A Proposed Solution for Reconciling the Cerebral Blood Flow Discrepancy Between PET and MRI

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Introduction Gray matter (GM) cerebral blood flow (CBF) values measured by arterial spin labeling (ASL) are generally larger than PET values, with the underlying reason(s) remaining a topic of discussion (1). More specifically, PET literature commonly reports CBF values for normal GM in the range 40–55 ml/100g/min (2), whereas ASL values as high as 100 ml/100g/min have been published for comparable subjects. We demonstrate two main causes of this discrepancy, one relating to GM CBF *overestimation* by ASL and one relating to GM CBF *underestimation* by PET. First, when using short inversion time (TI), ASL overestimates perfusion due to insufficient time for in-flowing labeled blood water to leave small arteries and enter tissue. Although this has been previously discussed (3), short TIs remain in usage at low field strength where the T1 of blood water is short (1350 ms) (4). We evaluate TI influence on CBF measurements at 3T, where the T1 of arterial blood water is much longer (1630 ms) (5). Second, due to the low resolution at which PET CBF measurements are acquired, partial volume effects between GM, white matter (WM), and cerebrospinal fluid (CSF) cause GM CBF to be underestimated. We evaluate the extent of these effects by analyzing ASL CBF maps as a function of resolution.

Methods Studies were performed at 3T (Philips Medical Systems) using body coil transmission and SENSE head coil reception. ASL measurements used the Transfer-Insensitive-Labeling-Technique (TILT) (6), a pulsed ASL technique similar to EPISTAR in that only blood proximal to the imaging slice is labeled, thereby reducing draining venous blood contributions. TILT is also insensitive to magnetization transfer effects which could be misinterpreted as perfusion. Multiple-resolution studies (n=5) were performed with FOV=240 mm, slice thickness=5 mm, SENSE=2.0 and in-plane isotropic matrix sizes=96, 80 64, 48, and 32. A gradient echo, single-shot EPI readout was used with TR/TI = 2000/1500 ms. For arterial flow analysis, TILT perfusion images for varying TI (100–2500 ms) were overlaid on phase-contrast MR angiography (MRA) images to identify precisely when and where arterial blood was contributing to perfusion signal. CBF quantification was performed by applying the solution of the Bloch equation modified to include perfusion and transit time to the TILT difference image; GM/WM segmentation was achieved by overlaying the CBF map on a T1-weighted anatomical image (FOV=240 mm, matrix size=256, slice thickness=5 mm, TR/TE=8.3/3.9 ms). CBF in GM and WM was calculated based on this total area approximation. This GM/WM delineation method is similar to the PET method.

Results and Discussion MRA (Fig. 1a) depicts the position of small arteries in a single slice; arterial contributions to the perfusion signal corresponding to this slice can be seen for low TI (Fig. 1b) as well as high TI (Fig. 1c). CBF dependence on TI is shown for five different regions of interest in Fig. 1d. At TI<1500ms, CBF is greatly overestimated due to arterial contributions, whereas the CBF roughly plateaus at TI≥1500 ms, once tagged arterial blood water has left the imaging slice and/or entered tissue. At 1.5T, the T1 of blood water is much shorter than at 3T and therefore perfusion imaging at TI≥1500 ms is difficult since most of the tag has already decayed. The longer T1 at 3T allows for longer TIs, thereby reducing arterial contributions to the ASL difference signal. This method of "waiting" is perhaps more effective than applying crusher gradients at low TI since crusher gradients have the potential of either not crushing all of the arterial blood water, or conversely, crushing some of the perfused tissue blood water. Fig. 2a-e shows perfusion images acquired for varying resolution, with the lowest resolution (7.5 mm^2 in-plane) roughly mimicking PET resolution. GM CBF decreases by approximately 20.5 % over this range (Fig. 2f) whereas WM CBF shows little change as a function of resolution. This WM pattern is expected due to the larger WM area and consequently fewer partial volumed voxels. It is important to note that at resolution 7.5 mm^2, GM CBF values appear to be close to those reported in PET literature, whereas at higher resolution (2.5 mm^2 in-plane) partial volume effects decline and CBF values converge to higher, more accurate values. These results largely reconcile the differences between PET and MRI CBF measurements and point toward a true measure of perfusion independent of imaging modality.

