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In vivo Venous Blood T_1 Measurement using Inversion Recovery TrueFISP in Children and Adults

Wen-Chau Wu^{1,2}, Varsha Jain³, Cheng Li³, Mariel Giannetta^{3,4}, Hallam Hurt⁴, Felix W. Wehrli³, and Danny J.J. Wang^{3,5,*}

¹ Graduate Institute of Oncology, National Taiwan University, Taipei 100, Taiwan, Republic of China

² Graduate Institute of Clinical Medicine, National Taiwan University, Taipei 100, Taiwan, Republic of China

³ Department of Radiology, University of Pennsylvania, Philadelphia, PA 19104, USA

⁴ Department of Pediatrics, the Children's Hospital of Philadelphia, Philadelphia, PA 19104, USA

⁵ Department of Neurology, University of California at Los Angeles, Los Angeles, CA, 90025, USA

Abstract

A time-efficient method is described for in-vivo venous blood T_1 measurement using multi-phase inversion-recovery prepared balanced steady-state free precession imaging (IR-TrueFISP). Computer simulations and validation experiments using a flow phantom were carried out to demonstrate the accuracy of the proposed method for measuring blood T_1 , by taking advantage of the continuous inflow of fresh blood with longitudinal magnetization undisturbed by previous RF pulses. In vivo measurement of venous blood T_1 in the sagittal sinus was carried out in 26 healthy children and adults aged 7 to 39 years. The measured venous blood T_1 values decreased with age as a whole ($p=0.006$) and were higher in females than males ($p=0.013$), matching the expected developmental changes and gender differences in human hematocrit level. The estimated mean blood T_1 values were highly correlated with normal hematocrit levels across age and gender groups (Spearman $r=0.93$, $p=0.008$). The longitudinal repeatability of this technique was 4.0% as measured by the within-subject coefficient of variation. The proposed multi-phase IR-TrueFISP method is a feasible technique for fast (< 1 min) and reliable in-vivo venous blood T_1 measurement, and may serve as an index of hematocrit level in individual subjects.

Keywords

Blood T_1 ; longitudinal relaxation; inversion recovery; TrueFISP

Introduction

The longitudinal relaxation time T_1 (or rate R_1) of blood is a critical physiological parameter. In vitro studies have demonstrated the linear dependence of both arterial and venous blood R_1 on hematocrit (Hct) level, in addition to its dependence on temperature and oxygenation level (1,2). Theoretically, in vivo measurement of blood T_1 may be used as a surrogate index of Hct, if both adequate accuracy and precision can be achieved. Human Hct

Corresponding Author: Danny J.J. Wang, Ph.D., UCLA Department of Neurology, Ahmanson-Lovelace Brain Mapping Center, 660 Charles E. Young Drive South, Los Angeles, CA 90025-7085, 310-983-3667 (voice) 310-794-7406 (fax), jwang71@gmail.com.

level varies considerably with age and gender across people's life span. It varies widely within the first month after birth (31–67%), gradually converges to 33–40% around the age of two, followed by a steady increase with age through childhood and adolescence. Gender difference in Hct emerges around the age of 12, and by the time of adulthood, the normal Hct range is 41–53% for males and 36–46% for females (3). A rapid and reliable in vivo measurement of blood T_1 therefore is desirable to account for individual variations of Hct in a number of MRI applications. For instance, hematocrit is a scaling factor for the widely adopted blood-oxygen-level-dependent (BOLD) signal (4). Hematocrit, through its effects on arterial blood T_1 , affects perfusion quantification using arterial spin labeling (ASL) (5) as well as the appropriate null point for black blood imaging (6) and vascular space occupancy (VASO) imaging (7).

