Robust T2 Measurements for Multi-TI Arterial Spin Labeling

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Introduction: Precise arterial spin labeling (ASL) T2 measurements are challenging because they are influenced by many factors like the choice of crusher gradients, the refocusing flip angle, and the number of fitted echoes. When extended to multi-TI T2 measurements, SNR at longer inflow times can be critical, especially if scan time limits the number of image averages. Here, a robust and reliable multi-TI T2 acquisition and fitting routine is presented that is fast enough to be implemented in clinical routine to determine T2 values for every individual patient. Later on the T2 values could be incorporated in recently presented two-compartment models for permeability quantification [1,2,3].

Material and Methods: Six healthy male volunteers participated in the study (age: 30-50 years). Written informed consent was obtained from all volunteers. The study was performed on a 3T scanner (Siemens Magnetom Verio) with a vendor provided 20 elements head coil.

A T1 weighted gradient echo sequence was obtained to generate masks for grey and white matter. Segmentation of grey and white matter was done in MeVisLab [4]. A voxel was assigned to grey or white matter when the probability for the particular case was 50% or higher.

For ASL imaging, an optimized FAIR PASL pulse scheme including background suppression with post labeling saturation (for T1 of 700 ms and 3500 ms) and Q2TIPS was used, in combination with a 3D-GRASE readout [5] with a spatial resolution of 6.0 x 6.0 x 6.0 mm. Images at six different TI were obtained (800 ms to 3300 ms with 500 ms spacing). The sequence was run twice. The first time with turbo factor (TF)1, imaging 12 contrasts, and the second time with TF3 and 8 contrasts. The refocusing flip angles were set to 180°. In both cases, eight partitions were encoded by using Partial Fourier of 6/8 and adaptive averaging [6] meaning that the higher inflow times are imaged more often than shorter ones. The distribution of the averages at the different inflow times was: 1, 2, 2, 3, 4, 6. The maximal bolus length was set to 1400 ms. The echo times and bandwidths were: TF1: first TE 20.49 ms, increment 20.49 ms, BW: 3551 Hz/Px; TF3: first TE 20.16 ms, increment 60.48 ms, BW: 3005 Hz/Px.

On the masked non selective, slice selective and perfusion weighted images, a voxel wise exponential fit was applied with the fitting routines of MPFIT [7] in MeVisLab. For TF1 five echoes were fitted in which the first echo was always excluded from the fit. This fitting approach was chosen because simulations showed that the first echo is mostly influenced by imperfect refocusing flip angels. For TF3 three echoes were fitted. Here, the first echo was included in the fit.

T2 values between 10 ms and 240 ms with a relative error below 20% were considered. The minimum threshold for signal intensity to apply a T2 fit was set to 10. The mean T2 of all volunteers of the perfusion weighted images was only calculated in the grey matter region. In white matter, the signal was not always high enough for a reliable analysis and might in several regions be confounded with partial volume effects of grey matter.

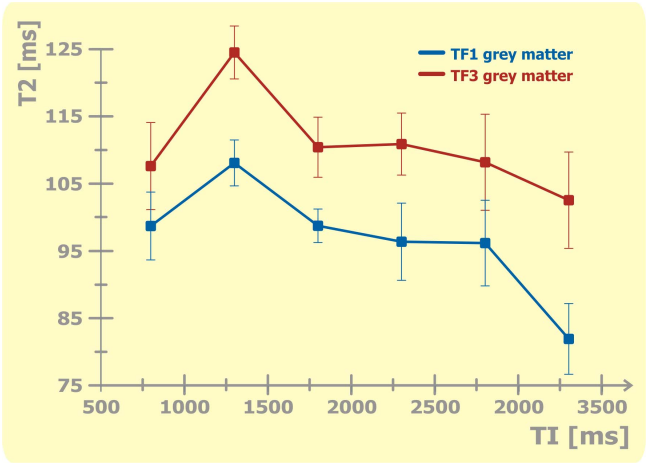

Figure 1: Fitted T2 values for grey matter of the perfusion weighted images at several inflow times for TF1 and TF3. For TF1, values for five fitted echoes are displayed, for TF3, values with three fitted echoes are shown. The error bars indicate the standard deviation among volunteers.

