Quantification of CBV Changes During Brain Activation Using IR Methods: A Model Based on Fast Water Exchange and Non-equilibrium Flow Effect

C-W. Wu[1], H-L. Liu[2], J-H. Chen[1]

[1]Dept. Electrical Engineering, Interdisciplinary MRI/MRS Lab, National Taiwan University, Taipei, Taiwan, [2]Dept. Medical Technology, Chang Gung University; MRI Center, Chang Gung Memorial Hospital, Kweishan, Taiwan

Synopsis

Negative response has been observed by non-slice-selective inversion recovery (NSIR) sequence during neural activation and the phenomenon was speculated to be relative to variation of cerebral blood volume (CBV) fraction. However, the quantification of CBV changes using this method is still hindered by the complicated biophysical implications in the signal. In this study, we introduce a model for NSIR signal based on fast water exchange and non-equilibrium flow effect. Simulations are performed to examine these effects on the estimation of CBV changes at both steady and transient state.

Introduction

Recent studies showed that the NSIR sequence with specified inversion time (TI) that suppress blood signals can be used to observe the regional dynamic CBV changes during brain activation [1]. However, estimated CBV change from NSIR was larger than observed CBV change by contrast enhanced MR techniques [2]. It was believed that CBV change may be overestimated without taking cerebral blood flow (CBF) effect in consideration. Fast water exchange effect, flow effect, and cerebrospinal fluid (CSF) effect were adopted to provide a more completed model for evaluating the CBV change in this study. Simulations were proposed to evaluate the influence of CBF, CSF to CBV variation during brain functional activities.

Methods

Bloch equation in consideration of flow effect is described as: $\frac{dM(t)}{dt} = \frac{M_0 - M(t)}{T_1^{mix}} + f_{in} \cdot M_a - f_{out} \cdot M_v$, where M is the voxel magnetization, T_1^{mix} is mixed T_1 containing CSF, blood, and tissue compartments, M_a and M_v represents for inflow and outflow spin density [3]. Since the water exchange rate in human brain is faster than the difference between CSF, blood and tissue relaxation rates, voxel T1 (T_1^{mix}) can be assumed to be a function of CBV: $1/T_1^{mix} = v/T_1^{blood} + (1-v_{CSF}-v)/T_1^{tissue} + v_{CSF}/T_1^{CSF}$ [4], where v is CBV fraction and v_{CSF} is CSF fraction in the voxel. Another assumption is that blood inflow is equal to outflow in steady state. The third assumption is in NSIR sequence, inflow spin density is from arteriole blood, which relaxed according to T_1^{blood} during inversion time and outflow spin density is proportional to tissue magnetization due to fast water exchange. Therefore, the solution of Bloch equation for NSIR signal is modified as:

$$M_{NSIR}(TI) = M_0 \cdot \left\{ \left(T_1^* \cdot \left(\frac{1}{T^{mix}_1} + \frac{f}{\lambda} \right) \cdot \left(1 - e^{-TI/T_1^*} \right) - e^{-TI/T_1^*} \right) - 2 \cdot \left(\frac{f}{\lambda} \right) \cdot \left(\frac{e^{-\frac{TI}{T_1^*}} - e^{-\frac{TI}{T_1^{blood}}}}{\frac{1}{T_1^{blood}} - \frac{1}{T_1^*}} \right) \right\} \quad [1]$$

$$\frac{1}{T_1^*} = \frac{v}{T_1^{blood}} + (1 - v_{CSF} - v) \cdot \left(\frac{1}{T_1^{tissue}} + \frac{f}{\lambda} \right) + \frac{v_{CSF}}{T_1^{CSF}} \quad [2]$$

