Published in final edited form as:

Magn Reson Med. 2009 July ; 62(1): 17–25. doi:10.1002/mrm.21970.

Short Echo Spectroscopic Imaging of the Human Brain at 7T Using Transceiver Arrays

NI Avdievich¹, JW Pan^{1,2}, JM Baehring^{1,2}, DD Spencer¹, and HP Hetherington¹

¹Department of Neurosurgery, Yale University

²Department of Neurology, Yale University

Abstract

Recent advances in magnet technology have enabled the construction of ultra high-field magnets (7T and higher) that can accommodate the human head and body. Despite the intrinsic advantages of performing spectroscopic imaging at 7T, increased SNR and spectral resolution, few studies have been reported to date. This limitation is largely due to increased power deposition and B_1 inhomogeneity. To overcome these limitations, we have utilized an 8 channel transceiver array with a short TE (15 ms) spectroscopic imaging sequence. Utilizing phase and amplitude mapping and optimization schemes, the 8 element transceiver array provided both improved efficiency, (17% less power for equivalent peak B_1) and homogeneity ($SD(B_1) = \pm 10\%$ versus $\pm 22\%$) in comparison to a transverse electromagnetic (TEM) volume coil. To minimize the echo time to measure J-modulating compounds such as glutamate, we developed a short TE sequence utilizing a single slice selective excitation pulse followed by a broad band semi-selective refocusing pulse. Extracerebral lipid resonances were suppressed with an inversion recovery pulse and delay. The short TE sequence enabled visualization of a variety of resonances, including glutamate in both a control subject and a patient with a grade II oligodendroglioma.

Introduction

With recent advances in magnet technology, ultrahigh field magnets (7T and higher) with bore sizes capable of accommodating the adult human head have become available from a variety of research and clinical vendors. The increased field strength confers the intrinsic advantages of increased SNR and for spectroscopic studies, increased spectral resolution and spectral simplification for J-coupled resonances. Although measurements of NAA, creatine and choline have been highly successful at conventional field strengths (3T and lower), the measurements of amino acids such as glutamate and glutamine in high resolution spectroscopic images has been difficult due to spectral overlap and SNR limitations. Thus, spectroscopic imaging at 7T should provide an ideal platform for evaluating these compounds.

However, despite the availability of 7 and 8T systems dating from late 1990s (1), (2), there have been few reports of their use in spectroscopic imaging studies. This limitation is largely due to the inherent disadvantages of high field which include: 1) a roughly linear increase in required power for equivalent B_1 generation (Watts/kHz of B_1), 2) substantial B_1 inhomogeneity for conventional volume coils (up to 42% (2)) and 3) larger errors in chemical shift dispersion when using slice selective pulses. All of these issues are especially problematic with sequences that use in-plane selection with slice selective refocusing pulses. Due to the decrease in efficiency of B_1 generation (2-4), maximal values for B_1 are limited,

typically ~1kHz for peak powers of ~3-4kW, resulting in small bandwidths and lengthy pulses. The long RF pulses in turn make J-modulating resonances such as glutamate and glutamine difficult to measure. Additionally, the limited B_1 exacerbates the substantial chemical shift dispersion errors when large volumes (e.g. 10×10 cm) are used for in-plane localization and suppression of extracerebral resonances. Further, even moderate B_1 inhomogeneity can result in substantial SNR losses when multiple refocusing pulses are used for volume selection. Although adiabatic refocusing pulses can compensate for B_1 inhomogeneity, due to the limited B_1 they are also lengthy and deposit substantial amounts of RF power, making short echo times difficult to achieve. Thus to date the majority of spectroscopic imaging studies reported at 7T using volume head coils have used long echo times (5-7) making measurements of glutamate and glutamine difficult.

Based on the issues of RF inhomogeneity and power deposition, there has been intense interest in developing alternative designs to coils which typically utilize a single transmitter input such as transverse electromagnetic (TEM) and birdcage coils for human brain imaging at 7T. One design, the transceiver array (3,4), consists of an array of independent coils circumscribing the head which can be used for reception and transmission. By independently varying the amplitude and phase of the RF delivered to each coil simultaneously during transmission, "RF shimming", the spatial distribution of the RF can be optimized to provide improved homogeneity and efficiency (8,9). Since the loading conditions and efficiency of each coil will vary from subject to subject and the phase relationships between the coils are spatially dependent, for optimal performance the phase and amplitudes at 7T and higher should be optimized for each study and location. With improved homogeneity, the transceiver array enables conventional refocusing pulses to be efficiently used while allowing echo times and power deposition to be minimized. To eliminate in-plane chemical shift registration errors due to spatially selective refocusing pulses with limited bandwidth, alternative methods to suppress extracerebral regions or resonances can be used, such as outer volume suppression (10) or inversion recovery methods (11). Since these methods occur prior to the excitation pulse, they minimize the required TE. Thus, the goal of this work was to develop a short TE spectroscopic imaging sequence with low power deposition at 7T using a transceiver array. In this work we describe: 1) an 8 element 7T transceiver array and compare its efficiency (kHz B_1 per kW of applied power) and homogeneity with a TEM coil; 2) a method for individualized optimization of the rf phases and amplitudes and compare it with another method for optimization; 3) a short TE sequence for spectroscopic imaging and 4) data acquired with this sequence in control subjects and a patient with a brain tumor.