Results: Figure 1 shows the mean grey matter T2 values for TF1 und TF3 of the perfusion weighted images, at the different inflow times for all volunteers. For TF1 the T2 values decrease with longer inflow time and lie between 80 ms and 110 ms. Just like for TF1, with TF3 the T2 values fall with longer inflow time. They lie between 92 ms and 128 ms. Comparing TF1 and TF3 reveals higher T2 values for TF3. The percental ratio of these T2 values is smallest for 800 ms, varies between 10 and 14 for inflow times from 1300 ms to 2800 ms, and has a value of 20 at the longest inflow time. For one representative volunteer, perfusion weighted T2 maps at four different inflow times obtained with TF1 are displayed in figure 2.

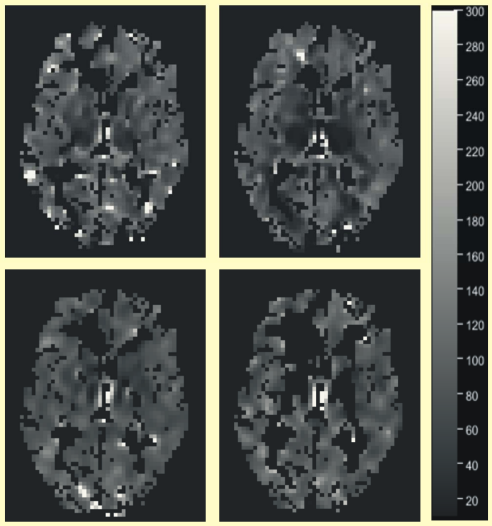

Figure 2: Perfusion weighted T2 maps of one representative volunteer at the different inflow times. Top: 1300 ms and 1800 ms. Bottom: 2300 ms and 2800 ms. Grey and white matter were fitted.

Discussion and Conclusion: As indicated by the standard deviation (error bars in figure 1), the T2 values of the different volunteers do not deviate much, underlining the robustness of the acquisition.

Figure 2 shows that the number of voxels with a good T2 fit is decreasing, especially in white matter and for higher inflow time. Further averaging would be beneficial. Independent of the turbo factor, the values for the longest inflow time are most critical. Even though TF3 reveals higher T2 values than TF1, the general decreasing development of T2 over TI is the same for both TFs. The values at 800 ms might be reduced due to inflow artifacts and should be taken with precaution. For TF3 the first three spin echoes are used to encode the first contrast. If the acquisition scheme is adjusted by using the first spin echo to encode a non central partition for TF3, the difference of the T2 values obtained with TF1 and TF3 will most probably decrease. If always the same scan protocols are used, the obtained values will be comparable although a difference among the T2 values persists.

It was shown that robust multi-TI T2 measurements are possible in a reasonable scan time. With TF1, the acquisition took less than 13 minutes and with TF3 less than five minutes. The T2 values were determined for every inflow time and could be directly incorporated in a two compartment model for perfusion quantification.

References: [1] Gregori et al., 2009 Assessment of Blood-Brain Water Transfer by Arterial Spin Labeling Based T2 Measurements. Proceedings: 17th Scientific Meeting ISMRM; [2] Gregori et al. 2012, T2-based Arterial Spin Labeling measurements of blood to tissue water transfer in human brain, JMRI [3] Liu et al., Magn Reson Med., 65(1):120–127. [4] Ritter et al., 2011. IEEE pulse, 2(6):60–70. [5] Günther et al., 2005. Magn Reson Med, 54(2):491–498. [6] Kramme, J et al. 2011. Adaptive averaging improves the Signal to Noise Ration in ASL experiments especially at high inflow times. Proceedings: 19th Scientific Meeting ISMRM. [7] Markwardt, 2009. Non-linear Least Squares Fitting in IDL with MPFIT. Proceedings: Astronomical Data Analysis Software and Systems XVIII, 2008 Quebec.