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SERIAL Transmit - Parallel Receive (ST_xPR_x) MR Imaging Produces Acceptable Proton Image Uniformity without Compromising Field of View or SAR Guidelines for Human Neuroimaging at 9.4 Tesla

Keith R. Thulborn¹, Chao Ma², Chenhao Sun⁴, Ian C. Atkinson¹, Theodore Claiborne¹, Reiner Umathum³, Steven M. Wright⁴, and Zhi-Pei Liang⁵

¹Center for Magnetic Resonance Research, University of Illinois at Chicago, 1801 West Taylor St., MC 707, Suite 1307, Chicago, IL 60612, USA

²Gordon Center for Medical Imaging, Department of Radiology, Massachusetts General Hospital, 125 Nashua Street, Suite 660, Boston, MA 02114, USA

³German Cancer Center (DKFZ), Division of Medical Physics in Radiology, Im Neuenheimer Feld 280, 69120 Heidelberg, Germany

⁴Department of Electrical and Computer Engineering, Texas A&M University, Wisenbaker Engineering Building, College Station, TX 77843-3128, USA

⁵Beckman Institute, University of Illinois at Urbana-Champaign, 405 North Mathews Avenue, Urbana, IL 61801, USA

Abstract

Purpose—Non-uniform B1⁺ excitation and high specific absorption rates (SAR) compromise proton MR imaging of human brain at 9.4T (400.5MHz). By combining a transmit/receive surface coil array using serial transmission of individual coils with a total generalized variation reconstruction of images from all coils, acceptable quality human brain imaging is demonstrated.

Methods—B₀ is shimmed using sodium MR imaging (105.4MHz) with a birdcage coil. Proton MR imaging is performed with an excitation/receive array of surface coils. The modified FLASH pulse sequence transmits serially across each coil within the array thereby distributing SAR in time and space. All coils operate in receive mode. Although the excitation profile of each transmit coil is non-uniform, the sensitivity profile estimated from the non-transmit receive coils provides an acceptable sensitivity correction. Signals from all coils are combined in a total generalized variation (TGV) reconstruction to provide a full field of view image at maximum signal to noise (SNR) performance.

Corresponding Author: Keith Thulborn, MD, PhD, Center for Magnetic Resonance Research, University of Illinois at Chicago, 1801 West Taylor Street, Chicago, IL 60612, mrix@ameritech.net, cell: 847 830 0725.

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Results—High-resolution images across the human head are demonstrated with acceptable uniformity and SNR.

Conclusion—Proton MR imaging of the human brain is possible with acceptable uniformity at low SAR at 9.4 Tesla using this serial excitation and parallel reception strategy with TGV reconstruction.

Graphical Abstract

Keywords

Ultrahigh magnetic field; 9.4 Tesla; proton MRI; human brain; surface coil arrays; SENSE reconstruction; specific absorption rate; image uniformity

Dedication

I had the good fortune to meet Joe Ackerman in 1979 when I arrived to work as a post-doctoral research fellow in the Biochemistry Department at Oxford University, UK. The laboratory, led by Sir George K. Radda, had one of the first wide bore (89mm diameter) magnets from Oxford Instruments that fostered in the era of *in vivo* multi-nuclear NMR spectroscopy. Joe was working with a cardiologist on *in vivo* cardiac ^{31}P NMR spectroscopy, implanting solenoidal RF coils around rat hearts. It was Joe who made the leap in RF coil design from volume coils to surface coils. Prior to his observations and insight, it was widely believed that volume RF coils, (i.e. solenoidal coils) with homogeneous B1 sensitivity profiles were mandatory for good spectra. Joe made a giant step forward when he pointed out that the end of a solenoidal coil gave excellent spectra despite the non-uniform B1 field. From this observation, he rapidly developed the concept of the surface RF coil. Such flat coils were much easier to use *in vivo* to localize spectra to an organ while still yielding high quality spectra despite their non-uniform excitation profiles [1]. After a highly successful fellowship, Joe returned to the USA to set up his own biologically oriented NMR laboratory at Washington University in St Louis.

