Theoretical investigation and optimization of the labeling process in continuous artery-selective spin labeling (CASSL)

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Introduction

Continuous artery-selective spin labeling (CASSL) [1] is one technique to image perfusion territories of cerebral arteries. To date, CASSL was only applied to major brain feeding arteries, but, recently, empirical findings have demonstrated that this method is even capable of selectively label the blood of individual branching intracranial arteries [2]. However, the exact behavior of spins moving through the labeling plane multiple times in a non-selected vessel may only be determined by computer simulations. In this study we performed extensive simulations to better understand the underlying labeling process to optimize key labeling parameters and demonstrate prospects and constraints of this method to selectively label the blood of individual branching intracranial arteries in-vivo.

Material and Methods

In CASSL selective labeling is obtained by a rotating labeling plane in conjunction with a frequency modulation of the labeling RF pulses that fixes the position of the labeling plane to the selected artery. A saturation of the magnetization in non-selected vessels depends basically on the angle θ between selected vessel and rotating labeling plane, its rotation frequency f_{rot} and the distance d from the labeling focus, respectively. The locus of resonance at a distance d to the labeling spot will vary in time and is given by [1] as

$S(t) = d \cdot \tan\theta \cdot \sin(2\pi f_{rot} t + \varphi)$

with φ as an arbitrary phase. It is hypothesized that this will lead to a pseudo-random behavior when a spin is moving through the labeling plane multiple times and that this will on average cause a saturation of the magnetization in a given voxel. The ratio of the maximum velocity of the labeling plane $v_{max,\,plane}$ and the velocity of the blood flow v_{blood} determines how many times the blood will move through the labeling plane and therefore how often the blood's magnetization experience a certain change in orientation. A computer program written in Matlab (The Mathworks) was used to optimize the labeling parameters in terms of spatial selectivity and labeling efficiency. The program calculates the magnetization of flowing spins after inversion by adiabatic fast passage. Orientation and strength of the labeling gradient as well as the amplitude of the RF pulse were chosen according to a previous study with a fixed labeling plane [3]. T_1 is set to 1000ms, T_2 to 200ms to represent human blood values at 1.5T. The simulation is based on a stepwise integration of the Bloch equations for a single magnetization vector using a 4th order Runge-Kutta algorithm. The magnetization starts with a value of M=[0,0,1] and travels with a fixed velocity along a linear trajectory. Labeling efficiency was computed as

$(M_{Control}-M_{Label})/2.$

Simulations are repeated and averaged for 36 different start-phases φ of the rotating labeling plane ranging from 0° to 360° and for 40 different flow speeds ranging from 20 cm/s to 60 cm/s. To take the laminar flow profile into account the total magnetization transported by the vessel per time unit is then calculated by numerically integrating the magnetization at the end of the trajectory, $M_z(v)$, over the velocity range

$$\langle M_z \rangle = \frac{2}{v_{max}^2} \int_0^{v_{max}} v \cdot M_z(v)\,dv$$

with v_{max} as the maximum velocity of the blood [3]. Finally, the vessel size is taken into account. A verification of the simulations was performed by shifting the location of the labeling focus in the right left direction at the internal carotids of five volunteers. MR images were acquired with a standard transmit/receive head coil on a clinical 1.5 T Philips Intera MR system. For planning of the labeling position, a 2D inflow angiography was performed. Labeling parameters were as follows: Labeling duration 2.2s; post labeling delay 0.8 s. Scanning parameters were: FFE-EPI acquisition; FOV 220x176 mm; scan matrix 80x71; TR/TE, 3600/42 ms; 5 to 7 slices; thickness, 8 mm; gap, 1 mm; 40 labeled and 40 non-labeled acquisitions; scan time 5:13 min; SAR, 1.87 W/kg.

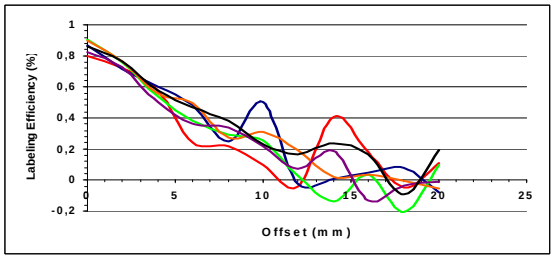

Figure 1: Simulated labeling efficiency of six selected flow speeds (20 (•), 24 (•), 30 (•), 42 (•), 52 (•) and 60 (•) cm/s) against the offset of the labeling focus (θ=10°, f_{rot}=120Hz).

