Validation of calibrated MRI using continuous-wave and time-domain near-infrared spectroscopic imaging

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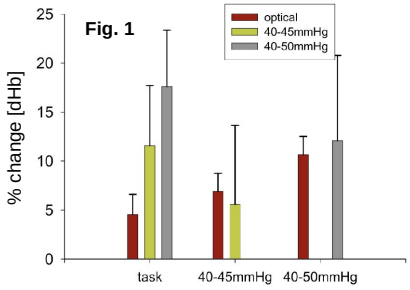
Introduction

Calibrated MRI seeks to estimate changes in tissue oxygen consumption during a task using BOLD and arterial spin-labeling (ASL) measurements acquired during a calibration procedure, typically hypercapnia, and the task of interest. A number of biophysical models have been employed [1,2,3] to translate MRI-observable measures into physiological parameters such as the fractional changes in blood volume fraction and concentration of deoxygenated hemoglobin in venous blood. These parameters, together with the fractional change in cerebral blood flow, can be used to estimate the relative change in the cerebral metabolic rate of O_2 consumption ($CMRO_2$) elicited by the task. Confidence in measures of $CMRO_2$ derived from this approach has been limited by the fact that a number of implicit assumptions used in the available models have been difficult to verify. The objective of the present study is to verify key aspects of the model proposed by Davis *et al* by using quantitative optical imaging methods to determine fractional changes in cerebral blood volume (CBV) and the tissue concentration of deoxygenated hemoglobin.

Methods

6 healthy subjects participated in the study. Hypercapnic modulation was achieved by mixing gases containing different concentrations of CO_2, O_2 and N_2 using a RespirAct blood gas control system (Thornhill Research, Toronto, Canada). End-tidal pCO_2 transitions from 40 to 45mmHg and 40 to 50mmHg were targeted. Four functional scans during a motor task were performed, consisting of alternating 20 s blocks of self-paced finger apposition and rest, starting with a rest period.

MRI data acquisition was carried out using a Siemens Trio 3 Tesla MRI system. Images of relative perfusion were acquired using a PICORE/Q2TIPS arterial spin-labeling acquisition. The spatial resolution was 3.75mm x 3.75mm on a 64x64 matrix, with 5 slices of 7 mm thickness. Other sequence parameters included TR/TE/α = 2s/25ms/90˚ and TI1/TI2 = 700ms/1400ms. A slab thickness of 200 mm was used, with a 10 mm gap between the top of the label slab and the most inferior image slice. The Q2TIPS stop time was 1350 ms. Flow crushing diffusion gradients with a velocity cutoff of 20 cm/s were applied. A T1-weighted structural scan was also acquired, for later use in spatial normalization, at 1mm isotropic resolution, using an MPRAGE sequence with TI/TR/TE/α = 900/2300/2.94/9˚. Voxel size was 1.0 x 1.0 x 1.2 mm. A continuous wave optical imaging system (CW5, TechEn Inc., Milford MA) with 8 sources and 8 detectors was used to measure task-induced changes in absorption on the left side of the head at 690 and 830 nm. A time-domain system with four sources (690, 750, 800, and 850 nm) and four detectors was used to measure baseline optical properties of the brain and head based on a two-layer model, permitting accurate determination of fractional change values for deoxygenated hemoglobin during activation [4]. Fractional changes in BOLD and ASL signals during the two levels of hypercapnia and the motor task were determined by linear modeling, and substituted in the MRI signal model used previously to estimate $CMRO_2$. The model implicitly estimates the fractional change in venous dHb concentration, which can be converted to an estimate of tissue concentration change by adjusting for cerebral blood volume (based on the Grubb relation used in this model). Implied values for fractional change in dHb were extracted for each subject and averaged, then compared against corresponding values based on the optical measures.



Results and Discussion

Fractional changes in dHb concentration indicated by the optical and MR measures were in close agreement for hypercapnia (Fig. 1). For the motor task, the optically derived values were significantly lower, to a degree consistent with the partial volume effect expected in this modality (less of an issue for hypercapnia, due to the global extent). We are presently working on solution of the forward optical propagation model based on tissue classification of the T1-weighted scan and focal activation patterns depicted by the functional MRI data. This should allow comparison of the measurements under more equivalent spatial conditions.

References

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