To date, however, the endeavor for in-vivo blood T_1 mapping in individual subjects has been dampened mainly by prohibitively long scan times. For instance, conventional T_1 measurement is typically achieved by sampling the longitudinal relaxation curve with varied inversion times (TI) following a global inversion pulse, resulting in a scan time that spans at least several minutes. Other challenges for in-vivo measurement of blood T_1 include partial volume effects and flow-related artifacts. Fast imaging sequences such as echo-planar imaging (EPI) in conjunction with the Look-Locker method (8–11) have been proposed for in-vivo blood T_1 mapping, by exploiting continuous refreshing of blood between excitations. However, the relatively long EPI readout may result in spatial shift, distortion and dropout of blood pool signal. An alternative approach has been proposed recently that utilizes ASL to isolate the blood signal while the inflowing blood magnetization is prepared with varied degrees of saturation recovery (12). However, this technique suffers from prolonged scan time and may be confounded by inflow effects in human studies.

TrueFISP is a balanced steady-state free precession technique (13) that offers high signal-to-noise ratio (SNR) and imaging efficiency. TrueFISP is ideal for blood vessel imaging as it highlights the blood signal because of its intrinsically high T_2/T_1 ratio (14,15). The technique is also inherently flow-compensated in the directions of slice selection and readout, and therefore exhibits minimal sensitivity to flow artifacts (16). It has been shown that an $\alpha/2$ RF pulse followed by a train of polarity-alternating α pulses provides a smooth signal evolution toward steady state and therefore is suitable for readout after magnetization preparation (17). Inversion recovery (IR) prepared TrueFISP has been applied to in-vivo T_1 mapping of static tissue (18). However, the estimated apparent T_1 is a complicated function of the flip angle α , and the relaxation time constants of the tissue, although correction schemes have been proposed for concurrent fitting of T_1 , T_2 and proton density images (19).

An interesting empirical observation based on our experience is that IR-TrueFISP signal of a volume element filled with blood exhibits little sensitivity to imaging parameters other than T_1 probably due to the continuous inflow of fresh blood with longitudinal magnetization undisturbed by previous RF pulses. In particular, the measured IR-TrueFISP curves from the mid-sagittal sinus exhibit excellent fit using the standard IR model, with estimated blood T_1 values matching those reported in the literature (20). In the present study, we conducted computer simulations and a phantom validation study to demonstrate the accuracy of the proposed IR-TrueFISP method. In vivo venous blood T_1 measurements were carried out in healthy children and adults aged 7 to 39 to investigate the reliability of the technique and the feasibility for detecting expected Hct change with age and gender.

Materials and Methods

Computer Simulations

The temporal evolution of the magnetization is calculated following the framework described by Hargreaves et al (21). For blood that flows through a slice of thickness d at a nominal velocity of v , the number of α pulses experienced by a flowing proton is $d/v/TR$. Once a proton flows out of the slice, another fresh proton enters and the replacement takes place in a first-in-first-out order. The slice can thus be divided into $d/v/TR$ segments to contain protons that had experienced monotonically increased number of RF excitations toward the downstream end. Finally, the overall magnetization in the slice is the vector sum over these segments.

We computed the magnetization evolution for three values of v (1, 10, and 20 cm/s) and three α 's (10° , 30° , and 50°), and compared the results to those with blood treated as static tissue (i.e., $v = 0$). The parameters used in the calculation were as follows: $d = 5$ mm, $TR = 5$ ms, $T_{1,\text{blood}} = 1600$ ms, $T_{2,\text{blood}} = 60$ ms. The values of α , d , and TR were chosen to match those used in actual scans. All simulations were implemented with MATLAB (Mathworks, Natick, MA, USA) using in-house programs. Given that the susceptibility difference between oxygenated and deoxygenated blood is 0.18 ppm (22), the off-resonance frequency is ~ 23 Hz at 3.0 T, which only generates an SSFP phase of 0.11° when $TR = 5$ ms. For simplicity and yet without loss of generality, we considered here the on-resonance condition ($\theta = 0^\circ$) unless otherwise noted.

