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Technical note

CNR improvement of MP2RAGE from slice encoding directional acceleration

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ABSTRACT

Purpose: While MP2RAGE shows the potential to generate B_1 insensitive T_1 contrast, the long TR of MP2RAGE (≥6 s at 7 T) is essential to provide the large dynamic range of apparent T_1 relaxation for dual inversion time acquisitions. We present a 2 direction (2D) accelerated MP2RAGE, which provides an increased flip angle while maintaining similar dynamic recovery as 1D accelerated MP2RAGE.
Method: Simulations were conducted to optimize 2D accelerated MP2RAGE parameters and healthy subjects were scanned with 1D and 2D accelerated MP2RAGE at 7 T. Images were compared visually and contrast to noise (CNR) between brain tissues was measured.
Result: Simulations showed that CNR is primarily determined by the TR, followed by the number of the first partition encoding steps in MP2RAGE. Keeping TR constant, a smaller number of partition encoding steps increases the achievable maximal CNR. In-vivo 2D MP2RAGE improves CNR between white and gray matters by 9% when compared to 1D accelerated MP2RAGE with identical voxel size.
Conclusion: We presented 2D accelated MP2RAGE at 7 T with the increased flip angle. We show that this leads to CNR improvement, and consequently a reduction of scan time to be compared to 1D accelerated MP2RAGE.

1. Introduction

Magnetization-prepared rapid gradient-echo (MPRAGE) [1] has been routinely used to provide 3D brain anatomical information based on T1 weighted tissue contrast. MPRAGE generates T_1 weighted contrast through inversion recovery (IR) preparation and segmented 3D gradient recalled echo (GRE) acquisition with appropriate time delay between the preparation and acquisition (TI). Typically, single k_z encoding is completed with linear view order along the k_z direction for each inversion preparation, which is repeated over different k_y values. TI, flip angle of excitation RF pulses (α) and the time gap between adjacent IR pulses (TR) are the key parameters to be optimized for maximum T_1 contrast among white matter (WM), gray matter (GM) and cerebral spinal fluid (CSF) [2]. It is known that MPRAGE is sensitive to B_1 inhomogeneity because the apparent relaxation during the readout ($1/T_1^* = 1/T_1 + \ln(\cos\alpha)/\tau$, where τ is the time gap between adjacent RF pulses) spatially varies, depending on the actual flip angle. At ultra-high magnetic field (UHF; ≥7 T) which involves large B1 variation, the B1-induced intensity variation is undesirable for quantitative analysis, e.g. segmentation of brain tissue types.

Magnetization prepared 2 rapid acquisition gradient echoes (MP2RAGE) [3] can reduce image intensity inhomogeneity due to B_1 variation by acquiring two image volumes with different flip angles and inversion times [3,4]. Fig. 1.A shows the timing diagram of the MP2RAGE sequence. After the IR pulse, the first series of low flip angle (α_1) excited GRE acquisitions are collected to complete a single phase encoding (k_y) over the corresponding N_z partition encoding (k_z) steps, followed by the second series of GRE acquisitions with flip angle (α_2) for the same k-space coverage. The dual GRE acquisition per IR pulse is repeated N_y times to obtain two sets of 3D k-space data. The total scan time for an MP2RAGE is thus N_y times TR for full k-space sampling, and can be reduced by undersampling in the k_y dimension using parallel imaging reconstruction (i.e. k_y-direction acceleration) [3].

Various methods have been proposed to reduce the scan time in MPRAGE-type acquisitions. One common technique is to skip and zero-fill parts of k-space, as in partial Fourier and elliptical sampling schemes. With recent hardware improvements, most importantly multi coil arrays, parallel imaging techniques are increasingly used, making use of the data redundancy when sampling the same object with multiple antennas [5–7]. The original 1D acceleration was extended to accelerate two dimensions [8–10]. Subsequently, the 2D undersampling patterns were optimized to maximize the distance of the point spread functions, known as Controlled Aliasing In Parallel Imaging (CAIPI) [11]. Recently, Brenner et al. proposed accelerated

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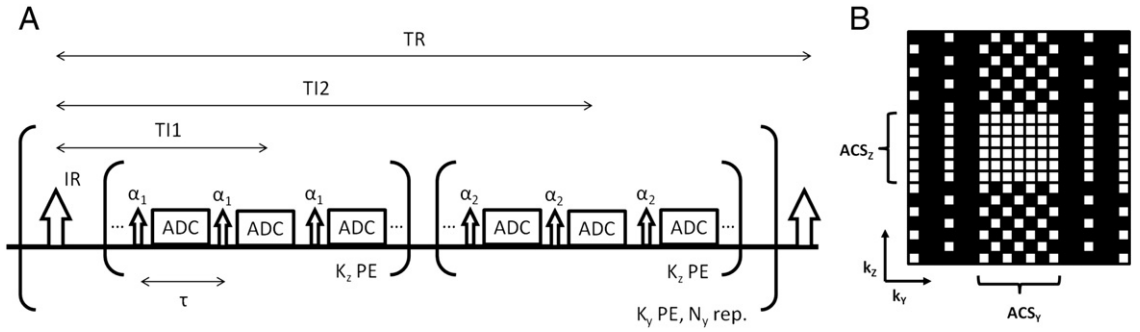

Fig. 1. A sequence diagram of MP2RAGE (A) and the example of the proposed 2D accelerated acquisition paradigm with acceleration (acc.) in k_y (=3) and k_z (=2) directions (B). White dots indicate the sampled k-space locations.

