McMPRAGE (multi-contrast MPRAGE): a novel sequence for generating multiple contrast images in a single scan

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Target audience: MR physicists, neuroradiologists, engineers involved in neuro image processing/segmentation

Purpose: Magnetization-prepared rapid gradient echo (MPRAGE) is an integral part of most neuroimaging protocols. A single, fixed inversion time is typically chosen to null cerebral spinal fluid (CSF). Recently, white-matter (WM) nulled MPRAGE has been reported for improved visualization of multiple sclerosis lesions [1] and thalamic nuclei [2]. Nulling the WM-GM interface has also been proposed for better brain segmentation. We report here on the development of a new multi-contrast MPRAGE sequence (McMPRAGE), which uses novel *k*-space sampling and view sharing to generate multiple image contrasts in a single, short scan. This sequence was optimized and tested on patients and healthy subjects at 7T.

Materials and methods: Sampling scheme- After skipping the corners of k-space, the elliptical region is divided into radial sectors, with the number of points per sector dependent on the desired "temporal resolution" i.e. spacing along the time (TI) axis of the desired phases. Each radial sector is further divided into a central A region and N interleaved peripheral B_i regions (Fig 1). Division into N B_i regions is achieved by first sorting the points according to increasing k_r ($= \sqrt{(k_y^2+k_z^2)}$) and selecting every N^{th} point, offsetting the starting point by 1 for each B_i region. Starting offsets are dithered across radial sectors to minimize clumping and produce an incoherent point spread function (PSF). The acquisition scheme is shown in Fig 2: following the IR module, regions $AB_1AB_2..AB_N$ are repeated [M/N] times, where M is the number of desired phases. A "phase" is reconstructed for each AB_i by view copying B_j ($j \neq i$) from the neighboring phases. The location of the A region along the time axis determines image contrast for that phase. Note that view copying is performed in data acquisition memory during the recovery period TD, eliminating any latency and requiring no change in image reconstruction software. Simulations- The MPRAGE signal equation was modeled in MATLAB and TI_0 adjusted until desired contrast was achieved in each phase. A synthetic phantom with multiple tissue compartments was used to simulate the effects of view sharing on image quality. Experiments- After informed consent, patients and healthy subjects were scanned on a 7T scanner (Discovery MR950, GE Healthcare) using a 32-channel receive array coil (Nova Medical) and the following scan parameters: TS/TI/TR/TE 2000/500/5.8/2.4ms, BW 25 kHz, 9º flip, 18 cm FOV, 160x160x160 matrix, 1.1 mm thick, 2X ARC parallel imaging, 7.5 min scan time.

Results: Based on simulations, N=3 B_i regions each with 12 points, and an A region with 11 points optimizes image quality (i.e. each phase is 23TR ~125 ms) The *k*-space signal weighting for a white-matter-nulled phase (TI=500ms) and the corresponding PSF are shown in Fig 3 for the dithered interleaved (a) as well as sequential case (c). Despite the modulation across *k*-space due to signal recovery from the null point (blue to red), the randomized nature of B_i minimizes any PSF coherence, which could result in streaking artifacts. Fig 4 shows M=6 phases obtained using McMPRAGE from a patient with Alzheimer's disease at 7T. TI_0 was chosen to produce a WM-null at phase 1 (ideal for thalamic nuclei segmentation), WM-GM interface null at phase 2, GM null at phase 3, GM-CSF interface null at phase 4 and CSF-null at phase 5. All 6 phases were acquired in 7.5 min.

Conclusion: We have developed a fast, fully Cartesian-sampled method for generating multi-contrast MPRAGE images in short scan times. Unlike previously proposed schemes with 3D radial acquisition [3], our method is robust to system imperfections, easily accelerated with parallel imaging and can be used on existing systems without changes in reconstruction. The multiple images can also be fitted to generate T_1 maps. The different contrast images are also inherently co-registered due to the interleaved acquisition, eliminating the need for registration when processing. This can be useful, for example, in resting state fMRI, with the CSF-nulled images used for registration to MNI space and the WM-nulled images used for single thalamic nucleus seed voxel placement.