Methods

Spectrometer

All data were acquired with a Varian Direct Drive system and a head only (68cm ID) actively shielded 7T magnet. The system uses a Magnex head gradient insert (50mT/m, 250 μ s rise times) equipped with a complete set of 2nd and 3rd order shims, with each of the shims powered by its own 20A supply (Resonance Research Inc.). For the transceiver array studies, 8 fully independent transmit channels were used to drive eight 1kW RF amplifiers (Communication Power Corporation). A T/R network was constructed to provide RF power to each of the 8 surface coils during transmission and to connect surface coils to preamplifiers during reception. The T/R network consisted of 8 preamplifiers (Advanced Receiver Research) and 8 homebuilt single-pole double throw (SPDT) PIN diode switches each capable of handling 1 kW of RF. The SPDT switches were constructed using lump-element quarter wave length transformers and provide greater than 55 dB of isolation and less than 0.15 dB of insertion loss. A PIN diode driver was also constructed to provide 50-100 mA per diode. The data were acquired using 8 independent receivers. For the TEM

comparison study, the RF amplifier was “switched” to high power mode in which the outputs of the 8 channels were phase matched and combined to a single output.

Transceiver Array

The transceiver array consists of eight rectangular surface coils circumscribing the head. The surface coils (9 cm longitudinal $\times$ 8 cm circumferential) were formed using 6.4 mm wide copper tape. The surface coils were placed on cylindrical acrylic former (25 cm diameter, 20 cm length) and distributed at 45° increments (1.8 cm gap circumferentially). To decrease radiation losses and improve the performance, the array was shielded (12). The shield (33 cm diameter, 20 cm length) was constructed from a 50 μ m polyamide film with a superficial 5 μ m copper layer. The thickness of the copper layer is only slightly larger than the skin depth at 298 MHz (^1H frequency at 7 T). This allowed us to preserve an unloaded Q -factor while suppressing gradient eddy currents (13). Each coil was individually tuned and matched and had six uniformly distributed capacitors (Fig. 1). To prevent wave propagation on the shield, a cable trap was used. Additionally, all of the tank circuits were “grounded” to the shield (Fig.1). To minimize inductive coupling between the surface coils, a pair of small ($\sim$ 12 mm in diameter) loops in series with each surface coil were used (14) (Fig. 1). Isolation between the coils varied depending on the size and the shape of a head in the range of -16 to -25 dB. Isolation between all the non-adjacent coils was in the same range and did not require any additional decoupling. The ratio of unloaded to loaded Q -factors (Q_U/Q_L) measured 350/75 on a human head. Although the inductive decoupling between coils (spacing between the 12mm loops) was not adjusted on a per subject basis, the tuning and matching of each coil was adjusted individually for each subject. This process takes approximately 2 minutes. This study was approved by the Yale Human Investigations Committee and all subjects provided informed consent prior to the study

TEM Coil

The quadrature TEM volume coil was built using tuneable capacitive coaxial elements with Teflon insulators similar to that previously described (2,13,15). The coil consists of 16 resonant elements constructed of copper tubes with the diameter of 12.5 mm and 6.4 mm utilized for the outer shells and the central conductors, respectively. The cylindrical RF shield was constructed from a polyamide film (50 μ m) with a 5 μ m layer of copper on the surface. The shield was closed at the end by a shielded back wall for improved homogeneity of the RF field in longitudinal direction (13,16). The RF shield diameter measured 29.8 cm. The coil's length measured 17 cm and the element ID (diameter measured at the element centers) measured 26.7 cm. The coil was driven in quadrature using capacitive matching and a four-port drive with the driven elements located at 45° in respect to vertical and horizontal axes similar to that previously described (16). The ratio of unloaded to loaded Q -factors (Q_U/Q_L) measured 600/45 on a human head.

B₁ Mapping

B₁ maps (amplitude and phase) of individual coils (single coil transmitting) or the combined array (all coils transmitting simultaneously) were acquired using a double excitation method (17). For these studies, the B₁ maps used a target B₁ of 1kHz and a nominal excitation pulse of 60° with a TR of 0.5 s. The images were acquired at 64 $\times$ 64 resolution over a FOV of 192mm using a 5mm slice thickness to maximize SNR. After selecting a target slice using default parameters (equal power with incremental 45° phase shifts), the amplitude and phase of each coil was mapped (8 coils $\times$ 0.5 s $\times$ 64 = 256 s). After numerical optimization of the B₁, a second B₁ map using the optimized parameters with all coils transmitting simultaneously was acquired (32 s), to confirm the target B₁ distribution was achieved.

