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Inversion Recovery with Embedded Self-Calibration (IRES)

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Abstract

With self-calibrated parallel acquisition, the calibration data used to characterize coil response are acquired within the actual, parallel scan. Although this eliminates the need for a separate calibration scan, it reduces the net acceleration factor of the parallel scan. Furthermore, this reduction gets worse at higher accelerations. A method is described for 3D inversion recovery gradient-echo imaging in which calibration is incorporated into the sequence but with no loss of net acceleration. This is done by acquiring the calibration data using very small ($\leq 4^\circ$) tip angle acquisitions during the delay interval after acquisition of the accelerated imaging data. The technique is studied at 3T with simulation, phantom and *in vivo* experiments using both image space-based and k-space-based parallel reconstruction methods. At nominal acceleration factors of three and four, the newly described Inversion Recovery with Embedded Self-calibration (IRES) method can retain effective acceleration with comparable SNR and contrast to standard self-calibration. At a net 2D acceleration factor of four, IRES can achieve higher SNR than standard self-calibration having a nominal acceleration factor of six but the same acquisition time.

Keywords

Parallel imaging; inversion recovery; self-calibration; gradient echo

INTRODUCTION

Inversion recovery (IR) pulse sequences employ long inversion and delay times to generate the desired contrast weighting and provide adequate signal recovery. 3D IR sequences such as magnetization-prepared rapid gradient-echo (MP-RAGE) (1) acquire multiple k-space lines in an IR cycle via fast gradient-echo (GRE) repetitions. However, the acquisition times can still be on the order of several to ten minutes long. Parallel imaging can be applied to further reduce acquisition times (2–4). To do so, coil sensitivities have to be determined. One option is to use self- or auto-calibrated methods (4), which in general use unaccelerated sampling in some central region of the otherwise undersampled k-space. However, the fraction of acquisition time devoted to calibration can be substantial, particularly at high acceleration factors, resulting in reduced net acceleration. Although a separate calibration acquisition may be made (3), the additional scan time can reduce the overall efficiency of the MRI examination.

The purpose of this work was to devise a means for performing self-calibration while retaining the nominal acceleration factor. The method described, referred to as Inversion Recovery with Embedded Self-calibration (IRES), embeds self-calibration acquisitions within the delay intervals of IR sequences. This specific work implements IRES at 3T in an MP-RAGE sequence, which readily lends itself to this self-calibration scheme. In standard

MP-RAGE, an inversion pulse precedes a GRE train of small nutation angles (8–10°), which is followed by a delay interval prior to the next inversion cycle. With IRES, calibration acquisitions are measured using even smaller nutation angles ($\leq 4^\circ$) during the delay interval. In addition to describing the IRES method, it is shown in this work how IRES can be used to provide higher net acceleration than standard self-calibration for the same nominal acceleration. Alternatively, IRES can be used to provide improved image quality over standard self-calibration for the same net acceleration. Nominal accelerations as high as six are studied. Because IRES calibration data are acquired at different signal levels from the imaging data, they may not be incorporated into the parallel reconstruction. The effect of this limitation on SNR is also studied in this work.

METHODS

Reduced efficiency with standard self-calibrated parallel imaging

Prior to presenting the IRES method it may be instructive to study the decrease in net acceleration due to self calibration. Although this trend for acceleration loss in self calibration is general, we make several specific assumptions to illustrate the behavior mathematically: (i) 3DFT acquisition is performed and acceleration is applied along two phase encode directions (5); (ii) with no acceleration the number of encodes along y and z is the same ($N = N_Y = N_Z$); (iii) the corners of k_Y - k_Z space are not sampled so as to provide approximate isotropic resolution in the y-z plane (6). In this case the number of phase encodes required for the unaccelerated scan is simply $\pi N^2/4$. If parallel acquisition is performed this number can be reduced, as shown schematically in Fig. 1a where undersampling of two-fold is done in each direction. If this nominal undersampling factor is R , where R is assumed to be the product of undersampling factors in y (R_Y) and z (R_Z), then the required number of phase encodes is reduced to $\pi N^2/4R$. The net acceleration R_{net} , defined as the ratio between the number of fully-sampled and actual under-sampled phase encodes, is simply R .

