Can 7T MPRAGE match MP2RAGE for gray-white matter contrast?

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A B S T R A C T

Ultra-High Field (UHF) MRI provides a significant increase in Signal-to-Noise Ratio (SNR) and gains in contrast weighting in several functional and structural acquisitions. Unfortunately, an increase in field strength also induces non-uniformities in the transmit field (B1+) that can compromise image contrast non-uniformly. The MPRAGE is one of the most common T1 weighted (T1w) image acquisitions for structural imaging. It provides excellent contrast between gray and white matter and is widely used for brain segmentation. At 7T, the signal non-uniformities tend to complicate this and therefore, the self-bias-field corrected MP2RAGE is often used there. In both MPRAGE and MP2RAGE, more homogenous image contrast can be achieved with adiabatic pulses, like the TR-FOCI inversion pulse, or special pulse design on parallel transmission systems, like Universal Pulses (UP).

In the present study, we investigate different strategies to improve the bias-field for MPRAGE at 7T, comparing the contrast and GM/WM segmentability against MP2RAGE. The higher temporal efficiency of MPRAGE combined with the potential of the user-friendly UPS was the primary motivation for this MPRAGE-MP2RAGE comparison. We acquired MPRAGE data in six volunteers, adding a k-space shutter to reduce scan time, a kt-point UP approach for homogeneous signal excitation, and a TR-FOCI pulse for homogeneous inversion. Our results show remarkable signal contrast improvement throughout the brain, including regions of low B1+ such as the cerebellum. The improvements in the MPRAGE were largest following the introduction of the UPS. In addition to the CNR, both SNR and GM/WM segmentability were also assessed. Among the MPRAGEs, the combined strategy (UP + TR-FOCI) yielded highest SNR and showed highest spatial similarity between GM segments to the MP2RAGE. Interestingly, the distance between gray and white matter peaks in the intensity histograms did not increase, as better pulses and higher SNR especially benefitted the (cerebellar) gray matter. Overall, the gray-white matter contrast from MP2RAGE is higher, with higher CNR and higher intensity peak distances, even when scaled to scan time. Hence, the extra acquisition time for MP2RAGE is justified by the improved segmentability.

1. Introduction

High-quality anatomical T1w images are essential for several MRI applications, notably to serve as an anatomical reference in fMRI and Gray Matter segmentation (Marques and Norris, 2018). Typically, 3D T1 weighted images are acquired with the Magnetization Prepared Rapid Gradient Echo (MPRAGE) sequence (Mugler and Brookeman, 1991, 1990). A 3D Gradient Echo (GRE) train is applied with short repetition times (TRs) and small flip angles (close to the Ernst angle), with the level of T1 weighting being predominantly controlled by the inversion time and the inversion efficiency of the applied inversion pulse (Mugler and Brookeman, 1991).

Anatomical, T1-weighted Magnetic Resonance Imaging (MRI) can benefit substantially from the use of Ultra-High Field (UHF, ≥ 7T). Compared to conventional, clinical, field strengths, imaging at 7T offers higher signal-to-noise ratio (SNR), which can be traded for an increased spatial resolution to reduce partial volume effects and more precise cortical measurements. Although T1 relaxation times at 7T are longer for both GM and WM, the difference is sufficient to result in significant gains in tissue contrast for structural MRI (Rooney et al., 2007; Wright et al., 2008). While there are benefits at ultra-high field (UHF), traditional MPRAGE data are also affected by transmit (B1+) and receive (B1-) radiofrequency field inhomogeneities. In addition, large static (B0) field variations in brain areas close to air-tissue boundaries also affect the inversion efficiency.

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Hence, some measures have to be taken to make MPRAGE images B1+ insensitive at 7T. Several possibilities exist. A widely used approach is that of the adiabatic pulses for inversion. These adiabatic pulses have lower or no dependency on B1+. Examples of adiabatic pulses are the flattened hyperbolic secant pulse (H88) (Garwood and DelaBarre, 2001), and time resampled frequency offset compensated inversion (TR-FOCI) pulse (Hurely et al., 2010). This is not sufficient to completely remove all B1+ contamination as it only improves the inversion homogeneity; the excitation homogeneity is still affected by B1+ variation.