To determine the amplitude of each coil, the excitation pulses were applied sequentially to each coil prior to incrementing the phase encoding value. A sum of squares reconstruction using all receive coils was used. Using this method the ratio of the amplitudes of the two images (S_2/S_1) is given by $\sim \cos(\theta)$ where $\theta = B_1(t) \cdot t$, (B_1 in units of Hz). Since the pulses are slice selective, the measured θ , is then corrected for the effects of the slice profile using a look-up table calculated from numerical simulations (17).

The relative phase of the B_1 generated by each coil was determined by reconstructing the phase sensitive images acquired using coil #1. By using a single coil for reception (coil #1, the reference coil), differences in signal phase due to reception phase differences between the coils are eliminated. The relative transmission phase for each coil, on a pixel by pixel basis, was then calculated by subtracting the phase of the acquired signal when coil #1 was used for both transmission and reception.

Optimization of the Phases and Amplitudes

The B_1 generated at any given point within the sample from a set of n independent RF coils driven simultaneously can be expressed as

$$B_1(r) = \sum_{j=1}^n a_j B_{1j}(r) \cdot \exp(-i(\Phi_j(r) + \Phi_{0j})) \quad [1]$$

where the amplitude ($B_{1j}(r)$) and phase ($\Phi_j(r)$) of the j th coil are both functions of spatial coordinates determined by coil geometry and loading conditions, and a_j and Φ_{0j} are amplitude scaling and phase parameters adjusted by the user. Since $\Phi_j(r)$ and $B_{1j}(r)$ vary throughout the sample for each coil and a_j and Φ_{0j} are spatially invariant, the phase of the RF generated by each coil can not be perfectly coherent over all regions of the sample simultaneously (18). Thus phase cancellation of the RF will occur in some regions, reducing the amplitude of the B_1 generated. The degree of phase coherence (fraction of B_1 generated divided by maximal B_1 if all coils were “in phase”) is given by

$$\eta(r) = \left| \sum_{j=1}^n a_j B_{1j}(r) \cdot \exp(-i(\Phi_j(r) + \Phi_{0j})) \right| / \sum_{j=1}^n |a_j B_{1j}(r)| \quad [2]$$

such that “perfect” constructive interference ($\eta(r) = 1$) occurs at a given location when at that location, the phase term $\Phi_j(r) + \Phi_{0j}$ is equal for all coils.

To minimize the power required to excite central brain regions, the degree of phase coherence, η , from central brain regions needs to be maximized. To achieve this, we selected a circular ROI of 60 mm diameter in the center of the brain within the target slice to optimize the phase. The mean Φ_i over this region for each coil j was determined from the B_1 maps. Setting the mean phase over the ROI for coil #1 (the reference coil) to 0, (i.e., $\Phi_{1\text{mean}} = 0$), the optimized applied driving phase of each coil, Φ_{0j} , is given by $-\Phi_{j\text{mean}}$. For each coil, this minimizes the difference in mean phase value between coils over the ROI, thereby maximizing η over the central ROI. Using these phases for $\Phi_j(r) + \Phi_{j\text{mean}}$, the amplitude scaling factor a_j was then optimized over a second ROI encompassing the majority of the brain within the slice. The optimization was performed using a least squares algorithm to minimize the difference between the target B_1 and the achieved B_1 .

Spectroscopic Imaging Pulse Sequence

To minimize the TE we implemented a sequence (Fig. 2) which utilizes a slice selective excitation pulse and a single broad band semi-selective refocusing pulse (9). By eliminating the in-plane selection a TE of 15ms was attained and artifacts due to chemical shift dispersion errors within the plane were eliminated. The slice selective excitation pulse used a 300Hz/mm gradient strength, resulting in a chemical shift mis-registration of 1mm between slices representing creatine and NAA. To provide additional water suppression we included a frequency selective (± 0.50 ppm bandwidth) low amplitude (~ 100 Hz) adiabatic inversion pulse in conjunction with an inversion recovery time of 1200ms. To suppress extracerebral lipids, we also used an adiabatic inversion pulse (± 6.7 ppm bandwidth) with an inversion time of 350ms. Thus in-plane localization in this sequence is provided by phase encoding, while artifacts resulting from 'bleed' of the extracerebral lipids into brain regions via the spatial response function of the encoding is minimized by the inversion recovery. A TR of 3 s was used resulting in a time average deposited power of 2.5-3.1 W (n=5 subjects) or ~ 1 W/kg. The data were acquired using 24×24 rectangular encoding over a FOV of 192×192 mm².

