Multi-parameter mapping of the human brain at 1mm resolution in less than 20 minutes

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Introduction:

So called "structural MRI" of the human brain reflects not only the individual morphology, but also the intrinsic properties of tissue water related to the imaging process. The latter may more directly reflect changes in brain development and pathology. The goal of this study was to devise an imaging protocol for obtaining the major contrast parameters in a clinically feasible time at 1 mm isotropic resolution. We present a 20 minutes long protocol using multi-echo 3D FLASH for mapping proton density (PD), T1, magnetization transfer (MT) and T2* at 1mm resolution.

Methods:

Principal design strategy

The proposed imaging protocol consists of three 3D FLASH acquisitions, since spoiled gradient echoes combine robustness with a high signal-to-noise-ratio (SNR). Maps of PD and T1 can be obtained by a dual-angle approach from PD-w(eighted) and T1-w volumes. These serve as references to correct the MT-w FLASH acquisition to obtain the MT-ratio (MTR) and the T1-corrected MT map (**1**). Higher SNR is achieved by increasing TR and thus the time fraction of signal acquisition, until the gain in steady state is used up by T2* losses. Splitting the acquisition into multiple echoes of higher bandwidth (BW) allows for mapping of T2* and B0, while averaging restores most of the SNR for mapping T1, PD, and MT (**2**).

Timing and system constraints

The goal was to achieve 1 mm resolution within 20 minutes. The repetitions were reduced by more than 60% using partial parallel acquisition (2x; to avoid g-map losses) in phase (AP) and 6/8 partial Fourier in partition direction (LR). An echo train length of 20 ms was chosen to trade off emerging T2* contrast (**3**) against shim-induced losses, at the same time providing T2* information and high SNR. For a volume of 256x224x176 pixels, the maximum number of echoes was limited by the image reconstruction system. The BW was 425 Hz/Px to avoid susceptibility-related distortions. Since the MT-w experiment includes an additional off-resonance Gaussian-shaped RF pulse and must be run with the same TR as the PD-w experiment to calculate MTR, 6 instead of 8 echoes were acquired. A rather short off-resonant MT-pulse was favored over a binomial design as the latter would interfere with local B0 offsets. Based on physical variation in the subjects, the power of the MT-pulse was chosen to keep heat deposition below the 75% SAR limit. The T1-w experiment was run at shorter TR as determined by the 6 echoes averaged to increase SNR. The flip angles were optimized empirically for high SNR in the parameter maps. The frequency of the MT-pulse was increased to reduce direct saturation effects.

Implementation

3D multi-echo FLASH MRI with non-selective excitation was performed at 3 Tesla (Siemens Magnetom TIM Trio; 12-channel receive head coil) on healthy adult volunteers. Magnitude and phase images were acquired at PD-w (TR= 23.7 ms, $\alpha = 6°$) from eight bipolar gradient echoes at equidistant TE between 2.2 ms and 19.7 ms (425 Hz/pixel BW). These yielded maps of B0 and T2* related to "shim" and susceptibility. Maps of PD, T1, and MT were obtained from two additional experiments: One with additional MT-w by a 4 ms Gaussian-shaped RF pulse (220° nominal flip angle, 2 kHz frequency offset) and one with predominant T1-w (6 echoes, TR = 18.7 ms, $\alpha = 20°$). Data acquisition took 6:47 and 5:21 minutes, respectively.

Processing routines were developed for use with SPM5 and FSL:

Linear regression of the log signals yielded T2*. To increase SNR, the first 6 echoes were averaged. Then, T1 and amplitude A were calculated from the PD-w and T1-w experiments by means of a rational approximation of the FLASH signal (**4**):

$$S = A\,\alpha\,TR/T1\,/\,[TR/T1 + \alpha^2/2]. \quad [1]$$

In the approximate signal equation of the MT-FLASH experiment:

$$S = A\,\alpha\,TR/T1\,/\,[TR/T1 + \alpha^2/2 + MT], \quad [2]$$

MT represents the additional percentage saturation of Mz due to a single MT pulse. It is calculated from Eq. [2] using A and T1 obtained from Eq. [1]. Since MT is approx. proportional to $B1^2$, receive and transmit profiles cancel out in the MT-map. Unlike the MTR, MT is not affected by T1 relaxation (**5**).

Results:

MT-maps provided a high contrast between CSF/GM/WM that was hardly affected by spatial variation. Accordingly, the histogram displayed three clearly separated modes across the whole brain (**Fig. 1**). In particular, the occipital, orbitofrontal and temporobasal cortex was well delineated. Due to the high BW, unwarping of odd and even echoes was not needed. The amplitude and T1 maps had a higher SNR than MT, but showed slight spatial heterogeneities due to residual B1 effects (**Fig. 2**). Maps of 1/T2* revealed the iron-containing nuclei in midbrain (**Fig. 2**, not visible on T1-w images) besides larger vessels and connective tissue.

Discussion:

The proposed method provides all standard contrast parameters (except T2) in clinically feasible time with minimal spatial distortions. It may be readily extended by mapping the flip angle and receive profiles to yield fully quantitative maps of T1 and PD (**4**). The amplitude maps can be extrapolated to the PD-signal at zero TE. A similar method has been proposed for T2 mapping (**6**). In particular, the MT and T2* maps appear to be valuable for visual diagnostics. Beyond this, MT maps promise to improve tissue segmentation. It is believed that quantitative changes of contrast parameters provide sensitive and specific information on changes in pathology and normal development; even in the absence of or prior to morphological alterations.

References:

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Acknowledgements:

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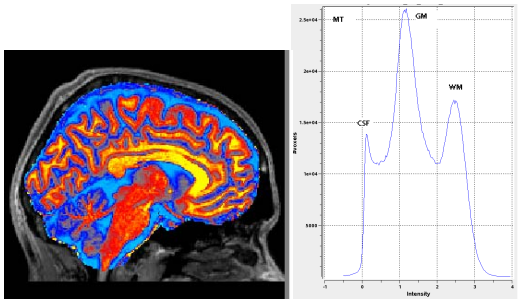

Fig. 1: Pseudocolor MT map and histogram of MT values. GM = gray, WM = yellow, CSF = light blue

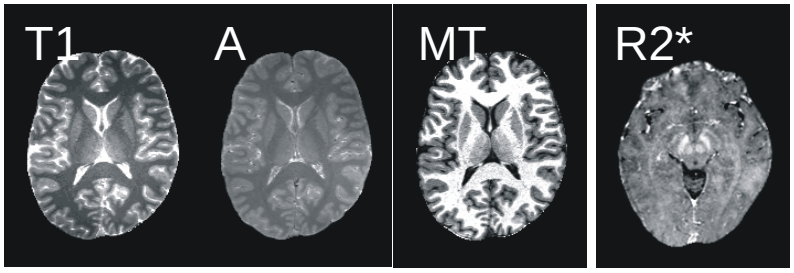

Fig. 2: Maps of T1, A, MT, and R2* (=1/T2*, different slice position).