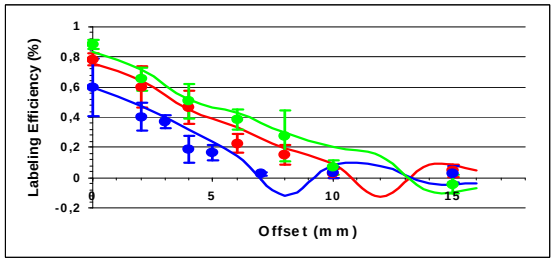

Figure 2: Simulated and measured labeling efficiency for θ =8° (•), θ =10° (•) and θ =15° (•) and a rotation frequency of 160Hz against the offset of the labeling focus.

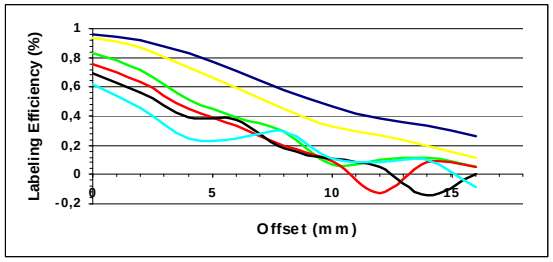

Figure 3: Simulated labeling efficiency for θ =10° and different rotation frequencies (80Hz (•), 100Hz (•), 140Hz (•), 160Hz (•), 180Hz (•), 200Hz (•) against the offset of the labeling focus.

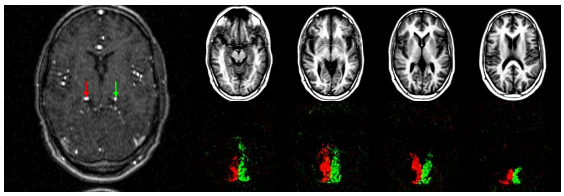

Figure 4: Color encoded perfusion territories of left and right posterior artery. Labeling parameters were θ=15° and f_{rot}=160Hz.

Results and Discussion

Figure 1 shows the simulated labeling efficiency of six selected flow speeds against the offset of the labeling focus. The labeling efficiency can vary strongly as a function of flow velocity and offset. This underlines the fact that the simulations cannot be performed for only a single average flow velocity but have to be averaged over the different flow velocities in the cardiac cycle. Figure 2 shows the labeling efficiency for three different angles θ of 8°, 10° and 15° and a rotation frequency of 160Hz against the offset of the labeling focus. Laminar flow and vessel size (Ø 4mm) is taken into account. Good agreement between simulations and measurements is provided and different saturation points can be identified. An angle θ =8° ensures a saturation of the magnetization in the non-selected artery when the offset exceeds approximately 13 mm. For θ =15° offsets greater than approximately 7 mm will guarantee no labeling in adjacent vessels. Figure 3 shows the labeling efficiency for an angle θ=10° and several rotation frequencies ranging from 80Hz to160Hz against the offset of the labeling focus. Higher rotation frequencies lead to a decreased labeling efficiency. This is due to the fast change in orientation of the labeling gradient in the direction of the flowing blood which impedes the process of adiabatic fast passage. However, it can be seen that for a frequency of 160Hz saturation occurs already at a distance of 11mm to the selection point. Lower and higher rotation frequencies shift the saturation point even further away from the labeling focus, e.g. for 200Hz the saturation point is 15mm away from the labeling focus.

Figure 4 shows the perfusion territories of both posterior arteries. Several small vessels are in distances ranging from 6.8 to 17.1mm to the selected vessels. In agreement with the theoretical findings, applying labeling parameters of θ=15° and f_{rot}=160Hz results in well delineated perfusion territories of the selected arteries.

References: [1] Werner et al, MRM 2005,53:1006-1012; [2] Helle et al, ISMRM 2008, abstract 3345; [3] Werner et al, MRM 2005,53:1096-1102