from Eq.[1] and Eq.[2], the CBV fraction can be iteratively computed with known NSIR signal change and CBF change. In our simulation, we used well-accepted physiological and mechanical parameters under 1.5T magnetic field into the fast water exchange model (FWE model): rest CBV fraction in a voxel is equal to 5%, rest CBF is 60 ml/100g-min, CBF change is 70%, λ is 0.9 ml/g, T_1^{blood} is 1.35 sec [1], T_1^{tissue} is 0.88 sec, TI is 0.92 sec to suppress blood signal, and TR >> T_1^{mix}. CSF effect was simulated under three conditions: 1. there is no CSF exists in the voxel, 2. CSF has constant 5% volume fraction in the voxel, and 3. CSF has 5% volume fraction which decreases during vasodilatation while tissue volume remains constant. To enhance the influence of flow effect in NSIR sequence, VASO model, which assumed that there is no relationship between NSIR signal and CBF, was also simulated in comparison with FWE model.

Results

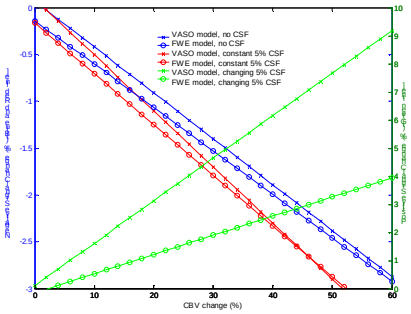

Figure. 1: The relationship between NSIR signal change and the CBV alteration.

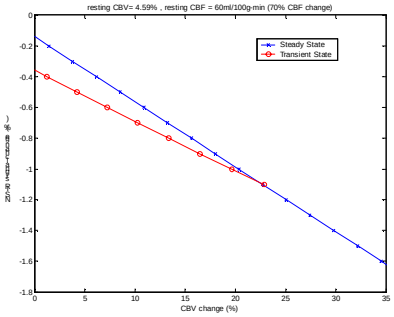

Figure. 2: CBV estimation from FWE model of steady state and transient state.

Fig. 1 shows the relationship between NSIR signal change and CBV change in steady state both for FWE model (circle line) and VASO model (cross line). The figure implies that VASO model may overestimate CBV change by 2.45% when -1.7% signal change is observed. Considering CSF effects, difference of CBV change was observed to be 6.7% between first and second CSF effects (blue and red lines respectively). The third CSF effect (green lines) is unlikely to be true because negative signal change was observed. Besides, flow effect at transient state was simulated. According to balloon model [5], outflow is a function of CBV rather than proportional to inflow. Thus the function: $f_{out}/f_0 = (v/v_0)^{1/\alpha}$ was used to represent outflow function which contains CBV information, where f_0 is the rest cerebral blood flow, v is CBV, and α is the constant of 0.38. Fig. 2 was derived to make comparison between transient (circle line) and steady state (cross line) and it shows that CBV change may be overestimated when the observed NSIR signal change is less than -1.08%.

Conclusion & Discussions

In this study, we provide FWE model for CBV estimation with NSIR sequence. Fast water exchange and CBF influence CBV change estimation slightly comparing to VASO model while CSF effect may be the determinant factor. Though the third CSF effect is not consistent with observed NSIR signal, both second and third CSF effects contribute to the signal change in real case. Furthermore, simulated FWE model at transient state speculates that overestimation of CBV may occur, which leads to the fact that CBV delay in time course may be steeper than the variation of NSIR signal change. Nevertheless, one problem of FWE model is that the selection of rest physiological parameters. We chose the most common values to construct the model, but it may not be suitable for all voxel in an image neither for all subjects. Further studies are needed to prove the accuracy and specificity of the constructed model.

References

[1]. Lu H et al., Magn Reson Med 2003; 50; 263-274 [2]. Francis ST et al., Magn Reson Med 2003; 50; 483-492 [3]. Kwong KK et al., Magn Reson Med 1995; 34; 878-887 [4]. Donahue KM et al., Magn Reson Med 1994; 32; 66-76 [5]. Buxton RB et al., Magn Reson Med 1998; 39; 855-864