Next assume that self-calibration is applied to the undersampled k-space of Fig. 1a. Define a unitless calibration ratio r_c as the fraction of k-space radius acquired for calibration, as shown in Fig. 1b, and assume the self calibration data is somehow interleaved into the parallel acquisition. The total number of phase encodes A required for the self-calibrated scan is equal to the sum of the fully sampled central area and the sparsely sampled peripheral area, and can be shown to be:

$$A = \pi N^2 (1 + r_c^2(R - 1)) / 4R. \quad (1)$$

As above, the net acceleration R_{net} is equal to the ratio between A in an accelerated scan versus that in an unaccelerated scan, or $R_{net} = R / (1 + r_c^2(R - 1))$. Finally, the case of separate calibration is shown in Fig. 1c, whereby the calibration acquisitions overlap in k-space with the accelerated acquisitions but are somehow measured separately.

Fig. 2 shows how R_{net} varies with r_c and R in self-calibration acquisitions. The fact that R_{net} falls below R reflects the loss of nominal acceleration due to self-calibration. The convexity of the curves means that this loss gets worse as R increases. Referring back to Fig. 1b, as the sparseness of sampling in the peripheral region increases, the area of the central, self calibration region becomes an increasing fraction of the total scan time.

Basic description of IRES

The IRES method was implemented using 3D MP-RAGE as shown in the sequence diagram of Fig. 3a, where the inversion time (TI), data acquisition using n_R GRE repetitions per cycle with flip angle α , delay time (TD), and overall repetition or cycle time (TC) are all defined. The parameters are often chosen to achieve the dual objectives of minimizing cerebral-spinal-fluid (CSF) signal and maximizing contrast between white matter (WM) and gray matter (GM). The cycle is assumed repeated n_c times until all desired acquisitions are made; hence A is the product of n_R and n_c .

With standard self-calibration, additional repetitions are incorporated into the overall sampling of α pulses, either by extending the number of α pulses within an individual cycle or by increasing the number of cycles. The fundamental concept of the proposed IRES technique is to acquire the data for self-calibration during the delay interval in which the magnetization is allowed to recover prior to the next inversion pulse. This concept is shown schematically in Fig. 3b with the inclusion of n_β radiofrequency pulses of flip angle β after acquisition of the accelerated data. The principal consequence of incorporating the β pulses is perturbation, and specifically reduction of the degree of longitudinal magnetization recovery during the TD interval. This is shown in Fig. 3c, results of a simulation comparing magnetization levels of standard (solid line) and IRES MP-RAGE (dashed line) for the same number of α pulses. This effect is studied in more detail later in this work, but because the calibration map can be of low spatial resolution, the β pulses can use small ($\leq 4^\circ$) flip angles and still allow magnetization recovery to occur. It can also be seen from Fig. 3c that in spite of large signal perturbation during the TD interval, the signal perturbation during the acquisition period is considerably smaller.

Technical parameters of IRES in MP-RAGE

The specific MP-RAGE sequence used as a starting point was developed for high resolution T1-weighted brain imaging for use at multiple institutions (7,8). A sagittal slab was excited, and parameters included a constant flip angle of $\alpha = 8^\circ$, $TC/TI/TR/TE = 2300/900/6.4/2.8$ msec, acquisition bandwidth ± 31.25 kHz, $FOV_X = 26$ cm (superior-inferior), $FOV_Y = 24$ cm (anterior-posterior), slab thickness (FOV_Z) = 20.4 cm (right-left), $N_X/N_Y/N_Z = 256/240/204$, and isotropic resolution = 1.0 mm^3 . The number of α pulses per cycle, n_R was set to 170, with small variations of ± 3 in different acceleration configurations that resulted from rounding off the number of cycles. As discussed, the corners of k_Y - k_Z space were excluded from sampling. In addition, a recessed-elliptical centric (EC) phase encoding order (9) was applied, which generally eliminates ghosting and flow artifacts *in vivo* and provides approximately isotropic k-space modulation in the phase-encoding plane (10). With this recessed order TI is defined as the delay from the initial inversion pulse to the central k-space view, as shown schematically in Fig. 3a.

To incorporate IRES into the MP-RAGE sequence, β pulses ($\leq 4^\circ$) with the same acquisition bandwidth, TR, and TE as the α pulses were added following the last α pulse. The number of β pulses, n_β is subject to two competing requirements. Firstly, to obtain the required number of calibration samples it can be shown that $n_\beta \geq \pi N^2 r_c^2 / 4 n_c$. Secondly however, the β calibration sampling period should not exceed the delay time, i.e. $n_\beta \leq TD/TR$. Both requirements were met in this work, given that $TD = 670$ msec, $TR = 6.4$ msec, and $n_\beta \leq 100$.