In the MP2RAGE sequence (Marques et al., 2010), two 3D GE trains are acquired at different inversion times (TI), producing a T1w, and an approximately proton density (PD) weighted image. The combination of both images results in a T1w and a T1 map, ideally both free of B1+, PD and T2+ contrast (Marques et al., 2010; Van de Moortele et al., 2009). Even in an MP2RAGE acquisition, very low local B1+ can cause a loss of SNR and contrast, usually in the cerebellum and temporal lobes (O’Brien et al., 2014b). If not adequately addressed, these non-uniformities can compromise the image quality, or even provide incorrect segmentation, inappropriate diagnostic or poor co-registration (Haast et al., 2018; O’Brien et al., 2014a, 2014b). Also, in the MP2RAGE, more homogeneous contrast can be achieved by applying better and more efficient adiabatic inversion pulses (O’Brien et al., 2014b). Previous studies showed that image contrast uniformity from TR-FOCI is higher compared to H88 (Hurely et al., 2010; O’Brien et al., 2014b).

A third alternative to improve contrast homogeneity is using multiple transmission channels, also known as parallel transmission (so-called pTx or Multix systems) (Padorno et al., 2016). Parallel transmission systems enable local decreases or compensation in B1+ inhomogeneity using multiple transmit RF coils which are independently driven and operating simultaneously (Deniz, 2019). Typically, pTx pulse design relies on calibration measurements performed before scanning subjects, which takes up valuable scan time. To address this issue, broadly-calibrated plug-and-play pulses were developed (Gras et al., 2017) for both inversion and excitation. This Universal Pulses approach involves first collecting the B1+ field information for several subjects, followed by calculating their optimal set of parameters. The resulting pulses are therefore computed to account for the average variations in B1+ field. Successful implementation in a variety of sequences has been demonstrated, including 3D T1 weighted MPRAGE (Gras et al., 2017), T2-weighted TSE, (Gras et al., 2018), fluid-attenuated inversion recovery (FLAIR) (Gras et al., 2019b), multi-band and 3D EPI for whole-brain fMRI (Gras et al., 2019a; Le Ster et al., 2019) and T2’ weighted 2D Gradient echo (Gras et al., 2017).

Regarding temporal efficiency, high resolution isotropic T1w MPRAGE can be acquired in 5–7 min, whereas a T1w MP2RAGE with the same spatial resolution usually requires 10–14 min, in both cases using a two-dimensional (ky-kz) under-sampling pattern with parallel imaging reconstruction. Various strategies have been employed to reduce scan time in both acquisitions. The most common approach is to skip and zero-fill parts of the k-space, whether applying partial Fourier or elliptical sampling schemes (k-space shutter).

The differences between both MPRAGE-like sequences has been studied before, although mainly for morphometric assessment (Fujimoto et al., 2014; Okubo et al., 2016; Seiger et al., 2015; Yan et al., 2020). To our knowledge, the influence of different bias-field correction strategies, specifically using UPs for whole-brain MPRAGE compared to MP2RAGE, has not been reported yet. Therefore, the present study compares different strategies to improve the homogeneity of contrast in whole-brain MPRAGE images. This comparison is motivated by higher temporal efficiency of MPRAGE, combined with the great potential of UPs to make UFH more accessible. To acquire the best possible MPRAGE on a 7T scanner we acquired T1-weighted images with (1) a k-space shutter to reduce scan time; (2) Universal k-point pulses to homogenize signal excitation and (3) a TR-FOCI to invert the signal uniformly. We expected that the combination of all advanced techniques would improve image SNR and CNR and improve the segmentation results. The MPRAGE data were compared to MP2RAGE data, which were acquired with a k-space shutter and a standard HS8 inversion pulse or a TR-FOCI.

2. Methods

2.1. Subjects

Six healthy volunteers (age 25–40 years, three women) participated in the present study. The local ethics committee approved the study, and all volunteers provided written consent after being informed of the experimental procedures.

2.2. MRI Sequences

All imaging measurements were performed on a 7T MRI scanner (Philips Healthcare, Best, The Netherlands) using an 8-channel transmit, 32-channel receive head coil (32RxSTx, Nova Medical Inc, Wilmington, United States) with a close to circularly polarized-mode achieved by B1-shimming over the entire brain of a separate group of volunteers.

