Pseudo-Continuous Arterial Spin Labeling based Dynamic Angiographic Imaging with Decreased Number of Acquisitions

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Purpose: To obtain Pseudo-Continuous Arterial Spin Labeling (pCASL) based dynamic angiographic images with decreased number of acquisitions, replacing the conventional control-tag pair imaging with more efficient labeling based on Hadamard encoding.

Background: General Kinetic Model (GKM) [1] describes ASL measurements by the following equation:

$$\Delta M(t) = 2M_{0B} f \int_0^t c(t')\, r(t-t') m(t-t') dt' \qquad [1]$$

where $\Delta M(t)$ is the magnetization difference between control and tag measurements, M_{0B} is the equilibrium magnetization of blood, f is the flow, *c(t)* is the delivery function (normalized arterial concentration of magnetization arriving at time *t*), *r(t-t')* is the residue function (fraction of labeled spins that arrived at time *t'* and still in the voxel at time *t*) and *m(t-t')* is the magnetization function (fraction of original longitudinal magnetization that arrived at time *t'* that remains at time *t*).

Methods: In the measured ASL signal *M(t)*, flow contribution $M_f(t)$, and static tissue contribution S_0 can be separated:

$$M(t) = S_0 + M_{0B} f \int_0^t c(t')\, r(t-t') m(t-t') dt' \qquad [2]$$

In continuous ASL (CASL) and pseudo-continuous ASL (pCASL) experiment, c(t) above can be written as $e^{\delta t/T1B}$w(t), where δt is transit time, *w(t)* is the arterial modulation function equals "*+1*" for control and "*-1*" for label images. Avoiding using an analytical expression for *r(t)* allows it to be redefined and include the transit time **δ**t effect, such that *r(t< δt)=0*. Discretization of Eq-1 results:

$$M(t_i) = S_0 + \overbrace{M_{0B} f \Delta t e^{\delta t/T1B}}^{\kappa} \sum_{j=1}^{i} w(t_j)\, r(t_i - t_j) e^{(t_i - t_j)/T1B} \qquad [3]$$

Luckily, one can use arbitrary $w(t_j)$ for arterial modulation. Forming *w* using Hadamard encoding is proposed as an efficient way of labeling [2-3]. Then, it becomes easy to obtain S_0 and *r(t)* by addition/subtraction of measured magnetizations:

$$0.25 \begin{pmatrix} 1 & 1 & 1 & 1 \\ 1 & -1 & 1 & -1 \\ 1 & 1 & -1 & -1 \\ 1 & -1 & -1 & 1 \end{pmatrix} \begin{pmatrix} M_1 \\ M_2 \\ M_3 \\ M_4 \end{pmatrix} = \begin{pmatrix} S_0 \\ \kappa R_3 e^{-3\Delta t/T1B} \\ \kappa R_2 e^{-2\Delta t/T1B} \\ \kappa R_1 e^{-\Delta t/T1B} \end{pmatrix} \qquad [4]$$

Imaging is performed with a healthy volunteer (35 yo, male) in a 3T Siemens (Tim Trio) system. PCASL is used with 24° flip angle, 520 μs Hanning pulses, 1 ms pulse separation, 7.5 mT/m gradient amplitude, 1.0 mT/m mean gradient during tag and zero gradient during control pulses. 6 segment Turbo-Flash is used for readout, TR=10 ms, TE=3 ms, 10° flip angle, 192x152 matrix size with 1x1x50 mm voxel size. Dynamic angiographic images of from the conventional (control-tag pairs) and the newly proposed sequences are obtained and compared. Former uses progressive durations of labeling (400 ms, 800 ms, 1200 ms) to obtain inflow effect [4, 5]. In the newly proposed (based on Hadamard encoding) sequence, 1200 ms bolus is split to three 400 ms sub-bolus, [1 1 1], [-1 1 -1], [1 -1 -1], [-1 -1 1] bolus profiles for respective measurements. These acquired images [M_1, M_2, M_3, M_4] were put in Eq-4 to obtain [S_0, R_1, R_2, R_3].

Results: Images obtained using control-tag pairs is shown in Figure-1a and demonstrates three different inflow phases. Images in Figure-1b are obtained by Hadamard method. R1, R1+R2 and R1+R2+R3 map are theoretically identical to ΔM_1, ΔM_2 and ΔM_3 maps respectively.SNR measurements for six images were 27.3, 30.3, 38.5, 30.4, 30.1 and 34.4.

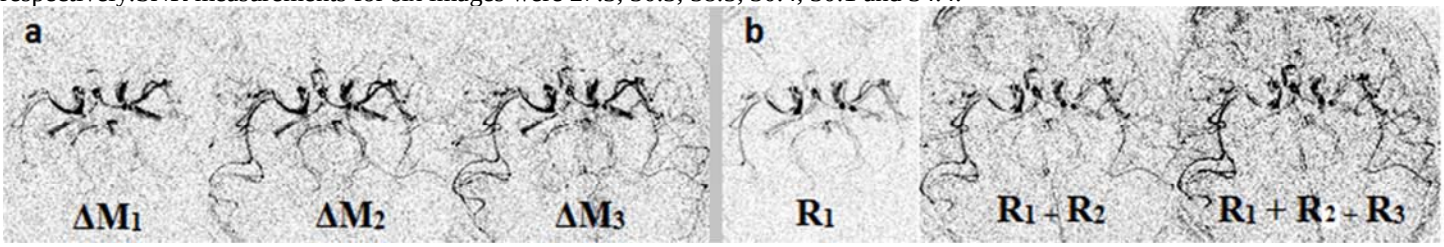

Figure-1a. Inflow phases obtained by conventional control-tag subtraction. **b.** Images reproduced by Hadamard method with 4 acquisitions instead of 6.

Discussion & Conclusion: Using Hadamard labeling scheme and performed analysis, inflow maps are obtained by reduced number of acquisitions. Three images of Fig1a are obtained by 6 measurements, on the other hand corresponding images of Fig 1b are obtained by only 4 measurements. SNR values were observed to be similiar for both methods for the first two phases. A relative decrease of SNR in Hadamard method was measured for the longest phase, possibly because of adding more noisy R_3 map (response that is subject to longest T_1 decay). Still, the SNR efficiency was better for all phases in Hadamard method. This scheme can also be extended such that 7 phases can be obtained with 8 acquisitions instead of 14. Previously, it is shown that longer bolus ASL signal curves can be reproduced by summation of short bolus signal curves [6]. To the best of our knowledge, it is first time that this scheme is applied in a dynamic angiographic study on human brain.

References: [1] R. Buxton et. al, MRM 40:383(1998), [2] M. Gunther, Proc. Intl. Soc. Mag. Reson. Med. 17 (2009), [3] J.A.Wells et. al MRM 63:1111(2010), [4] P.M. Robson et. al, Radiology (2010), [5] O.Ozyurt et. al, Intl. Soc. Mag. Reson. Med. (2010), [6] M. Gunther, Proc. Intl. Soc. Mag. Reson. Med. 15 (2007)