A novel method to estimate labeling efficiency for pseudo-continuous arterial spin labeling imaging

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Introduction: Labeling efficiency is one of the most important factors affecting quantification of pseudo-continuous ASL (pCASL) perfusion imaging and is conventionally assumed as constant for different subjects and obtained from simulations[1]. However, labeling efficiency of pCASL is vulnerable to factors, such as field inhomogeneity and flow velocity, which may vary across scans, subjects and arteries. Therefore, acquisition of labeling efficiency for each artery and scan might improve the accuracy of pCASL perfusion quantification. Previously, it was proposed to use phase-contrast MRI (PC-MRI) to estimate the labeling efficiency[2], but this approach showed unsatisfying repeatability[3]. In this study, we proposed a novel method to estimate the labeling efficiency by performing multi-phase pCASL imaging distal to the labeling plane.

Theory: The ASL-signal (i.e. the subtracted signal) in an arterial cross section is made up of labeled spins that have various velocity (Fig 1A). Both labeling efficiency and the transit time are velocity-dependent. Therefore, to calculate the labeling efficiency a separate T1 decay correction should be done for each velocity. The labeling efficiency in labeling plane can be denoted as: $ASL(t)=\int_0^{v_{max}} P(v)D(v)E(v,t)dv$ (1), where M_0 is the total equilibrium blood magnetization in the arterial cross section; $\alpha(v)$ and $M(v)$ are the labeling efficiency and longitudinal magnetization of spins with velocity v. The ASL-signal in the imaging plane can be described as: $ASL(t)=\int_0^{v_{max}} P(v)D(v)E(v,t)dv$ (2), where $D(v)=e^{-TT(v)/T_{1b}}$ is T1 decay experienced by the spin; $E(v,t)=rect(t-TT(v))$ denotes the presence of spin in the imaging plane, which is assumed to be a rectangle window pulse with a width equal to the labeling duration and a time shift equal to the spin's transit time $TT(v)$. The above problem of obtaining $\bar{\alpha}$ can be solved numerically, whereas the transit time of the fastest spin is detected as the latest time-point with maximum ASL intensity (e.g. indicated by the arrows in Fig 1B). This prior information is utilized to improve the numerical solving algorithm.

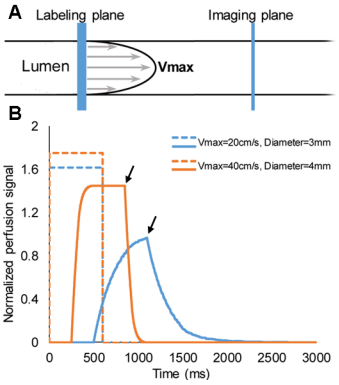

Fig 1. A) Schematic plot of the simulations. B) Simulated ASL-signal curves of one pixel in labeling (dotted line) and imaging plane (solid line) for two arteries with different diameter and blood velocity.

Methods: The ASL-signal as detected in the arterial cross section is simulated using a laminar flow model[4] for the velocity distribution (Fig 1A). Subsequently, labeling efficiencies are derived using equation (1) and (2). In vivo multi-phase single slice pCASL imaging was performed in two healthy subjects on a 3T MR scanner (Philips healthcare) equipped with a 32-channel head coil. The imaging parameters were as follows: post labeling delay 5ms, 2D EPI readout with 90° excitation RF pulse, half Fourier factor 0.71, number of phases 42, phase interval 37.5ms, number of averages 15. For the first subject, the labeling plane was placed just below the carotid bifurcation and the imaging plane about 25mm below skull base. Slice thickness of the imaging plane was 5mm. The imaging was performed for labeling durations of 400, 600, 800 and 1000ms. Then two additional scans were performed with labeling duration 400ms, but with the imaging plane either moved to the skull base or to the level of the basilar artery. These 3 imaging locations were denoted as low, middle and high, respectively. For the second subject, slice thickness was reduced to 3mm with a labeling duration of 600ms. These scans were performed with 6 different flip angles of the pCASL RF labeling pulse: 5°, 7.5°, 10°, 15°, 25° and 30°. Furthermore, quantitative flow imaging was conducted in the labeling plane to obtain the mean blood velocity of each artery, which was used for simulation of the theoretical relationship between pCASL labeling efficiency and pCASL RF flip angle. Cardiac triggering with pulse oximeter was used for all above labeling efficiency imaging. An additional scan without cardiac triggering was also performed in the second subject. The M_0 was obtained from the mean of the last ten control phases and was used to normalize the ASL-signal time curve before averaging over the ROI containing the artery. The average normalized ASL-signal curve was them smoothed and interpolated, followed by manual detection of the transit time of the fastest spin and calculation of labeling efficiency.