The whole-brain T1-weighted MPRAGEs and MP2RAGEs were acquired within the same session with matched voxel sizes and acceleration factors for all six sequences to ensure a fair comparison between all sequences. The common set of parameters for both techniques were: matrix size = 288 × 288 × 232, Field of View (FOV) = 230 × 230 × 186 mm³, with isotropic voxel-size of 0.8 mm, with SENSE under-sampling factor in two directions (Left-Right and Anterior-Posterior), 2D SENSE = 1.8(LR) × 1.8(AP), a slice oversampling of 20% and sagittal orientation. All the MPRAGEs shared the same TR/TE = 12/3.3 ms, TI = 1000 ms, TR_volume = 3000 ms, FA = 8° and BW_readout = 235.2 Hz. This protocol was based on the ADNI T1-weighted anatomical (http://adni.loni.usc.edu/), albeit with longer TR to accommodate the longer T1 at 7T. For the MP2RAGE, Bloch simulations were used to optimize the inversion times, TR_volume and flip angles following (Marques et al., 2010), resulting in a TR/TE = 6.2/2.3 ms, T1l = 800 ms and T12 = 2700 ms, TR_volume = 5500 ms, FA = 7°/5° and BW_readout = 401.9 Hz. The length of a readout block for the MPFRAGE is 196 ms and for the MP2RAGE 1016 ms. The total scan duration and the specific parameters for the individual scans are given in Table 1.

The standard MPRAGE was defined with an HS8 inversion pulse, standard excitation pulse (no UPs), a fully sampled matrix (without the k-space shutter), and called MPRAGE1. The 159 readout lines correspond to a single plane in k-space for MPRAGE1. For all other scans, a vendor-supplied elliptical k-space shutter was added. With the k-space shutter the edges of the k-space are skipped in an ellipsoidal fashion, reducing scan time by ∼25%. A similar number of lines per gradient echoes (GRE) readout block was used (164) as for the MPRAGE1 (159). For MPFRAGE3 the standard excitation pulses were additionally replaced by the UPs. For the UPs, the k-point method (Cloos et al., 2012) was used to generate the appropriate pulse shapes. A small tip angle pulse with 1ms of duration with five k-points was designed on 16 subjects with an interleaved greedy-local optimization (Grisson et al., 2012) as in (Roos et al., 2019). And finally, for MPRAGE4 the HS8 adiabatic inversion pulse was replaced by the TR-FOCI pulse, with an amplitude adjusted to have a minimum of 15 μT and inversion pulse duration of 13 ms. The MP2RAGEs were acquired with the standard HS8 and with a TR-FOCI. Both MP2RAGEs were acquired solely with k-space shutter and standard excitation pulses, to keep the scan session reasonably short. The vendor-supplied bias-field removal correction CLEAR (Constant Level Appearance) (Harvey et al., 2015) was used for both MPRAGE and MP2RAGE data, though this is cancelled out again in the MP2RAGE T1-weighted images.

2.3. MP2RAGE reconstruction

In the MP2RAGE sequence, after the simultaneous acquisition of the first inversion (GRE_T11) and the second inversion (GRE_T12), a uniform
T1-weighted image is obtained by computing the real component of the normalized complex ratio from both volumes (Marques et al., 2010), where \( G_{R_{T1}} \) stands for the complex conjugate of \( G_{R_{T1}} \). Once the T1-weighted image was generated, a background noise removal was applied using layNii toolbox (Huber et al., 2021).

\[
S = \frac{\text{Re}(G_{R_{T1}} \times G_{R_{T1}}^*)}{|G_{R_{T1}}|^2 + |G_{R_{T1}}^*|^2}
\]

### 2.4. Registration and Segmentation

All T1w data were registered to the MPRAGE1 space using the six degrees of freedom rigid body co-registration of SPM12 (www.fil.ion.ucl.ac.uk/SPM/software/spm12/) to allow further comparisons. Segmentation of gray matter (GM), white matter (WM), and cerebrospinal fluid (CSF) were performed for each subject separately using the computational anatomy toolbox CAT12 (www.neuro.uni-jena.de/cat/) for SPM with bias field correction in medium level (0.5). The segmentation procedures resulted in 6 different GM and WM segmentation volumes, which were combined, resulting in a single GM and WM combined ROI for each subject and later used to define the regions of interest’s (ROI’s) voxels. Voxels were assigned GM if they were GM in at least two datasets, the same approach was used for WM, and overlapping voxels were excluded from the final combined ROI. A cerebellum mask was created using the Nighres brain region extraction function (Hantenburg et al., 2018) and further separated into GM and WM cerebellum masks using the GM and WM mask from the unified segmentation. Another two cubic ROIs (50 × 50 × 50 mm) in the right temporal lobe and centred on the PCC (Posterior Cingulate Cortex) were assessed as additional examples of low (temporal) and high (PCC)-B1+ areas.

