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BRAIN IMAGING WITH IMPROVED ACCELERATION AND SIGNAL-TO-NOISE RATIO AT 7 TESLA OBTAINED WITH 64 CHANNEL RECEIVE ARRAY

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Abstract

PURPOSE: Despite the clear synergy between high channel counts in a receive array and magnetic fields 7 Tesla, to date, such systems have been restricted to a maximum of 32 channels. Here, we examine SNR gains at 7 Tesla in unaccelerated and accelerated images with a 64 receive channel (64Rx) RF coil.

METHODS: A 64Rx coil was built using circular loops tiled in two separable sections of a close fitting form; custom designed preamplifier boards were integrated into each coil element. A 16-channel transmitter arranged in two rows along the z-axis was employed. The performance of the 64Rx array was experimentally compared to that of an industry standard 32 channel receive (32Rx) array for SNR in unaccelerated images and for noise amplification under parallel imaging (PI).

RESULTS: SNR gains were observed in the periphery but not in the center of the brain in unaccelerated imaging compared to the 32Rx coil. With either one and two dimensional undersampling of k-space or with slice acceleration together with one-dimensional undersampling of k-space, significant reductions in g-factor noise was observed throughout the brain, yielding effective gains in SNR in the entire brain compared to the 32Rx coil. Task-based fMRI data with 12-fold two-dimensional (slice and phase-encode) acceleration yielded excellent quality functional maps with the 64Rx coil but was significantly beyond the capabilities of the 32Rx coil.

CONCLUSIONS: The results confirm the expectations from modeling studies and demonstrate that whole brain studies with up to 16-fold, two-dimensional acceleration would be feasible with the 64Rx coil.

INTRODUCTION

Introduction of 7 Tesla (7T) for human imaging was initially motivated by the push to achieve high resolution functional maps in the human brain at the mesoscopic scale of

elementary computational units, such as cortical columns and layers (1). This objective is increasingly pursued by a multitude of research centers (e.g. reviews (1–5) and references therein) as the number of installations of commercially available 7 Tesla instruments grows rapidly. Such high-resolution functional imaging (fMRI) studies are typically performed by restricting the field-of-view (FOV) aggressively to a small subsection of the cortex using techniques like inner volume selection (e.g. (6–8)). However, there is mounting interest in expanding high resolution, ultrahigh magnetic field (UHF)¹ studies to the entire human brain for fMRI based detection of task induced activation and functional networks (9–11), and for elucidation of cortical connectivity using diffusion weighted imaging (dMRI) (12–14). Similarly, high resolution UHF imaging of brain anatomy, covering the entire brain or large sections of the brain, is finding increasing use in clinical and research applications (e.g. reviews (15–17) and references therein), enabled by the higher SNR (18,19) and, in many cases, higher contrast-to-noise ratio (20,21) provided by UHF.

Increases in ultimate intrinsic SNR (uiSNR) (22,23) expected with higher magnetic fields (22,24,25) are critical for the afore-mentioned high-resolution studies. However, capturing this SNR advantage optimally requires array coils composed of multiple distributed small receive elements (see review (26)). There are strong synergies between image SNR, ultrahigh magnetic fields, and parallel imaging (PI) performance achieved through the use of array coils, particularly in whole brain investigations where acceleration through PI techniques becomes indispensable for efficient volume coverage (e.g. (9,12,13,27–30) and references therein). The use of PI naturally degrades the SNR due to g-factor noise (31); this degradation, however, is less for the same acceleration at ultrahigh magnetic fields (24,32–34). Larger number of elements in the array can also mitigate the g-factor noise; magnetic field magnitude also plays a critical role in this approach (32–34): As the size of the coils progressively decreases with increasing channel count in the array, thermal noise from the coil become dominant, thus restricting the approach to uiSNR; this problem however, is less pronounced at UHF as the thermal noise from the sample can remain the predominant contribution (35).

Despite these well-recognized synergies between UHF and high channel count receive arrays, until recently, UHF instruments were available only with 32 receive channels, restricting the experimental exploration of higher number of receive elements to lower field strengths (26,36,37). In this paper, we report results for the first time from a 64 channel receive (64Rx) coil for brain imaging at 7 Tesla, enabled by a recent 64 channel receiver implementation on our 7 Tesla system (38). The results confirm the expectations from modeling studies with spherical phantoms, demonstrating SNR gains in the periphery but *not* the center of the brain in unaccelerated imaging compared to an industry standard 32 channel receive (32Rx) coil while yielding substantial SNR gains in the entire brain, including in the center, with accelerated imaging.

¹For human imaging, ultrahigh magnetic field (UHF) is defined as 7 Tesla and higher.

METHODS

64 channel Receiver:

All data were acquired on a 7 Tesla system based on a standard MAGNETOM 7T (Siemens Healthcare, Erlangen, Germany) console interfaced to a 90 cm bore 7T magnet (Magnex Scientific Limited, Abington UK (the first 7T human magnet produced)). This console is normally equipped with up to 32 receivers. This receiver system was expanded to 64 channels by adding an additional cabinet in the equipment room and a second 32-channel RF switch matrix at the service end of the magnet to provide power for preamplifiers and secondary gain (38). Similar modifications have been demonstrated previously at 3T (39). For improved usability compared to previous implementations, however, a custom controller area network (CAN) device was developed and installed to automatically route the system CAN bus signals such that the component modifications are transparent to the operator. Digital receiver cards were added to the standard image reconstruction computer to receive the digitized signal, and software patches were implemented to support simultaneous acquisition with all receivers, and to allow for the image reconstruction processes to function fully in real time with any combination of active receivers (38).

