Measurement of multi-slice cerebral blood flow with T1-normalized arterial spin labeling MRI using a volume RF labeling coil
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Introduction Arterial spin labeling (ASL) MRI tags arterial spins as endogenous tracer and the signal change is sensitive to CBF\(^1\). However, the RF tagging pulse may also attenuate tissue signal through magnetization transfer (MT), direct RF saturation and off-resonance spin locking effects, particularly when a volume RF coil is used. Importantly, ASL MRI contrast scales with \(T_{1app}\) the apparent \(T_1\) under the RF tagging pulse, which varies with its amplitude and offset \(^{4,5}\). Nevertheless, a single \(T_{1app}\) value is often used for CBF calculation. As \(T_{1app}\) map is heterogeneous, such a global \(T_{1app}\) approach may be oversimplified for mapping regional CBF.

Materials and Methods Animal experiments were carried out in accordance with institutional guidelines. MRI was obtained at 4.7 Tesla. We obtained single slice \(T_1\) (slice thickness 3mm) and \(T_1\) MRI with an inversion recovery sequence using identical recovery time in 10 adult male Wistar rats. We had \(B_1=4.7\) μT, a labeling distance of 15 mm (\(\Delta\omega=10,000\) Hz), modulation frequency of 250 Hz, and post-labeling duration of 300 ms. In addition, CBF was obtained with amplitude modulated (AM)-ASL (TR/TE=6,500ms/28ms, NA=32) in two Wistar rats\(^6\) in two Wistar rats\(^6\-7\). CBF was calculated as CBF=\(\lambda\cdot(I_{ref}-I_{tag})/(2\alpha I_0)\cdot C\), where \(I_{ref}\) and \(I_{tag}\) are image intensities when RF labeling and reference pulses are applied, respectively, and \(I_0\) is control image without RF irradiation. In addition, \(\lambda\) is the brain/blood partition coefficient, \(\alpha\) is the inversion efficiency. In addition, \(C=a e^{\delta/T_1}\cdot T_{1app}\), and \(\lambda=0.9\) ml/g and \(\alpha=0.65\).

Results and Discussion Fig. 1 shows single-slice \(T_1\), \(T_{1app}\) and \(T_{1app}/T_1\) maps of a representative normal animal. \(T_1\) and \(T_{1app}\) maps were heterogeneous, being 1.56 ± 0.15 s and 0.83 ± 0.09 s (\(B_1=4.7\) μT and \(\Delta\omega=10\) kHz), respectively. Despite their spatial heterogeneity, the parametric \(T_{1app}/T_1\) map was reasonably homogeneous, being 0.53 ±0.02. Ventricle appeared hyperintense, likely caused by cerebral spinal fluid (CSF) partial volume effect. We found \(T_1\), \(T_{1app}\) and \(T_{1app}/T_1\) to be 1.55 ± 0.03 s, 0.84 ± 0.01 s and 0.54 ± 0.01, respectively (\(n=10\)).

Fig. 2 compares the single-slice CBF map calculated from \(T_{1app}\) map, single \(T_{1app}\) value (\(T_{1app}=0.84\) s) and the scaled \(T_1\) map (\(T_{1app}=\eta\cdot T_1\), with \(\eta=0.54\)), respectively. In addition, CBF\(^*\) calculated from a single \(T_{1app}\) value and \(T_1\) map were found to be 0.98 ± 0.31 and 1.02 ± 0.31 ml/g.min, respectively. Whereas the CBF values were reasonably close to one another, subtle difference in the CBF map can be detected. This is because by using the mean \(T_{1app}\), CBF calculated from the single \(T_{1app}\) value can approximate the mean CBF measurement from the \(T_{1app}\) map and scaled \(T_1\) map, it may not fully account for \(T_{1app}\) heterogeneity-induced regional CBF difference. This can be better appreciated in Fig. 3, which is an overlaid scatter plot of CBF and CBF\(^*\), per voxel. The proposed \(T_1\)-map-normalized CBF\(^*\) closely correlated with the \(T_{1app}\)-map-normalized CBF (black square), where CBF\(^*\)=0.92 CBF + 0.04 ml/g.min and the coefficient of determination \(R^2\) is 0.94. In comparison, the correlation between CBF\(^*\) calculated using the single \(T_{1app}\) value and CBF (red circle) was CBF\(^*\)=0.81 CBF + 0.12 ml/g.min, \(R^2=0.71\).

The proposed approach can be extended to study multi-slice CBF. Due to concomitant RF irradiation effects, \(T_{1app}/T_1\) depends on RF offset and power. We serially varied RF offset and power and showed that \(T_{1app}/T_1\) as a function of RF irradiation offset can be described by linear regression relationship. This allows us to scale multi-slice \(T_1\) map with pre-determined coefficient to derive \(T_{1app}\) map for multi-slice CBF calculation using AM-ASL MRI (Fig. 4). Such an approach is applicable for studying heterogeneous neurological disorders such as acute stroke.