Flow Measurement using Arterial Spin Labeling with Flow Discrimination by Cumulative Readout Pulses

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Background: The Pennes bioheat transfer equation (BHTE) is the most widely used equation to model the effects of heat deposition and dissipation in tissues. The formulation includes terms for thermal conductivity and an effective perfusion, which represents the rate at which blood flow removes heat from a local tissue region. MR thermometry has allowed accurate estimations of these subject-specific thermal properties [1]. Using these estimated parameters enables more accurate treatment planning. However, tissue properties, particularly perfusion, are known to change over the course of a thermal therapy treatment [2, 3]. Detecting perfusion changes during a thermal therapy treatment would allow for the adjustment of treatment parameters to achieve a more efficacious therapy. In this work, we present a method to use arterial spin labeling (ASL) to determine the rate at which flow passes through a point. The pulse sequence combines the turbo-FLASH (TFL) imaging and Look-Locker-like [4] readout at multiple inversion times (*TI*) in a single scan. The data obtained from this newly developed sequence approximates the average velocity of blood (fluid) passing through a thin slice, providing a surrogate for the Pennes' perfusion term. This method is independent of MR thermometry, decoupling the blood flow measurement from the MR temperature maps, allowing the perfusion changes to be monitored throughout the thermal therapy session.

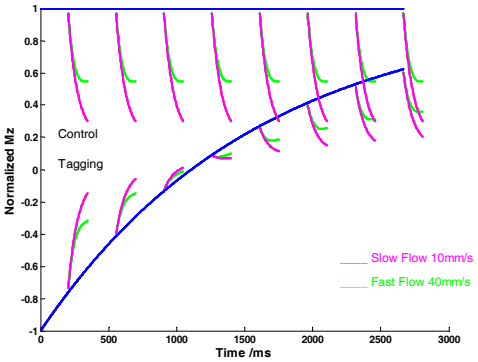

Fig. 1 Normalized longitudinal magnetization evolution of inflowing tag and control signals at TI of [200, 552, 904, 1256, 1608, 1960, 2312, 2664] ms for flow velocities of 10 mm/s(magenta) and 40 mm/s (green). 64 phase-encoding lines are acquired, and a linear k-space ordering, i.e., k-space center being acquired at 32nd line, shows velocity dependency — the slower the velocity, the further the signal deviates from the main inversion recovery curve.

Theory: IDOL magnetization preparation [5] is performed, and then, unlike conventional methods, multiple TFL readouts with linear k-space ordering are employed to capture the flow information encoded in the train of readout excitation pulses. Different flow velocities experience a different number of alpha readout pulses, which can be used to estimate velocity. Based on the Bloch equations, the simulation of the magnetization evolution of labeled and unlabeled inflowing fluid is presented in Fig. 1. In the modeling, each slice is subdivided into *N* partitions, where *N=D/(v·TR)*, *D* is the slice thickness, *v* is the flow velocity, and *TR* is the time interval between the excitation pulses. Each *TR*, the magnetization goes through excitation, recovery and shifting to the next sub-slice due to flow. The resulting signal is averaged from all *N* sub-slices. A hemodialyzer, which has thousands of fibers with a diameter on the order of hundreds of microns each, was selected as a phantom to mimic human tissue perfusion and was tested with a wide range of flow rates. Due to fluid exchange, signal from intra- and extra-fiber compartments should be taken into account. Mathematically, we derived the tag signal as a weighted average of signal from stationary fluid (S_s) and flowing fluid (S_f), i.e., $S_{tag} = f \cdot S_s + (1-f) \cdot S_f$,

with $S_s(n) = \frac{M_0(1-E)[1-(EC)^{Ny/2}]}{1-EC} + (EC)^{Ny/2+(n-1)Ny} M_0 E^{n-1} + \frac{M_0(1-E)\cdot(EC)^{Ny/2}\cdot(1-[E\cdot(EC)^{Ny}]^{n-1})\cdot(1-EC+E[1-(EC)^{Ny}])}{[1-E(EC)^{Ny}](1-EC)}$;

$S_f(n) = \frac{M_0(1-E)}{1-EC} + \frac{N-Ny/2}{N}(EC)^{Ny/2}[\frac{M_0E(1-C)}{1-EC} - 2E_1] + \frac{M_0E(1-C)[1-(EC)^{Ny/2}]}{N(1-EC)^2} - \frac{2M_0E_n[(EC)^{Ny/2}-E^{Ny/2}]}{N(C-1)}$

where $E = \exp(-TR/T_1)$, $E_1 = \exp(-TI(1)/T_1)$, $E_n = \exp(-TI(n)/T_1)$, $C = \cos(\alpha)$, *Ny* is the number of phase encoding lines, *n* is the index of *TI*, α is flip angle, and *f* is the fraction of the stationary fluid.

