PASL-like Pseudo Random Amplitude Modulation: measure transit time distribution

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Introduction

Arterial Spin Labeling (ASL) [1], is an important alternative methodology to traditional perfusion imaging that can quantify Cerebral Blood Flow (CBF) without exogenous contrast agents. However, a problem in CBF quantification by ASL is the uncertainty in transit times. Therefore, Continuous ASL (CASL) and pseudo-continuous ASL (pCASL) insert a post-labeling delay before starting data acquisition [2] to reduce the transit-time sensitivity, while Pulsed ASL (PASL) specifies time windows for perfusion, assuming a certain range of transit times [3]. Several transit-time mapping techniques have been proposed to address this problem, but are generally limited to measuring a single value, i.e. mean transit time [4, 5]. In this work, we present the theory and application of PASL-like Pseudo Random Amplitude Modulation (PRAM) [6, 7] to measure the distribution of transit times on a 3T scanner. We show that the transit time distribution in a single slice of human brain can be measured on a clinical scanner within a short time (~ 17s).

Theory

Following the indicator-dilution theory [8], the input arterial magnetization $M_a(t)$ to each voxel can be written as:

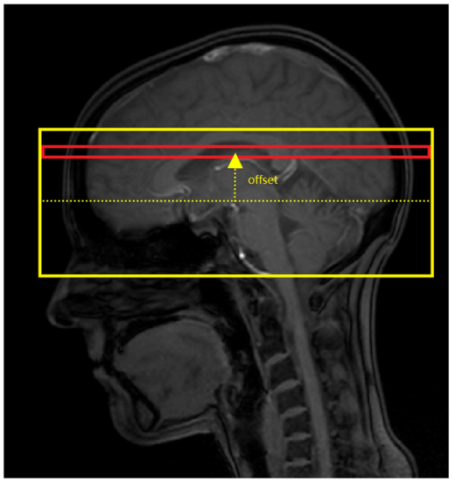

Fig. 1: Labeling scheme of pulsed PRAM: image slice (red) is within labeling region (yellow)

$$M_a(t) = \int_{-\infty}^{t} M_{deliver}\left(t, t', M_l(t')\right) h(t, t') dt'$$

where $M_l(t')$ is the magnetization at system entrance, $h(t, t')$ is the fraction of spins entering the system at t' and arriving at the voxel at t, i.e. the transit time distribution, and $M_{deliver}\left(t, t', M_l(t')\right)$ is the evolved magnetization of $M_l(t')$ at time t. For pulsed PRAM sequence (Fig. 1), t' is the time spins enter the labeling region, and $M_l(t') = M_a^0$. When spins travel between t' and t, they experience a pseudo random series of ON-and-OFF inversion RF pulses denoted by A_i (A_i=1 if inversion pulse ON and 0 if OFF) applied at $t_i \in (t', t)$. As a result, $M_{deliver}$ is a function of blood longitudinal relaxation T_{1a}, repetition time TR, and $\{A_i\}$. For any $t^*_{n-m} \in (t_{n-m}, t_{n-m+1})$, the delivery function in the pulsed PRAM sequence is:

$$M_{deliver}(t_n^+, t^*_{n-m}) = M_a^0 \left\{ E^{m-1}(-1)^{\sum_{j=n-m+1}^{n} A_j} + (1-E)\left[\sum_{j=0}^{m-2} E^j (-1)^{\sum_{k=n-j}^{n} A_k} \right] \right\}$$

where $E = \exp(-T_{1a}/TR)$. The matrix representation of the input magnetization then becomes $M_a(t_n^+) = TR \cdot W \cdot H$, where $W_{nm} = M_{deliver}(t_n^+, t^*_{n-m})$, $H(m) = h(t_n^+, t^*_{n-m})$. Furthermore, if a time-invariant system is assumed, $H(m) = h(t_n^+, t^*_{n-m}) = h(t_n^+ - t^*_{n-m})$, and t^*_{n-m} is usually interpreted as the center of the (t_{n-m}, t_{n-m+1}) period, i.e. $(t_{n-m} + t_{n-m+1})/2$. The solution to the Bloch equation of brain tissue is:

$$M_b(t^-_{n,img}) = M_b(t^-_{n-i,img}) \cos^i(\theta) E_{T_{1app}}^i (-1)^{\sum_{j=n-i+1}^{n} A_j} + \left\{ \sum_{j=1}^{i} \left[f \cdot TR \cdot \delta_{T_{1app}T_{1a}} E_{T_{1app}} \Delta M_a(t^*_{n-j}) + M_0\left(1 - E_{T_{1app}}\right) \right] \cos^i(\theta) E_{T_{1app}}^{j-1} (-1)^{\sum_{k=n-j+1}^{n} A_k} \right\}$$

where θ is flip angle, $\frac{1}{T_{1app}} = \frac{1}{T_{1b}} + \frac{f}{\lambda}$, $\delta_{T_{1app}T_{1a}} = \exp\left[(t^*_{n-j} - t_n)\left(\frac{1}{T_{1app}} - \frac{1}{T_{1a}} \right) \right]$, $\Delta M_a(t) = M_a(t) - M_a^0$.

Methods

Both 15-cycles and 31-cycles PRAM sequences (data not shown) were implemented and tested with human brain and flow phantom (data not shown) on a 3T Siemens Trio scanner. A 15.36 ms hyperbolic secant RF pulse was used for the inversion pulse, followed by a 20 ms gradient spoiler. The labeling block was executed before gradient-echo echo-planar imaging acquisition. The other imaging parameters were: TE = 52 ms, PRAM labeling slab thickness =130 mm, FOV = 220 x 220 mm^2, matrix size = 64 x 64, number of averages = 4. For TR = 200 ms and 500 ms, the total scan time was 17s and 39s, respectively. Multiple datasets were acquired with the labeling region offset 45, 0, -45 mm with respect to the image slice center. To enhance the signal from the vasculature, 90-degree excitation angle was employed and two pre-saturation pulses separated by 25 ms were applied every TR to the image slice to suppress the non-flowing spins.

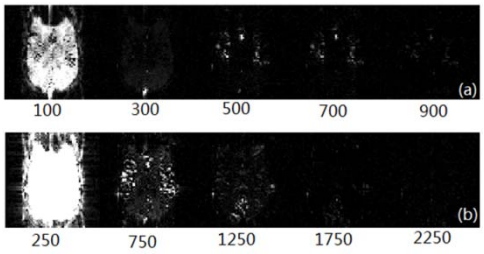

Fig. 2: First five images of reconstructed H, 15-cycles PRAM with offset = 45. Axis unit: ms. (a) TR = 200 ms; (b) TR = 500 ms.

Results

Reconstructed images of H at various TR with the 15-cycles PRAM sequence clearly show arterial structures at consistent absolute times (Fig. 2). Fig. 3 confirms the fact that spins reach the image slice at different times as the offset is varied. The initial images contain a large signal arising from static tissue modulated by PRAM inversion pulses.

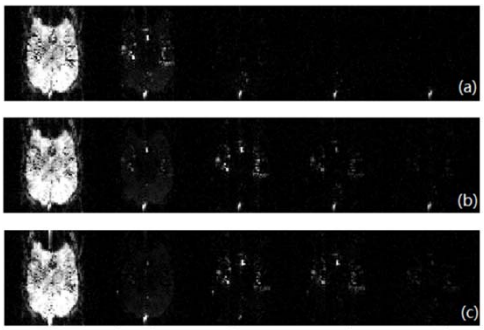

Fig. 3: First five images of reconstructed H, 15-cycles PRAM with TR = 200 ms. (a) offset = -45 mm; (b) offset = 0 mm; (c) offset = 45 mm.

Discussion

Pulsed PRAM echo-planar sequences can measure the arterial transit time distribution, but current methods suffer from low SNR partially due to the large static tissue signal and T_2^* effect from the long TE gradient echo readout. Future work will address these issues by utilizing spin-echo parallel imaging techniques and pCASL-like PRAM.

References

[1] Detre et al, MRM, 23(1): 37-45 (1992). [2] Alsop et al, JCBFM, 16(6): 1236-1249(1996). [3] Wong et al, MRM, 39(5): 702-708(1998). [4] Gunther et al, MRM, 46(5): 974-984(2001). [5] Wang et al, MRM, 50(3): 599-607(2003). [6] Brown et al, ISMRM 2006. [7] Taei-Tehrani et al, JMRI, 2011.[8] Martin et, Am. J Physiology, 207: 162-168(1964).