Thanks to Joe's transformative work on the potential of surface coils more than three decades ago, this current report uses an array of surface coils as a means to facilitate safe ultrahigh field MR imaging of the human brain. The approach avoids the self-resonance limit of large volume coils at ultrahigh frequencies while reducing the non-uniform specific absorption rate (SAR) and the non-uniform excitation profile that also limit ultrahigh field imaging. Combined with the SENSE reconstruction techniques, the non-uniform but

localized excitation profile of the surface coil does not compromise large field of view ^1H MR imaging, at least for the human brain at 9.4 Tesla.

Introduction

9.4T MR scanners for human imaging provide the expected signal to noise ratio (SNR) improvement for quantitative metabolic imaging of non-proton nuclei (^{23}Na , ^{17}O , ^{31}P), thereby providing access to new information about the human brain [2–5]. ^1H MR imaging at 9.4T has not received the same emphasis for anatomical imaging as at 7T where the parallel transmission approach has been the focus of efforts to overcome the non-uniform excitation that results when the transmission wavelength approaches the dimensions of the sample [6, 7]. This report demonstrates the application of a previously published and patented approach, but not yet reported for in vivo human imaging, using serial excitation of individual coils within a surface coil array with parallel reception in all coils to perform proton MR imaging at 9.4T [8–12]. Orzada published an alternative approach in which different modes were sequentially excited from different phase shifts applied to a set of eight transmit array elements [12]. That method produced excellent homogeneity at 7T, but required additional hardware for rapidly changing the applied phase to the array elements after each acquisition. Additionally, because of the different phase shifts, there may be a small SAR penalty due to summation of the electric field from different elements. As the proton imaging is to be used to provide correction for partial volume effects in lower resolution metabolic images obtained from non-proton imaging, the full range of ^1H image contrast used in clinical imaging is not the goal of this approach.

We report on the feasibility of ^1H MR imaging of the human brain at 9.4T (400.4 MHz) in which acceptable image uniformity across the entire brain can be achieved well below the SAR guidelines using 2D or 3D FLASH (fast low angle shot) sequences. The proposed approach uses an array of transmit/receive surface coils in which the coils are multiplexed to transmit serially (ST_x) rather than in parallel (PT_x) but all coils receive simultaneously. Although each RF coil has a very non-uniform B1^+ excitation profile, the non-transmit coils are used to form a virtual volume coil with acceptable total image uniformity. The final image is reconstructed using the SENSE method with sensitivity maps derived from the virtual coil. Although image reconstruction develops a sensitivity profile from the non-transmit elements, the data from all coils are used in the final total generalized variation SENSE reconstruction [13]. We present human brain images with significant contrast between gray matter, white matter and cerebrospinal fluid (CSF).

Materials and Methods

Human Studies

Human imaging was performed at 9.4T under a protocol approved by the institutional review board (IRB) using an investigational device exemption (IDE) from the Food and Drug Administration (FDA). Informed signed consent was obtained from all human volunteers (N=6).

Phantom Studies

The phantom was a plastic sphere of 16cm outer diameter (OD) with an array of solid plastic rods (5mm OD) with five rows of decreasing separation (6, 5, 4, 3, 2mm edge to edge). The sphere was filled with aqueous saline solution (40mM sodium chloride) with a small bubble to identify the orientation of the phantom within the FOV.

