by fresh arterial blood signal within T_{inv} and thus be less visualized in the arteriograms or not filled-in by nulled arterial blood signal and thus contaminate venograms; similarly, with slow venous outflow, large venous vessels would be less visualized in the venograms or contaminate arteriograms. From the clinical experience in NCE abdominal MRA (44–48), one needs to consider the balance of the blood inflow time and the signal recovery of the tissue background. T_{inv} could be lengthened to allow more fill-in of the incoming arterial blood, but at the cost of more recovery and thus less suppression of unwanted signal.

For the proposed venography protocol, the observed arterial contamination (Fig. 5e) is a result of incomplete inversion of the incoming arterial blood. Despite the use of an advanced adiabatic SSI pulse to reduce sensitivity to large B_0 and B_1 inhomogeneities over the chest, the inversion is still limited by the length of the body coil (65 cm). In our recent quantification work for arterial blood T_1 (30), we had to reposition the subjects to align the center of the body coil from the subject's eye level to his/her clavicle in order for proper inversion recovery of all the feeding blood covered by the RF coil.

The dependence of the blood signal on the velocity-encoding direction of the VSS pulse train was demonstrated in our earlier work (22) and is also evident for certain vessels which run perpendicular to the foot-head direction applied in this study, e.g. middle superior sagittal sinus (Fig. 6 and Table 1). Multidirectional flow-signal suppression has been achieved in flow-sensitive dephasing (FSD) magnetization prepared MRA by employing two-FSD modules consecutively with velocity-encoding gradients applied along orthogonal directions (60). Unfortunately, this strategy does not apply to our VS MRA technique since the flow-signal saturated by one VS pulse train with a perpendicular velocity-encoding direction would be dephased and not be affected by following VSS pulse train.

Horizontal stripe artifact observed in the background of both arteriograms and venograms (Fig. 6) is a combined result of imperfections of both VS gradient (e.g. eddy currents) and refocusing pulses (B_0/B_1 inhomogeneities) in the VSS pulse trains (22,23). In each velocity-encoding step, ideally, tissue spins dephased by the 1st unipolar gradient should be rephased by the last unipolar gradient if the refocusing between them works perfectly. However, imperfect unipolar gradient and 180° rotation hamper perfect rephrasing, and leave a fraction of spins still dephased at the end of each encoding step. Since the stripes can be approximated as sinusoidal with a period determined by the area of the unipolar gradient, a recent study (23) proposed to apply two VS preparation pulses which have the same excitation profile over velocity but spatially shifted by half the period of the stripes. This

approach has shown to substantially suppress the stripes but at the cost of increased SAR and reduced arterial signal intensity.

CONCLUSION

We have improved velocity-selective saturation (VSS) pulse trains in terms of reducing sensitivity of background tissue suppression to B1+ inhomogeneity in non-contrast-enhanced MRA spanning both for the intracranial and cervical spine regions. Compared to existing techniques, the proposed OCP-based VSS pulse trains rendered improved contrast of vessels relative to background. Moreover, we obtained arteriograms or venograms separately by placing spatially selective inversion pulses before the VSS-prepared acquisition to selectively null signals from venous or arterial blood. The feasibility of these technical advances for VS MRA was demonstrated on both healthy volunteers and patients at 3T, which efficiently delineated tortuous and complex neurovascular tree with large spatial coverage. The clinical impact of these techniques needs to be evaluated among a large group of patients with various cerebrovascular disorders.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.

Diagram of the Fourier transform based velocity-selective saturation (VSS) pulse train using a hard pulse or an optimized composite (OCP) pulse as the low-flip-angle excitation pulse (green) for each velocity-encoding step, along with paired and phase-cycled refocusing pulses (blue) and surrounding gradients with alternating polarity for velocity-sensitization (red).

Figure 2.

(a) The VS MRA sequence diagram with the VSS pulse train preceding the acquisition with whole-brain coverage (b, the orange box). When either arteriogram or venogram is desired, the spatially selective inversion (SSI) pulse is applied at the intracranial (c) or cervical (d) regions (inside the green dashed box) with an inversion delay (T_{inv}) to null the downstream venous blood or upstream arterial blood, respectively.

Figure 3.