MPRAGE using CAIPI sampling pattern and elliptical acquisition window [12].

In this study, we hypothesize that the number of partition encoding steps acquired for each preparation, or N_z, significantly affects the MP2RAGE signal intensity due to its connection to the apparent T_1 relaxation during the GRE blocks. A simulation was conducted to test the effects of N_z on contrast to noise ratio (CNR) among different brain tissues. Healthy subjects were scanned using MP2RAGE with 1D (k_y direction only) and 2D (k_y and k_z direction) accelerations, and the reconstructed images are compared.

2. Material and methods

2.1. SNR and CNR calculation in MP2RAGE signal intensity

Marques and his colleagues defined the MP2RAGE signal using two different TIs and investigated the noise propagation in the previous study [3]. Here, the mathematical concept of SNR and CNR calculation in MP2RAGE is summarized.

When GRE_{TI1} and GRE_{TI2} are signal intensities at inversion times TI1 and TI2, then the MP2RAGE signal is described as

$$MP2RAGE = GRE^{*}_{TI1} GRE_{TI2} / \left(GRE^{2}_{TI1} + GRE^{2}_{TI2} \right) \quad (1)$$

where * is a complex conjugation operation, which makes the signal scaled between −0.5 and 0.5.

Let x, A, and B represent random variables with variance of σ_x^2, σ_A^2, and σ_B^2, respectively.

If x is function of A and B, and if covariance between A and B is zero, then the general noise propagation equation of x can be expressed using the partial derivatives as,

$$\sigma_x^2 = \sigma_A^2 (\partial x/\partial A)^2 + \sigma_B^2 (\partial x/\partial B)^2 \quad (2)$$

Replacing x in Eq. (2) with a MP2RAGE signal, given in Eq. (1), then subsequently, A and B will be replaced with GRE_{TI1} and GRE_{TI2}. Assuming that σ_{noise} is the standard deviation in real numbers of GRE_{TI1} and GRE_{TI2}, then the noise standard deviation in the MP2RAGE signal is simplified as

$$\sigma_{MP2RAGE} = \sigma_{noise} \sqrt{ \left(GRE^{2}_{TI1} - GRE^{2}_{TI2} \right)^2 / \left(GRE^{2}_{TI1} + GRE^{2}_{TI2} \right)^3 } \quad (3)$$

CNR between tissues can be defined as the dynamic range of two different tissues divided by the sum of squares of noise on the MP2RAGE signal of each tissue. For example, CNR between WM and GM is;

$$CNR_{WG} = \frac{MP2RAGE_{WM} - MP2RAGE_{GM}}{\sqrt{\sigma^2_{WM,MP2RAGE} + \sigma^2_{GM,MP2RAGE}}}. \quad (4)$$

CNR between GM and CSF can be described in the similar way. Note that the previous study [3] employed CNR efficiency, divided by square root of TR, but we used CNR between tissues, not including a time term because the direct comparison of MP2RAGE image at different TR's is also of interest in this study.

2.2. Simulation

A computable Bloch simulation of MP2RAGE signals of WM, GM, and CSF was conducted using sequence parameters modified based on ref. [2,3]. The initial longitudinal magnetization, M_0 was set to 100. The signal intensity was calculated with the longitudinal magnetization at each TI, multiplied by a sine term of the flip angle. The exponential T2 term and the different T2 of each tissue, respectively, were not considered. Flip angles were varied from 1° to 12°, and τ was set to 7 ms. N_z was varied from 80 to 180 in increments of 10. Assuming linear acquisition in the partition encoding step, TI_1/TI_2/TR was increased in 100 ms increments, up to a maximum TR = 7 s. T_1 of WM, GM and CSF at 7 T were assumed to be 1.05, 1.85, and 3.35 s, respectively [3]. $\sigma_{MP2RAGE}$ in Eq. (3) was normalized by σ_{noise} and, CNR between WM and GM (CNR_{WG}) and between GM and CSF (CNR_{GC}) were calculated using Eq. (4). Also, MP2RAGE signals of each tissue were calculated with a B_1 inhomogeneity of ±40%, which was used in [3]. The optimized parameters for MP2RAGE were determined based on three criteria: 1) Screening the parameter sets that generate MP2RAGE signal variation of less than ±0.07 in WM, GM, and CSF with ±40% B_1 inhomogeneity, 2) $MP2RAGE_{WM > GM > CSF}$ and 3) maximizing CNR_{WG} or CNR_{GC}, whichever is smaller.

2.3. Two dimension accelerated MP2RAGE

Fig. 1. A shows the MP2RAGE sequence diagram, as described in the introduction. TI_1 and TI_2 are defined as the time intervals between the IR pulse and the centers of the 1st and 2nd GRE acquisition, respectively. While the acceleration in k_y-direction reduces the scan time, reducing N_z (or k_z-direction acceleration) relaxes the saturation effect of the repetitive readout excitations, thereby leaving room for larger flip angles to obtain higher SNR with similar dynamic range of relaxation recovery as the original N_z. Since MP2RAGE needs to keep a constant TI_1 or TI_2 in each TR, the k_z-direction acceleration is not as flexible as in a GRE sequence