Specific Absorption Rate (SAR) Power Simulations

Although the low power of the FLASH sequences was not expected to produce SAR issues for either STx or PTx modes, typical SAR simulations were performed on phantoms (cylindrical and spherical) filled with normal saline (dielectric constant of 78, conductivity of 0.4S/m) using both an in-house developed 2D Full-wave EM solver using Finite Difference Time Domain (FDTD) [14] and a commercial 3D FDTD program (RESCOM XF7, Hershey, PA) for purposes of internal validation. Human SAR simulations were done using the adult male model provided with XF7 (AF “MAN” 1mm). A 4-coil configuration described below was modeled with four hexagonal coils (11cm O.D.) centered at 45°, 135°, 225° and 315° around the circumference of a cylindrical support (27cm O.D.) with the sample centered within the coils. The SAR calculations for PTx mode assumed experimentally realistic parameters with an excitation hard RF pulse of 1ms duration and repetition time (TR) of 250ms to give a duty cycle of 0.4% with a 90° flip at the center for the four-coil array. Coil phases were set according to coil position around the head, with a 90° degree phase delay for each successive coil. Each coil element was excited with equal power, and then the total field scaled to provide a 90° flip angle. For STx mode, identical coil excitations were used, with the same duty cycle. For purposes of comparison between PTx and STx configurations, excitations for STx mode were sequential rather than simultaneous as in PTx mode but all within the same 250ms TR as used in the parallel excitation case. Practically, a 22.5° flip angle in the center of the head cannot be achieved from a single coil on the current hardware so that the simulation overestimates the SAR producing a conservative result for human safety.

MR Imaging

All 9.4 Tesla imaging was performed on a customized 80 cm diameter clear bore 9.4 Tesla superconducting magnet (GE Medical Systems Oxford Ltd, Abingdon, UK) fitted with a head sized asymmetric gradient set (clear bore diameter = 37.9cm, maximum amplitude = 98mT/m, maximum slew rate = 392mT/m/s, GE Medical Systems Oxford Ltd, Abingdon, UK) [15]. The scanner employs commercial electronics and software (Biospec, Paravision 5.1, Bruker Biospin, Billerica, MA) that have been customized with electronics that divide the RF power (4kW) from a single RF excitation channel into four separate channels (each 1kW). Signals (TTL levels) from the modified gradient echo pulse sequences are used to serially select one excitation channel from these 4 possible channels. These control signals from the pulse sequence determine the ordered serial selection of each of four or eight RF surface coils (11cm OD hexagonal shaped loop formed from 0.63cm wide strips of printed circuit board; Taconic, Petersburg, NY, RF-35-0200-C2/C2) for excitation. During the transmission period, control signals select a single resonant coil for transmission and switch the other three RF coils from resonant mode (400.5MHz) to a grounded open circuit with an

isolation of approximately 50 ± 5 dB from the excitation RF coil in a response time of less than 0.01 ms using the circuit for a single coil shown in Figure 1.

Two different ^1H coil configurations were tested. The first coil configuration used four small RF surface transmit/receive coils, spaced uniformly around the outside circumference of a cylinder (27 cm OD plastic tube). This configuration covered the cerebral hemispheres of the human brain. Whole brain coverage including the posterior fossa used a configuration of two rings of 4 RF surface coils separated by 12 cm (coil center to coil center). The coils in one ring were rotated (45°) to those of the other ring to maximize independent coverage of the head. All surface coils were tuned to 400.5 MHz and matched to the electrical loading of a human head.

Image Acquisition

Proton MR imaging was performed using either 2D or 3D FLASH pulse sequences modified so that the RF pulses were directed serially through each of the RF coils while the other non-transmitting RF coils were switched to the non-resonant decoupled state. The pulse sequences were standard 2D and 3D FLASH sequences that had been modified to produce control signals that manage the SERIAL hardware to serially select the transmission RF coil while detuning the remaining coils during excitation. All coils, switched to the resonant mode during receive mode, detected the MR signal via its own receiver. As the scanner has only 4 receivers, data from the two loops of four coils were acquired separately to demonstrate feasibility of whole human brain coverage but, in principle, an 8-receiver configuration would require only a single, albeit longer, acquisition.

As optimizing the B_0 homogeneity over the sample is not possible with the inhomogeneous ^1H signal, shimming was performed using the sodium signal detected by a separate sodium tuned birdcage volume coil (105.4 MHz). This sodium coil can be exchanged for the proton coil without moving the patient's head that is positioned on a counterbalanced support that centers the head without touching the RF coil support.

Offline Data Management

The single data file, produced from the 4-channel array receive configuration generated from the serial excitation of the four RF coils, was taken offline from the scanner and sorted into 16 independent data files. Each file was reconstructed by either 2D or 3D fast Fourier transform (FFT). The images from each non-excitation coil were used to generate a sensitivity map for correction of a total generalized variation SENSE reconstruction of all 16 files to provide a single final image.