To test IRES over a range of accelerations, three specific acceleration values were used: (i) $R = R_Y = 3$; (ii) $R = R_Y \times R_Z = 2 \times 2 = 4$; (iii) $R = R_Y \times R_Z = 3 \times 2 = 6$. That is, the first of these was 1D acceleration along y while the latter two were 2D accelerations (5) along y and z. Reconstruction of the undersampled data was done using both SENSE (3,5) and GRAPPA

(4). As mentioned before, IRES calibration data were not incorporated into the GRAPPA-reconstructed image. However, for the standard self-calibrated GRAPPA implemented here, against which IRES was compared, the self-calibration data were used in the image reconstruction and thus can provide some incremental benefit to SNR.

Simulations

Simulations were developed based on the standard Bloch equations for MP-RAGE (11) to examine the effects of the IRES train of β pulses on the MP-RAGE signals of white matter (WM) and gray matter (GM). The assumed T1/T2 relaxation times (msec) were 850/85 (WM) and 1300/110 (GM). Other parameters used in the simulation were described in the previous section, with $n_\beta = 100$ unless otherwise stated.

Hardware and software

Imaging was performed at 3T (GE Healthcare, Signa version 14M5) with a head coil (Invivo Corp, Orlando, FL, USA) with eight equally sized elements oriented in a 1×8 array with circumference 75 cm that circumscribes the head. All simulations and parallel imaging reconstructions were performed offline on a dual, quad-core (Intel E5472 processor), 3 GHz Xeon computer (32 GB RAM) with Matlab R2008a (The Mathworks, Inc., Natick, MA, USA) running two threads.

With GRAPPA reconstruction, data were first transformed to the x - k_Y - k_Z hybrid space and were subsequently reconstructed on a slice-by-slice basis in the x -direction. A range of 2D training kernel sizes (from 3×3 to 20×20 in the k_Y - k_Z plane) was tested to determine specific sizes that provided the best SNR for each acceleration value. These kernels were ellipse-shaped and the ones chosen for this work had 18 ($R = 3$), 18 ($R = 4$) and 38 points ($R = 6$).

Phantom imaging

A tri-compartment, cylindrical phantom (diameter = 18 cm, height = 10 cm) was constructed with Carrageenan gels (Sigma-Aldrich Inc, St Louis, Missouri, USA) (12). The relaxation parameters as measured using spin-echo experiments were T1/T2 = 1090/150, 1370/160, 420/70 msec, which approximated those for WM, GM and fat respectively. The MP-RAGE parameters were kept the same as discussed above, subject to minor modifications in imaging resolution for efficiency (FOV = 20 cm, $N_X/N_Y/N_Z = 256/168/168$, resolution = $0.8 \times 1.2 \times 1.2$ mm³). The phantom was first imaged at different β values ($1^\circ \leq \beta \leq \alpha$) with $n_\beta = 100$ to determine a value for β for subsequent experiments. β values smaller than or equal to α were desired to minimize perturbations to longitudinal magnetization of the MP-RAGE signal.

For hypothesis testing, the phantom was imaged separately with standard self-calibration and IRES at nominal acceleration factors of $R = \{3, 4, 6\}$. Moderate calibration ratios of $r_c^2 = 0.09$ to 0.12 ($r_c = 0.3$ to 0.35) were applied, resulting in n_β in the range of 47 to 63. The calibration ratios used were found to produce image reconstructions of reasonable image quality. To facilitate comparison, the phantom studies were grouped into Set I, composed of results using $R = 3, 4$, and Set II ($R = 4, 6$) as shown in Table 1.

T1-weighted brain imaging

For *in vivo* human studies, six subjects between the ages of 25 and 45 were recruited in accordance with a protocol approved by the institutional review board. Subjects were all imaged at nominal acceleration factors of $R = 3, 4$, and 6 . The same calibration ratio of $r_c^2 = 0.12$ ($r_c = 0.35$) was applied to all acquisitions, resulting in n_β in the range of 61 to 82. A

random scan order was applied per subject. Similar to the phantom studies, the *in vivo* studies were grouped into Set III ($R = 3, 4$) and Set IV ($R = 4, 6$) as shown in Table 1.

