Introduction

Blood $T_1$ is a critical parameter for black blood imaging (1) and perfusion quantification with arterial spin labeling imaging (2). Conventional $T_1$ measurement is time-consuming as the longitudinal relaxation curve is sampled with varied inversion times (TI) in separate scans. Characterized by high imaging speed while preserving a high signal-to-noise ratio, TrueFISP (3) has been proposed for in-vivo $T_1$ mapping of static tissue with inversion recovery (IR) preparation (4). The apparent $T_1$ of static tissue estimated with IR-TrueFISP, however, exhibits a complicated dependence on the flip angle and the relaxation time constants ($T_1$/$T_2$) of the tissue (5). An interesting observation from our experiment is that IR-TrueFISP may provide an efficient yet accurate approach for in-vivo blood $T_1$ mapping, due to the replenishment of blood spins with longitudinal magnetization unperturbed by the TrueFISP pulse train. Therefore, blood pool signals in IR-TrueFISP acquisitions generally follow the conventional $T_1$ recovery model. In this study, we conduct experiments and computer simulations to investigate the feasibility and reliability of using IR-TrueFISP for blood $T_1$ measurement.

Materials and Methods

All MR imaging was in accordance with the Institutional Review Board guidelines, and performed on a 3.0 T whole body scanner (Siemens Trio, Erlangen, Germany) with a standard setup of body coil transmission and phased-array head coil reception. Six healthy volunteers (age = 16-25 years, F/M = 2/4) were imaged after written informed consent was obtained from each of them. For IR-TrueFISP scans, the $\alpha/2$-($\alpha/2$) scheme was adopted for efficient signal stabilization, and phase encoding advanced in a centric order (TR = 5 ms, TE = TR/2, $\alpha = \{10^\circ, 30^\circ, 50^\circ\}$, in-plane matrix = 128x128, FOV = 220 mm). Following a spatially nonselective hyperbolic-secant inversion pulse and 20 dummy scans, 50 phases of TrueFISP readout were carried out with 19 lines of k-space data obtained during each phase. The TI values corresponding to the 50 phases thus ranged from 100 ms to 4850 ms. At the end, the magnetization was restored to the $+z$ axis using a $-\alpha/2$ pulse. The procedure was then repeated for the next 19 k-space lines and so on with a total scan time of 48 sec. Images were obtained from a 5-mm axial slice where the sagittal sinus was perpendicular to the imaging slice to avoid signal from the skull base. Signals were slightly saturated at the slice. Signals were imaged after written informed consent was obtained from each of them. For IR-TrueFISP scans, the $\alpha/2$-($\alpha/2$) scheme was adopted for efficient signal stabilization, and phase encoding advanced in a centric order (TR = 5 ms, TE = TR/2, $\alpha = \{10^\circ, 30^\circ, 50^\circ\}$, in-plane matrix = 128x128, FOV = 220 mm). Following a spatially nonselective hyperbolic-secant inversion pulse and 20 dummy scans, 50 phases of TrueFISP readout were carried out with 19 lines of k-space data obtained during each phase. The TI values corresponding to the 50 phases thus ranged from 100 ms to 4850 ms. At the end, the magnetization was restored to the $+z$ axis using a $-\alpha/2$ pulse. The procedure was then repeated for the next 19 k-space lines and so on with a total scan time of 48 sec. Images were obtained from a 5-mm axial slice where the sagittal sinus was perpendicular to the slice. Signals were extracted from regions-of-interest at gray matter and mid-sagittal sinus, and fitted to a three-parameter model: $k_0(1-k_0*\exp(-t/k_0))$. The effects of $\alpha$ and flow velocity ($v$) were estimated with numerical simulations of Bloch equations. Phase-contrast (PC) MRI was acquired at the same imaging slice to estimate the mean venous blood flow velocity in sagittal sinus (FOV = 220 mm, matrix = 128x128, flip angle = 15°, TR = 25 ms, TE = 3 ms, VENC = 60 cm/s along z axis, scan time = 3 s). For comparison, single-phase IR-TrueFISP acquisitions were performed at 16 different TI’s (100-5000 ms).

Results and Conclusion

Figs. 1 and 2 show the simulated signal evolution of blood signal with respect to $v$ and $\alpha$, in which multi-phase IR-TruFISP signals deviate from the theoretic value when the tissue is static and $\alpha$ increases. Listed in Table 1 are the predicted accuracy and variation of measurement based on 50 sets of computer-generated IR-TrueFISP data (SNR = 10 and Gaussian noise). Experimental data show good compatibility with the numerical results in that when $\alpha$ increases from 10° to 30°, and 50°, the $T_1$ of venous blood ($v ~ 18 \pm 41$ ms, 655±41 ms, and 1652±103 ms, respectively), whereas the measurement in gray matter (static tissue) remarkably varies (1112±42 ms, 655±41 ms, and 428±35 ms, respectively). Repeatability was tested on 4 subjects and coefficient of variation was ~2%. The measured blood $T_1$ shows a linear relationship with age in the studied span of age (Fig. 3). As shown in Fig. 4, $T_1$ seems to increases with $v$ when males and females are considered separately. Females tend to have slower venous flow although no $T_1$ difference has been observed between genders. It is noted that single-phase scheme consistently measures $T_1$ longer than multi-phase scheme (Fig. 5). The discrepancy could be because the interval between single-phase scans was not long enough and thus longitudinal signals were slightly saturated after the scan of the first TI. In conclusion, we have demonstrated the feasibility of using multi-phase IR-TrueFISP for fast (< 1 min) and reliable (2%) repeatability in-vivo blood $T_1$ mapping.

References