Image Reconstruction

To provide phase and amplitude scaling for the reconstruction of the metabolite SI data, an unsuppressed water SI was acquired using the same sequence with the exceptions that: 1) the initial IR pulses and their associated recovery delays were deleted; 2) the broadband semi-selective pulse was replaced by a non-selective pulse (500 μ s) and 3) the TR was shortened to 0.5 s. The data acquired from each receiver coil for the water SI was then reconstructed using a 3DFFT (1D spectral, 2D spatial), including both spatial (cosine filter, 1cc effective volume) and spectral (3Hz Gaussian broadening) filtering. The water resonance from each pixel was then automatically phased and the phase parameters for each pixel were saved. The water resonance was then integrated to determine the relative signal strength of each coil at each pixel location. The relative noise from each coil was determined by automatically measuring the standard deviation of the noise from a region outside the brain. Spectroscopic images from each coil were then reconstructed. The metabolite spectrum from each pixel from each coil was then phased using the values determined from the water reference image. To optimize the SNR of the metabolite SI, the metabolite spectrum from each pixel of each coil was scaled by the SNR of water SI for that pixel from each corresponding coil. A single metabolite image was then generated by summing the phased and SNR weighted spectra from each coil for each pixel.

Results

Coil Performance

Due to differences in head size, placement and shape, the amplitude and phase distributions of each RF coil varies between subjects. To optimize the performance on each subject the phase and amplitude of each coil was mapped. Displayed in Fig. 3a are amplitude, $B_1(r)$, and phase, $\Phi(r)$, maps from the 8 element transceiver array from a human head at 7T. Displayed in Fig. 3b is the ROI used for optimization. To quantify the degree of homogeneity we selected a 150×120 mm ROI centered within the brain on the axial slice. Using a peak power (sum of the power applied to all coils) of 2.9 kW, the mean B_1 achieved over the 150×120 mm² ROI was 1032 Hz with a standard deviation of 101Hz (10%) with a maximal value of 1249 Hz in the center of the head (Fig. 3c). Displayed in Fig. 4a and 4b are maps of the amplitude of B_1 using the TEM coil. Using a peak power of 3.5 kW for this ROI, the mean B_1 was 856 Hz with a standard deviation of 192 Hz (22%) with a maximal value of 1231 Hz in the center of the head. Thus in addition to improving the homogeneity of the B_1 , significant improvements in transmit efficiency were also obtained when using the

transceiver array as opposed to a TEM coil. The analogous slices for the 8 channel transceiver array after B_1 optimization is displayed in 4c and 4d.

B_1 Optimization

Conventionally, many investigators have used fixed relative phases (approximately $360^\circ/n$ increment per coil, where n is the number of coils) and amplitudes to drive transceiver arrays (3,4). To assess the extent to which the determined values are sensitive to head geometry and size we acquired data from $n=5$ subjects selecting an axial plane at the level of the cingulate. To provide an accurate comparison we also calibrated the phase and amplitude of the RF at the RF coil for each transmit channel by acquiring B_1 maps from coil #1 using the output of each transmit channel sequentially. We then used this data to normalize the amplitude of the required drive voltage and relative phase determined from the B_1 optimization in each of the 5 subjects. In this group (Fig 5) the coils directly under the posterior portion of the head (coils 1 and 2) show greater efficiency (lower required drive voltage) than the coils over the lateral (3, 4, 7, 8) and anterior portions of the brain (5, 6). This reflects the closer proximity of the back of the head to coils #1 and #2 than other brain regions to their nearest coils. Additionally coils #1 and #2 show less variability across individuals. This reflects the fact that the distance from the posterior portion of the head is fixed relative to coils #1 and #2 (patient head support of ~ 1 cm thickness). Thus the generally greater variability seen from the lateral and anterior coils reflects different head sizes and shapes. Fig. 5b displays the mean and standard deviation of the transmit phase used relative to coil #1 for each coil. In comparison to the theoretical values of incremental 45° shifts, the mean phase shifts across the group varied from 36° to 53° depending upon the specific coil.

As described, the phase parameters, Φ_{0j} , were chosen to be optimal for the center of the brain (6 cm diameter ROI). A plot of η over a 12×16 cm² elliptical ROI is shown in Fig. 6. There is excellent phase coherence over the central target ROI ($0.8 < \eta < 1$), while η drops to ~ 0.4 on the periphery of the large ROI, in accordance with the constructive/destructive interference effect described previously by Van de Moortele et al (18). To evaluate the importance of optimizing the efficiency of the coil in the center and the relative penalty in loss of achievable homogeneity, we carried out a least squares optimization where both Φ_{0j} and a_j were allowed to vary simultaneously. This resulted in a small improvement in homogeneity $\sigma B_1(r) = 8\%$ as opposed 10% when maximizing η from the central brain region. However, when both Φ_{0j} and a_j are allowed to vary simultaneously, η is reduced (15-20%) in central brain regions (Fig. 6) and (5-10%) in peripheral brain regions. Effectively this equalizes the RF amplitude by reducing the efficiency of B_1 generation from the central brain region relative to the periphery. Unfortunately, to achieve this modest improvement in homogeneity, η is reduced throughout the brain, including the periphery. To regain the lost amplitude from the entire ROI, the overall amplitude must be increased, resulting in a 21 % increase in power deposition. Thus maximization of B_1 homogeneity is not necessarily equivalent to the maximization of power efficiency.