MRI Experiments

We performed 3 MRI experiments: 1) phantom validation study; 2) initial feasibility study in adult subjects; and 3) validation study in children to assess the reliability and to detect expected developmental changes in blood T_1 . Detailed parameters for each experiment are listed below. Experiment 1&2 were performed on a 3.0 T Siemens Trio whole body scanner (Erlangen, Germany), and experiment 3 was performed on a 3.0T Siemens Verio whole body scanner, both using the body coil for transmission and phased-array head coil for reception.

The IR-TrueFISP scan adopted a segmented multi-phase acquisition scheme (see Fig. 1). Fifty phases of TrueFISP readout were carried out following a spatially nonselective hyperbolic-secant inversion pulse during each segment. Twenty dummy scans (flip angles = $\alpha/21$, $-2\alpha/21$, $3\alpha/21$... $-20\alpha/21$) were applied in place of a single $\alpha/2$ RF pulse to minimize transient oscillation of static tissue signal (23). The MR signals were then continuously acquired using the ($\pm\alpha$) scheme, and phase encoding advanced in a centric order. Imaging parameters included: $TR = 5$ ms, $TE = TR/2$, $\alpha = 30^\circ$, matrix size = 128×128 , field-of-view (FOV) = 220 mm, 6/8 partial Fourier, 19 lines of k-space data per segment and bandwidth=977Hz/pixel. The TI values corresponding to the 50 phases thus ranged from 100 ms to 4850 ms in steps of 95 ms. Residual transverse magnetization after the last phase was restored to the +z axis using a $-\alpha/2$ pulse followed by a delay of 3 s for full recovery of magnetization. The procedure was then repeated for the next segment. The total scan time was only 48 sec. A phase-contrast (PC) sequence was used to estimate the mean flow velocity (FOV = 220 mm, matrix size = 128×128 , flip angle = 15° , $TR = 25$ ms, $TE = 3$ ms, VENC = 60 cm/s along z axis, scan time = 3 s). For comparison, a single-phase IR-TrueFISP measurement was performed at 16 TI's (100–5000 ms) to estimate blood T_1 .

1. Phantom Validation Experiment—A flow phantom was constructed using a container (~ 1500 ml) with feeding and draining silicon tubes (4 mm inner diameter) driven by a variable-flow peristaltic pump (Thermo Fisher Scientific Inc., Pittsburgh, PA). The

phantom was filled with distilled water doped with 0.05 mM/L MnCl_2 and 0.2 mM/L NaCl . IR-TrueFISP scanning was performed with 3 flip angles (10, 30, 50°) at 4 flow velocities ($v=6, 10, 13, 17\text{cm/s}$ estimated by PC-MRI) and without flow ($v=0$), respectively. A 5-mm axial slice perpendicular to the feeding and draining tubes was imaged. The region-of-interest (ROI) was drawn within the draining tube (or the “vein”). The T_1 of doped water was estimated using the single-phase IR-TrueFISP sequence at $v=0$, and T_2 was estimated using a segmented spin-echo EPI sequence with $\text{TR}=1\text{s}$ and 9TEs from 10 to 200ms.

2. Feasibility Study in Adult Subjects—Eight healthy volunteers (mean age=24.9yr, range=19–39yrs, 4 males) were recruited in this study and all gave informed written consent before participation, according to the protocol approved by the Institutional Review Board. Images were obtained from a 5-mm axial slice where the sagittal sinus was perpendicular to the slice. In 5 subjects, the IR-TrueFISP scans were performed with 3 flip angles (10, 30, 50°). PC MR images were acquired in seven subjects at the same slice to estimate the mean venous blood flow velocity. In addition, The single-phase IR-TrueFISP measurement was performed on 5 of the 8 subjects and at 16 T_1 's (100–5000 ms).

3. Validation Study in Children—Eighteen normal pediatric subjects (mean age=11.5yr, range=7–17 yrs, 10 males) were recruited after informed consent was obtained from a parent or legal guardian, and assent was obtained from the pediatric subject. Fourteen of the 18 subjects were scanned twice within 2–4 weeks. A 5-mm axial slice perpendicular to the sagittal sinus was imaged using the same multi-phase IR-TrueFISP sequence. Due to the use of a slightly lower bandwidth (867Hz/pixel), the duration of each phase was 104ms.

