Impact of equilibrium magnetization of blood on ASL quantification

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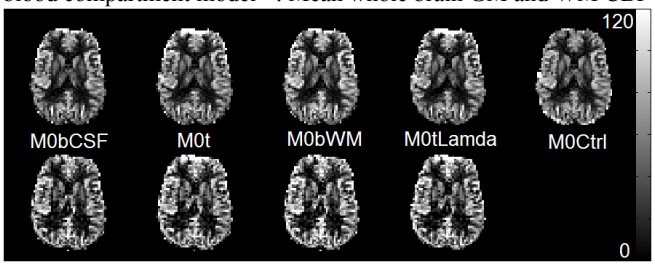
Introduction Arterial spin labeling (ASL) is a noninvasive technique capable of measuring cerebral blood flow (CBF) in absolute physiological units. In the single blood compartment ASL model[1], the term M_{0b} represents the equilibrium magnetization of blood, which is difficult to measure given the low resolution of ASL images. Most ASL studies are divided into two groups regarding M_{0b}: one group calculates M_{0b} based on either cerebrospinal fluid (CSF) or white matter (WM) signal on a separately acquired M_0 image, the other uses the entire M_0 image to generate a M_{0b} map based on λ, the partition coefficient of tissue-to-blood. Continuous ASL (CASL) studies have the additional choice of using the control images as an estimate of M_0, thus eliminating the need for a separate acquisition. To date, no systematic comparison between the methods has been presented. In this study, we compare the quantitative CBF values generated using these different M_{0b} estimates.

Methods Ten healthy subjects (age 25±6, 3F, 7M) were scanned on a whole-body 3.0T MR scanner (Siemens TIM Trio, Erlangen Germany) equipped with a 32-channel receive-only head coil. All subjects were scanned with both FAIR[2] (TI1/TI2=600ms/1600ms) and pseudo-continuous ASL[3] (pCASL, labeling duration 1.5s, postlabeling delay 1s) using gradient-echo EPI (TR/TE/FA=4s/18ms/90°, 18 axial slices of 6mm thickness and 1.2mm gap), 60 pairs of control/tag. An additional M_0 image was acquired using the same EPI parameters without any ASL preparation. A minimum contrast image was also acquired for coil sensitivity correction. Finally, high resolution MPRAGE images were collected for co-registration purposes.

The mean difference images and M_0 images were corrected for inhomogeneities in coil sensitivity by division with the minimum contrast image. To assess the effect of different M_{0b} calculations on CBF estimates, the following M_{0b} values were used: 1) M0b_CSF: M_{0b} estimated using the equation $M_{0b}=R*M_{0CSF}*\exp(TE/T^*_{2CSF}-TE/T^*_{2b})$, where R was set to 0.87[4], M0CSF was signal of CSF voxels (designated as those with top 20% of signal in the center of the image), T^*_{2CSF} and T^*_{2b} were set to 74.9 and 43.6[4]. 2) M0t: The M0 image was divided by 0.9, designated as mean λ between gray (GM) and white matter (WM). 3) M0b_WM: M_{0b} estimated using the same equation as M0b_CSF with R=1.19 and T^*_{2WM}=44.7[4]. 4) M0tLamda: The M0 image was segmented into gray and white matter, gray matter voxels were divided by λg=0.98, white matter voxels divided by λw=0.84[5]. 5) M0ctrl: Mean of the control images divided by 0.9. This was only available for pCASL. CBF maps were generated using the single blood compartment model[1,6]. Mean whole brain GM and WM CBF were generated using the segmented tissue masks thresholded at 0.75.



Results Sample CBF maps calculated using the different M_{0b} values are shown to the left (top: pCASL, bottom: FAIR). No apparent difference was observed between the various CBF maps. Barplots of mean (±standard error) values of M_{0b}, GM CBF and pCASL GM:WM CBF ratio are shown below. Map-based M_{0b} values were extracted from whole-brain GM masks. Although mean M0b_CSF was the highest of all M_{0b} estimates, resulting in the lowest CBF value, the effect size was small. However, the use of different M_{0b} estimates altered the GM:WM contrast of the qCBF maps, as is evident in the 3rd barplot. While both M0bCSF and M0bWM preserved the intrinsic contrast of the difference images, both M0t and M0tCtrl resulted in lower contrast as a result of using a single λ for the entire brain. The contrast was restored when tissue-specific λ was used in M0tLamda.

Discussion Our results demonstrate minor differences between the various M_{0b} estimates. The largest discrepancy was observed between M0bCSF and M0bWM, likely due to the assumed values of T_2^* which is known to be highly variable. Interestingly, M0tCtrl was very similar to M0t, suggesting that the concern about magnetization transfer effects of the long labeling pulse train is minimal, possibly due to the long labeling duration + post labeling delay used in the current study.

Although the GM CBF values appear similar, visible differences were observed in the GM:WM ratio between the various M_{0b} methods. Notably, the map-based methods: M0t and M0tCtrl which use a single λ for the entire brain reduce the GM:WM contrast. Although the correct contrast can be restored by using tissue-specific λs, this approach requires additional tissue segmentation step, which is subject to partial volume effects at the image resolution of typical ASL studies. In conclusion, the M0bCSF and M0bWM approaches may be more accurate, though care must be taken to eliminate coil inhomogeneity and accurately measure T_2^*. Though the two methods are interchangeable in healthy subjects, patient populations with WM pathology may be limited to the M0bCSF method.

References (1)Buxton, R. B. *J Magn Reson Imaging* **2005**, *22*, 723-6. (2)Kim, S. G. *Magn Reson Med* **1995**, *34*, 293-301. (3)Dai, W.et al., *Magn Reson Med* **2008**, *60*, 1488-97. (4)Cavusoglu, M. et al., *Magn Reson Imaging* **2009**, *27*, 1039-45. (5)Herscovitch, P.; Raichle, M. E. *J Cereb Blood Flow Metab* **1985**, *5*, 65-9. (6)Alsop, D. C.; Detre, J. A. *J Cereb Blood Flow Metab* **1996**, *16*, 1236-49.