Short TE Spectroscopic Imaging

Displayed in Fig. 7 are data acquired using 24×24 encodes over a FOV of 192×192 mm with a 10mm slice thickness. Due to the long times associated with the lipid IR pulse, a 3s TR was used to minimize losses to the metabolites, resulting in an acquisition time of 29 min. Excellent spectral quality was obtained throughout the slice, including the most anterior brain regions. By eliminating the need for in-plane selection during the echo, we were able to use a semi-selective refocusing pulse for water suppression. The duration of the pulse was 5ms allowing the echo time to be reduced to 15ms. To augment the water suppression a frequency selective (± 150 Hz) adiabatic inversion pulse (60 ms, 0.5 W deposited power)

was followed by a 1200 ms recovery delay. This achieved greater than 2000 fold suppression of the water resonance despite the short TE used.

As described, suppression of extracerebral lipids is achieved by the use of a non-selective adiabatic inversion pulse followed by a 350 ms delay. To assess the extent of lipid suppression we acquired 16×16 spectroscopic images with and without the application of the lipid inversion pulse. Eight locations (~45 degree increments about the head) overlying the skull and scalp were selected and the magnitude spectra from those locations was integrated from 1.1 to 1.7ppm to provide an estimate of the lipid signal. In this case a factor of 20.8 ± 5.0 was achieved. Although the IR pulse provides excellent suppression of resonances with T1s ~500 ms, it also provides a flat response for resonances with T1s between 1.0 and 2.0 s, $0.36 < |M_0| < 0.46$. However this does reduce the SNR of the metabolite signals by ~60%. Despite this reduction in SNR, in addition to the up-field resonances of glutamate and glutamine, the high sensitivity of 7T also permits the 7.9ppm resonance of NAA to be observed in the downfield portion of the spectrum (Fig. 7).

Displayed in Fig. 8 are data from a tumor patient referred for evaluation following an increase in FLAIR signal intensity without Gd enhancement (clinical imaging performed at 3T) adjacent to the resection site. Using the short TE sequence we acquired a spectroscopic image of the plane containing the resection cavity, including the 3T FLAIR abnormality. Following the 7T study, additional resective surgery was performed and histological evaluation of the excised tissue revealed a grade II oligodendroglioma. Although the tumor ROI was adjacent to a resection cavity and in an anterior location, sufficient B_0 homogeneity was obtained for this site and throughout the brain within the slice for the 7T study. Although this low grade tumor shows no Gd enhancement visually there are reductions in NAA/creatine and increases in lactate, choline/creatine and myoinositol/creatine in this region. Thus, in addition to providing the ability to measure amino acids such as glutamate and glutamine and non-modulating singlets (NAA, creatine and choline), the short echo sequence also enables the measurement of lactate when elevated, such as in primary brain tumors. In this case, suppression of extracerebral lipids by use of the inversion recovery sequence will also suppress any lipids within the tumor, improving visualization of lactate.

Discussion and Conclusions

RF Coil Design and Optimization

Typically a transceiver array for brain imaging consists of multiple resonant elements positioned symmetrically on a cylinder circumscribing the head which is parallel to the axis of the static magnetic field. These arrays can be further categorized based on the orientation of their current paths relative to the cylinder circumscribing the head. In the first category the current loops are positioned within a plane perpendicular to the surface of the cylinder and the return current path is provided by the coil shield or a ground plane of the microstrip (3,4). In the second category the current loops are located parallel to or on the surface of the cylinder. The perpendicular design facilitates mutual inductive coupling between the elements and, therefore, strong coupling exists not only between nearest neighbors but also between non-adjacent elements (19). However, in the parallel design, such as used in this work, positioning the surface coils parallel to the surface of the cylindrical holder decreases mutual inductive coupling between the coils, such that strong mutual coupling is only present between adjacent coils. This allows the shield or ground plane to be displaced away from the coils, improving efficiency. In this work we have also used inductive decoupling between adjacent coils to further minimize the coupling.