The 64 channel receive coil:

We utilize a 16 channel, dual row, loop transmit only array (16Tx) combined with a close fitting 64 receiver array coil (64Rx) (Fig.1), with the coils distributed to cover the brain (Fig. 1B), allowing superior visual presentation and in-bore eye and motion tracking (Fig.1A). In this paper, we focus only on the receive performance of this coil, particularly targeting SNR gains in highly accelerated imaging with increased number of receive elements; as such, the transmitter does not necessarily represent an optimal design, but rather a sufficient design for the purpose at hand. Evaluation and optimization efforts on the transmitter are currently ongoing and will be presented in the future. Nevertheless, the transmitter configuration is also described. The parameters calculated for SNR and g-factor noise and the comparisons presented are calculated to be independent of the transmitter capabilities.

The coil consists of a lower section and a removable upper section (Fig.1), each with separate layers of transmit only and receive only loops (Supporting Figure S1). Supporting Figure S2 illustrates the circuit diagram for both the receive and transmit elements, with additional information on the construction and performance of the coils given in the legend of this figure. The 16 transmit loops, built from 1/4" copper tape, are arranged in a dual row configuration (40–43) with the lower section containing 11 (6+5) overlapping loop transmitters (~10 cm × 11 cm each) and the upper section 5 (3+2) transmitters. The dual row design was chosen because it has been shown to enable the most effective SAR minimization with slice-selective parallel transmit spoke pulses optimized with SAR constraints (44,45). Typical values for unloaded to loaded Q (Q_U/Q_L) for the transmitter coils were ~225/20.

The influence of the receive array on the transmitter was evaluated by collecting data with and without the receiver inserted in the transmitter, in the latter case using the transmitter loops also as receivers; we observed minimal increase in required transmit power (< 10%).

When inserting the (detuned) receive array we consistently observed a ~2-3 MHz shift in resonance frequency as well as a modest change in matching characteristics; this is illustrated in Supporting Figure S3 using S11 plots corresponding to a typical (Supp. Fig. S3A) and worst case (Supp. Fig. S3B) channel response to loading with the detuned receive array.

The 64 receiver loops are mounted directly on the inner former closest to the subject and varied in size between ~4 cm to ~5.5 cm in diameter. There are 48 loops in the lower former and 16 in the upper former (Fig. 1C, 1D and Supporting Fig. S1). Loops are constructed from a single segment of enamel coated 18 AWG copper wire and tuned at the feed point with variable and fixed capacitors. Balanced impedance matching is achieved using a lattice balun transformer. Loops are actively detuned during transmit via a pair of MA4P7470F-1072T (MACOM) PIN diodes. The 48 lower section loops are arranged in 7 overlapping rows (8-9-10-9-7-3-2); the upper receiver loops are configured in 3 rows (7-6-3). All receive elements are arranged as overlapping loops (36,37) and the housing seam between the upper and the lower sections was designed to allow for overlap decoupling between lower and upper section coils. To support on-coil pre-amplification, a small tune/match/decoupling board was developed utilizing a low noise preamplifier (WanTcom, WMA7RA, Minneapolis, MN) with a low input impedance of 1.5 ohms. Shielding of the preamplifier board allowed for operation within close vicinity of the transmit arrays and suppressed oscillation. A phase shifter realized in lumped elements on the preamplifier board supports fine-tuning for preamplifier decoupling (46,47). Supporting Figure S4 displays a Noise correlation matrix among the 64 channels, illustrating the degree of coupling among these receive elements. Most of the coil correlations were low, in the range 0.25 or less, with majority of these in the 0.1 or less range. However, there were coil pairs that displayed significantly higher (~0.5 - 0.6) near neighbor correlations. Temporal stability of the receive array was also employed as a routine evaluation tool to check the coil performance. An example of this is shown in Supporting Figure S5.

We achieved receiver detuning on the order of -40dB when compared between tuned and detuned stated and measured through the preamplifier output.

Data acquisition:

MRI data were collected from six subjects (5 male, ages 24 to 44) after they provided written informed consent. The experimental protocols conformed to safety guidelines for MRI and were approved by the Institutional Review Board of the University of Minnesota. An industry standard 1-channel transmit, 32-channel receive (1Tx/32Rx) coil by Nova Medical Inc. (Wilmington, MA) was employed for comparisons. The data were acquired both with the 64Rx and the 32Rx coils on each individual using the same acquisition parameters. The only difference was that the 64Rx coil, having a 16Tx transmitter, needed B1 shim solutions or, slice specific spoke pulses (Fig.3).