### 2.5. Data Analysis

The k-space shutter blurriness evaluation metric was the local Point Spread Function (PSF). A signal profile was taken along the LR axis through the ventricles for both MPRAGE1 and MPRAGE2. The full width half maximum (FWHM) of the ventricles along this line was calculated. The FWHM estimation was performed for both MPRAGEs, per subject, and the differences were assessed using a paired t-test. Additionally, a sigmoid function, equation [2], was fitted to part of the profiles covering the ventricle wall, and the slope parameter ‘\( d \)’ was taken as a measure of the steepness of the slope (Bazin et al., 2020).

\[
y = a + \frac{b}{1 + e^{-\frac{x}{\sigma}}}
\]

The quantitative assessment was based on Signal-to-Noise ratio (SNR) and Contrast-to-Noise ratio (CNR) measurements. The SNR was calculated using equation [3], where \( \mu_{\text{foreground}} \) is the mean and \( \sigma_{\text{foreground}} \) is the standard deviation of the signal over the foreground mask, with \( n \) equal to the number of voxels within the mask. The foreground mask corresponds to the combination of GM and WM masks. The CNR was calculated using equation [4], where \( \mu_{\text{GM}} \) and \( \mu_{\text{WM}} \) are the mean over GM and WM, respectively and \( \sigma_{\text{GM}} \) (standard deviation of GM) and \( \sigma_{\text{WM}} \) (standard deviation of WM). One-way ANOVAs with Tukey post-hoc analyses were used to test the SNR and CNR differences between all the six sequences. All generated boxplots, and the repeated measures ANOVA estimation with Bonferroni Pairwise correction was performed using R (R Core Team, 2020), where \( p \)-values < 0.05 were considered statistically significant.

\[
\text{SNR} = \frac{\mu_{\text{foreground}}}{\sigma_{\text{foreground}}} \times \sqrt{n/(n-1)}
\]

\[
\text{CNR} = \text{abs}((\mu_{\text{WM}} - \mu_{\text{GM}})) \times \sqrt{\sigma_{\text{WM}}^2 + \sigma_{\text{GM}}^2}
\]

The GM and WM ROI masks were used to assess the histograms of image intensity. Receiver operating characteristic (ROC) curves and area under the curve (AUC) were used to evaluate the robustness of separation from the GM and WM histogram intensity (GM and WM tissue fraction) peaks, reflecting which sequence yielded better segmentation. Both alternative metrics were calculated per subject.

We calculated the Dice Similarity Coefficient (DSC) (Dice, 1945) of the GM probability tissue mask (with threshold = 0.59) to compare the segmentation’s spatial similarity among all six sequences. DSC is an overlap-based metric commonly used to validate segmentation boundaries, and in the present study, the DSC was used to verify that the segmentation is consistent.

### 3. Results

Fig. 2 shows the sagittal, coronal and axial views of all six T1-w images of a single representative subject (subject #2). Visual inspection suggests that there are no significant differences between MPRAGE1 and MPRAGE2. However, a less intense central brightening and more signal in the cerebellar cortex can be seen in MPRAGE3&4 compared to MPRAGE1&2 (Fig. 2, white arrow). Generally, the UPs recovered the signal in the low-B1 regions. Another observation was the more homogeneous WM signal throughout the brain in MPRAGE3&4. The MP2RAGEs (Fig. 2, right column) were likewise visibly more homogeneous than MPRAGE1&2 (Fig. 2, left column). Example slices of T1w images from all six subjects can be seen in the supplementary material (Supplementary Fig. 1.). Noteworthy, the bias field gradually reduced from MPRAGE1/2 to MPRAGE3/4 and the MP2RAGEs (Supplementary Table. 2). With the used parameters, all protocols are expected to have some B1+ dependent spatial variation, even the MP2RAGEs (Haast et al., 2018; Marques et al., 2010). In addition, a residual B1- variation is expected in the MPRAGE data, even with the vendor-supplied CLEAR correction (Harvey et al., 2015).