Total Generalized Variation (TGV) SENSE Reconstruction

In a typical experiment, multi-channel data are collected from PQ virtual coils: P transmit coils (SERIAL transmit mode) $\times$ Q receive coils (PARALLEL receive mode). Images are reconstructed using the SENSE method [16] with a penalty on the generalized total variation (TGV) [13] to promote piecewise smooth reconstructions.

Let $\mathbf{S}_{p,q}$ be a diagonal matrix containing the sensitivity map and $\mathbf{d}_{p,q}$ be a vector concatenating the measured k-space data when the p -th coil is used for transmission and the q -th coil is used for receiving. The desired image is reconstructed by solving the following optimization problem:

$$\boldsymbol{\rho}^* = \arg \min_{\boldsymbol{\rho}} \frac{\beta}{2} \sum_{p=1}^P \sum_{q=1}^Q \|\mathbf{d}_{p,q} - \mathbf{F} \mathbf{S}_{p,q} \boldsymbol{\rho}\|_2^2 + \text{TGV}(\boldsymbol{\rho}) \quad \{1\}$$

where the first term describes data consistency, the second term penalizes the second-order TGV of the reconstructed image, β is a regularization parameter, and $\mathbf{F}$ is the measurement matrix that models Fourier transform and k-space sampling. The second-order TGV term is defined as {2}

$$\text{TGV}(\boldsymbol{\rho}) = \min_{\mathbf{u}} \alpha_1 \|\nabla \boldsymbol{\rho} - \mathbf{u}\|_1 + \alpha_0 \|\varepsilon(\mathbf{u})\|_1 \quad \{2\}$$

where $\nabla \boldsymbol{\rho}$ is the gradient of $\boldsymbol{\rho}$, $\varepsilon(\mathbf{u}) = \frac{1}{2}(\nabla \mathbf{u} + \nabla \mathbf{u}^H)$, and α_1 and α_0 are parameters to balance the first and second-order derivatives.

To estimate the sensitivity map of each virtual coil, we need a set of calibration images, which can be collected from a low-resolution auxiliary scan for the accelerated imaging case or can be simply the fully sampled data of each channel. Let $\rho_{p,q}^c(\mathbf{x})$ be the acquired calibration image when the p -th coil is used for transmission and the q -th coil is used for receiving. We first reconstruct a virtual body coil image that is considered to be homogeneous by simple sum-of-square (SOS) reconstruction and omitting the images received from the excitation channel:

$$\rho^b(\mathbf{x}) = \sqrt[2]{\sum_{p=1}^P \sum_{q=1, q \neq p}^Q |\rho_{p,q}^c(\mathbf{x})|^2} \quad \{3\}$$

The sensitivity of each virtual coil is then estimated as:

$$S_{p,q}(\mathbf{x}) = \frac{\rho_{p,q}^c(\mathbf{x})}{\rho^b(\mathbf{x})} \quad \{4\}$$

Our experience shows that reconstructions are fairly robust to the choice of α_1 and α_0 . In this paper, we chose $\alpha_0 = 1$ and $\alpha_1 = 4$. The regularization parameter β is related to the noise level of measured data and is chosen based on the discrepancy principle [17]. The optimization problem in Eq. {1} can be efficiently solved using the alternating direction method of multiplier (ADMM) method as described in [13].

Results

The time average SAR (W/kg) maps for the same coil configuration operated in PTx and STx modes at the same duty cycle are shown in Figure 2. Figures 2(a–d) are single coil SAR maps for the 0.4% duty cycle described earlier. The peak scale corresponds to 0.18 W/kg over a 10 g average voxel volume, with an input power for each coil corresponding to approximately 22.5° flip angle in the center of the brain, requiring approximately 225 watts. Figures 2e and 2f compare the peak SAR (10 g average) from PTx (**2e**) and the sum of the peak 10g SAR maps in STx mode (**2f**). In both PTx and STx configurations, the total tip angle is set to 90 degrees in the center of the brain. The peak local SAR is reduced by approximately a factor of 1.5 in STx mode due to the absence of summing E fields in STx compared to PTx mode. Total imaging time was the same in both PTx and STx modes for these simulations.