Anatomical Imaging: Whole brain T2*-weighted (T2*w) Images were acquired at 1-mm isotropic resolution using a 2D multislice GRE sequence and with the following parameters: repetition time (TR) = 2450 ms, echo time (TE) = 15.0 ms, maximum flip angle = 19°,

matrix size = 220×220, field of view (FOV) = 220×220 mm², 128 interleaved oblique axial slices ($T > C - 22.0$), bandwidth = 320 Hz/pixel and total acquisition time = 8 min 59 sec. To improve flip angle uniformity, RF shimming was conducted following the slab-wise design framework (48). Specifically, eight sets of 16-channel RF shim values (corresponding to RF magnitude and phase modulations) were calculated to improve the uniformity of the RF distribution across the brain. This was done by performing multichannel B1+ mapping in 32 oblique axial slices, contiguously covering the brain in the slice direction. These 32 B1+ mapping slices were further grouped into eight slabs, with each slab having four neighboring B1+ mapping slices. A slab-specific RF shim set was calculated for each slab to create uniform RF distribution inside the slab by considering jointly the multichannel B1+ information obtained in all of the four B1+ mapping slices. During image acquisition, each of the eight slab-specific RF shim sets were applied repeatedly to excite the image slices residing in the corresponding slab.

The multichannel multislice B1+ mapping was obtained using a hybrid B1+ mapping method (49,50). The eight slab-specific RF shim sets were calculated jointly by formulating a regularized magnitude least squares problem (51) and solving it based on a two-step procedure for improved robustness, an approach similar to that in (13). Further, the RF shim calculation only considered brain-extracted B1+ maps serving as the ROI, defined from T1-weighted (T1w) whole-head MPRAGE images (52) by applying the FSL's brain extraction tool (53). The T1w image was acquired with matched FOV with the multichannel B1+ mapping, using 1-mm isotropic resolution, TR/TE/TI=2300/3.25/1300 ms, nominal flip angle=6°, 2-fold in-plane GRAPPA, partial Fourier=6/8 (PE), bandwidth=180 Hz/pixel and total acquisition time=3 min 56 sec. The inversion pulses of the MPRAGE sequence employed adiabatic pulses (54,55). The eight calculated RF shim sets were subsequently utilized to acquire the whole brain T2*w image

The 1-mm whole brain T2*w images were obtained with a standard 2D multislice, gradient-recalled echo (GRE) sequence. During image acquisition, the sequence sent a TTL signal from shot to shot for a phase/gain controller to dynamically adjust the RF magnitudes and phases of individual transmit channels (56,57) consistent with the order by which the interleaved slices were excited. As a result, each of the eight slab-specific shim sets was applied 16 (number of slices per slab) × 220 (number of phase encodes per slice) times to excite a total of 128 slices covering the whole brain. The flip angle was set to 19° at the maximum in the brain.

Data for SNR calculations: Images for SNR calculations were acquired with a 2D GRE sequence, using a long TR to virtually reach full relaxation in gray and white matter at 7T in 80 interleaved axial slices covering the whole brain (TR=10,000 ms, TE=3.8 ms, full bandwidth=87.04 kHz, matrix size=84×256, field of view=256×168 mm, voxel size=2.0×1.0×2.0 mm). Following recommendations in (58), noise images were acquired using identical parameters except the RF excitation that was off (flip angle=0°) and a shorter TR of 1000 ms was employed. At UHF, covering the whole brain with a multi-channel transmit array while avoiding local dropouts in transmit B₁ (B₁⁺) is often obtained by combining two data sets acquired with complementary B₁ shim patterns (e.g. a CP-like and a CP2+ like modes, with high RF efficiency in the center and in the periphery of the brain, respectively).

In practice, it was possible with the 16Tx/64Rx RF coil to only include data acquired with one B_1 shim solution aiming at a tradeoff providing sufficient RF efficiency in the center while mitigating B_1^+ dropout in the periphery of the imaged slices (59). With only one transmit channel, the 32Rx Nova RF coil was not subject to B_1 shimming. For both RF coils, prior to the acquisition of the GRE data for SNR calculations, an absolute transmit $|B_1^+|$ map was obtained, using the 3D Actual Flip Angle (AFI) sequence (60). For the 16Tx/64Rx RF coil, the same B_1 shim setting was used for B_1 mapping and GRE acquisition for SNR calculations. For both RF coils, the RF voltage was then adjusted to reach a nominal flip angle of 90° in the central region of a brain with the GRE sequence. Complex raw data were exported to an external server for computational purpose.

Data for g-Factor maps: Images were acquired with a 3D GRE sequence (repetition time = 6.7 ms, echo time = 3.0 ms, flip angle = 5° , matrix size = $256 \times 176 \times 256$, FOV = $256 \times 176 \times 256$ mm, voxel size = 1.0 mm isotropic) covering the whole brain. In order to avoid the possibility of having regions of very low flip angle resulting in low SNR “dark” spots in the image, data used for g-factor calculations for the 64Rx coil were preemptively acquired with two fully sampled complementary B_1^+ shim solutions where such dark spots, if present, would be in different locations in the sample. This is similar to the concept employed in TIAMO approach (61), except that we did not follow the TIAMO approach of using a GRAPPA reconstruction treating the two sets of images as virtual channels. Rather, in our case, the data from these two solutions were added together in image space after accounting for the spatially varying relative phase difference between the two acquisitions. The two complementary solutions were a CP-like and a CP2+-like solutions, calculated on a central slice using phase only shimming (62); however, g-factor noise calculated using these two solutions was not different from those calculated with a good CP-like solution alone, as in the case of the data acquired for calculating unaccelerated SNR (see section above).