Fig. 3a and b show the sampling pattern of the k-space shutter and the resulting point spread function (PSF). The observed profiles (Fig. 3B-G) show minimal changes; the most noticeable is the signal intensity reduction in MPRAGE2, related to the reduced scan time. We calculated the FWHM and the steepness of the slope for both MPRAGEs. No differences in FWHM in these profiles were observed (21.2±/− 1.0 and 21.3
Fig. 1. Flowchart representation of the registration and segmentation used to create the unified GM and WM mask. A) Registration performed in SPM12. The panel B) shows the output of CAT12 segmentation, for a single sequence; after performing the segmentation in all six sequences, the GM and WM probability tissue maps were averaged, resulting in a single GM and WM masks. The voxels were assigned GM if they were GM in at least two datasets the same approach was used for the WM, and overlapped voxels were excluded from the final combined ROI.

Fig. 2. Sagittal, coronal and axial T1w example slices of subject #2 for all six acquisition protocols. The cerebellar area of low B1+ yields low signal in MPRAGE1 and MPRAGE2. A signal increase is seen in the same regions by introducing the UP in MPRAGE3 (white arrow), where the central brightness also appears less intense. The bias field gradually reduced from MPRAGE1/2 to MPRAGE3/4 and the MP2RAGEs (Mean Percentage Bias Field gradually increases from MPRAGE1 to MP2RAGE6). Note that Mean Bias Field percentage is the estimated intensity effect (Bias Field) averaged across subjects. The percentage values represent the data quality estimated before the bias field correction in CAT12 (higher is better, representing the data quality).
Fig. 3. Panel A) shows the sampling pattern in k-space of the k-space shutter used on the k-space coverage for MPRAGE2-4 & MP2RAGE5 and 6. From B) to G) we show the signal profile along the Left-Right axis through the ventricles in individual subjects, the green signal profile represents MPRAGE1 signal profile, in which no elliptical shutter was employed and the yellow signal profile representing the MPRAGE2 signal, with k-space shutter yielding 25% scan time reduction. There was no difference in the estimated sharpness of the PSF.

Fig. 4. Boxplots for each sequence showing the SNR and CNR (between Gray and White matter) in the whole brain, cerebellum, the right temporal area and the ‘PCC’ (a cube centred on the posterior cingulate cortex). In all cases, the SNR was higher after introducing Universal pulses, MPRAGE3 (blue box) and MPRAGE4 (dark orange) and higher after combining with the TR-FOCI (MPRAGE4, dark orange). In contrast, the CNR was higher for MP2RAGEs for all example areas. ‘PCC’ area resulted in higher inter-subject variability due to anatomy differences (gray/white matter ratio) across subjects.

\(+/-.0.1\), Paired two-sided t-test \(p = 0.36\) and the slope parameter of the sigmoid fit was also stable \((0.46+/-0.04\) and \(0.45+/-0.04\), Paired two-sided t-test \(p=0.52\)), suggesting that the edges were preserved with the application of the k-space shutter.

The SNR and CNR assessment results of all sequences are shown in the boxplots in Fig. 4. Four distinct regions were assessed: a whole-brain ROI (including the cerebellum), a cerebellum ROI, a right temporal lobe and a PCC, as example regions affected by B1+ inhomogeneities. For all ROIs, the UPS yielded higher SNR (all ROIs with \(p < 0.05\)), and when combined with TR-FOCI pulse, the SNR was even higher, notably for whole-brain and cerebellum (higher median, dark orange boxes). MP2RAGE also showed similar results; the introduction of the TR-FOCI pulse also led to a higher median SNR than the HS8 pulse in the whole-brain and cerebellum masks. It is worth mentioning that the low SNR values found here are also related to the contrast contributions to the signal variation, as the CNR of the gray/white matter difference showed a somewhat different pattern. CNR was higher for MP2RAGEs than the MPRAGEs, for both whole-brain and the cerebellum masks (both with \(p < 0.05\)), with the TR-FOCI in MP2RAGE6 leading to a slightly higher CNR (higher median) than the HS8 inversion in MP2RAGE5, though this difference was not statistically significant. Among the MPRAGEs, the measured CNR was similar, with only a slightly higher value for MPRAE 3&4 (higher median) than MPRAE 1&2, although also with higher variability between volunteers. It is also important to mention that the higher inter-subject variability in the right temporal and PCC masks are most likely due to differences in anatomy (gray/white mat-
Fig. 5. Signal distribution in MPRAGE and MP2RAGE acquisitions for both GM and WM tissues. In A) Whole-brain and B) cerebellum ROIs for one representative subject. The Gray matter peaks are the first ones (upper part, graphics A and B), while below is the White matter. Note that the MP2RAGE intensities are divided by a factor of 10 to match the x-axis. Panel C and D are the ROC curves of the same representative subject for whole-brain and cerebellum, respectively. The curves show better separability of GM and WM for MP2RAGE compared with MPRAGE for both masks. Areas under the curve (AUC) for whole-brain (Panel E) and cerebellum (Panel F). MP2RAGE shows better GM/WM segmentability.