3. Data processing and analysis—Complex data of TrueFISP scans were reconstructed to magnitude images and PC scans to phase images. The images were then analyzed using custom software written in IDL (RSI, Boulder, CO, USA). Regions of interest of mid-sagittal sinus (mean $\pm$ SD ROI volume=66 $\pm$ 21mm³) and gray matter were manually generated. Time courses of MR signals were extracted from the ROI's of blood pool and gray matter respectively, and then fitted to a three-parameter model:

$$S(t)=k_1 \cdot \left(1 - k_2 \cdot \exp\left(-\frac{t}{k_3}\right)\right) \quad [1]$$

where k_3 represents blood T_1 . The model fitting was also applied to computer-generated multi-phase IR-TrueFISP data (superimposed with Gaussian noise and $\text{SNR} = 10$) as well as the flow phantom data.

Statistical analysis was performed using the Stata 10.0 software (College Station, TX, USA). The Friedman's two-way analysis of variance was used to compare the T_1 values measured with three different flip angles. Spearman's rank correlation was calculated to examine the relationship between the T_1 values measured with multi-phase and single-phase IR-TrueFISP, as well as the relationship between blood T_1/R_1 , flow velocity and Hct level. Test-retest repeatability of the measurements was estimated using within-subject coefficient of variation (wsCV) and intraclass correlation coefficient (ICC). Lastly, multivariable linear regression was carried out to derive the model of blood T_1 change with age and gender.

Results and Discussion

Simulation Results

Figure 2 shows the simulated temporal evolution of blood signal with respect to flow velocity (v) and flip angle (α). The signal of multi-phase IR-TrueFISP deviates from the

standard IR curves for static tissue or very slow flow ($v \leq 1$ cm/s; Figure 2a), or when α is increased (Figure 2b). Listed in Table 1a are the predicted accuracy and variation of the model fitting based on 50 sets of computer-generated multi-phase IR-TrueFISP data at an SNR of 10 (Gaussian noise). The fitted T_1 's are highly consistent with the assumed T_1 value when v is larger than 10 cm/s, especially in conjunction with a small α ($\leq 30^\circ$). The results provide confidence in the validity of the subsequent in-vivo blood T_1 measurements based on two rationales. First, the flow velocity measured by PC MRI was 17 ± 4 cm/s ($n = 7$) at the sagittal sinus from which blood T_1 was estimated in this study. At this level of flow velocity, the temporal evolution of multi-phase IR-TrueFISP signals closely approximate the standard IR model, even with an α as large as 50° . Second, we observed that with our experimental setting the SNR of multi-phase IR-TrueFISP was always above the simulated level when $\alpha = 30^\circ$. We thus chose to use 30° pulses in following studies to examine the dependence of blood T_1 on gender, age and flow velocity.

Exp 1: Phantom Validation

Table 1b shows the flow phantom results. The estimated mean T_1 of doped water was 1493 ± 16 ms within the velocity range of 6–17 cm/s and across 3 flip angles. The mean error was only 1.4% (maximum 2.5%) compared to the T_1 value (1473 ms) estimated by the single-phase IR-TrueFISP scan. Consistent with simulation results, the apparent T_1 was underestimated for stationary fluid ($v=0$) especially with large α . The estimated T_1 increased slightly with larger flow velocities (0.2% per cm/s increase in v). However, such effect was not statistically significant ($p=0.39$) using repeated scans at different flow velocities (data not shown).

Exp 2: Feasibility Study in Adults

Figure 3 shows the IR-TrueFISP images as a function of TI as well as the extracted time courses in the ROI's of sagittal sinus and gray matter from a representative subject. The blood pool signal in the sagittal sinus is clearly visible in later phase images due to the saturation of tissue signals. This allows easy identification of the ROI's to avoid partial volume effect of surrounding tissue. The extracted time courses show that the null point is virtually identical for blood signal across three α 's (i.e., similar blood T_1 values). In contrast, the null point of gray matter signals is shifted toward shorter TI values as α is increased.