Figure 3 shows the matrix of the sixteen individually images obtained from each coil of the 4-coil array acquired in the ST_xPR_x mode. The highest signal intensity for the excitation coil is spread heterogeneously across the limited FOV as expected for a single surface coil. Significant but much lower signal is received on the non-transmitting coils but can still be used to calculate the sensitivity map to correct for the final TGV regularized SENSE reconstruction.

Figure 4 compares images of the resolution phantom obtained with the array of four surface coils transmitting in parallel and receiving in parallel (PT_xPR_x mode) with images obtained from the ST_xPR_x mode but made with different sensitivity corrections and reconstruction strategies. The PT_xPR_x image shows the non-uniformity expected at 400MHz across the 20×20 cm² FOV even with a uniform saline phantom. For comparison, the equivalent images from the same array used in ST_xPR_x mode show greater uniformity (Figure 4b–e). Images using a sensitivity correction based on all 4 coils with a SOS reconstruction (Figure 4b) and with a TGV-SENSE reconstruction (Figure 4d) show different distributions of signal variations from the nonTx 3-coil correction for the SOS (Figure 4c) and TGV-SENSE (Figure 4e) reconstructions but large differences are not appreciated for this saline phantom.

When applied to the human brain, this approach using the 2D FLASH in the axial plane yields images shown in Figure 5 where the two different sensitivity correction strategies are compared for the SOS and TGV-SENSE reconstructions. The signal variations across the FOV for the 4-coil correction are greater for both the SOS (Figure 5a) and TGV-SENSE (Figure 5c) images when compared to the proposed nonTx 3-coil correction used for both SOS (Figure 5b) and TGV-SENSE (Figure 5d) images. The increased variation in signal intensity closest to the coils when all four surface coils are used for the sensitivity correction map is readily appreciated, irrespective of the reconstruction method. In comparison, the images using only the non-transmit coils for sensitivity correction show tissue contrast over the entire image with a single setting of the display parameters. Any loss of SNR from this sensitivity map is recovered in the final image by using data from all coils in the final image.

Figure 6 shows a sagittal image of the human brain using the 8-coil array to cover both the cerebrum and cerebellum. The decreased signal at the vertex of the cerebrum but high signal

at the craniocervical junction indicates that the isocenter of the coil array was positioned lower than would be optimal but whole brain coverage was achieved. The marked susceptibility signal loss along the anterior cranial fossa is expected due to the air-tissue interface of the paranasal sinuses at 9.4T. The in-plane resolution of 0.4mm is sufficient to acquire images that visualize the stria of Gennari in the primary visual cortex in 2 minutes (Figure 6b,c). The acquisition of these images produced no significant SAR (<5% FDA guideline limit).

Discussion

The electric field calculations confirm the experimental result of decreased SAR when each coil of the array is operated independently in time and space in the ST_x mode. As shown in Figure 2, the SAR map is symmetric for the ST_x mode as in-phase addition of electric fields from multiple coils does not occur as in PTx mode. Since power is proportional to the square of the electric field, the PTx mode can produce locations at which the electric fields from different coils are in phase with resultant locally increased SAR. In contrast, the STx mode uses each coil independently in time so that electric fields cannot add or have asymmetric, phase dependent peaks in local SAR. Thus, the derating of SAR to allow for potential “hotspots” due to electric field alignments during B₁ shimming in PTx mode is avoided in STx mode.

This sparse small surface coil array limits pulse sequences to small tip angles across the field of view. There is insufficient power for a small coil to achieve the simulation conditions. Tissue contrast available from clinical sequences requiring high flip angles cannot be achieved with this configuration of small surface coils. However, image contrast between cortex, white matter and cerebrospinal fluid across the human brain is readily appreciated without varying display parameters (Figures 5 and 6). Contrast is adequate for defining these tissue boundaries for automated segmentation. This can be used for correction of partial volume effects of co-registered quantitative metabolic imaging of non-proton MR signals.

