Imaging functional decrease of the cerebrospinal fluid volume fraction with a spin-locking fMRI technique

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Introduction

A voxel of magnetic resonance imaging often contains blood, tissue water, as well as the cerebrospinal fluid (CSF). Recent studies have suggested that brain vascular activation could induce a change in the volume fraction of the CSF compartment that serves as a buffer for the brain cortex [1-2]. This effect may have important implications to fMRI, especially when the tissue signal is manipulated and the CSF signal contribution becomes significant, such as in the case of VASO-fMRI. Current detection of CSF volume fraction and its functional change, however, requires multi-compartment data fitting, the robustness of which relies on the underlying theoretical model and the signal to noise of the data acquisition. In this work we aimed to image the CSF compartment directly using a spin-locking (SL) technique at 9.4 T. With a long SL preparation, the parenchyma signal can be suppressed and a functional decrease of CSF volume fraction can be robustly detected during cat visual stimulation.

Materials and methods

All experiments were performed on a 9.4T MRI (Varian) system. Six female adolescent cats were anesthetized and scanned using a 1.6-cm diameter surface coil. A coronal slice was chosen for fMRI study and a T_1 weighted image with 128×128 matrix size was obtained for anatomical reference. Other imaging parameters were: 2×2cm^2 FOV, 2mm slice thickness, 2-segmented EPI with TR 2.5s/segment, and 96×96 matrix size which was zero-filled to 128×128 for reconstruction. The pulse sequence for SL experiment was a double spin-echo EPI sequence with adiabatic SL preparation [3]. Following the non-selective SL preparation, transverse spins were refocused using two adiabatic full-passage RF pulses with slice-selection gradients. At 9.4 T and with a SL frequency of about 500 Hz, the spin-lattice relaxation time in the rotating frame ($T_{1\rho}$) of the CSF water was measured to be ~450 ms, much longer than that of tissue water (~50 ms). $T_{1\rho}$ of blood is expected to be only slightly longer than the T_2 of blood (~40 ms for arterial and 6-9 ms for venous blood, respectively) at a low SL frequency used in this study (400 –600 Hz depending on the position of voxel in the area of interest). Hence, the MR signal of an imaging voxel with very long SL preparation (≥ 200 ms) can be simplified into a single CSF compartment. The fractional signal change during brain activation can be expressed as:

$$\Delta S / S \approx \Delta V_{csf} / V_{csf} - TSL \cdot \Delta R_{1\rho,csf} - TE \cdot \Delta R_{2,csf} \qquad (1)$$

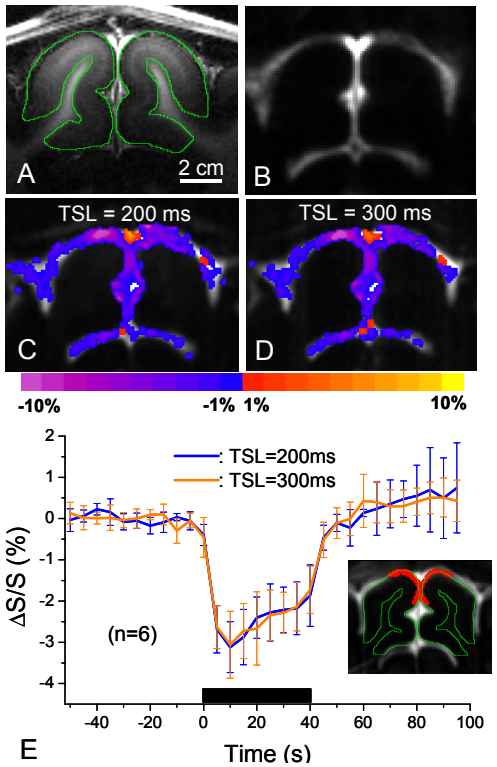
where V_{csf} is the volume fraction of CSF, TSL is the duration of the SL pulse, and TE is the echo time of the double spin-echo EPI readout. Equation (1) takes into account the functional change of the CSF volume fraction as well as residue contributions from the change of $R_{1\rho}$ and R_2 of CSF, if any. Two sets of experiment parameters were chosen to extract the fMRI CSF volume fraction change: (i) TSL/TE = 200/15 ms, and (ii) TSL/TE = 300/22.5 ms. The visual stimulation paradigm is 10 (60s) control, 8 (40 s) stimulation and 12 control images for each fMRI run, and 45 s resting time between runs. To match their SNR, runs with TSL = 300 ms is averaged ~40% more than that with TSL = 200 ms.