The RF waveform of the (a) 10° hard pulse and (c) 10° OCP pulse with its x and y components of the amplitude generated using an optimal control routine. The simulated M_z responses to different B0 off-resonance frequencies and B1+ scales are shown for (b) hard pulse and (d) OCP pulse, respectively.

Figure 4.

The simulated Mz-velocity response after applying VSS pulse trains with (a, c) 10° hard pulses and (b,d) 10° OCP pulses at different B0/B1 conditions: (a, b) B0 off-resonance from -200 Hz to 200 Hz, B1+ scale = 1.00; (c, d) B0 off-resonance = 0 Hz, B1+ scale from 0.8 to 1.2. The OCP pulses lead to much less dependence on various B1+ scales within the saturation band for stationary tissue (c, d).

Figure 5.

Representative MIP images of angiograms with 3 orthogonal views: (a) using 10° hard pulses as the excitation pulses in VSS pulse train; (b) same as (a) but with two successive VSS preparations; (c) using the 10° OCP pulses; (d) arteriogram and (e) venogram acquired using the inserted SSI pulses and 10° OCP pulses.

Figure 6.
The MIP images of the arteriograms (top row) and venograms (bottom row) from all 6 subjects using the VSS pulse train with the OCP pulses.

Figure 7.

Compared to MIP images of TOF MRA (top row) from all six subjects, VS MRA extracted from the whole-brain arteriogram (bottom row) with the same spatial coverage as TOF delineate more small distal branches with slow flow.

Figure 8.

Axial MIPs of a 48-year-old female patient with moderate P2 stenosis. Compared to (a) TOF MRA, right distal P2 branches (arrows) are better depicted by (b) VS MRA.

Figure 9. Coronal MIPs of the extracted MCAs of a 46-year-old male patient with focal lesion within the left M1 segment (arrows): (a) TOF MRA; (b) VS MRA.

Table 1

Quantitative measurement of relative contrast ratios in major cerebral vascular segments for different VS MRA protocols. The ROIs are labeled in the corresponding vessels as shown in Supporting figure S1.

	Angiogram		Arteriogram		Venogram	
	hard pulses ×1	hard pulses ×2	OCP pulses	OCP pulses	OCP pulses	OCP pulses
ICA	0.83+/-0.04	0.91+/-0.01	0.91+/-0.01	0.93+/-0.01		
VA	0.76+/-0.07	0.87+/-0.03	0.89+/-0.02	0.91+/-0.01		
BA	0.81+/-0.04	0.88+/-0.03	0.91+/-0.02	0.93+/-0.01		
ACA	0.75+/-0.03	0.84+/-0.02	0.83+/-0.05	0.86+/-0.03		
MCA	0.77+/-0.04	0.83+/-0.03	0.86+/-0.04	0.88+/-0.03		
PCA	0.71+/-0.05	0.82+/-0.03	0.85+/-0.03	0.89+/-0.02		
Average+/-Std	0.77+/-0.04	0.86+/-0.03	0.87+/-0.03	0.90+/-0.03		
IJV	0.74+/-0.05	0.84+/-0.02	0.87+/-0.01		0.86+/-0.01	
TS	0.55+/-0.12	0.64+/-0.06	0.79+/-0.04		0.78+/-0.04	
SS	0.68+/-0.07	0.79+/-0.05	0.85+/-0.04		0.84+/-0.03	
ISS	0.46+/-0.16	0.59+/-0.06	0.82+/-0.04		0.79+/-0.06	
ASSS	0.60+/-0.11	0.69+/-0.08	0.75+/-0.07		0.76+/-0.03	
MSSS	0.39+/-0.14	0.54+/-0.07	0.66+/-0.09		0.62+/-0.14	
PSSS	0.68+/-0.06	0.83+/-0.03	0.84+/-0.04		0.84+/-0.04	
Average+/-Std	0.59+/-0.13	0.70+/-0.12	0.80+/-0.07		0.78+/-0.08	

ICA = internal carotid artery; VA = vertebral artery; BA = basilar artery; ACA = anterior cerebral artery; MCA= middle cerebral artery; PCA = posterior cerebral artery; IJV = internal jugular vein; TS = transverse sinus; SS = straight sinus; ISS = inferior sagittal sinus; ASSS = anterior superior sagittal sinus; MSSS = middle superior sagittal sinus; PSSS = posterior superior sagittal sinus.