Hypotheses and evaluation

The first hypothesis (Hypothesis A) was that for a fixed nominal acceleration R , IRES could provide comparable signal-to-noise ratio (SNR) and contrast but with a shorter acquisition time (higher net acceleration) than standard self-calibration. This was tested using nominal accelerations of $R = 3$ (Sets I, III) and $R = 4$ (Sets II, IV).

The second hypothesis (Hypothesis B) was that for the same acquisition time (equal net acceleration) IRES could provide higher SNR and contrast than standard self-calibration. This was tested with IRES $R = 3$ against self-calibration $R = 4$ (Sets I, III), and at IRES $R = 4$ against self-calibration $R = 6$ (Sets II, IV).

For the phantom studies, statistics were not measured within regions-of-interests (ROIs), but rather it was possible to estimate SNR for each voxel using the statistics across multiple independent acquisitions. Specifically, the phantom was imaged $n = 15$ times with each acquisition method. It was found beforehand that $n > 15$ did not materially affect SNR estimation. For each voxel the signal S was defined by the mean value of the 15 samples at that voxel, σ was defined by the observed standard deviation across those 15 values, and SNR was defined by the ratio S/σ . The normalized difference between the SNR of the IRES and standard self-calibrated acquisitions, ΔSNR was used for statistical analysis:

$$\Delta\text{SNR} = \frac{\text{SNR}_{\text{IRES}} - \text{SNR}_{\text{SC}}}{(\text{SNR}_{\text{IRES}} + \text{SNR}_{\text{SC}})/2} \quad (2)$$

This was repeated for all voxels (in excess of 10^5 per tissue type) within each tissue type (WM, GM, Fat) in the entire volume of the phantom. A histogram of the normalized SNR difference from each comparison was generated, and a one-sided matched pairs t-test (13) was performed to test for a significant difference of the result from zero. The p-value was determined for each comparison, and $p < 0.05$ was taken as being statistically significant. It should be noted that Hypothesis A is proven if ΔSNR is not significantly different from zero; i.e. $p > 0.05$, thereby indicating that the SNR of standard self-calibration and IRES are not statistically different. On the other hand, Hypothesis B is proven if ΔSNR is significantly different from zero with $p < 0.05$.

For *in vivo* imaging it was not practical to image each subject 15 times as for the phantoms. Rather, analyses of SNR and contrast-to-noise ratio (CNR) were performed with a ROI method. In this case for each tissue type (WM and GM) 21 regions of uniformly appearing tissue were selected from across the typical 60 axial sections, and each region was greater than 36 mm^2 in area. For each region, S was taken as the mean of the reconstructed values within the ROI, σ was taken as the standard deviation of those reconstructed values, and SNR was taken as S/σ . Similar to the phantom studies the normalized SNR (Eq. 2) was determined for each region. This was repeated for each of the six subjects, and the pooled results provided a total of 126 comparisons for each tissue type. Similar to the phantom results, a one-sided matched pairs t-test was performed on the set of normalized SNR values. It is noted for this ROI-based method that SNR measurements are compromised due to tissue inhomogeneities within the ROI, which cause the observed σ to exceed that due solely to statistical uncertainties (14). This increases the degree of overlap of SNR distributions between two populations, making it more difficult to establish significance. Therefore, care was taken to select only regions of tissue that were as homogeneous as possible.

For comparing the WM vs. GM CNR, the ROIs of WM and GM could not be used as pairs, and so the inputs were the average CNR from each subject ($n = 6$). A Wilcoxon signed rank test (13) was used for comparing CNR.

Evaluation of incorporation of calibration data

As mentioned earlier, compared to standard self-calibration, IRES does not benefit from the SNR improvement from incorporation of calibration data into the GRAPPA reconstruction. To study this, fully-sampled acquisitions of the phantom were made using the MP-RAGE acquisition that included IRES calibration. The acquired data were subsequently undersampled to mimic parallel acquisition for $R = 3, 4$, and 6 and with a range of degrees of self-calibration of $0.1 < r_c < 1.0$. GRAPPA reconstructions were done both with and without the incorporation of the self-calibration data into the image reconstruction, the latter mimicking IRES. This was repeated for a total of $n = 20$ trials. The SNR was determined using the voxel-based method described earlier and the results were compared.