Functional Imaging Acquisition: Data were collected using GRE EPI at 0.8 mm isotropic resolution (matrix size = 162×200 ; 62 coronal slices with Right(R)-Left(L) phase; Multiband = 2; TR = 2 s; TE = 23 ms; 6/8 Partial Fourier; echo-spacing = 1.0 ms; k-space undersampling of factor of 6 (i.e. Reduction factor $R = 6$) along the PE direction. For the 16Tx/64Rx, the relative phases of the coil were adjusted to achieve a B_1 solution (62) that optimized uniformity and efficiency throughout early visual regions. For both coils, flip angles were mapped at the beginning of the scanning session and the transmit voltage was adjusted to provide the target flip angle in occipital cortex.

In each experiment the participant freely viewed clips of an animated movie (12 s on/12 s off, 20.5 cycles per scan). This 8-minute scan (246 volumes) was repeated 4 times in each experiment while fMRI data were collected. During each experimental session, GE EPI images (30 volumes) were also acquired with reversed phase encode to enable B_0 -induced distortion compensation during pre-processing.

SNR calculations:

The two receive arrays were compared for unaccelerated SNR, and g-factor noise for accelerated acquisitions.

Unaccelerated SNR: Reconstructed images of fully sampled data of the 32Rx or 64Rx were merged using the standard square root of the sum of the squares (SRSS) formulation (46). Prior to SRSS combination, a 32×32 or 64×64 noise covariance matrix, derived from the complex noise data, was used to de-correlate all SNR data, subsequently expressed in standard deviation units (63). The rician distribution of magnitude MR images, a well-known source of bias (64,65) yielding an SNR offset, is a non-linear function of the number of channels. However, in the current study, all voxels presented a sufficiently large SNR, resulting in a relatively small offset bias, which was ignored as previously suggested (46).

The MR signal of a GRE sequence in steady state is given by the standard Bloch equation:

$$SNR \propto \rho \sin \alpha \frac{(1 - e^{-TR/T1})}{(1 - \cos \alpha e^{-TR/T1})} e^{-TE/T2^*}. \text{ Here, we use the "proton density" variable } \rho \text{ to}$$

represent the fraction (unit-less) of tissue volume occupied by water detected in our GRE acquisition. Typical values for ρ are 1.00, ~0.82 and ~0.7 for CSF, gray matter and white matter respectively (66).

With a TR of 10 seconds, $e^{-TR/T1} \ll 1$, so that the impact of the $|B_1^+|$ spatial heterogeneity on SNR observed at 7T comes only from $(\sin \alpha)$ where α is the measured actual flip angle map rescaled to the voltage applied during the GRE acquisition; then SNR (still in standard deviation units) is extrapolated to the nominal value it would achieve when $\alpha = 90^\circ$ with $SNR_{90} = \frac{SNR}{\sin \alpha}$, which can be directly compared between RF coils for a given subject,

keeping all acquisition parameters the same. As previously described (67), and based on (68), SNR_{90} (in standard deviation units) can be further converted to estimate the intrinsic SNR in (Hz)/ml units, thereby providing further generality enabling comparisons to acquisitions obtained with different field of view, resolution, bandwidth and number of average. The estimated intrinsic SNR in Hz/ml units, Ψ , of the ensemble [RF receive array + preamplifiers] can be expressed as:

$$\Psi = \frac{SNR}{\sin \alpha} \cdot \frac{F\sqrt{BW}}{V\sqrt{N_{RO}N_{PE}NEX}} \cdot \frac{1}{\rho e^{-TE/T2^*}} = SNR_{90} \cdot \frac{F\sqrt{BW}}{V\sqrt{N_{RO}N_{PE}NEX}} \cdot \frac{1}{\rho e^{-TE/T2^*}}$$

where SNR is the measured signal-to-noise ratio in standard deviation units, F is the unitless noise figure of the rest of the receive chain (after the ensemble [RF receive array + preamplifiers]), V is the voxel volume in ml, N_{RO} and N_{PE} are the number of samples along readout and phase encoding axis (considering a 2D acquisition), NEX the number of acquisitions and BW the sampling bandwidth in Hz (covering $\pm$ Nyquist frequency). In practice, we did not correct for $\rho e^{-TE/T2^*}$, so our SNR results are shown as $\Psi_{TE\rho F}$ where

$$\Psi_{TE\rho F} = \frac{\Psi \cdot \rho \cdot e^{-TE/T2^*}}{F} \text{ with TE=3.8ms. The noise figure } F, \text{ which is independent of}$$

which RF coil is used, does not alter the SNR comparison between the two arrays.

g-factor noise calculations: These were performed using the fully sampled whole brain data acquired with the 3D GRE sequence. The various different undersampling schemes were generated by post-processing. The multi-channel k-space data were first decorrelated

using a Singular value decomposition of thermal noise acquired with RF pulse magnitudes set to zero. Sensitivity profiles S_j were extracted from the decorrelated, channel specific reconstructed images I_j as $S_j = I_j / \sqrt{\frac{1}{N} \sum_n |I_j|^2}$, filtered with a Gaussian filter using a full width half maximum (FWHM) of 8 pixels applied to S_j and furthermore defined to be zero for all voxels where the SNR of $\sqrt{\frac{1}{N} \sum_n |I_j|^2}$ was less than 3.

