Flow-Sensitized Dephasing Prepared SSFP: A New Noncontrast MRA Technique

Z. Fan[1,2], J. Sheehan[1], X. Bi[3], R. Dharmakumar[1], R. Jerecic[3], J. Carr[1], and D. Li[1,2]

[1]Radiology, Northwestern University, Chicago, IL, United States, [2]Biomedical Engineering, Northwestern University, Evanston, IL, United States, [3]Siemens Medical Solutions USA, Inc., Chicago, IL, United States

Introduction: While catheter angiography remains the gold standard for evaluation of vascular disorders, 3D contrast-enhanced MRA is becoming the method of choice due to its high diagnostic accuracy. However, safety concerns with contrast administration in patients with renal insufficiency have triggered a renaissance of interest in non-contrast MRA (NC-MRA). The conventional NC-MRA techniques, including time-of-flight and phase contrast, fail to gain clinical acceptance due to their lengthy scan times and possible overestimation of degree and length of stenosis.

Recently, a flow-sensitized dephasing (FSD)-prepared SSFP was developed for vessel-wall imaging [1]. The flow-void effect of the FSD module is based on intravoxel dephasing in moving spins as determined by flow velocity and the first-order gradient moment, m_1, imparted by the FSD gradient pulses. Due to the marked velocity difference between arterial and venous flows at systole, we hypothesized that an appropriate m_1 could selectively suppress the arterial blood signals while having little effect on the venous blood and static tissues. Subtraction between bright-artery scan at mid-diastole using SSFP and dark-artery scan at systole using FSD-prepared SSFP would result in an artery-only data set. This work aimed to investigate the feasibility of this new NC-MRA technique on multiple vascular territories.

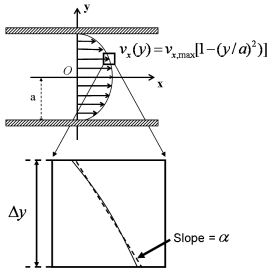

Fig. 1. 1D laminar flow model. Flow velocity profile is described by a parabolic equation. Velocity varies along y-axis linearly. if the pixel is small.

Theory: An optimal m_1 may exist for selective suppression of arterial blood signal according to a 1D laminar flow model as shown in Fig. 1 [2]. Specifically, if velocity only varies in y-axis and the pixel is small enough so that the velocity variation within a pixel is approximately linear, the reconstructed signal will be given by $|\rho| = |\rho_0| \cdot |\mathrm{sinc}(\gamma \alpha m_1 \Delta y/2)|$, where ρ_0 is the signal being imaged, α is the slope of velocity variation within a pixel, and Δy is the pixel size. As $\gamma \alpha m_1 \Delta y/2 = 1$ (i.e. at the first zero-crossing point of the sinc function), $\rho = 0$, meaning that flow signal is completely eliminated. The greater flow velocity in the arterial blood results in a greater α, and thus $\gamma \alpha m_1 \Delta y/2$ approaches 1 more rapidly in arterial flow. In addition, the bipolar FSD module may outperform the conventional unipolar module when m_1 is small (Fig. 2). Both modules employ 90^0x-180^0y-90^0-x driven equilibrium. In unipolar module, the net phase $\varnothing$ in the stationary spins accumulated before and after the 180^0-pulse cannot be canceled out with each other due to imperfect radio-frequency (RF) response of the hard pulse, potentially resulting in a spatially periodic signal modulation, or stripe artifacts. The artifacts become more pronounced as m_1 is reduced. Bipolar FSD provides a solution by making $\varnothing$ to be 0.

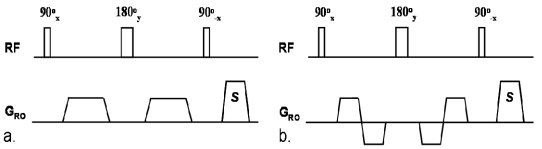

Fig. 2. Unipolar (a) and bipolar (b) FSD modules. S: spoiling.

Materials and Methods: Healthy volunteers were imaged using a 1.5T MR system (MAGNETOM Avanto, Siemens). The proposed method was tested on the major arteries at lower extremities, feet, hands, and carotids. Relevant imaging parameters are shown in Table 1. In each session, a phase-contrast flow scan was first performed at the arteries of interest to derive the arterial flow peak time *T*. Subtraction MRA was subsequently conducted, consisting of one bright-artery scan at mid-diastole and one dark-artery scan triggered at *T* employing various m_1's as estimated by the 1D laminar flow model. Subtraction of the two matched scans created a subtraction data set from which maximum intensity projections (MIP's) were created for qualitative evaluation of optimal m_1.