Figure 4 shows the fitted T_1 values of gray matter and blood at three α 's of 10° , 30° , and 50° in 5 subjects. The model fitting was successful in all subjects ($R^2 > 0.9$). As predicted, the estimated apparent T_1 of gray matter was clearly underestimated and this effect increased with α . By contrast, the measurement of blood T_1 showed no apparent dependence on α and remained constant within the α range investigated. The Friedman's test indicated that the measured T_1 varied with α in gray matter ($p < 0.01$) whereas no significant differences in the fitted blood T_1 values were observed across the three flip angles examined ($p > 0.99$). The relationship between blood T_1 and flow velocity was not significant ($r=0.04$, $p=0.94$) by Spearman correlation test in 7 adult subjects.

The multi-phase method was compared with the conventional single-phase method in 5 subjects. As listed in Table 2, the single-phase measurement consistently yielded larger T_1 's. Given the accuracy of multi-phase measurement demonstrated in the flow phantom study, the observed discrepancy is likely due to cerebrospinal fluid (CSF) contamination in the sagittal sinus known as arachnoid granulations, small protrusions of the arachnoid into the venous sinuses that allow CSF to enter the blood stream (24). By applying bi-exponential fitting for blood and CSF, assuming $T_{1,CSF} = 4300$ ms, $T_{2,CSF} = 1442$ ms (25), the mean partial volume of CSF is $14.9 \pm 3.5\%$ in the 5 subjects (see Table 2). If the arachnoid protrusion explanation were true, the multi-phase IR-TrueFISP method might provide more

accurate estimation of blood T_1 than the single-phase measurement due to the saturation of CSF signal. Assuming a mean CSF partial volume of 15% and blood T_1 of 1600ms, the accuracy of multi-phase and single-phase IR-TrueFISP measurements is within 5% and 9%, respectively, based on our simulation.

Exp 3: Validation Study in Children

The wsCV of venous blood T_1 values was 4.0% and the ICC was 0.55 in the 14 of the 18 healthy children who underwent repeated scans 2–4 weeks apart. Regression analyses were performed using the mean venous blood T_1 as the dependent variable, age and gender as the independent variables. Venous blood T_1 was found to decrease significantly with age (slope= -21.5ms/yr , $p=0.02$), while there was a trend for gender difference (females longer than males by 73.3ms, $p=0.169$) in 18 children aged 7 to 17 years (see Fig. 5a). This result is in good agreement with expected Hct change in children and the emergence of gender difference around the age of 12. Repeating the regression analysis on the adult data (Exp 2) revealed a significant gender effect (females longer than males by 148.6ms, $p=0.033$) while the T_1 decrease with age showed a trend (slope= -6.4ms/yr , $p=0.144$, Fig. 5b). This result again is consistent with literature data since Hct stabilizes during adulthood with higher level in males than in females. When we pooled the pediatric and adult data together, regression analysis revealed both significant gender (females longer than males by 106.1ms, $p=0.013$) and age effects (slope= -7.8ms/yr , $p=0.006$, Fig. 5c).

Table 3 shows the mean values and standard deviation (SD) of estimated venous blood T_1 as well as the normal Hct range (3) in 3 age groups and 2 genders respectively. As shown in Fig. 6, the mean R_1 values were highly correlated with mean normal Hct levels across the 3 age groups and 2 genders (Spearman's $r=0.93$, $p=0.008$). The mean blood T_1 was $1862\pm105\text{ms}$ and $1755\pm126\text{ms}$ for females and males in the age range of 7–39 yrs, respectively. Our estimate is within the upper range of reported in-vivo values at 3T (e.g. $1932\pm85\text{ms}$ by Stanisiz et al (26), $1550\pm85\text{ms}$ by Noeske et al (27), $1608\pm62\text{ms}$ by Varela et al (9), $1911\pm55\text{ms}$ in females and $1718\pm62\text{ms}$ in males by Qin and van Zijl (8)). As shown earlier, CSF contamination through arachnoid granulations may be a potential confounding factor. Assuming a mean CSF partial volume of 15% in the sagittal sinus, the corrected venous blood T_1 is $1769\pm100\text{ms}$ in females and $1667\pm120\text{ms}$ in males, respectively (95% of estimated value).

