Eliminating the Partition Coefficient from ASL perfusion quantification with a homogeneous contrast reference image

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Introduction: Arterial spin labeling (ASL) perfusion measurements are conventionally quantified with Eq. [1], where λ is the tissue-to-blood partition coefficient of water, M_t^0 is the fully relaxed equilibrium magnetizations of tissue, and $g(T_{1a},T_{1t},\alpha,\tau,w,\delta)$ is a function containing the relaxation, labeling, and transit time related terms. The tissue-to-blood partition coefficient of water, λ, is typically assumed to be a constant in the measurement of perfusion. However, it is well known that the partition coefficients are substantially different between normal gray matter (GM) and white matter (WM) (1) and they may differ greatly in pathology. Roberts et al. (2) suggested measuring λ. It has been emphasized by Buxton (3), however, that division by the proton density image and multiplication by λ multiplies and divides by M_t^0 unnecessarily. If one instead creates a map of the sensitivity, sen(x), to pure water magnetization, M_w^0, at each voxel [3], the tissue partition coefficient becomes unnecessary [4]. When the coil sensitivity is very uniform, as for transmit receive coils at low field, a measurement of sensitivity in a reference phantom or in the CSF of the ventricles (4) can be used for quantification. Unfortunately this approach is not effective for modern imaging configurations with receive coil arrays and even transmit field inhomogeneity at high field. Here we propose a quick scan to measure the sensitivity map by generating a single image with homogeneous contrast between GM, WM, CSF and potentially other pathological tissues.

$$f=\lambda\cdot\frac{\Delta M}{M_t^0 g(T_{1a},T_{1t},\alpha,\tau,w,\delta)}\quad[1]\qquad \lambda=\frac{M_t^0}{M_b^0}\quad[2]\qquad \Delta S=sen(x)\cdot\frac{\Delta M}{M_w^0}\quad[3]$$

$$f=\frac{\Delta S M_w^0}{sen(x)M_b^0}g(T_{1a},T_{1t},\alpha,\tau,w,\delta)\quad[4]\qquad R_i=R_i^{CSF}+(1-PD)r_i\quad i=1,2\quad[5]$$

Theory: Our goal is to design a preparation to make the magnetization of different brain tissues equal to the same target value V_t. A saturation pulse and several inversion pulses are used to prepare the homogeneous contrast. Restriction to B1 insensitive inversion and saturation pulses helps to avoid B1 dependence of the resulting magnetization. The magnetization after saturation and a series of inversion pulses is a function of the pulse timings, T_1, PD and TE of the subsequent imaging sequence. To a reasonable approximation in most tissues, T_1 and T_2 can be expressed as a function of PD (Eq. [5] (5)). Therefore, for a given TE and the maximum time (T) allowed for the preparation sequence, optimized pulse timings could be calculated such that the magnetization for a range of T_1 ($T_{1min} < T_1 < T_{1max}$) are close to the target value V_t.

Methods: T_{1min} and T_{1max} were set at 0.6 and 4.2 respectively. The signal target value V_t selected was 20% of the fully relaxed CSF signal. TE was assumed to be the minimum echo time, 16.7 ms, of our spin-echo EPI sequence (24 cm FOV, 128x128 matrix). Because the results were predicted to be insensitive to r2, a fixed value of 75 s^{-1} was used. To determine the best r1 value to approximate tissue T1 relaxation as a function of PD [5], we performed pulse preparation timing optimizations for a number of r1 values and for maximum prep time, T, of 3 s, 6 s, and 10 s.

Six subjects were studied on a GE 3 Tesla images with a receive-only 8-channel head array coil. First we acquired images with T=3 s optimized preparations with varied r_1 from 1.35 to 6.25. These were used to determine the optimal r1. Multi-slice homogeneity images were then acquired with a 64 × 64 matrix size to match an ASL acquisition. The inversion times for the multi-slice prep were calculated for r_1=3.75. Multi-slice proton density images were also acquired with spin-echo EPI. ASL images were acquired with pulsed-continuous labeling (PCASL) (6) ($B1_{ave}$ =14 mG, G_{ave}= 0.07 G/cm, and G_{max}/G_{ave}= 10, Δt =1.5 ms). Forty pairs of label and control images were acquired for ASL signal averaging.

