Inversion-Prepared Pulsed ASL with Single-Shot FSE Readout for the *In Vivo* Measurement of the T_1 of Arterial Blood

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Introduction

To fully quantify cerebral blood flow (CBF) with arterial spin labeling (ASL), it is necessary to have an accurate estimate of the longitudinal relaxation time of arterial blood (T_{1a}) (1). However, it is very challenging to measure T_{1a} in vivo because of the high flow velocity in arteries and the relatively small artery diameters, particularly in small animals. Even with adequate physiological gating, these measurements are subject to artifacts from cardiac pulsatility and respiration, and this leads to high inter- and intra-subject variability in T_1 estimates (2). Although *in vitro* methods avoid these problems, they do not replicate the exact condition *in vivo*. To avoid the problems of isolating signal from fast flowing intravascular blood, a pulse arterial spin labeling approach for measuring T_{1a} has been suggested (3). In this approach, a global saturation pulse precedes a flow-alternating inversion recovery ASL technique with a single-shot echo-planar imaging readout. By keeping all parameters constant and using different saturation pulse delays, a measurement of T_{1a} can be made. In this study, we have suggested a novel approach to improve this method, which increases the dynamic range, avoids possible venous signal contamination, and maximizes the signal-to-noise ratio (SNR) in the presence of static field inhomogeneities. We validated this technique by imaging rat brain on a clinical scanner.

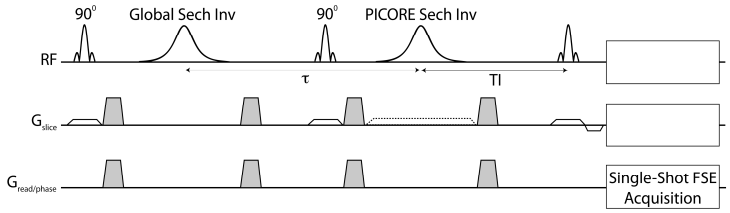

FIG.1. Pulse sequence diagram of the PICORE sequence with a global inversion preparation.

Materials and Methods

All images were collected on a whole-body clinical 3T MRI scanner (Siemens Trio; Siemens Healthcare, Erlangen, Germany) with a body coil transmitter and small (35 mm ID) receive-only surface coil passively decoupled during transmit. All data were obtained on adult male Sprague-Dawley rats (n=5; 430-470 g; Charles River, MA, USA). Anesthesia was induced with 3% isoflurane in oxygen for approximately five minutes, and the animals were then maintained on spontaneous inhalation anesthesia consisting of 1.5% isoflurane in 30% oxygen delivered at 1.5 L/min through a tight-fitting nose cone. Head movement was restricted during imaging using a custom built MR-compatible animal cradle with attached earbars.

A schematic of the pulse sequence is shown in Fig. 1. A proximal inversion with control for off-resonance effects (PICORE) inversion pulse was used to label arterial blood (4). This pulse has the advantage over FAIR in that it avoids possible contamination of the venous blood returning to the heart from distal locations. We used a global inversion pulse instead a saturation pulse to prepare the magnetization, which has the advantage of doubling the dynamic range. It is important to note that because the transmitter coil is much larger than the rat, there is no potential for error in the measurement due to the inflow of uninverted blood into the imaging region. Image acquisition was performed with a single-slice, single-shot fast spin echo (FSE) sequence. The use of FSE allows for a very thick slice to be used without signal losses due to background field inhomogeneities, maximizing SNR. The imaging parameters were: TE/TR: 25/8000 ms, slice thickness = 8 mm, FOV = 64 × 64 mm, and matrix size = 64 × 64.

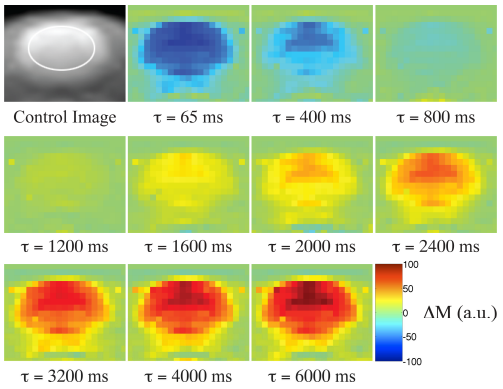

FIG. 2. A representative set of PICORE signal difference ΔM (control minus tag) images in arbitrary units (a.u.) from the pulse sequence shown in Fig. 1.

If TI is kept at a constant value while the inversion preparation time τ is varied, the PICORE signal difference can be expressed as: $\Delta M(\tau)=A(1-2\,\alpha\,\exp(-\tau/T_{1a}))$, where A is a voxel-specific constant independent of τ and α is the inversion efficiency. To allow for a robust, overdetermined fit in each animal, ten values of τ (65, 400, 800, 1200, 1600, 2000, 2400, 3200, 4000, and 6000 ms) were used with a constant TI of 1500 ms. A randomized acquisition order was used, and tag and control images were interleaved for each value of τ. Ten repetitions of each value of τ were obtained with a total imaging time of twenty-seven minutes. The data were fit for α, A, and T_{1a} using a non-linear least squares fit in MATLAB.

Results

A representative set of ΔM images during normoxia with ten values of τ is shown in Fig. 2. As predicted from the model, the signal increased exponentially from an initial negative value of ΔM and changed globally to the final expected positive value of ΔM as τ increased. The experimental data from the whole-brain ROIs from each animal, along with the fits to the model are shown in Fig. 3. From visual inspection, it is clear that the recovery of the magnetization is well described by a monoexponential recovery. The average value of α was found to be 0.94. The average value of T_{1a} in all five rats was found to be 1.62 ± 0.07 sec.

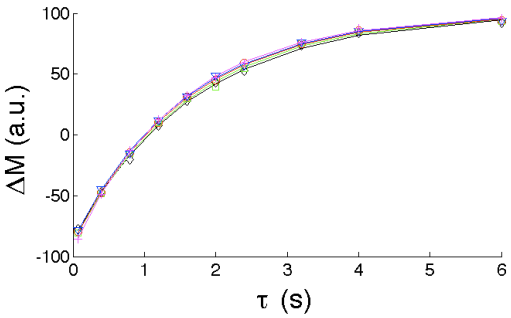

FIG. 3. PICORE signal difference (ΔM) in arbitrary units (a.u.) versus inversion preparation time τ in a whole brain ROI (see Fig. 2) in five rats.

Discussion

In this study, a novel approach to the *in vivo* measurement of T_{1a} was implemented which avoids a number of artifacts and the requirement for physiological gating, and when compared to a similar approach in the literature, approximately doubled the dynamic range with an inversion preparation, avoided possible venous signal contamination with a PICORE inversion, and maximized the signal-to-noise ratio (SNR) in the presence of static field inhomogeneities with a FSE readout. The values measured for T_{1a} are in close agreement to prior *in vivo* (1623 ms; (5)) and *in vitro* (1664 ms; (6)) literature values.

References

(1) Buxton, et al. *MRM* (1998); (2) Guo, et al. *Med. Phys.* (2005); (3) Thomas, et al. *MRM* (2006); (4) Wong, et al. *NMR Biomed.* (1997); (5) Nichols, et al. *JMRI* (2008); (6) Lu, et al. *MRM* (2004).

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