We also attempted the quantification of arterial blood T_1 using IR-TrueFISP by placing the imaging slice through the Circle of Willis. However, the measured T_1 value was in the range of 1200–1400 ms with poorer fitting than that of venous blood. This apparent underestimation may be related to the faster flow that is likely to incur inflow of un-inverted spins, and greater pulsatility in major arteries. Turbulent flow is also more likely to occur in large arteries at branch points especially with high flow. By contrast, the estimated venous blood T_1 appeared to be very robust irrespective of the flip angle, acquisition bandwidth (362–977 Hz/pixel), inter-segment delay time (0–3s), and the number of lines acquired in the k-space per segment (11–31) of the IR-TrueFISP sequence (data not shown), which may be attributed to the relatively stable venous flow. Additionally, the brain acts like a “reservoir” to ensure global inversion of blood magnetization for venous blood T_1 measurements. The proposed IR-TrueFISP method takes advantage of the “time-of-flight” principle in conjunction with the unique characteristics of the TrueFISP sequence including inherent flow compensation and high imaging efficiency.

To develop IR-TrueFISP for arterial blood T_1 mapping which is more relevant for ASL, VASO and black-blood imaging, a few obstacles need to be overcome, including inflow and pulsation effects etc. Assuming little inflow occurs before “globally inverted” arterial blood reaches the null point ($\sim 1\text{s}$, as in VASO), the initial portion (including the null point) of the

IR-TrueFISP curve may be used for arterial blood T_1 fitting in conjunction with cardiac gating to control for variations in flow velocity. Such measurement may be performed in the heart and extremities, in addition to the brain.

One caveat of the IR-TrueFISP technique is the potential confounding effect of laminar flow in blood vessels. Laminar flow inevitably includes very slow flow that may cause underestimation of blood T_1 . Nevertheless, our phantom results showed no measurable underestimation as long as the mean flow velocity was faster than 6 cm/s (see Table 1b). Considering that the measured mean flow velocity within the sagittal sinus ROI was 17 ± 4 cm/s in 7 human subjects (range 9–29 cm/s across pixels in a representative subject obtained using high-resolution PC-MRI), it is unlikely that our estimation of venous blood T_1 was underestimated due to very slow laminar flow. Nevertheless, high resolution PC-MRI may be performed in conjunction with IR-TrueFISP in future studies to exclude pixels with very slow flow for accurate blood T_1 measurement. In the present study, although we mainly related blood T_1 data in healthy children and adults with known developmental changes of Hct, blood T_1 may also be affected by oxygen saturation level (e.g. there is approximately 100ms difference between arterial and venous blood T_1 at 3T)(1). In vivo estimation of venous oxygenation level has been performed recently using susceptometry (28) and blood T_2 measurement (24). Such methods for blood oxygenation measurement may be combined with the proposed IR-TrueFISP method for accurate quantification of Hct and oxygen saturation level. Finally, a rigorous study needs to be carried out to validate MRI measurements of Hct and blood oxygenation levels using lab test results in individual subjects.

Conclusion

In conclusion, we have demonstrated the feasibility of using multi-phase IR-TrueFISP for fast (< 1 min) and reliable (4% repeatability) in-vivo venous blood T_1 measurement. The method is promising for serving as an index of Hct level in individual subjects, although potential confounding factors such as blood oxygenation level need to be adjusted for in future studies.