Results and discussions

Fig. A shows the T_1-weighted anatomical image of the cat visual cortex where the gray matter is outlined in green. When an echo planar image is acquired with a long SL preparation of 350 ms, the parenchyma signal dropped to less than 1% before contrast, whereas a significant portion (~50%) of the CSF signal remained (Fig. B). The image intensity is thus proportional to the CSF volume fraction. A CSF volume fraction map can consequently be obtained with proper normalization process (data not shown), without the necessity of multiple-compartment data fitting. This effective tissue-nulling method utilizes the large difference between the $T_{1\rho}$ of tissue and CSF water (50 vs. ~450 ms) therefore has good sensitivity. As comparison, the differences at 9.4 T in the T_1 (2 vs. 4.3 s), T_2 (40 vs. 180 ms), and the water diffusion coefficient (0.75 vs. 2.5 x 10^{-3} mm^2/s) are much smaller.

In contrast to conventional BOLD functional map that shows increase of signal mainly located at the parenchyma, with a long SL preparation the parenchyma is suppressed, and the MR signal decreases at the boundary of the parenchyma/CSF interface (Figs. C and D). The activation pattern and the number of activation pixels were very similar for the two maps. Note some pixels show signal increase (red to yellow pixels), which is probably caused by the inflow effect from spins outside the surface coil coverage. From a cortical surface ROI (Fig. E, inset), an averaged 2.5% decrease of the signal was observed during a 40-s stimulation, and the time courses show that the signal changes are nearly independent of the TSL/TE value (Fig. E), suggesting that the observed signal change is dominated by the change of V_{csf}, and the residue $R_{1\rho}$ and R_2 contribution would be small. The averaged baseline V_{csf} is ~24% for this surface ROI, i.e. the CSF compartment experiences an absolute decrease of 0.6% in pixel volume.

Volume redistribution between parenchyma (including tissue and blood) and CSF may have important implication to fMRI signal depending on the relative weighting of the parenchyma and CSF signal in a voxel, and thus also on experimental technique and parameters. In a previous multi-TE spin-echo fMRI study with similar parameters as in this work [4], a percentage signal decrease of 0.27% was obtained at the cortical surface area when extrapolated to TE = 0, which can mainly be attributed to this volume redistribution effect. Generally, this effect of volume redistribution may be negligible in conventional gradient-echo BOLD experiments with very short TR that suppresses the CSF, but may become significant when the tissue signal is suppressed as compared to CSF, e.g. in the case of $T_{1\rho}$ or VASO fMRI.



Figures: (A) The T_1-weighted image. (B) The $T_{1\rho}$ weighted image with TSL = 350 ms. (C and D) The percentage signal change maps of a representative animal obtained with TSL/TE = 200/15 and 300/22.5 ms, respectively. (E) The time courses of the signal measured with TSL = 200 and 300 ms were obtained from the ROI shown in the Inset.

Acknowledgments: This work is supported by EB008717, EB003324, EB003375, and NS44589.
References: [1] Piechnik SK et al., MRM 61:579 (2009). [2]. Scouten A et al., MRM 58:308 (2008). [3] Grohn HI et al., MRM 54 :19 (2005). [4] Jin T et al., MRM 55 :1281 (2006)