Histograms

Whole-brain

| ROI | Intensities |
|-----|-------------|
| GM  | ~100 to ~130|
| WM  | ~200 to ~300|

Cerebellum

| ROI | Intensities |
|-----|-------------|
| GM  | ~100 to ~130|
| WM  | ~200 to ~300|

ROC

AUC

E) AUC for whole-brain ROIs.

F) AUC for cerebellum ROIs.

The SNR ($SNR\sqrt{T_{\text{acq}}}$) and the CNR ($CNR\sqrt{T_{\text{acq}}}$) per unit of time were also assessed to compensate for differences in acquisition time. For the SNR, a significant difference was observed ($p < 0.05$) for the UPs (MPRAGE 3&4) when compared with other MPRAGEs and MP2RAGEs for all masks. As expected, the CNR increased when UPs were employed, however, with no significant differences to the MP2RAGEs, although it is worth mentioning that the CNR values are higher for MPRAGE3 in the right temporal and both MPRAGE3 & MPRAGE4 in the PCC mask. A detailed description of the SNR and CNR per unit of time can be seen in Supplementary Fig. 2.

Fig. 5 (Panels A and B) shows the histograms of the signal distributions in MPRAGE and MP2RAGE acquisitions. The gray matter (~100 and ~130) and white matter (~200 and ~300) signal intensities for the whole-brain (Fig. 5 Panel A) underwent a substantial shift after the introduction of the UPs for all subjects, relatively higher in the gray matter than in white matter. For the cerebellum (Fig. 5 Panel B) the same behavior was observed, it also presented a shift towards higher intensities values after introducing the UPs. Note that the MP2RAGE intensities were put on a scale of 0-409.6 here, rather than the usual 0-4096. The intensity histograms for all six subjects are depicted in the supplementary material (Supplementary Fig. 3).

ROC curves and, subsequently, the AUC were calculated per scan and per subject to quantify the gray-white matter separability since the GM and WM intensity peaks are not entirely separated in the histograms. Fig. 5 (Panels C and D) shows the GM and WM fraction’s ROC curves for one representative subject. The ROC curves from all six subjects are depicted in supplementary Fig. 4. For the example subject, as for the other individuals, the ROC curves were higher for the MP2RAGE than for the MPRAGEs, with only small differences between MPRAGE acquisitions. To summarise the segmentability in a single number, the area under the curve, or AUCs, for the whole brain ROI and the cerebellum were generated and shown in violin/boxplot format (Fig. 5, Panels E and F). AUCs were consistently higher for the MP2RAGEs. Among the MPRAGE acquisitions, MPRAGE3&4 showed slightly higher values than MPRAGE1&2 for the whole-brain ROI (Fig. 5 Panel E) while, on the other hand, for the cerebellum ROI (Fig. 5 Panel F) the MPRAGE 1&2 show higher values than MPRAGE3&4.
Fig. 6 shows the Dice coefficients of the GM probability tissue, averaged over all six subjects. Panel A for whole-brain and Panel B for cerebellum. The DSC per subject can be seen in supplementary material (Supplementary Table 1.a-b). The purpose of this metric is to verify how similar the segmentations are. DSC coefficient shows that the segmentation results from MPRAGE1 and MPRAGE2 were highly similar. Even higher similarity was observed between MPRAGE 3&4 and between MP2RAGE 5&6. It is worth pointing out that the segmentation performance of MPRAGE3&4 scored higher values than the MPRAGE1&2 when the reference was the MP2RAGE5 or MP2RAGE6.

4. Discussion

In this study, we compared different strategies to improve whole-brain MPRAGE acquisitions at 7T. A k-space shutter was used to reduce the scan time further. Universal k-t point pulses were used to homogenize signal excitation, and the TR-FOCI was used to invert the signal more uniformly. All these MPRAGE variations were compared to MP2RAGE data acquired with two different inversion pulses (HS8 and TR-FOCI). We anticipated that combining all advanced techniques would yield higher SNR and CNR, resulting in a better separability of Gray and White Matter for MPRAGE with comparable results with the MP2RAGE acquisition which is now very widely used at 7T. Interestingly, we found that the UPs and the TR-FOCI pulse provide a robust uniformity and an increase in SNR; however, it does not translate into better segmentability than MP2RAGE.