RESULTS

The simulation results in Fig. 4a–b showed that increasing the β flip angles progressively increases both WM and GM signals (a) but at the cost of reducing WM-GM contrast (b). The overall increase in WM and GM signals also appeared to flatten the shape of the MP-RAGE signal. The images in Fig. 5 of the phantom acquired with IRES at different β values (b–d) were visually similar to the standard self-calibrated result (a), except when $\beta = 1^\circ$ (b), which had unacceptably high artifact. The β value chosen for all subsequent experiments was 4° , which struck a compromise between increase of tissue signal and reduction of tissue contrast.

Table 2 shows results of the tests of Hypotheses A and B. Note that evaluations were done and results are presented for both SENSE and GRAPPA reconstructions. For Hypothesis A the p-values exceeded 0.05 for all cases, and thus the hypothesis was proven for these cases considered. Hypothesis B was proven for all cases at the higher acceleration ($R = 4, 6$). For accelerations of $R = 3, 4$, Hypothesis B was proven only for CNR of the *in vivo* studies using SENSE. Fig. 6 illustrates the results from the sets of higher acceleration, which shows histograms of Δ SNR in the phantom (a) and *in vivo* (b) studies.

Fig. 7 shows images from Sets III and IV of a subject from the *in vivo* studies. At the same nominal R , the IRES images appear visually similar to the standard self-calibrated images (b vs. a, and e vs. d). When acquired at a smaller R but with the same acquisition time, the IRES images appear less noisy than the standard self-calibrated images, and this difference is more pronounced at the higher acceleration of $R = 4$ vs. 6 (e vs. f) than at $R = 3$ vs. 4 (b vs. c). The contrast between WM and GM appears similar in all images.

Fig. 8 shows phantom results comparing the effects of incorporating calibration data for image reconstruction with GRAPPA. As seen in (a), for all accelerations as r_c^2 increases the SNR of standard self-calibration increases, and all three curves approach the same limit as r_c^2 approaches unity, which corresponds to no net acceleration. For this work, r_c^2 (r_c) values of 0.09 (0.3) and 0.12 (0.35) were used, as noted in (a), and for this range of r_c^2 the mean SNR differences between standard self-calibration and IRES are small ($<10\%$). Furthermore, these SNR differences were not statistically significant in the actual experiments as summarized in Table 2. Fig. 8b shows the same data as (a) but plotted against R_{net} , and shows that the R_{net} of IRES stays invariant with respect to r_c^2 .

DISCUSSION

We have shown the feasibility of acquiring calibration data within an MP-RAGE pulse sequence using the technique we call “Inversion Recovery with Embedded Self-calibration” or IRES. The basis of the method is to acquire the calibration data during the recovery interval after acquisition of the imaging data, an interval normally used simply for magnetization recovery. As such, IRES does not require use of a separate calibration scan. At the same time, it is not subject to the loss of acceleration that is intrinsic to standard self-calibration. The feasibility of IRES was demonstrated by showing either comparable SNR with time-savings or higher SNR at equal acquisition time relative to standard self-calibration. The time-savings and SNR benefits were more significant at higher R factors (Sets II and IV).

The IRES method was tested in the two scenarios of what we call “moderate” acceleration of $R = 3$ to 4, and “high” acceleration where R ranged from 4 to 6. Both hypotheses were confirmed to be true for this latter case. Specifically, Hypothesis B was shown for a net acceleration of four. One factor contributing to this positive result was that to attain a net acceleration of four, the nominal acceleration for a standard self-calibrated scan must be higher (six), imposing a typically higher geometry- or g-factor (3). Specifically, in the *in vivo* acquisitions the g-factor of $R = 6$ results were on average 76% higher than that of $R = 4$ (2.76 vs. 1.57); the maximum g-factors of $R = 6$ scans were at least three times higher than that of $R = 4$ scans (45.2 vs. 12.2). Further, when Hypothesis B was not met at $R_{net} = 3$, that was a comparison between 1D acceleration (IRES $R = 3$) and 2D acceleration (standard self-calibration $R = 4$), whereby 2D accelerations can be better conditioned than 1D ones (5). The *in vivo* g-factor of $R = 4$ results were higher than that of $R = 3$ by only 22% on average and by only 75% for maximum g-factors.

