Robust $T_1$ mapping in the presence of partial volume effects

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Introduction: The longitudinal relaxation time ($T_1$) of tissues plays an important role both as a contrast mechanism for anatomical imaging and as a disease marker. To date, several techniques have been proposed for measuring it (e.g. [1,2]). Most of them are based on modeling monoexponential behavior for individual voxels. Even though brain tissue is considered to exhibit monoexponential $T_1$ behavior, deviations from this are expected both due to partial volume effects and tissue changes due to disease. This is bound to lead to errors in $T_1$ estimation unless multiple exponentials are taken into account. However, fitting multiple exponentials is well-known to be a mathematically ill-defined problem [3], requiring high signal to noise ratio.

This study addresses partial volume problems when a voxel contains both cerebrospinal fluid (CSF) and another tissue type, as it occurs in the cortical ribbon and periventricular white matter. With a $T_1$ value in the order of 4400 ms at 3.0T, CSF interference can cause significant errors in $T_1$ estimation. Here, a new pulse sequence is presented that allows measuring $T_1$ while keeping the signal from substances with CSF-like $T_1$ values suppressed, thus minimizing interference and estimation error.

Theory & Methods: The proposed sequence is shown in figure 1. It starts with a saturation pulse, followed by an inversion pulse at time $t_a$ and an imaging 90º pulse at time $t_p$. Single-shot echo planar imaging (EPI) is chosen for the image acquisition.

Based on the Bloch equations, signal amplitude after the imaging pulse can be calculated as:

$$\text{Signal} = M_0 \left[ 1 + \left( 1 - 2 e^{t_p/T_1} \right) e^{-t_p/T_1} \right]$$

(1)

If we assume that the CSF signal can be represented by an average $T_1=\overline{T_1}_{\text{CSF}}$, then it is possible to suppress the CSF signal if we choose $t_b = \overline{T_1}_{\text{CSF}} \ln \left( 2 e^{t_p/T_1_{\text{CSF}}} - 1 \right)$

(2)

Hence, for every $t_a$ there is a $t_b$ that keeps the CSF signal suppressed. By acquiring the signal values for different values of $(t_a,t_b)$ it is possible to fit Eq. 1 to the experimental data so as to obtain the $T_1$ value of the tissue under examination.

Selection of a set of sampling points $(t_a,t_b)$ was the result of an optimization process. A cost function was created by modeling the fitting process based on a D-optimum design [5], which was extended into a Bayesian model in order to cover a $T_1$ range of interest between 900 and 2100 ms. The optimization process was repeated for different numbers of sampling points for a total sequence duration of 3 minutes.

Monte Carlo simulations were used to assess the stability of the fitting scheme over a range of $T_1$ values, both for the target tissue as well as for the “interfering” tissue. The proposed sequence was implemented on a 3.0 T GE scanner. Adiabatic FOCI pulses were used for the inversion pulses. Saturation pulses were implemented as a series of two 90º pulses separated and followed by spoiler gradient pulses in order to minimize effects of transmit RF variations.

Results: The optimization process showed that gains from additional sampling points leveled at approximately 35 points (figure 2); a 40-point implementation was selected as a trade-off between signal-to-noise ratio and implementation complexity. The combination of the time constrain and the Bayesian nature of the cost function made the final design deviate from the two-point optimal designs for exponential fitting, giving the following set of $t_a$: [1285, 1374, 1380, 1474, 1487, 1499, 1512, 1534, 1557, 1558, 1565, 1565, 1570, 1583, 1596, 1606, 1620, 1648, 1648, 1664, 1687, 1690, 1692, 1706, 1715, 1720, 1733, 1736, 1777, 1798, 1924, 2296, 9021, 9404, 9593, 9717, 9786, 10087, 10114] ms. Figures 3 and 4 show the precision and the accuracy of the proposed method compared to an inversion recovery (IR)-based technique with 12 exponentially-spaced fitting points. In terms of precision, the proposed sequence is comparable up to 1500 ms and acceptable up to 2100 ms; however in terms of accuracy it clearly outperforms single-exponential IR fitting even if misestimating the interfering $T_1$.

Discussion & Conclusion: Partial volume effects can present a serious problem in $T_1$ measurement in the interfaces between brain tissue and CSF, especially in the cortex and the periventricular regions. As shown, $T_1$ estimation using single exponential fitting of the IR curve shows significant error even for a moderate percentage of partial volume contamination. The presented method gives a reliable measurement over a range of $T_1$ values normally expected in the human brain. It provides an easy way to minimize partial volume induced error if there is an estimation for the $T_1$ of the second tissue type.

Acknowledgement: This research was supported by the Intramural Research Program of the NIH, NINDS.

References