Flow-weighted arterial transit time mapping of the human brain

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Introduction

The knowledge of the arterial transit time, δ_a, from the labeling plane into the brain is essential for blood-flow quantification. In previous work, the MATISSE (Mapping of Arterial Transit times by Intravascular Signal Selection) approach for the mapping of δ_a was introduced (1). It relies on the use of short repetition times, *TR*, and the application of an amplitude-modulated shape of the labeling function. The results were found to agree well with the vascular territories of the brain-feeding arteries (1). Under these conditions, flow-through signals, i. e. the signals of arterial vessels permeating the voxel of interest, are a major source of the MATISSE signal. For CBF quantification, however, δ_a is defined as the time the label needs to travel from the labeling plane to the voxel assuming the endpoint of the arterial vascular tree in the same voxel (2). In the current work, it is shown that a mild flow-weighting (FW) is suitable for an almost complete removal of flow-through signals in MATISSE. The goal was to closer meet the demands for CBF quantification by the corresponding δ_a maps.

Method

All experiments were performed using a 3-T whole-body scanner (MedSpec 30/100, *Bruker BioSpin, Ettlingen, Germany*). For image acquisition, a microstrip helmet resonator of 23-cm inner diameter and 18-cm length was used (3). For imaging, a gradient-echo EPI sequence with an echo time of 24 ms, a FOV of 174×192 mm^2, an acquisition matrix of 58×64, an acquisition bandwidth of 150 kHz, and a slice thickness of 4 mm was employed. For continuous arterial spin labeling (CASL) at the common carotid artery, a labeling coil consisting of the perpendicular combination of two circular coil loops (6-cm diameter each) was employed (4). The labeling gradient strength was 2 mT m^{-1}. A variable labeling RF power was used between subsequent repetitions in order to create a smooth variation of the labeling efficiency. The labeling RF pulse was applied during the first 400 ms of each *TR* interval (500 ms), and two slices were acquired without further delay during the last 100 ms. 273 repetitions were acquired within one scan, during which the labeling RF power was cycled in 21 steps. Whole-brain coverage was achieved by a total of 18 slices which were acquired in 9 scans. MATISSE data with and without a mild FW (a bipolar gradient pulse along the *z*-axis with *b*-values between 2 and 4 s/mm^2) were obtained from 6 healthy volunteers (2 male, 4 female, age 23-31 years). Maps of δ_a and the MATISSE signal change were obtained from a Gaussian fit of the perfusion-weighted time series. The standard deviation of the Gaussian fit with respect to the MATISSE signal change was used as threshold.

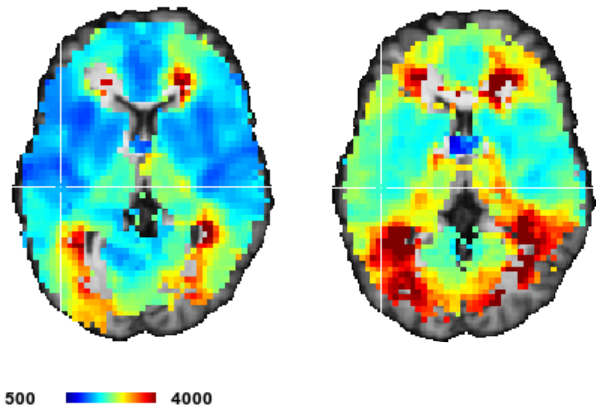

Fig. 1. Averaged δ_a maps of 6 subjects in milliseconds with (right) and without (left) the application of a mild FW with *b*-values between 2 and 4 s/mm^2.

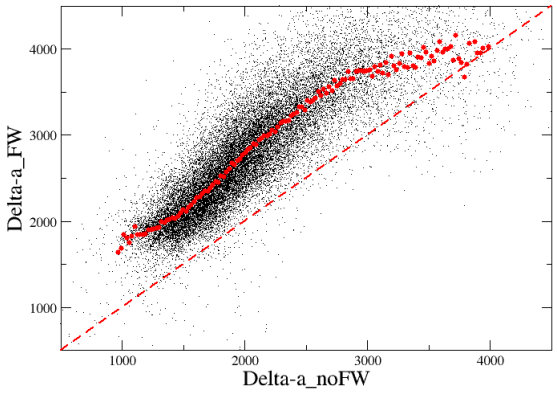

Fig 2. Scatter plot for the datasets shown in Fig. 1 with the dataset without FW on the *x* and the dataset with FW on the *y* axis. Red circles represent mean values along the *y* axis. The dashed line marks the line of identity.

Results

Figures 1 & 2 show results of MATISSE obtained with and without FW. Both, the maps in Fig. 1 and the scatter plot in Fig. 2 indicate that mild FW increases the observed δ_a's drastically. In the range of δ_a without FW between 1000 ms and 3000 ms a nearly constant difference of about 700 ms was obtained. For values above 3000 ms the differences decrease and approach zero at about 4000 ms. On the other hand, the comparison between data obtained in a single subject with a FW of 2 and 4 s/mm^2 showed nearly insignificant differences. Simulations of the signal loss in arterial vessels of varying diameter (5) suggest that, for diameters above 0.2 mm, a *b*-value of 2 s/mm^2 already destroys most of the arterial signal. This diameter coincides well with the value of 0.1 mm estimated for the smallest arteries capable of permeating the voxel of interest at the spatial resolution of 3 mm. We therefore conclude that MATISSE with mild FW removes flow-through signals almost completely. This comes to the expense of the reduced signal amplitude which on average roughly was halved from 1.4 % to 0.8 %. The remaining MATISSE signal still is of intravascular origin from vessels with the endpoint of the arterial vascular tree in the same voxel. CBF quantification based on two-compartment models as in Ref. (2) might therefore be improved by applying δ_a maps obtained with FW-MATISSE. These δ_a values can easily be above 2 s, and suggest that, for typical parameters applied in dual-coil CASL, the perfusion signal might contain a large fraction from the vascular compartment, because pure tissue perfusion is observed only for post-labeling delays with $w > \delta_a + \Delta\delta_a$ with $\Delta\delta_a$ being the mean residence time in the vascular/microvascular bed (6).

References

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