The calculation of g-factors was performed using the equation derived by Pruessmann (31) as $g = SNR^{full} / (\sqrt{R} SNR^{red})$, where R is the reduction factor relative to a fully-sampled acquisition.

For simulated Simultaneous Multi Slice (SMS)/Multiband (MB) experiments, the sensitivity profiles were shifted relative to the FOV in the phase encode dimension (29,69,70), and the same equation was used for g-factor evaluation. The relative FOV shift was 1/3 the FOV between adjacent slices, except for a phase-encoding undersampling of 3 where the relative shift was 1/4 the FOV between adjacent slices.

After calculation of the g-factors for the 3D volume, g-factor histograms were determined, and the mean, median and 98 percentile maximum calculated. For g-factor maps, a g-factor_{MIP} images was extracted as a Maximum intensity projection in the LR direction, over a 80mm thick slab in the sagittal plane and displayed over a silhouette of the central sagittal slice. For group averaging, g-values for the calculated mean, median and 98 percentile maximum were averaged over all subjects.

Calculation of Functional Images:

Pre-processing was done using tools provided by AFNI (<https://afni.nimh.nih.gov/>). Motion-compensation parameters were calculated by registering (3dvolreg) each volume to the first volume of the scanning session. A voxel displacement map was computed from the average of the first functional scan and the EPI images acquired with opposite phase encode directions (3dQwarp, -plusminus flag). Distortions due to known gradient non-linearities were calculated using the grad_unwarp.py toolbox provided at <https://github.com/Washington-University/gradunwarp>. The B₀-compensation and gradient nonlinearity-compensation WARP maps were combined with motion-compensation matrices to create preprocessed data that were motion- and distortion-compensated in a single resampling step.

Responses to visual stimuli were estimated with a General Linear Model (AFNI: 3dDeconvolve), after pre-processing the data and smoothing the images with a 2D Gaussian kernel (FWHM = 0.8 mm). First, the four identical scans were averaged together. Then, coherence (unsigned correlation) with visual stimulus presentation was computed for each voxel as the ratio of the stimulus-related Fourier component (20 cycles/scan) to the square root of the integrated power spectrum of the time series. Activation was visualized after subjecting coherence maps to two thresholding steps: first, an uncorrected threshold of $p < 0.001$ was applied at each voxel; next, a Monte Carlo simulation (3dClustSim), which took into account the smoothness of the data (estimated by 3dFWHMx, using the new ACF method) was used to determine a minimum cluster size of 10 that corresponded to a false-positive rate of 10^{-6} .

RESULTS

Figure 2 displays SNR data for unaccelerated imaging over the brain for two subjects obtained with the 64Rx and 32Rx channel coils. For a more quantitative comparison of the differences, Figure 3 illustrates a Ratio of SNR images for a central slice for one subject (subject A shown in Figure 2), as well as plots from two lines traversing the head in the AP direction in this central slice located in each hemisphere (Figure 3B and 3C). Figure 3D shows the same data for one of the lines in a second subject. The SNR is similar in the two coils in the center of the head, showing variations of approximately $\pm 5\%$ as a function of space (Figure 3). Unlike the center, however, there were clear gains in the periphery (Figure 3); in a ~ 1 cc volume of interest located in the visual cortex, the SNR gains were 36% and 51% for subject A and B (Fig.2), respectively.

Figure 4 demonstrates the coverage provided by the 64Rx coil over the human brain; it displays representative slices of the 1-mm isotropic whole brain T2*w images obtained with the 64Rx coil with slice specific single spoke pulses designed for the 16 channel transmitter.

Parallel imaging performance is illustrated for the two coils in Figures 5 through 9 and in Table 1. Figure 5 displays 1/g-factor maps calculated for axial slices and presented as maximum intensity projections (MIPs) in the sagittal orientation from an 80 mm central sagittal slab; 1/g-factor maps are presented for a single subject for k-space undersampling of 4 and 5 in the left-right (LR) and anterior-posterior (AP) directions, and for two dimensional undersampling (LRxAP). The mean and the minimum 1/g-factor values, calculated from the average and maximum (98 percentile) g-factors, *averaged over all subjects*, are also given for each case in the lower left corner of the panels. Clearly, 64Rx coil outperforms the 32Rx coil substantially for the same acceleration rate. The 32Rx coil has acceptable performance with R=4 for phase encode undersampling in the AP direction, with mean and minimal 1/g factor of 0.88 and 0.67 respectively (mean and maximal g-factors of 1.14 and 1.5, respectively); 64Rx coil has clearly lower g-factor noise for the same acceleration and undersampling but in addition demonstrates similarly acceptable performance for R=5 for the AP undersampling (mean/max. g factors of 1.2/1.6). Both coils have a better performance (lower g-factor noise) when undersampling is in the AP direction.

Figures 6, 7 and 8 illustrate the acceleration capabilities of the two coils for simultaneous multi slice (SMS)/Multiband (MB) acquisition for slice acceleration in the presence of k-space undersampling in the PE direction.