Although serial imaging has a time penalty that increases with the number of coils, the global SAR is reduced to barely measureable levels and local hot spots are avoided. The use of TGV-SENSE reconstruction of images offers an opportunity to use both undersampling and multi-band techniques to reduce this time penalty.

Shimming the B₀ field with sodium imaging produces acceptable homogeneity with sodium linewidths of about 30Hz on a human brain. Although this step may be perceived as a further time penalty, we plan to use proton imaging for correction of partial volume effects in conjunction with quantitative metabolic imaging. Thus, this step has no additional time impact for this application [18–20].

The use of surface coils avoids the challenge self-resonance of large volume coils at high frequencies while the use of a single transmission channel saves costs. Although the SAR reduction is probably not important for low flip angle gradient echo imaging of protons even at 9.4T, SAR is a significant problem for lower gamma nuclei. This STxPRx approach can take advantage of the short T1 relaxation times (e.g., sodium, oxygen and potassium) by

using very short repetition for signal averaging to improve signal-to-noise performance without T1 saturation effects that would otherwise compromise signal quantification.

In practice, the ST_xPRx mode has a time penalty or reduced volume coverage due to the increased number of RF pulses required to encode the full FOV. In principle, shorter TR values between coil excitations can be used while still avoiding T1 saturation effects. Similarly, array coils and TGV-SENSE reconstructions offer potential undersampling and multiband techniques to decrease acquisition times. As the regional RF exposure is limited to the FOV, safety concerns for heating of implanted medical devices away from this FOV are eliminated. Such heating of implanted medical devices has been a major concern of the FDA for ultra-high field human imaging.

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Highlights

Surface coil arrays achieve uniform human head imaging at 9.4T.

Serial transmission-parallel reception (ST_xPR_x) produces human brain images at 9.4T.

SENSE sensitivity correction of ST_xPR_x mode uses signal from non-transmit coils.

SENSE reconstructed ST_xPR_x acquisitions produce quality human brain images at 9.4T.

Surface coil arrays in serial mode avoid SAR limits for human brain imaging at 9.4T.

Figure 1.

Schematic diagram of one of the set of identical surface coils tuned to 400.5MHz with distributed capacitance ($C1 = 3.6\text{pF}$ American Technical Ceramics, Huntington Station, NY, P/N 100C3R6CMN2500X; $Ct1 = 3\text{--}10\text{pF}$, Polyflon Co., Norwalk, CT, P/N RP-VC10-6-05) with tuning and matching circuits ($C2 = 1.5\text{pF}$ American Technical Ceramics, Huntington Station, NY, P/N 100B1R5CBMN500X; $Lt2$ = adjustable inductor, Coilcraft, Cary, IL, P/N 148-04A12L, P/N 148-05A12L; $Lt3$ = adjustable inductor, Coilcraft, Cary, IL, P/N 148-04A12L, $Ct2 = 1\text{--}25\text{pF}$, Voltronics Corp., Denville, NJ, P/N NMAT25HV) with a DC switching circuit (Sw) using a variable inductor ($Lt1$ = adjustable inductor, Coilcraft, Cary, IL, P/N 148-04A12L) in series with PIN diode P1 (MACOM, Lowell, MA, P/N MA4P7446-1091T) and in parallel with the same PIN diode P2. A LC Band Stop filter ($L1 = 100\text{nH}$ inductor, Bourns Inc., Riverside, CA, P/N 9230-94-RC in parallel with a series pair of capacitors $C3 = 3.3\text{pF}$, AVX Corporation, Greenville SC, SQCB7M3R3BATME\500) in the switching control line (Sw) isolates the switching control signals on this input/output line from the MR signal. The capacitor ($C4 = 100\text{pF}$, American Technical Ceramics, Huntington Station, NY, P/N 100101GMN2500X) isolates the Sw circuit from the transmit (Tx) and receive (Rx) circuits. The T/R switch protects the preamplifier (A) during transmission.