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Figure 1.
The sequence diagram of the multi-phase IR-TrueFISP used in this study.

Figure 2.
Simulated signal-time curves for multi-phase IR-TrueFISP imaging.

(a)

(b)

Figure 3.

IR-trueFISP images (a) and time courses (b) from a representative subject. In (a), the numbers below images are the corresponding inversion times (TI's) in the unit of ms. In (b), time courses are plotted for the regions-of-interest at sagittal sinus and gray matter respectively, and for different flip angles ($\alpha = 10^\circ$, 30° , and 50°).

Figure 4.
Comparison of the T₁ measurements with different flip angles.

Figure 5.

Measured blood T_1 as a function of age and gender with fitted lines for each gender in 18 children aged 7–17 (a), in 8 adults aged 19–39 (b), and in 26 children and adults aged 7 to 39 (c) (Red diamond and line = female; Black square and line = male; gender=0 for female and 1 for male).

Figure 6. Scatter plot of mean venous blood R1 values and mean normal Hct values based on literature across 3 age groups and 2 genders. Error bars indicate SD.

Table 1a

Summary of the results of fitting the computer-generated multi-phase IR-TrueFISP data to the three-parameter model (see Eq. [1]). Combinations of three flow velocities (v 's) and three flip angles (α 's) were considered. Fifty sets of simulated data were generated for each condition and used to evaluate the reliability of the fitting procedure. The fitted T_1 's are in the unit of ms (mean $\pm$ SD).

Assigned $T_1 = 1600$ ms	$v = 0$ cm/s	$v = 10$ cm/s	$v = 20$ cm/s
$\alpha = 10^\circ$	1498 $\pm$ 16	1599 $\pm$ 9	1599 $\pm$ 7
$\alpha = 30^\circ$	1012 $\pm$ 65	1592 $\pm$ 10	1598 $\pm$ 11
$\alpha = 50^\circ$	587 $\pm$ 8	1588 $\pm$ 11	1595 $\pm$ 11

Summary of flow phantom results using distilled water doped with 0.05 mM/L MnCl₂ and 0.2 mM/L NaCl. Combinations of 4 flow velocities (measured by PC MRI) and three flip angles (α 's) were measured.

Table 1b

$T_1=1473\text{ms}$ $T_2=164\text{ms}$	$v = 0 \text{ cm/s}$	$v = 6 \text{ cm/s}$	$v = 10 \text{ cm/s}$	$v = 13 \text{ cm/s}$	$v = 17 \text{ cm/s}$
$\alpha = 10^\circ$	1399	1472	1499	1466	1509
$\alpha = 30^\circ$	1070	1490	1489	1499	1518
$\alpha = 50^\circ$	785	1479	1493	1499	1510

Comparison of blood T₁ values (ms) measured with single-phase and multi-phase IR-TrueFISP as well as estimated partial volume of CSF in 5 adult subjects. Multi-phase IR-TrueFISP signals of CSF were modeled based on ref (19).

Table 2

Subject	#1	#2	#3	#4	#5	Mean±SD
Single-phase IR-TrueFISP	2052.4	1876.0	1750.5	1793.7	1920.2	1878.6±117.9
Multi-phase IR-TrueFISP	1914.8	1752.5	1690.0	1597.8	1783.9	1747.8±117.3
CSF partial volume (%)	14.1	15.2	11.6	20.8	13.0	14.9±3.5

Estimated venous blood T1 values (mean±SD) and literature values (mean±SD) of normal Hct range in 3 age groups and 2 genders. Negative correlation between mean blood T1 and Hct values is observed (Spearman's $r=-0.93$, $p=0.008$).

Table 3

	6 < Age ≤12		12 < Age ≤18		Age > 18	
	Male	Female	Male	Female	Male	Female
Blood T1 (ms)	1855±109	1887±110	1684±97	1855±152	1677±68	1827±96
Normal Hct Range (%)	40±2.5		43±3.5	41±2	47±3	41±2.5