Figure 6 displays data for different slice orientations and PE directions for a fixed two-dimensional acceleration of MBxR=7x2. As in Fig. 5, images are from a representative subject but the numbers given in the figure are 1/g values averaged over all subjects. Again, the 1/g-factor maps were calculated for the slices corresponding to the orientation specified and subsequently projected onto the sagittal orientation for display purposes. The 32Rx coil displays unusably poor performance in all orientations displayed (Fig.6), achieving in the best case (sagittal orientation) mean and minimum 1/g factors of 0.65 and 0.41, respectively (corresponding to factors of 1.54 and 2.44, respectively). However, 64Rx coil supports this

MBxR=7×2 acceleration with the best performance along the sagittal direction with g-factors of 1.24 and 1.72 (average and maximum, respectively).

An axial-coronal oblique plane is often used for slice orientation in whole brain imaging studies since it minimizes the number of slices needed (e.g. the Human Connectome Project (HCP) (9,12,30,71)). Figure 7 illustrates 1/g-factor maps for SMS/MB with increasing MB factors for R=2 calculated in this slice orientation and then projected onto the sagittal plane. MBxR=5×2 acceleration depicted in the left most column (Fig. 7) was employed in the HCP 7T resting state fMRI (rfMRI) data acquisition with the 32Rx Nova coil (12), and serves as a reference. Clearly, a significant reduction in g-factor noise is seen with the 64Rx receive coil. The mean, maximum (98 percentile) and the median g-factor noise calculated for oblique slice orientation as employed in Figure 7, averaged over all participants, are given in Table 1.

A more commonly used undersampling factor in the phase encoding dimension at 7T with 32Rx coils is R=3. Figure 8 illustrates what MB factors would be achievable with R fixed at 3 for the three different orientations selected for slice acceleration. Again both coils perform better with sagittal and coronal slices and 64Rx coil outperforms significantly the 32Rx coil.

Finally, the ability to achieve high two-dimensional accelerations for SMS/MB acquisitions is demonstrated further in Fig. 9 with task-based BOLD fMRI data obtained with GRE EPI using MBxR = 2×6. The figure shows fMRI images in color obtained with visual stimulation, superimposed on the gray scale GRE EPI images. The 32Rx coil shows *incomplete* unaliasing of the two slices that were excited simultaneously with the MB=2 pulse, so that the two slices appear superimposed on each other. This is not the case for the 64Rx coil. In addition, due to the increased g-factor noise, the 32Rx fMRI data appears much noisier with extensive non-specific apparent activation appearing outside the brain.

DISCUSSION

Particularly with the development of parallel imaging methods (31,72,73), the use of multi-channel receive coils (46,74) have become standard practice in clinical systems while significant research efforts continue to be focused on the benefits of increasing the number of elements in such arrays (review (26) and references therein).

Early modeling studies with spherical phantoms predicted that for unaccelerated imaging, a limited number of channels ranging from ~8 to ~32 captures most of the uiSNR in the center of the object; at the periphery, however, where uiSNR is substantially higher (24,34,35,75), the approach to the uiSNR requires increasing the number of elements significantly beyond 32 (35). The impact of such an increase in the number of elements is particularly beneficial at UHF (35). Consistent with these predictions, in our comparisons we have observed that central SNR did not change when going from the 32Rx to the 64Rx coil in unaccelerated imaging, while there were large gains observed in the periphery (Fig. 2 and 3).

Of course, this SNR comparison can at best be considered semi-quantitative because most likely there exists differences between the coils beyond the number of loops in the arrays, the loop dimensions, and their geometrical distribution. Nevertheless, the current data

demonstrate that there are gains to be had for unaccelerated imaging on the periphery of the brain relative to an excellent, industry standard 32Rx coil by going to a 64Rx “brain” array, consistent with what is expected from theory and simulations.

With the use of PI achieved through k-space undersampling, SNR is naturally reduced by $R^{1/2}$ due to the R-fold fewer points collected in k-space; in addition, however, the array SNR suffers degradation with PI due to the g-factor noise (31), which is particularly pronounced where the aliasing of the image is high; for a spherical object, this is located in the center. Increased number of coil elements is then required to minimize this noise amplification in order to approach uiSNR (35,75–77). Thus, for *accelerated imaging with k-space undersampling*, the achievable SNR even in the center of the object should improve with increasing number of elements beyond 32 in the array. Indeed, this is where the 64Rx coil substantially outperforms the 32Rx coil (Fig. 5). Although the g-factor noise tends to be lower at the periphery in accelerated imaging, it is still very high for the 32Rx coil with $R=4$, 5 or 4×4 ; in contrast, the 64Rx coil demonstrates substantial improvements over the entire head, reducing the g-factor to ~ 1 in the periphery, and suppressing the magnitude and the spatial extent of higher g-factors in the central regions. Especially for studies targeting the cortex rather than the central regions of the brain, $R=5$, or 4×4 or likely somewhat higher accelerations would be feasible with g-factor ~ 1 with the 64Rx array.