Results: Fig 1B showed simulated ASL-signal curves for two arteries with different diameter and blood velocity. The labeling efficiency can be perfectly restored from the simulated ASL-signals (data not shown). ASL-signal curves with different labeling durations of the first subject's left internal carotid artery (LICA) are shown in Fig 2. The calculated labeling efficiencies of the same artery at three different imaging locations are shown in Table 1. The ASL control signals with and without cardiac triggering were shown in Fig 3. Measured in vivo labeling efficiencies for different pCASL flip angles are shown in Fig 4A. The Fig 4B showed simulated curves based on quantitative flow measurements by PC-MRI.

Table 1 Estimated labeling efficiencies for different imaging locations

	RVA	LVA	RICA	LICA
Low	0.531	0.813	0.686	0.684
Middle	0.370	0.760	0.568	0.657
High	0.518		0.508	0.635

* RVA=right vertebral artery, BA=basilar artery, LVA=left vertebral artery, RICA=right internal carotid artery, LICA=left internal carotid artery. "Low" means about 25mm below skull base, while "Middle" means skull base, "High" means the level of basilar artery.

Discussion and Conclusions: The proposed labeling efficiency estimation method works well on simulated data. The shape and amplitude of the measured ASL-signal as depicted in e.g. Fig 2 are largely consistent with the expected curves from simulations. Furthermore, the dependency of the in vivo measured labeling efficiencies for different pCASL flip angles are in line with theory as well. However, the measured labeling efficiency of the same artery measured at different imaging locations varied too much and in some measurements (see Fig 4A) unrealistic high labeling efficiencies were obtained. It is postulated that these measurement errors arise due to imperfect estimation of M_0, which may be caused by cardiac pulsation (see Fig 3) affecting spin velocity and transit times or partial volume effects that result from the low resolution or non-perpendicular sections through some arteries. In summary, a new, fast method for estimation of the labeling efficiency is proposed that is promising, although further improvements in robustness and quantification are needed.

References: 1. Alsop, MRM epub, 2014; 2. Aslan *et al*, MRM 63:765-771, 2010; 3. Dolui *et al*., Proc. Intl. Soc. Mag. Reson.Med. 22(2014). 0212; 4. Gallichan *et al*, MRM 60:53-63, 2008.

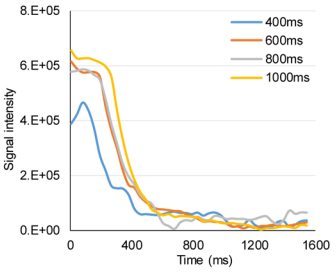

Fig 2. ASL-signal curves of one pixel within LICA acquired with different labeling durations.

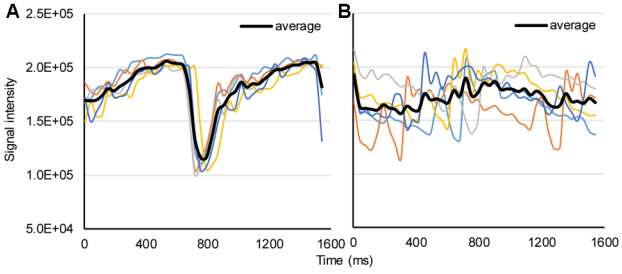

Fig 3. ASL control signals acquired with A) and without B) cardiac triggering. Signals of 5 individual dynamic scans were shown with the color curves, and their average was shown with black bold curve. Distinct effect of cardiac pulsation on the ASL signals can be seen.

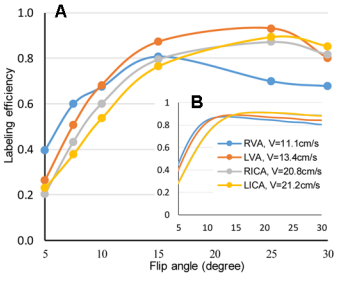

Fig 4. Labeling efficiency for different pCASL RF flip angles. A) In vivo labeling efficiency; B) Simulated labeling efficiency with blood velocity obtained from phase contrast MRI as input.