As described, the transceiver array provided superior performance in both homogeneity and efficiency in comparison to the TEM coil. The performance of our TEM coil of 1231Hz in

the coil center at 3.5kW is virtually identical to that reported by Vaughan (2) (1267Hz at 3.5kW), such that we believe this represents a fair and accurate comparison of the different designs. Recently Adriany and colleagues (4) reported performance specifications for a 16 element transceiver array at 7T. Averaging peripheral and central brain locations they reported a 90° pulse duration of 440 μ s at 1 kW for a 18×23 cm² array. This equates to a required power of 3.1 kW to achieve 1 kHz of peak B₁. In comparison, our circular transceiver array achieved a mean B₁ of 1 kHz for the entire ROI at 2.9 kW. Although an exact comparison between the two coils is not possible due to a variety of factors including: 1) number of elements (16 versus 8), 2) diameter (23 v. 25 cm), 2) length (18 v. 9 cm), and 4) shape of the array (elliptical versus circular); they provided similar performance. Notably the array described by Adriany (4), was made in a segmented fashion which allowed its geometry and size to be adjusted for individual heads. However, this also requires that the decoupling between the elements also be adjusted for each configuration. In our case the use of a fixed circular geometry results in variable performance for the individual coils with generally higher drive voltages required for coils which are more distant from the brain. Thus use of an elliptical array where the lateral coils can be brought into closer proximity would be of significant benefit. Likewise a split coil with accommodation for the nose could make patient positioning easier and allow the distance from the brain to the anterior coils to be minimized, thereby also improving efficiency.

Pulse Sequence

In 2001, Tkac reported a short TE spectra from the human head at 7T (20), using a single volume STEAM sequence with a surface coil demonstrating the advantages of increased spectral resolution and available SNR. By selecting a single moderately small volume (8 cc) with a sequence which does not use 180° refocusing pulses, the authors avoided the problems associated with power deposition and B₁ inhomogeneity. However for many applications where substantial portions of the brain need to be evaluated at high spatial resolution, such as in brain tumors, spectroscopic imaging (SI) acquisitions provide substantially more information.

To overcome the B₁ inhomogeneity issues, adiabatic pulses which compensate for inhomogeneity can be used (21). Recently Scheenen (5) reported a “semi-laser” sequence (22) for volume localization at 7T. Although the investigators (5) reduced the number of adiabatic pulses to 4 (two for the slice selection and 1 each for the two directions of in-plane localization), and used a peak B₁ of only 650 Hz, the resulting average power deposition was still 3.2W/kg. The decreased peak B₁ resulted in adiabatic pulses of 10ms in length and a minimum echo time of 50 ms for a volume head coil. The lengthy echo period resulted in near complete suppression of the glutamate and glutamine resonances. Furthermore, the 2.5 kHz bandwidth of the refocusing pulses resulted in large chemical shift mis-registrations, 45% for 3.8ppm bandwidth or an in-plane spatial shift of 11mm between slabs representing creatine and NAA. Using an alternative approach to overcome the chemical shift mis-registration problem, Balchandani reported a novel spectroscopic imaging sequence using spectral spatial pulses that excite narrow spectral bands, allowing NAA and creatine and choline to be imaged separately (6). Unfortunately, the pulses were long, resulting in an echo time of 90 ms, such that glutamate and glutamine were not seen in the spectra.

Due to the relatively good homogeneity provided by the transceiver array (~10% SD), we were able to reduce SNR losses associated with the B₁ dependence of the spin echo sequence without using adiabatic refocusing pulses. This has two advantages: 1) it reduces the power deposition markedly (from 9.6 W per pair of adiabatic pulses to 2 W for the semi-selective refocusing pulse) and 2) substantially decreases the minimum TE (~30 ms to 15 ms), allowing J-modulating compounds such as glutamate and glutamine to be detected. To minimize the chemical shift mis-registration errors in the slice direction, a slice selective

excitation pulse (3 kHz bandwidth at 750Hz B_1) was used in place of slice selective refocusing. To suppress extracerebral lipids, a non-selective adiabatic inversion pulse was used. Since no in-plane localization is used, spectra from the periphery of the brain can be acquired enabling analysis of cortical gray matter locations and peripheral tumors.

Unfortunately, the use of the non-selective inversion recovery pulse results in a ~60% reduction in SNR for NAA at 7T due to its state of partial inversion after the inversion recovery period. This is larger than the ~40% reduction seen at 4T when using similar sequences (23) due to the lengthening of the lipid T1 (~500 ms). Although in principle the inversion pulse could be made frequency selective to minimize perturbation to the metabolite signals, overall lipid suppression is typically degraded. The degradation in lipid suppression arises from: 1) failure to suppress lipid resonances arising from lipid groups adjacent to double bonds and containing double bonds (down field of NAA) and 2) susceptibility shifts at the periphery of the brain that tend to shift the lipid resonances downfield such that they reside on the transition bands (incomplete inversion) of the inversion pulse. Similarly, multiple slice selective pulses (e.g. 6 or 8) could also be used to suppress an octagonal band about the head (10). This would eliminate the SNR loss associated with the inversion pulse. However, depending on the bandwidth of these pulses, the power deposition could be significantly higher than 1W per pulse, exceeding the 4.8W deposited from the single adiabatic inversion pulse. Furthermore, the extent of suppression can be limited by the remaining inhomogeneity. Despite this significant reduction in SNR, the glutamate resonance and even the downfield NAA resonance are clearly visible from individual pixels.