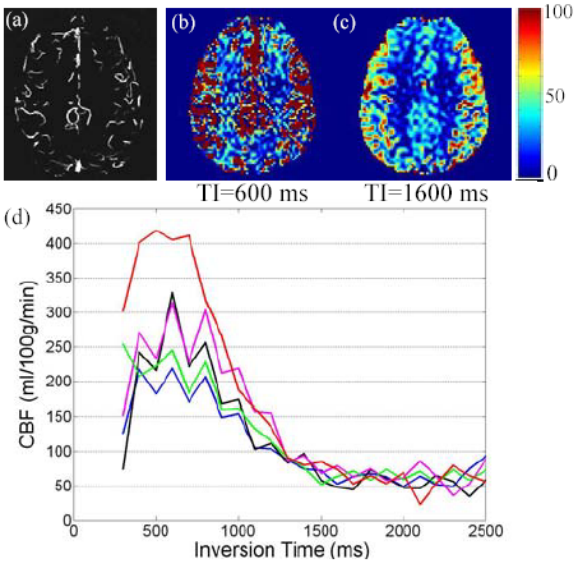

Fig. 1. (a) MRA depicting regions of arterial flow. Corresponding CBF map for (b) TI=600 ms and (c) TI=1600 ms. The CBF time course for five different regions of interest (d) shows a largely steady-state CBF at 1500 ms≤TI≤2000 ms.

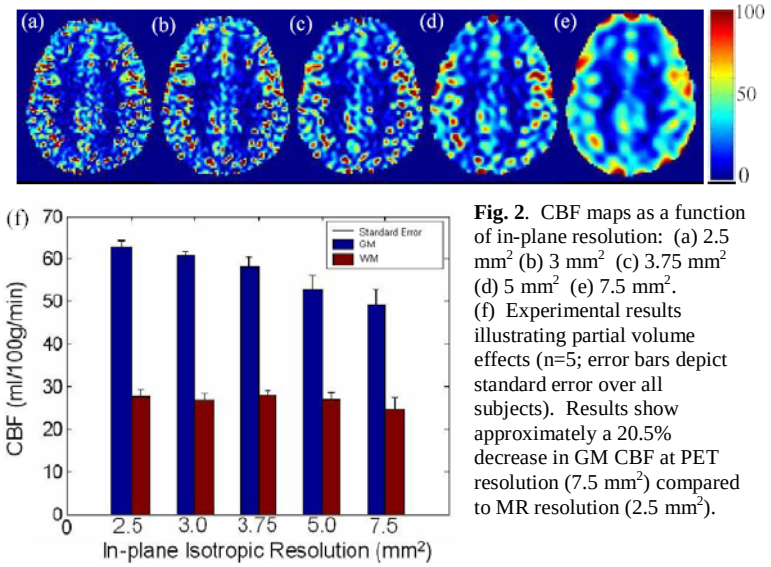

Fig. 2. CBF maps as a function of in-plane resolution: (a) 2.5 mm^2 (b) 3 mm^2 (c) 3.75 mm^2 (d) 5 mm^2 (e) 7.5 mm^2. (f) Experimental results illustrating partial volume effects (n=5; error bars depict standard error over all subjects). Results show approximately a 20.5% decrease in GM CBF at PET resolution (7.5 mm^2) compared to MR resolution (2.5 mm^2).

References [1] Ye F. MRM. 2000;44:450. [2] Iida H. JNM. 1998;39:1789. [3] Wong E. NCNA. 1999;9:333. [4] Lu, MRM 2003;40,263, [5] Lu, MRM 2004;52:679, [6] Golay, X. JMRI. 1999;9:454