Can Two Wrongs Make A Right? B_1-insensitive T_1-weighted Imaging of the Human Brain at 4.7T using 3D MDEFT with a Standard Non-Adiabatic Preparation Pulse

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Introduction

T_1-weighted MRI of the whole brain is problematic at high field strength due to the increased T_1 values of brain tissue, the reduced T_1 contrast, and the greater B_1 inhomogeneity (due to RF field/sample/coil interactions[1]) relative to low field. This means that image contrast is low and/or non-uniform across the brain if sequences that are commonly employed at low field strength to acquire T_1-weighted images (such as FLASH and MP-RAGE) are used. For this reason, MDEFT has been proposed as a method for achieving high quality T_1-weighted images at high field strength[2,3]. However, while efficiently achieving good contrast, when implemented with fast gradient echo methods and adiabatic preparation pulses[4], MDEFT remains sensitive to variations in the local B_1 field. In this paper, we show that by using a standard non-adiabatic hard RF pulse with a nominal flip angle of 130º as part of the magnetisation preparation, rather than an adiabatic inversion pulse, the combined B_1 sensitivities of this preparation pulse and the subsequent excitation pulses provide auto-compensation for B_1 inhomogeneity, resulting in images with uniform grey matter (GM), white matter (WM) and cerebrospinal fluid (CSF) signal intensity and contrast.

Methods

The standard MDEFT scheme is: $90°_{saturation}$ - τ_1 - $180°_{inversion}$ - τ_2 – [image], where τ_1 and τ_2 are delay times which allow a controlled amount of T_1 contrast. In our implementation, 2-shot centre-out phase encoded spoiled 3D FLASH imaging was used for image acquisition (TE=5.1ms; TR=13.1ms). The 2-shot centre-out phase encoding approach meant that each 3D k-space line was acquired in two segments: one, following the preparation step, in which the positive section of k-space was covered using a FLASH acquisition from the centre to the edge of k-space, and the other, following the next preparation, in which negative k-space was covered in the same way. The image acquisition matrix size was 256 (read; 2x oversampled) x 176 (2D phase encode) x 224 (3D phase encode) with an image resolution of 1x1x1mm. Based on optimisation using computer simulations of the Bloch equations[5], the following sequence parameters were chosen: τ_1=250ms, τ_2=350ms, RF flip angle=23°, acquisition bandwidth=50kHz. The total scan time for whole brain coverage was 12min 20s. To demonstrate the efficacy of our approach, MDEFT imaging was performed in 3 ways: using an adiabatic hyperbolic secant pulse for inversion; using a standard nominal 180° rectangular hard pulse for inversion; and using a standard rectangular hard pulse with the power reduced to correspond to a nominal 130° pulse instead of 180° (chosen based on the simulation results).

Results

Example slices using the three acquisition methods are shown below in Figure 1. The standard MDEFT acquisition (Figure 1a) achieves good tissue contrast but there is some signal non-uniformity caused by B_1 inhomogeneity, especially evident in the coronal (lower) image. Using a hard nominal 180° pulse the 'RF hotspot' effect is more pronounced (Figure 1b), causing even greater signal non-uniformity. However, when a hard nominal 130° pulse is used in place of an inversion pulse (Figure 1c), the signal uniformity is improved (*e.g.* note the much higher signal level in the lateral sections of the cerebellum on the coronal image) while maintaining good GM-WM contrast (see Table 1 below).

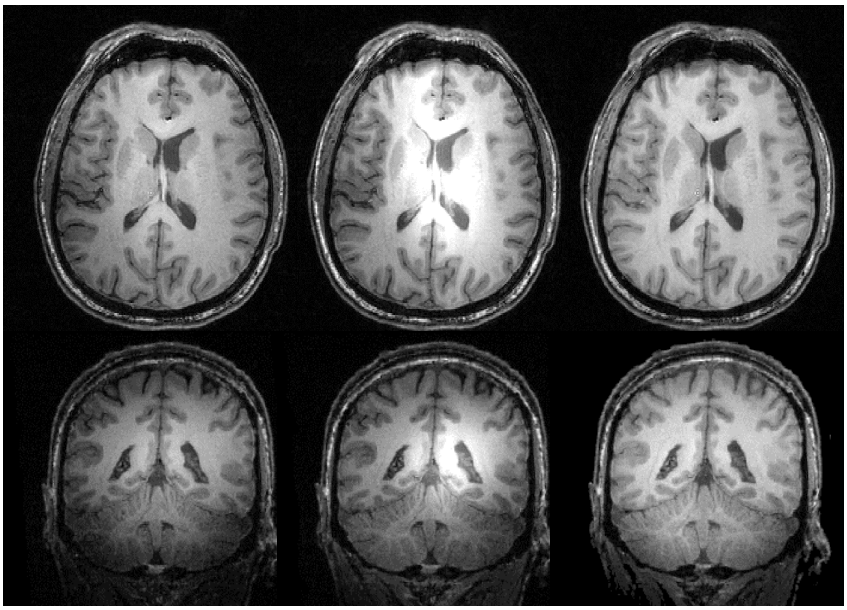

Fig 1a MDEFT with adiabatic inversion

Fig 1b MDEFT with hard nominal 180° inversion

Fig 1c MDEFT with hard nominal 130°

ROI location	Adiabatic	Hard 180º	Hard 130º
GM frontal	9358	13765	10660
GM occipital	5917	6758	10047
WM frontal	15043	20163	15494
WM occipital	10424	11740	14586

Table 1 Comparison of ROI signal intensities from each MDEFT sequence (arbitrary units)

Conclusions

We have proposed a modification to the MDEFT method using standard non-adiabatic hard preparation pulses which allows T_1-weighted imaging to be performed at high field strength with minimal non-uniformity in signal intensity and contrast. The sequence takes advantage of the complementary B_1 sensitivity of the preparation and excitation pulses, rather than seeking to minimise or eliminate the effect. This modification is easy to implement and allows more straightforward differentiation of tissue types, thereby reducing problems associated with automated tissue segmentation.

References **1** Hoult, D.I. JMRI 12:46-67 (2000) **2** Ugurbil, K. *et al.* Magn Reson Quart 9:259-277 (1993), **3** Lee, J.-H. *et al.* Magn Reson Med 34:308-312 (1995), **4** Norris, D.G. *et al.* MAGMA 9:92-96 (1999) **5** Deichmann, R. *et al.* NeuroImage 12:112-127 (2000)