The primary limitation of IRES is the alteration of MP-RAGE signals and contrast by about 10% as shown in Fig. 4. This was caused by the addition of β pulses that perturb longitudinal magnetization. To restore the original signals and contrast level to that of unaccelerated MP-RAGE, results from simulations show that it is possible to increase TC (by increasing TD) and hence the IRES acquisition time also by about 10%. However, the results from Table 2 and the images shown in Fig. 7 suggest that these perturbation effects were not significant. Furthermore, in the experiments the maximum n_β was 82, which was smaller than the maximum allowable n_β of 100. For R values larger than the highest $R = 4$ used here for IRES, it may be necessary to extend TD to accommodate the increase in n_β .

Because the calibration data are acquired separately in IRES and are excluded from the raw image data, IRES does not benefit from the incremental SNR provided by such points compared to standard self-calibrated GRAPPA. What this means is that for the same nominal acceleration R , IRES should have inferior SNR to standard self-calibration as seen in the results of Fig. 8a. Expressed another way, Hypothesis A should in general be false. However, for this work on MP-RAGE, even with the number of trials used in the experiments, the SNRs of IRES and standard self-calibration were indistinct (Table 2 and Fig. 6). Two factors contributing to proving Hypothesis A might be the slightly higher signal levels and flatter signal modulation with IRES (Fig. 4a, c). Also, as noted above, small r_c^2 values were used for which case expected SNR differences between standard self-calibration and IRES are small.

With 2D parallel acceleration and improved coil designs for parallel imaging, acceleration factors of four and above have been shown to be clinically feasible (15–18). While IRES with a ‘high’ acceleration of four was demonstrated, further time-saving benefits of IRES can be evinced at even higher accelerations. Also, IRES may be applicable to other 3DFT,

IR-based imaging in applications such as carotid plaque characterization (19), liver imaging (20–21) and non-contrast-enhanced MR angiography (22–25). As with 3D MP-RAGE, only one 3D data set is formed in each of these applications (19–25), and acquisition of calibration data is a major fraction of the total scan time. Elimination of this time as provided by IRES can be significant. An example of IRES applied to plaque characterization is shown in Fig. 9, in which an MP-RAGE sequence was modified for axial, black-blood imaging at 3T (TC/TI/TR/TE = 1950/900/12.8/3.2 msec, $\alpha/\beta = 12^\circ/4^\circ$, 6-element phased array) and fat suppression pulses were added. In contrast to the brain MP-RAGE of this work, because acquisitions are axial, only 1D parallel imaging is possible. The results follow the same trends demonstrated above for MP-RAGE, with IRES having similar image quality vs. standard self-calibration for less acquisition time (b vs. a) or superior quality for equal time (b vs. c).

CONCLUSION

IRES embeds self-calibration acquisitions within the delay times of inversion recovery sequences to maintain the nominal acceleration in parallel imaging. At nominal accelerations in the range of four to six, it was shown experimentally that IRES as integrated into MP-RAGE can provide either time-savings at comparable SNR, or SNR benefits at the same acquisition time.

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Figure 1.

Illustrations of sampling patterns with different calibration schemes for $R = 4$, $R_Y = R_Z = 2$.

(a) Parallel imaging without calibration, where the red squares indicate acquisitions of the accelerated sequence within the shaded area that excludes the corners of k-space. (b) Standard self-calibration, where the central shaded region indicates the self-calibrated k-space center with calibration ratio or radius r_c . (c) Separate calibration or IRES calibration, where crosses indicate acquisitions in the calibration sequence, which overlap with some of the accelerated acquisitions.

Figure 2. Net acceleration R_{net} as a function of the nominal acceleration factor R and the calibration ratio r_c in standard self-calibration.

Figure 3.

Pulse sequence diagrams comparing (a) standard MP-RAGE that has a constant flip angle α , and (b) IRES MP-RAGE that has an additional set of repetitions with flip angle β for calibration. (c) Longitudinal magnetization of white matter simulated with $\alpha = 8^\circ$ and $\beta = 4^\circ$, showing that IRES closely matches the transient behavior of standard MP-RAGE.

Figure 4. MP-RAGE signal simulations with different β . (a) Individual WM (dashed line) and GM (solid line) signals, and (b) WM-GM contrast are plotted against the α repetition number.

Figure 5. Phantom images in the 2D phase-encoding plane, reconstructed with GRAPPA, acquired with nominal $R = 4$ and (a) standard self-calibration; (b) IRES $\beta = 1^\circ$; (c) IRES $\beta = 4^\circ$; and (d) IRES $\beta = 8^\circ$.

Figure 6.