Table 1 Relevant imaging parameters (* optimal values (the range tested))

	Leg	Hand	Foot	Carotid
Coils	Peripheral & spine	Body matrix	Head	Head & Neck
Spatial resolution (mm)	0.8-1.2	0.98	1.0	0.83
m_1 (mTms^2/m)	35 (17-87)*	98 (58-156)	292 (195-390)	28 (10-50)
SSFP parameters	TE/TR=1.5-1.9/3.1-3.8 ms, BW=825-965 Hz/pixel, 3 shots/partition, FA=70-90°			

Results: Stripe artifacts with unipolar FSD module interfered with arterial visualization on MIP images, which were removed by the bipolar FSD module (Fig. 3). Isotropic high spatial-resolution MRA was successful in multiple vascular territories with an optimized m_1 as estimated using a 1D laminar flow model (Fig. 4-7). As m_1 increased, more veins appeared contaminating arterial delineation in MIPs.

Conclusions: Preliminary results have demonstrated the feasibility of this NC-MRA approach. Systematic optimization of m_1 is warranted for various clinical applications. The flexibility of adapting the FSD strength to individual flow conditions of the patient for different vascular territories and degrees of disease is a major advantage of this new approach. This technique has the potential to be used as a screening tool for whole-body vascular examination.

References: 1. Koktzoglou I, et al. JCMR 9:33 (2007). 2. Haacke EM, et al. Magn. Reson. Imag. physical principles and sequence design. (1999).

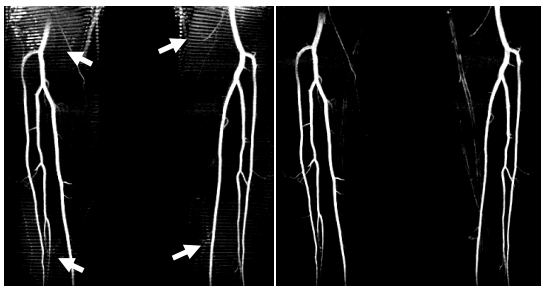

Fig. 3. MIPs of subtraction images. The stripe artifacts (arrows) shown on unipolar FSD images (left) are eliminated using bipolar FSD (right). m_1= 35 mTms^2/m.

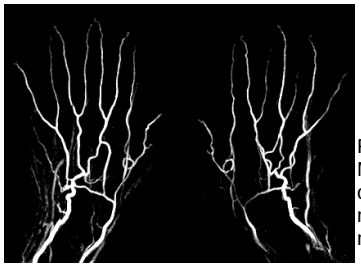

Fig. 5. MIP MRA of hands. m_1 = 98 mTms^2/m.

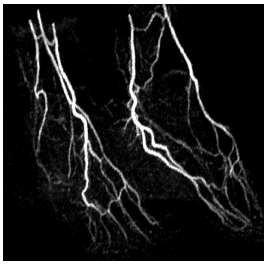

Fig. 6. MIP MRA of feet with 3D reformation. m_1= 292 mTms^2/m.

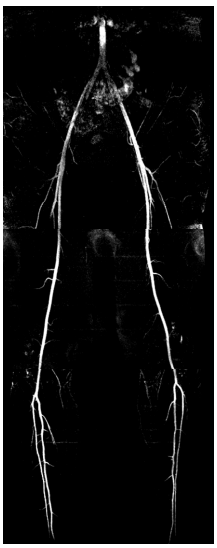

Fig. 4. MIP of three-station MRA of lower extremities. Major arterial banches are depicted. Minor venous signal are observed in the thigh. Signal loss at pelvis is most likely due to respiratory motion and bowel peristalsis. m_1=17-52 mTms^2/m.

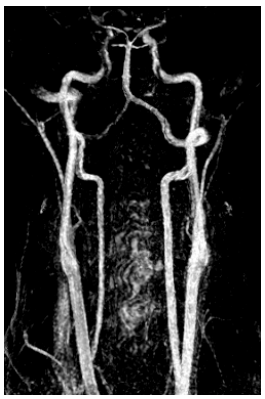

Fig. 7. MIP MRA of carotids. m_1= 28 mTms^2/m.