References: [1] Bluestein KT et al. *MRI* 2012 [2] Tourdias T et al. *Neuroimage* 2013 [3] Kecskemeti et al. ISMRM 2013 (p452)

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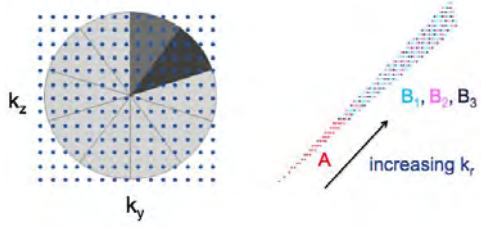

Fig 1. *k*-space divided into radial sectors (left) and each sector (right) divided into central A and peripheral B_i regions. Acceleration is achieved by sharing the B_i regions within each sector.

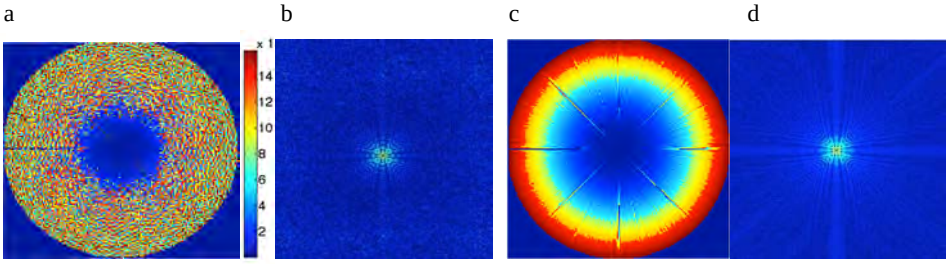

Fig 3. *k*-space signal weighting for a white-matter-nulled phase (a) and the corresponding PSF (b) for the proposed dithered interleaved scheme and for the sequential scheme where the B_i regions are contiguous (c,d). Note that the dithered interleaving of the B_i regions results in an incoherent PSF while the sequential scheme leads to a more coherent PSF due to the coherent signal modulation in *k*-space.

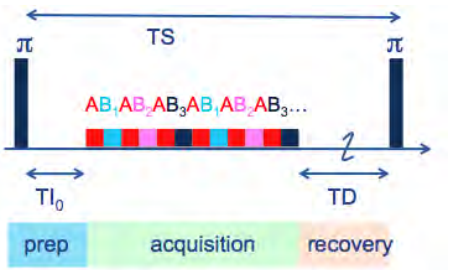

Fig 2. Multi-contrast MPRAGE sequence: following the IR pulse and delay TI_0, M AB_i regions are acquired to sample the recovery curve M times (in this e.g., M=6)

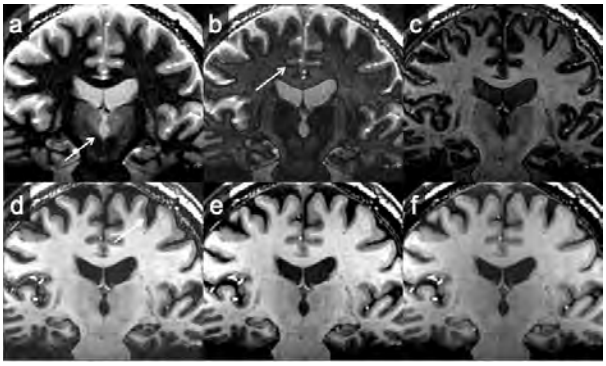

Fig 4. Six consecutive phases of McMPRAGE acquired on a patient with Alzheimer's disease on a 7T scanner in 7.5 min. The first phase (a, WM-nulled) clearly depicts the thalamic nuclei. The second phase nulls the boundary between WM and GM (b, arrow). GM is suppressed in phase 3 (c). The boundary between GM and CSF is nulled in phase 4 (d, arrow). The last two phases are CSF-nulled. Note that the streaking artifacts get progressively smaller in amplitude as the TI increases (a->f)