Image accelerations achieved through SMS excitation using MB pulses and subsequent unaliasing exploiting the spatial information inherent in an array has become the method of choice for slice-based rapid coverage of the human brain using fast sequences like Echo Planar Imaging (EPI) (27–30,78). While this approach can be employed without k-space undersampling at 3T, increased B_0 inhomogeneities at 7T necessitates the use of undersampling in the PE direction together with slice acceleration (30). In this approach, the MB factor becomes limited either by power deposition (i.e. Specific Absorption Rate (SAR)) or by the overall unaliasing capability of the receive array employed. In the latter case, if R needs to be >1 as is the case at 7T, it may be desirable to keep R to a low number (e.g. 2) and maximize MB in order to increase the temporal sampling rate, for example, to better extract the resting state networks from the spontaneous fluctuations in the brain (9,79,80) or to capture their temporal dynamics (81); such a strategy was employed in the 7T rfMRI component of the HCP (9) acquired with the 32Rx Nova coil, where the MB factor was maximized to a level compatible with power deposition limits, achieving $MB=5$, with R fixed at 2. Figs. 6, 7 and Table 1 demonstrate that using either the same oblique slice orientation employed in the HCP (fig. 7) or straight sagittal or coronal slices (fig.6), $MB\times R$ of 7×2 would be possible with the 64Rx coil with the g-factor noise constrained below 2. Relaxing the maximal g-factor threshold slightly would allow even $MB\times R=8\times 2$ (Table 1). Fig. 8 indicates that, with the 64Rx coil, it would also be possible to keep MB at 4 or 5 while increasing R to 3, thereby reducing the echo train length in an EPI acquisition and thus reducing distortions and signal losses associated with areas of high B_0 inhomogeneity.

RF power deposition can impose a severe limitation in slice acceleration in SMS/MB with power demanding spin echo sequences as employed in dMRI. In the HCP 7T dMRI acquisition, for example, MB was restricted to 2 due to power deposition; this enabled the use of a higher R factor of 3 with the 1Tx/32Rx Nova coil. The limitation on slice

acceleration can be significantly relaxed by the use of parallel transmit (pTx) MB pulses (82,83) explicitly designed to reduce power deposition (13,84–86). However, keeping $R=3$ while MB is increased with the use of a multi-channel transmitter again requires, in addition, a high performing receive array in order to handle the increased aliasing. Indeed, this is possible with the 64Rx array; MBxR of 3×3 , 4×3 , and even 5×3 with coronal and sagittal slices would be usable with maximum g-factor <2 . Favorable unaliasing performance with coronal and sagittal slices is fortuitous since the same slice orientations also provide the best performance in minimizing SAR and B_1 inhomogeneity in transmit arrays with azimuthally distributed elements (44).

In comparison to the 64Rx coil, MBxR of 5×3 acceleration is not supported by the 32Rx Nova coil; 3×3 would be feasible in the 32Rx coil equipped with multichannel transmit capability that allows the use of pTx pulses in order to reach MB=3; this has been demonstrated previously using sagittal slices with the 8Tx/32Rx version of this coil (13); although MB=4 with sagittal slices was also feasible with the Nova 8Tx/32Rx coil, the resulting dMRI data showed perceptible deterioration in tractography performance in the center of the brain even for MBxR= 4×2 (13).

Finally, the utility of 64-channel receiver capability is illustrated for high-resolution task based fMRI with an MBxR= 2×6 in the visual cortex (Fig. 9). This acceleration is simply not supported by the 32Rx NOVA coil, which shows improperly unaliased slices (indicated by the arrows in Fig. 9, lower panel) of the MB=2 excitation and increased false positives (clearly identifiable outside the brain) due to the increased g-factor noise. There are similar “activation” patterns recognizable in the two coils but the spatial location of the activation is confounded by the incomplete aliasing and g-factor noise.

Further improvements in this first generation 64Rx coil is possible. For example, suboptimal decoupling seen for some of elements in this coil (Supporting Fig. S4) leading to high noise correlation is expected degrade both SNR and parallel imaging performance. Efforts to improve the decoupling are currently being pursued, for example, by developing lower impedance preamplifiers, and more optimized cabling and coil layouts.

CONCLUSION

Image acceleration is critical for high resolution imaging enabled by inherent SNR gains provided by UHF, especially when coverage of the entire brain or a large subsection of the brain is desired, as in functional and anatomical connectivity studies. Despite this clear need, and the synergy between high channel counts in an array and UHF, 7T data to date has been restricted to a maximum of 32 receive channels. Using a 64 channel modification of the receiver chain in a 7T system, and 64Rx “brain” array coil we demonstrate that a 64Rx array provides significant advantages in SNR in the entire brain for acceleration in one or two dimensions, as well as SNR gain in the periphery of the brain in unaccelerated images compared to an industry standard commercial 32Rx coil. Nevertheless, these initial results ascertain the importance of going to such high channel counts at 7T, and anticipate, as do a recent study that employed the available 32 receiver channels to simulate a 256 channel receive coil (87), the feasibility of further potential gains with even larger number of

channels employed at UHF. Consequent increases in SNR in highly accelerated images promise major improvements in high spatio-temporal resolution data collected with the aim of understanding brain function, anatomy and connectivity in health and disease.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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FIGURE 1:

Panel A illustrates the assembled coil with a subject in the coil, and the coil-mounted, adjustable, wide-field-of-view screen for visual presentation. The two separable upper and lower sections of the coil are clearly visible in this picture. Panel B shows the receiver coil layout over a model of the human head. Panels C and D show the receiver coil layout in the upper and lower sections, respectively. Panels E and F illustrate the transmitters for the upper and lower sections, respectively. The transmitter coils in the upper section (Panel E) are on the distant side of the former indicated by the arrow; as such they are not visible in this picture.

