Non-contrast-enhanced peripheral MRA using velocity-selective saturation preparation

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Introduction: Flow-sensitive ECG-gated subtractive methods have been applied to non-contrast-enhanced (NCE) lower leg MRA with great promise [1-4]. While these methods have shown excellent artery-background contrast, they require two data acquisitions and may be problematic in patients with elevated diastolic flow or arrythmia. We aim to develop a novel NCE lower extremities MRA method which allows high artery-background contrast from a single 3D acquisition. This was achieved by employing velocity-selective (VS) magnetization preparation at systole which saturates all tissues except arterial blood. Excellent depiction of arteries is demonstrated on healthy volunteers.

Methods: VS saturation pulse design: Based on the excitation *k*-space formalism, a spatially non-selective, VS excitation can be achieved by playing RF sub-pulses between a series of bipolar gradients, each of which induces phase as a periodic function of spin's velocity [5,6]. The envelope of the RF sub-pulses can be designed by the Shinnar-Le Roux transform for high velocity selectivity with large flip angles [7]. Figure 1 shows an example of a VS excitation pulse sequence used in this study (a), and Bloch simulation of the resultant longitudinal magnetization displayed as a function of velocity and off-resonance (b). Designed with a velocity saturation bandwidth of 20 cm/s and a velocity FOV of 60 cm/s, this pulse saturates stationary tissue while barely affecting arterial blood with a velocity of 20-40 cm/s in the inferior direction. Note that the velocity profile is shifted as a linear function of off-resonance.

In-vivo experiments: The pulse sequence consisted of a VS saturation pulse played at the time of peak systolic flow, immediately followed by a fat saturation pulse and a balanced SSFP readout with a centric view order. VS pulses with six velocity saturation bandwidths (12, 16, 20, 24, 28 and 32 cm/s) were tested on healthy volunteers on a GE 1.5 T scanner with a quadrature birdcage coil. The imaging parameters were coronal imaging slab, spatial resolution = 1.1×1.1×1.1 mm^3, FOV = 28×28×7.7 cm^3, TR = 5.4 ms, and FA = 90°. Signal intensities were measured from 4 arterial ROIs (denoted by blue asterisks in Fig. 3) and neighboring background ROIs per leg (8 pairs of arterial and background ROIs per image).

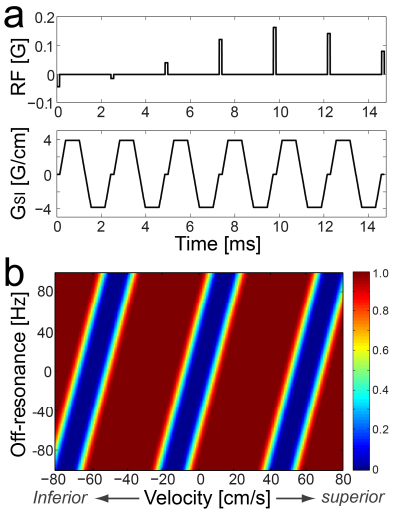

Figure 1. (a): VS saturation pulse sequence; (b): Bloch simulation of normalized longitudinal magnetization M_z/M_0.

Results: Figure 3 contains partial coronal maximum-intensity projection (MIP) images along with simulated longitudinal magnetization profiles of the corresponding VS pulses. With a narrower velocity saturation bandwidth, arterial signal appear brighter but venous and background signals are also increased (open arrow). Inhomogeneous background suppression is seen due to the velocity profile shifting caused by field inhomogeneity. As the velocity saturation bandwidth increases, background signals are more uniformly suppressed, but at the expense of a slight loss of arterial signal (solid arrow). The mean and standard deviation of arterial and background ROIs show high artery-background contrast over most of the saturation bandwidths tested (Fig. 2).

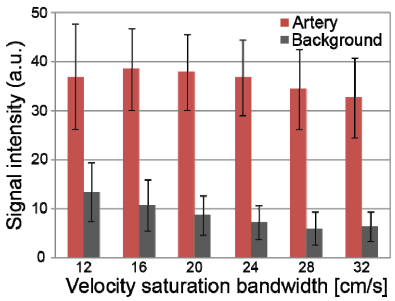

Figure 2. Signal intensities of arterial and background ROIs over a range of velocity saturation bandwidths.

Discussion: Unlike the motion-sensitizing gradient used in recent studies [3,4], the VS pulse used in this study modulates the amplitude of each magnetization with a desired velocity profile and thus enables a direct acquisition of a bright artery image without subtraction. The proposed method is robust to gating errors due to the use of magnetization saturation and also to arrhythmia due to the use of systolic phase which is less sensitive to R-R variability [8]. While a range of velocity saturation bandwidths yielded high artery-background contrast, the optimal bandwidth needs to be further investigated in a large cohort of subjects. Field inhomogeneity-induced velocity profile shifting may be a practical issue and its mitigation by incorporating refocusing pulses is under investigation.

References: [1] M Miyazaki et al., JMRI 12: 776-783, 2000. [2] M Miyoshi et al., ISMRM 2007:429. [3] Z Fan et al., MRM 62:1523-1532, 2009. [4] A Priest et al., ISMRM 2011: 90. [5] L de Rochefort et al., MRM 55:171-176, 2006. [6] T Shin et al., MR Angio club 2011:12.8. [7] J Pauly et al., IEEE TMI 10:53-65, 1991. [8] AM Gharib et al., JMRI 26: 921-926, 2007.

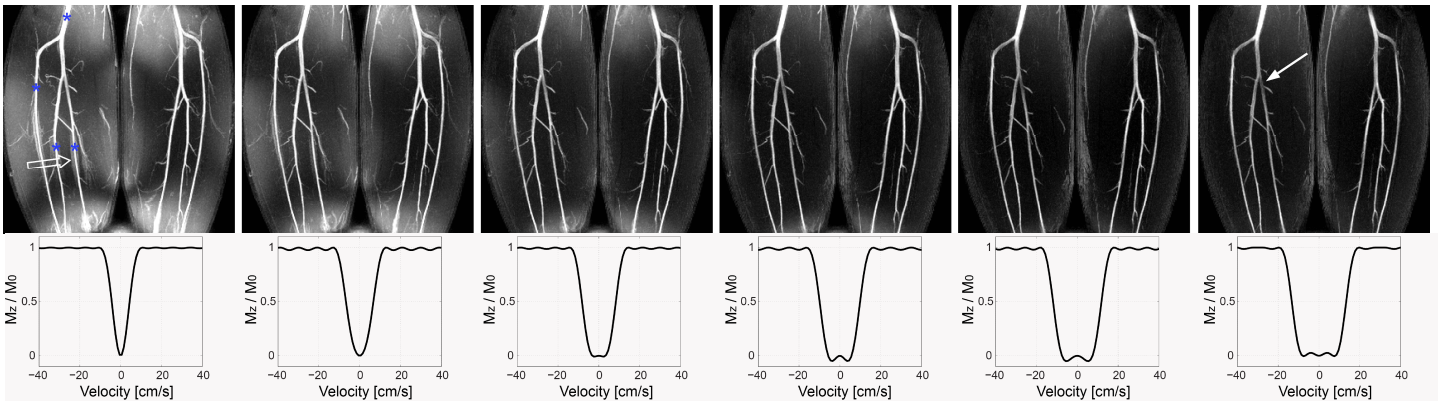

Figure 3. Coronal MIP images of lower leg MRA (top row) and simulated velocity profile of corresponding VS pulses on resonance (bottom row).