In conclusion, the combination of a transceiver array and RF shimming to optimize both the phase and amplitude of the driven coils provides for the efficient generation of a relatively homogenous B_1 field for the human head at 7T. Using the good homogeneity we have developed a short TE imaging sequence that provides excellent suppression of both the water resonance within the brain and extracerebral lipids while minimizing power deposition to ~1 W/kg for TR=3s. We have used this sequence to demonstrate the ability to detect glutamate in spectroscopic images in both controls and a patient with recurrence of a grade II oligodendroglioma.

References

1. Robitaille PM, Abduljalil AM, Kangarlou A, Zhang X, Yu Y, Burgess R, Bair S, Noa P, Yang L, Zhu H, Palmer B, Jiang Z, Chakeres DM, Spigos D. Human magnetic resonance imaging at 8 T. *NMR Biomed.* 1998; 11(6):263–265. [PubMed: 9802467]
2. Vaughan JT, Garwood M, Collins CM, Liu W, DelaBarre L, Adriany G, Andersen P, Merkle H, Goebel R, Smith MB, Ugurbil K. 7T vs. 4T: RF power, homogeneity, and signal-to-noise comparison in head images. *Magn Reson Med.* 2001; 46(1):24–30. [PubMed: 11443707]
3. Adriany G, Van de Moortele PF, Wiesinger F, Moeller S, Strupp JP, Andersen P, Snyder C, Zhang X, Chen W, Pruessmann KP, Boesiger P, Vaughan T, Ugurbil K. Transmit and receive transmission line arrays for 7 Tesla parallel imaging. *Magn Reson Med.* 2005; 53(2):434–445. [PubMed: 15678527]
4. Adriany G, Van de Moortele PF, Ritter J, Moeller S, Auerbach EJ, Akgun C, Snyder CJ, Vaughan T, Ugurbil K. A geometrically adjustable 16-channel transmit/receive transmission line array for improved RF efficiency and parallel imaging performance at 7 Tesla. *Magn Reson Med.* 2008; 59(3):590–597. [PubMed: 18219635]
5. Scheenen TW, Heerschap A, Klomp DW. Towards 1H-MRSI of the human brain at 7T with slice-selective adiabatic refocusing pulses. *MAGMA.* 2008; 21(1-2):95–101. [PubMed: 18210177]
6. Balchandani P, Pauly J, Spielman D. Interleaved narrow-band PRESS sequence with adiabatic spatial-spectral refocusing pulses for 1H MRSI at 7T. *Magn Reson Med.* 2008; 59(5):973–979. [PubMed: 18429014]