The use of a k-space shutter did not significantly affect the MPRAGE in terms of SNR (Fig. 2), and no difference in blurring at the high contrast edge of the ventricles was found (Fig. 3). These results were expected since the outer regions of the k-space only have small signal amplitude compared with the amount of noise; removing the corners of k-space results in a minimal loss of spatial resolution and considerable time saving for MPRAGE (~25%, here approximately two minutes). It seems safe to use a k-space shutter to save scan time where the loss in SNR because of the reduced scan time is not harmful. For these medium-resolution anatomical images, the loss in SNR does not affect the segmentation quality (Fig. 6).

Quantitative assessment was performed using SNR and CNR as metrics over the different acquisitions. The increased SNR offered by introducing the UPs (MPRAGE 3&4) shows a remarkable improvement compared to the other sequences used, including the MP2RAGE, for both whole-brain and cerebellum. The TR-FOCI pulse introduction resulted in a higher inversion efficiency and SNR, especially in the cerebellum, as expected (O’Brien et al., 2014b). However, the difference was not as large as in the O’Brien study, which used a single-channel transmit coil.

A possible explanation might be the TX8 Nova coil’s performance, which already provides good B1+ coverage over the brain. However, without a direct hardware comparison, this remains speculation. The increase in SNR achieved in MPRAGE3&4 did not translate to higher CNR values than MP2RAGEs (Fig. 4). This result is most likely due to the balance of tissue types in the areas most affected by the B1+ inhomogeneity. These contain mainly gray matter, and with increasing brightness becomes closer in intensity to the overall white matter peak.

Regarding the signal distribution (Fig. 5), both GM and WM signals increased with the introduction of the UPs; the dark orange/blue curves are moving to the right in the histogram for all subjects in the supplementary material (Supplementary Fig. 3). Similar behavior was observed in previous work when UPs also induced a signal increase compared with CP mode (Gras et al., 2017). Again, as the gray matter peak moved more than the WM peak, the signal intensity distribution of gray and white matter did not become better separable in MPRAGE3 or MPRAGE4 (Fig. 5A-B). This effect is especially clear in Fig. 5E-F, where the AUC improved a bit in the whole brain acquisition, probably because of the higher SNR, but decreased for MPRAGE3&4 compared to MPRAGE1&2 in the cerebellum, where the increase in GM signal intensities was largest. With the two signal compartments closer together in the image intensity histogram, they overlapped more. Nevertheless, the (cerebellar) segmentations in MPRAGE3&4 were more consistent than in MPRAGE1&2, as shown by the off-diagonal values in Fig. 6B.

We also compared the segmentation of gray and white matter, using the Dice coefficient to measure the segmentation stability. A high overlap was observed between MPRAGE 1 (no shutter) &2 (shutter), meaning that the k-space shutter’s introduction does not compromise the MPRAGE contrast. Even higher dice coefficients were observed between MPRAGE 3&4 and between MP2RAGE 5&6, which means that these pairs of acquisitions yielded very consistent segmentation of GM/WM. Interestingly, the segmentation performance of MPRAGE 3&4 scored higher values than the MPRAGE 1&2 when the reference was one of the MP2RAGEs, which means that the introduction of the UPs and the TR-FOCI pulse lead to better segmentation than a standard MPRAGE.

Nevertheless, all our results and extracted image quality metrics show that the MPRAGE does not yet match the MP2RAGE for gray and white matter contrast or segmentability. We achieved higher signal homogeneity using Universal Pulses (UPs) and TR-FOCI inversion and shorter scan time with k-space shutter. However, the increase in SNR obtained for MPRAGE was insufficient to translate directly into better CNR values or better separability of GM and WM than MP2RAGE. Hence, in their current implementation, MP2RAGE still merits the additional scan time for a T1-weighted 7T anatomical scan. A further advantage from MP2RAGE acquisitions is that a T1-map can be obtained from the data at
Data availability anatomical

The data that support the findings of this study are available from the corresponding author, upon reasonable request.

Supplementary materials

Supplementary material associated with this article can be found, in the online version, at doi:10.1016/j.neuroimage.2021.118384.

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