Results & Discussion: The optimized pulse timing from the simulation gave uniform signals across T_1 ranges (Fig. 1). Homogeneous contrast was obtained when optimizations for r1=3.75s^{-1} were employed (Fig. 2). Residual large scale intensity variations remained but reflected coil sensitivity profiles. The signals of neighboring GM, WM and CSF were well matched (Fig. 3) at r_1=3.55 (GM:1610 ± 148, WM:1653 ± 127, CSF:1663 ± 166) and r_1=3.75 (GM:1579 ± 153, WM:1604 ± 123, CSF:1646 ± 144). The multislice reference images also showed very homogeneous contrast (Fig. 4a). The perfusion images quantified with homogeneous images (Fig. 4c) showed greater contrast ratio between gray and white matter (ratio=4.24) than the images quantified with proton density images (Fig. 4d, ratio=3.30). Better agreement between the conventional and homogeneous contrast image quantification was achieved when separate gray and white matter partition coefficients (λ_g= 0.98, λ_w= 0.82) (1) were used in the conventional quantification (Table 1). With the homogeneous image approach, no assumptions or mapping of partition coefficient are required.

References: 1. Herscovitch et al, J Cereb Blood Flow Metab 1985;5:65-69. 2. Roberts et al, J Magn Reson Imaging 1996;6(2):363-366. 3. Buxton. J Magn Reson Imaging 2005;22(6):723-726. 4. Chalela et al, Stroke 2000;31(3):680-687. 5. Halle. Magn Reson Med 2006;56(1):60-72. 6. Dai et al, Magn Reson Med 2008;60(6):1488-1497.

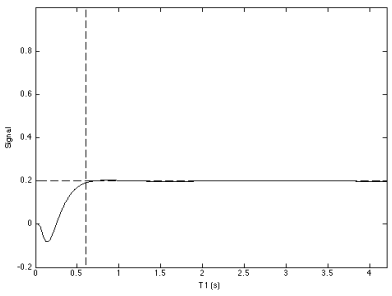

Fig 1. The simulated magnetization using the optimized inversion times as a function of T1 for r_1=3.75 and V_t=0.2 (a dashed horizontal line) at tsat = 6s. T_{1min} and T_{1max} used in the optimization were 0.6s (a vertical dashed line) and 4.2s.

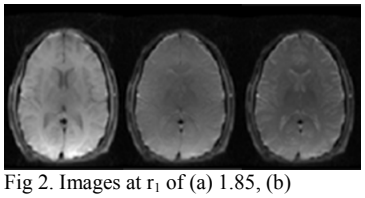

Fig 2. Images at r_1 of (a) 1.85, (b) 3.75 and (c) 6.25.

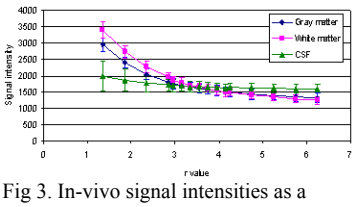

Fig 3. In-vivo signal intensities as a function of r_1 values.

Table 1. Perfusion of GM and WM from different methods.

Perfusion (ml/100g.min)	GM	WM
quantified with PD and λ	76.3 ± 5.7	23.2 ± 6.0
quantified with PD, λ_g, λ_W	83.1 ± 6.2	21.1 ± 5.5
quantified with homogeneous image	84.1 ± 5.5	19.8 ± 4.6

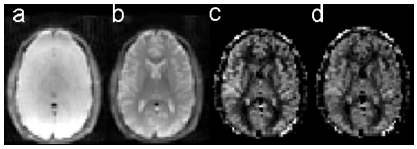

Fig 4. (a) homogeneous contrast image, (b) proton density, (c)perfusion image quantified with (a), and (d) perfusion image quantified with (b).