Measuring Blood T1 in the Jugular Vein: Juggling Size, Speed and Precision

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INTRODUCTION: Knowledge of the absolute value of the blood T_1 is important for a number of quantitative MRI applications, including determination of cerebral blood flow (CBF) using arterial spin labeling (ASL) (1,2), calculation of the inversion time (TI) in vascular space occupancy (VASO)-dependent fMRI (3), and estimation of kinetic parameters in dynamic contrast-enhanced MRI (4). Current T_1 information for blood at 3T has come from physiological phantom studies on animal blood (5). Measurement of blood T_1 *in vivo* could provide more specific information, e.g. for individuals with abnormal blood composition. T_1 of *in vivo* blood could be measured using a fast inversion recovery technique in which multiple inversion time points can be acquired rapidly due to constant refreshing of blood. This idea has previously been demonstrated in renal arteries and veins for evaluating kidney extraction fractions (6,7). Recently, it was applied to measure T_1 in the sagittal sinus (8). Here we applied this approach to the internal jugular vein (IJV), allowing determination of *in vivo* blood T_1 values in a time of only 1 min.

METHODS: Experiments were performed on a 3T Philips Intera scanner using the body coil for transmission and an 8-channel head coil for reception. A total of 16 healthy volunteers (age:24~49yrs) were enrolled with informed consent. Scan planning: phase contrast MRA (PCA) images were acquired in both the coronal and sagittal planes to visualize the location of major neck vessels.

In Fig. 1, the pulse sequence diagram for measuring T_1 in IJV (6-8) is described. It is composed of a non-selective (NS) adiabatic inversion pulse (10ms, BW=1250Hz) followed by an initial waiting time (TI(1)=50ms) and series of (N_{TI}=50) slice-selective (SS) 90º excitation pulses (THK=5mm) separated by a short time interval (ΔTI=200ms). The average blood velocity measured from all our healthy subjects was about v=16cm/s in IJV. It is assumed that all spins entering the imaging slice have experienced the inversion pulse, but none of the previous SS excitations (ΔTI>THK/v). Flow compensation (FC) gradients are used to reduce the phase loss caused by through-plane flow (9). For data acquisition (6-segment gradient echo EPI with SENSE factor=2, TE=15ms), the same fractions of the *k*-space are sampled with different TIs after each inversion. Using a rapid series of excitation pulses, static spins will be saturated and show little background signal. Other parameter: FOV=200x150mm^2, acquisition matrix=192x132, in-plane resolution after reconstruction was 0.78x0.78mm^2. Total measurement time≈10s(TR) x 6(N_{shots})=60s.

RESULTS and DISCUSSION: To allow proper T_1 relaxation, it is essential to assure whole-brain efficiency of the NS inversion. The effect of shimming on this was checked for 4 subjects. Different areas of the brain (top, middle, and bottom, all above the IJVs, Fig. 2a, dashed orange lines) were imaged using the same NS 180º pulse and either adjusting both 1st/2nd order shims locally (green box, condition 1) or only 1st order shims locally (condition 2). Blood T_1 measurement sequence was also performed under the same two shimming conditions (solid orange line). ROI locations for T_1 measurements from both IJVs were shown in Fig. 2b (TI≈10s, L: blue, R: green). IR images from the middle of the brain were acquired at TI=600ms with condition 1 (Fig. 2c) and with condition 2 (Fig. 2d). IR curves from the two ROIs with fitted T_1s were shown using condition 1 (Fig. 2e) and using condition 2 (Fig. 2f). There is a clear left/right difference in Fig. 2e and both left and right T_1 values are much lower than in Fig. 2f. The reason is that when shimming locally with higher orders, the field adjustment may be totally off for other parts of the brain, moving the resonance frequency in such regions outside of the bandwidth of the adiabatic 180º pulse.

With the proper shimming condition (only 1st order shims locally following global shim), a total of 12 healthy subjects (6 females, 24~44yrs; 6 males, 24~33yrs) were studied. To check for reproducibility, scans were repeated every 15 minutes for three times. The average of the three reproduced $T_{1,IJV}$ of each subject are shown in the table. The averaged blood T_1s of female subjects were about 100~200ms higher than those of male subjects, which we tentatively attribute to a hematocrit (Hct) difference between females and males (female: 0.36 ~ 0.44; male: 0.41 ~ 0.50).

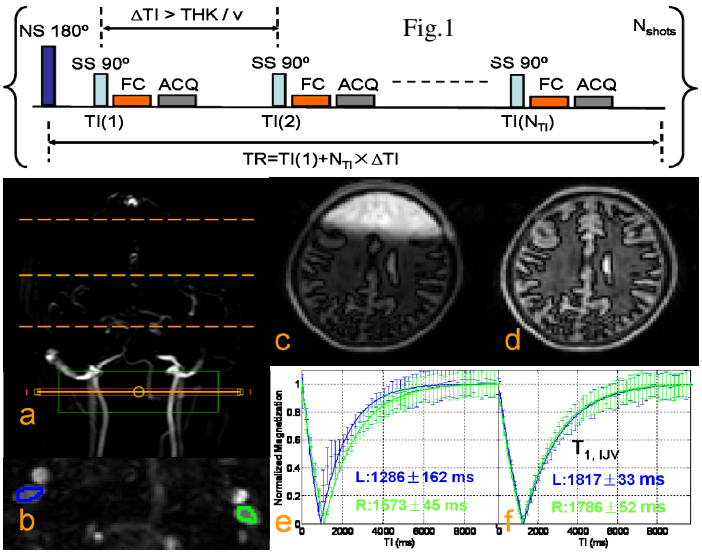

Fig.1

Fig. 2

Female	$T_{1,IJV}$ (ms)	Male	$T_{1,IJV}$ (ms)
1	1917±38 (18 voxels)	1	1707±5 (21 voxels)
2	1942±66 (37 voxels)	2	1702±15 (20 voxels)
3	1814±42 (48 voxels)	3	1759±51 (31 voxels)
4	1901±40 (12 voxels)	4	1704±36 (32 voxels)
5	1912±29 (34 voxels)	5	1625±37 (24 voxels)
6	1980±19 (13 voxels)	6	1809±41 (35 voxels)
Average	**1911±55**	**Average**	**1718±62**

CONCLUSION: We have shown that blood T_1 values can be measured efficiently *in vivo* (within 1 min) exploiting the inflow of fresh blood into the jugular veins. The proposed fast method can be performed in combination with ASL, VASO, or dynamic Gd experiments for a subject-based blood T_1 determination, which is important based on the strong dependence of T_1 on Hct and the possibility of different T_1s in patients with blood abnormalities

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