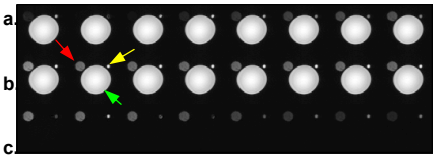

Fig. 2 Example axial (a) tag, (b) control, (c) difference images acquired at TI of [200, 600, 1000, 1400, 1600, 2000, 2400, 3000] ms. Each image is composed of cross-sections of 4-cm diameter dialyzer (red), static water phantom (green) and thin tubing with opposite directional flow (yellow).

Methods: A single axial slice of a commercially available hemodialyzer (Baxter Xenium-190) was imaged using a head coil on a Siemens 3T scanner with the 2D IDOL-prepared TFL sequence at a series of *TI* = [200, 600, 1000, 1400, 1600, 2000, 2400, 3000] ms in one scan. A non-pulsatile pump circulated tap water (*T1*=2.8s) at rates of [150, 300, 450, 600] cc/min through the semi-permeable dialyzer fibers. A static water phantom was included to test the complete cancellation of static signal between tagging and control. Linear phase-encoding ordering was employed to capture the difference caused by four flow rates. Other imaging parameters were *TR* = 3 ms, *FA*=α =15°, voxel size = 3.3x3.3x5 mm^3. Velocities and stationary fluid fraction were estimated from an exponential fit to the average signal over the dialyzer cross-section at each *TI*.

Results: Example tag (a), control (b) axial images acquired at multiple *TI* are shown in Fig. 2. In each image, the largest area is a static water phantom, with the dialyzer at the left and thin tubing at the right. A complete cancellation of static signal in the difference images (c) indicates that the signal is purely flow-dependent. The measured averaged difference signal from the hemodialyzer vs. *TI* plotted in Fig. 3(a) is consistent with the simulations given in Fig. 3(b) for the four flow rates. The flow velocities estimated from curve fitting are [30, 37, 48, 75] mm/s and the estimated stationary fluid fraction is [0.77, 0.73, 0.78, 0.78]. These velocities are consistent with an expected increase in flow in the stationary compartment with increasing volume flow.

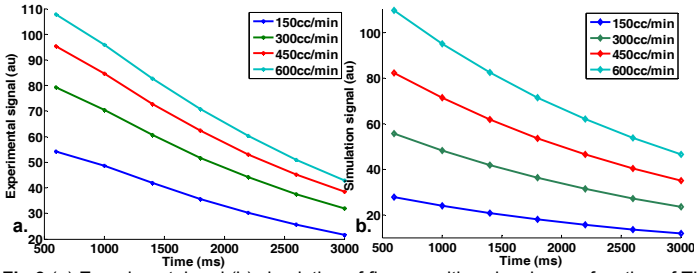

Fig.3 (a) Experimental and (b) simulation of flow-sensitive signals as a function of TI at four flow rates. The simulations agree reasonably well with the experimental results.

Conclusion: This work systematically examines ASL as a flow measurement technique using theoretical modeling and a hemodialyzer experiment. The results provide evidence that the velocity sensitivity from linear phase encoding in this proposed ASL technique may provide the ability to estimate the Pennes' perfusion term independent from heating and MRI thermometry.

References:
1. Dragonu I, de Oliveira PL, et al., NMR in Biomedicine, 2009; 22(8):843–851. 2. Song CW, Rhee JG, et al., J Natl Cancer Inst 1980; 64(1):119–124.
3. Wu F, et al., Ultrasound Med Biol, 2002; 28(4):535–542. 4. Gunther M, et al., MRM, 2001; 46, 974–984. 5. Jahng GH, et al., Med. Phys, 2007; 34(11):4519–4525.
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