Figure 2.

Simulation of peak SAR distributions over a 10g volume element for the transmission configurations of a four RF surface coil configuration. (a,b,c,d) are the single RF coil transmission profiles of each coil distributed equally around the circumference of the head sample. (e) is for the PT_xPR_x configuration and (f) is the ST_xPR_x configuration. All calculations assumed the same duty cycle, equivalent to the same total imaging time in both configurations. The maximum scale is 0.18 W/kg in a–d, and 0.4 W/kg in e and f. Maximum 10g average volume SAR in the figures is 0.176, 0.181, 0.173, 0.174 W/kg in a–d respectively, and 0.431 and 0.32 W/kg in e and f, respectively.

Figure 3.

Representative central partition of the resolution phantom from the ST_xPR_x 3D FLASH acquisition for each coil in the matrix of sixteen individual coil images (rows and columns 1, 2, 3 and 4 starting in the top left position correspond to coils 1, 2, 3 and 4, respectively). The diagonal images show highest signal intensity in the phantom closest to the transmission coils (T_x1, R_x1 at 45° position; T_x2, R_x2 at 135° position; T_x3, R_x3 at 225° position; T_x4, R_x4 at 315° o'clock position). The off diagonal images from the non-transmit coils in receive only mode show low signals from which a more uniform sensitivity correction was calculated using the proposed TGV SENSE method to yield the total image. The individual coil images have not been scaled so that the variation in signal intensity can be appreciated between transmit and receive coils (diagonal elements) and across the non-transmit receive coils (off diagonal elements). The lower signal intensity of the non-transmit receive coil images across the FOV build a more uniform total sensitivity correction albeit at lower signal-to-noise than when signals from the transmit coils are included.

Figure 4.

Representative axial partitions from a 3D FLASH acquisition of the resolution phantom (40mM NaCl) at 9.4T using an array of 4 surface RF coils operated (a) in PT_xPR_x mode without sensitivity correction. Using the ST_xPR_x acquisition, the image reconstruction has been performed with the sum of squares (SOS) method (b,c) and the TGV-SENSE method (d,e) using either the 4-coil uniformity correction (b,d) or the nonTx 3-coil sensitivity correction (c,e). Variation in signal intensity across the FOV is greater for the 4-coil correction. The 3-coil correction improves image uniformity for both SOS and TGV-SENSE reconstructed images.

Figure 5.

Comparison of the sensitivity correction for the SOS (a,b) and TGV-SENSE (c,d) reconstructions of a representative human brain axial image using sensitivity corrections performed with either signals from all 4 surface coils in the array including the Tx coils (a,c) or only signals from the nonTx 3 coils (b,d). The images using the proposed nonTx 3-coil correction with either SOS or TGV-SENSE reconstruction allow the gray matter, white matter and CSF contrast to be appreciated across the FOV without altering the display level and window. The 4-coil correction images contain the same information but with a greater variation in signal intensity that requires regional dependent display adjustments to see all features. The SOS images are displayed showing the regions closest to the coils with contrast settings to resolve cortex and white matter (Figure 5a) or as hot spots (Figure 5c) but no single display setting allows the same contrast to be appreciated over the entire image as with the SOS and TGV-SENSE reconstructions with the 3-coil sensitivity correction.

Figure 6.

(a) Sagittal partition through the head of a normal volunteer using ST_xPR_x 2D FLASH with two rings of four surface RF coils with the TGV SENSE reconstruction at an in-plane resolution of $0.4 \times 0.4 \text{ mm}^2$, 4 mm slice thickness and two-minute total acquisition time. Image uniformity is acceptable through most of the cerebrum and cerebellum. Reduced signal contrast is present at the vertex and in regions with magnetic susceptibility effects arising from the paranasal sinuses at the skull base. The SNR is sufficient to magnify the images to see fine structures such as the stria of Gennari in the primary visual cortex along the inferior bank of the calcarine fissure (white box), as shown magnified in (b) and labeled in (c) where the primary visual cortex is delineated by white dots and the white line is along the Stria of Gennari.