FIGURE 2:
 SNR (in units of $(\sqrt{\text{Hz}} / \text{mL})$) over the human brain for the 64- and the 32-channel receive coils in two different subjects for unaccelerated imaging.

FIGURE 3:

A) Unaccelerated SNR Ratio (64Rx/32Rx) image for a central slice for Subject A shown in Figure 3. SNR values in the same central slice are also plotted along two lines running in the A-P direction, located in each hemisphere near the hemispheric fisher for the same subject (B and C); approximate positions of these lines are indicated by the arrows in (A). The same plot for one of those lines is displayed for a second subject in (D).

FIGURE 4:

Representative slices of the whole brain T2* weighted images obtained at 1-mm isotropic resolution with the 64Rx coil. The images were acquired using a standard 2D GRE sequence (TR/TE = 2450/15 ms, maximum flip angle = $\sim 20^\circ$) and slab-wise RF shimming. Eight slab-specific RF shim sets were calculated to create uniform RF distribution across the brain and were applied in real time through a phase/gain controller to excite a total of 128 oblique axial slices covering the entire brain.

FIGURE 5:

g-factor maps (shown as MIPs from an 80 mm central sagittal slab) superimposed on a silhouette of a sagittal slice for one dimensional 4- and 5-fold undersampling of k-space (i.e. reduction factor (R) = 4 and 5) in one phase-encoding dimension when the phase encode is running left-right (LR) and anterior to posterior (AP). Right-most column shows a 3×3 and 4×4 fold, two-dimensional undersampling of k-space. The mean and the maximal (at 98% level) g-factor numbers calculated as the average of all participants are given in the lower left corner for each figure. . The darkest red color in the g-factor maps represents g-factor of 3 and greater.

FIGURE 6:

g-factor maps for 7×2 fold 2D accelerated images with 7-fold slice acceleration (i.e. MB=7) together with 2-fold undersampling of k-space along the phase encode (PE) (i.e. R=2). Designation as Sagittal, Coronal and Axial indicate the slice acceleration direction. The associated phase encode undersampling s AP PE (for phase encoding in the Anterior-Posterior direction)), or as LR PE (for Left-Right dimension). The g-factor maps are shown as a MIP over an 80 mm sagittal slab superimposed on a silhouette of a single sagittal slice. The mean and the maximal (at 98% level) g-factor numbers are given in the lower left corner for each figure. Images are from a representative subject, and the mean and maximum g-factor numbers are averages over all participants.

FIGURE 7:

SMS/MB acceleration for different MB factors for simultaneously excited and acquired slices with $R=2$ (i.e. the one-dimensional undersampling of k-space in the phase encode dimension is kept at reduction factor of 2). The g-factor calculations are performed for oblique axial-coronal slices as employed in whole brain data acquisition in the Human Connectome Project and are presented as a MIP over an 80 mm sagittal slab superimposed on a silhouette of a sagittal slice. The mean and the maximal (at 98% level) g-factor numbers are given in the lower left corner for each figure. Images are from a representative subject, and numbers are averages over all participants.

FIGURE 8: SMS/MB acceleration for different MB factors for simultaneously excited slices with R=3 (i.e. the one-dimensional undersampling of k-space in the phase encode dimension is kept at 3). The g-factor calculations are performed for sagittal, coronal and axial slices and slice accelerations (labeled on the left side of the figure) with k-space undersampling in the pE direction performed on the Anterior-Posterior direction for sagittal and axial slices and in the Left-right direction for the axial slices. The data are presented as a MIP over an 80 mm sagittal slab superimposed on a silhouette of a sagittal slice. The mean and the

maximal (at 98% level) g-factor numbers are given in the lower left corner for each figure. Images are from a representative subject, and numbers are averages over all participants.

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FIGURE 9:

Task based fMRI for visual stimulation obtained with 6-fold undersampling in k-space along the phase-encode dimension ($R=6$) and 2 fold slice acceleration ($MB=2$) at two different slice locations. Note that the slices shown are not simultaneously excited by the MB2 pulse. However, each of the two slices shown were excited together with another slice simultaneously; these slices are fully unaliased for the 64Rx coil but not the 32Rx coil. The green arrows in the 32Rx data point to the boundaries of the brain in the two simultaneously excited slices in SMS/MB sequence with $MB=2$ that are not fully unaliased from each other in the 32Rx data.

TABLE 1:

The mean, maximum (98 percentile) and the median of the g factor calculated for the oblique slice orientation used in Figure 7 for a fixed acceleration of factor R of 2 along the phase encode direction, for different MB factors

		MB5	MB6	MB7	MB8
NOVA 32Rx	Mean	1.35	1.52	1.80	2.02
	Max	2.00	2.68	3.52	4.62
	Median	1.31	1.48	1.67	1.88
CMRR 64Rx	Mean	1.17	1.19	1.31	1.32
	Max	1.53	1.78	2.00	2.23
	Median	1.14	1.19	1.25	1.30