7. Balchandani P, Spielman D. Fat suppression for 1H MRSI at 7T using spectrally selective adiabatic inversion recovery. *Magn Reson Med*. 2008; 59(5):980–988. [PubMed: 18429027]
8. Mao W, Smith MB, Collins CM. Exploring the limits of RF shimming for high-field MRI of the human head. *Magn Reson Med*. 2006; 56(4):918–922. [PubMed: 16958070]
9. Ibrahim TS, Tang L. Insight into RF power requirements and B1 field homogeneity for human MRI via rigorous FDTD approach. *J Magn Reson Imaging*. 2007; 25(6):1235–1247. [PubMed: 17520721]
10. Henning A, Schar M, Schulte RF, Wilm B, Pruessmann KP, Boesiger P. SELOVS: brain MRSI localization based on highly selective T1- and B1-insensitive outer-volume suppression at 3T. *Magn Reson Med*. 2008; 59(1):40–51. [PubMed: 18050349]
11. Hetherington HP, Pan JW, Mason GF, Ponder SL, Twieg DB, Deutsch G, Mountz J, Pohost GM. 2D 1H spectroscopic imaging of the human brain at 4.1 T. *Magn Reson Med*. 1994; 32(4):530–534. [PubMed: 7997121]
12. Harpen MD. Radiative losses of a birdcage resonator. *Magn Reson Med*. 1993; 29(5):713–716. [PubMed: 8505911]
13. Vaughan JT, Hetherington HP, Otu JO, Pan JW, Pohost GM. High frequency volume coils for clinical NMR imaging and spectroscopy. *Magn Reson Med*. 1994; 32(2):206–218. [PubMed: 7968443]
14. Avdievich NI, Hetherington HP. 4 T Actively detuneable double-tuned 1H/31P head volume coil and four-channel 31P phased array for human brain spectroscopy. *J Magn Reson*. 2007; 186(2):341–346. [PubMed: 17379554]
15. Avdievich NI, Hetherington HP. 4 T actively detunable transmit/receive transverse electromagnetic coil and 4-channel receive-only phased array for (1)H human brain studies. *Magn Reson Med*. 2004; 52(6):1459–1464. [PubMed: 15562466]
16. Vaughan JT, Adriany G, Garwood M, Yacoub E, Duong T, DelaBarre L, Andersen P, Ugurbil K. Detunable transverse electromagnetic (TEM) volume coil for high-field NMR. *Magn Reson Med*. 2002; 47(5):990–1000. [PubMed: 11979579]
17. Pan JW, Twieg DB, Hetherington HP. Quantitative spectroscopic imaging of the human brain. *Magn Reson Med*. 1998; 40(3):363–369. [PubMed: 9727938]
18. Van de Moortele PF, Akgun C, Adriany G, Moeller S, Ritter J, Collins CM, Smith MB, Vaughan JT, Ugurbil K. B(1) destructive interferences and spatial phase patterns at 7 T with a head transceiver array coil. *Magn Reson Med*. 2005; 54(6):1503–1518. [PubMed: 16270333]
19. Tropp J. Mutual Inductance in the Bird-Cage Resonator. *J Magn Reson*. 1997; 126(1):9–17. [PubMed: 9252272]
20. Tkac I, Andersen P, Adriany G, Merkle H, Ugurbil K, Gruetter R. In vivo 1H NMR spectroscopy of the human brain at 7 T. *Magn Reson Med*. 2001; 46(3):451–456. [PubMed: 11550235]
21. Tannus A, Garwood M. Adiabatic pulses. *NMR Biomed*. 1997; 10(8):423–434. [PubMed: 9542739]
22. Valette J, Park JY, Grohn O, Ugurbil K, Garwood M, Henry PG. Spectroscopic imaging with volume selection by unpaired adiabatic pi pulses: theory and application. *J Magn Reson*. 2007; 189(1):1–12. [PubMed: 17851103]
23. Hetherington HP, Mason GF, Pan JW, Ponder SL, Vaughan JT, Twieg DB, Pohost GM. Evaluation of cerebral gray and white matter metabolite differences by spectroscopic imaging at 4.1T. *Magn Reson Med*. 1994; 32(5):565–571. [PubMed: 7808257]

Fig. 1.
Schematic of two elements of the 8 element transceiver coil.

Fig. 2.

The short echo sequence includes the following pulses: a single slice selective excitation (red), a broad band refocusing pulse (green), a low amplitude (~ 100 Hz) frequency selective (± 150 Hz) adiabatic inversion pulse (green) and an adiabatic inversion pulse (yellow) with a recovery delay of 350ms to suppress lipids. Gradient dephasing is used (light purple). Phase encoding is applied in 2 dimensions (orange).

Fig. 3.

Displayed in 3A are B₁ maps (amplitude and phase) relative to coil #1 from each of the 8 coils. The image in 3B displays the two ROIs used for optimizing the transmit phase (60mm diameter) and transmit amplitude (150mm×120mm). Displayed in 3C is the B₁ map obtained using simultaneous transmission from all 8 coils after optimization of the phase and amplitude.

Fig. 4.

Displayed are a coronal (4a) and axial (4b) image of B₁ acquired from a TEM using a peak power of 3.5kW. Displayed in 4c and 4d are the corresponding B₁ images acquired using an 8 element transceiver array and a peak power of 2.9kW.

Fig. 5. Displayed in 5a and 5b are the mean and standard deviation of the relative drive voltage (5a) and phase (5b) for each coil in $n=5$ subjects. The mean drive voltage and phase has been referenced to the mean value of coil #1. Displayed in the inset is a cartoon indicating the orientation of the B₁ of each coil to the head.

Maps of Phase Coherence

Fig. 6.

Displayed are plots of η for the two optimization schemes: a) where the phase is calculated and the amplitude is allowed to vary and b) where both the amplitude and phase are allowed to vary.

Fig. 7.

Displayed is the scout image showing the SI grid (24×24 over a FOV of 192×192, 10mm thickness) and spectra from 7 locations. Displayed below the scout image is a representative spectrum (location G) showing the resonances of creatine (Cr), myo-inositol (ml), choline (Ch), glutamine (Gln), glutamate (Glu) and N-acetylaspartate (NAA). The full spectra from the 7 locations are shown to the right, including the residual water resonance. No post processing of the water resonance was necessary. The inverted down field resonance at 7.8ppm is NAA.

Fig. 8.

Displayed on the left are the scout images acquired at 7T and FLAIR and post Gd injection images acquired at 3T. Displayed in the central column are 7T scout images displaying the spectral locations selected for display from the tumor (top) and a control region (bottom). The boundaries of the ROIs selected for display are shown in green on the scout images. Displayed to the far right are spectra corresponding to the ROIs selected for display. The position of the lactate resonance is denoted by Lac.