Histograms of normalized difference in SNR of GM (GRAPPA reconstruction) used to test Hypotheses A and B from (a) phantom studies Set II and (b) *in vivo* studies Set IV. These histograms correspond to the entries in *italics* in Table 2. In both (a) and (b) ΔSNR was not significantly larger than zero for Hypothesis A but was significantly larger than zero for Hypothesis B.

Figure 7. MP-RAGE axial SENSE images of a subject acquired with standard self-calibration (SC) and IRES ($r_c = 0.35$, $r_c^2 = 0.12$ in all acquisitions). Images from Set III and Set IV are in the top and bottom row respectively.

Figure 8.

SNR characterization of standard self-calibration (SC) (+) and IRES (o) for various acceleration factors and calibration ratios r_c . (a) SNR against r_c , and (b) SNR against net acceleration of phantom-simulated GM. Arrows in (b) point in the direction of increasing r_c .

Figure 9. Accelerated, axial, black-blood imaging showing the right internal carotid artery just distal to the carotid bifurcation (a, arrow) as acquired with (a) standard self-calibration (SC) $R = 3$, (b) IRES $R = 3$ and (c) SC $R = 4$.

Sets of acquisitions, parameters of standard self-calibration (SC) and IRES calibration, and the indicated pairs of acquisitions for hypothesis testing. The calibration ratio is denoted by r_c .

Table 1

		Acquisitions			$r_c^2 (r_c)$	Hypotheses	
		1 (SC)	2 (IRES)	3 (SC)		A	B
Phantom	Set I	$R = 3$ Time=2.1 min	$R = 3$ Time=1.7 min	$R = 4$ Time=1.7 min	0.12 (0.35)	Acq. 2 vs Acq. 1	Acq. 2 vs Acq. 3
	Set II	$R = 4$ Time=1.6 min	$R = 4$ Time=1.3 min	$R = 6$ Time=1.3 min	0.09 (0.3)		
Human <i>In vivo</i>	Set III	$R = 3$ Time=3.6 min	$R = 3$ Time=2.9 min	$R = 4$ Time=2.9 min	0.12 (0.35)		
	Set IV	$R = 4$ Time=2.9 min	$R = 4$ Time=2.2 min	$R = 6$ Time=2.2 min	0.12 (0.35)		

Table 2

P-values from one-sided matched pairs t-tests of phantom and *in vivo* human studies. The four results shown in italics correspond to the histograms presented in Fig. 6.

		p-values											
		Set I (Phantom, $R = 3, 4$)						Set II (Phantom, $R = 4, 6$)					
Tissue	(SENSE/GRAPPA):	Hyp. A		Hyp. B		Hyp. A		Hyp. B		Hyp. A		Hyp. B	
		S	G	S	G	S	G	S	G	S	G	S	G
Fat SNR		0.41	0.44	0.34	0.40	0.40	0.40	0.42	0.42	0.42	0.42	<0.01*	<0.01*
WM SNR		0.44	0.44	0.33	0.35	0.46	0.47	0.47	0.47	0.47	0.47	<0.01*	<0.01*
GM SNR		0.41	0.45	0.33	0.34	0.43	0.49	0.49	0.49	0.49	0.49	<0.01*	<i>0.016*</i>
Set III (<i>In vivo</i> , $R = 3, 4$)													
Set IV (<i>In vivo</i> , $R = 4, 6$)													
Tissue	(SENSE/GRAPPA):	Hyp. A		Hyp. B		Hyp. A		Hyp. B		Hyp. A		Hyp. B	
		S	G	S	G	S	G	S	G	S	G	S	G
WM SNR		0.48	0.39	0.13	0.25	0.46	0.49	0.49	0.49	0.49	0.49	<0.01*	0.044*
GM SNR		0.40	0.43	0.18	0.24	0.33	<i>0.34</i>	0.33	<i>0.34</i>	0.33	<i>0.34</i>	<0.01*	<i>0.048*</i>
CNR [^]		>0.05	>0.05	<0.05*	>0.05	>0.05	>0.05	>0.05	>0.05	>0.05	>0.05	<0.05*	<0.05*

Hypothesis A: $\Delta\text{SNR} = 0$ at the same nominal R

Hypothesis B: $\Delta\text{SNR} > 0$ at different R but at the same R_{net}

* Statistically significant ($p < 0.05$)

[^] Wilcoxon signed rank test with critical values