

Case Report:

Brain Spectroscopic Imaging at 3T with the 32-Channel Phased-Array Head Coil

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Introduction

The newly developed 32-channel phased-array head coil [1, 2] has shown clear advantages for high resolution imaging, functional MRI (fMRI), perfusion and diffusion imaging of brain [3]. Known for its low intrinsic sensitivity, MR spectroscopic imaging (MRSI, also named chemical shift imaging or CSI [4]) is an obvious application that can benefit from the increased signal-to-noise ratio (SNR) provided by the 32-channel technology. Increased SNR will improve the current performance and reliability of MRSI, which will help in the realization of its full clinical potential. MRSI can study a wealth of in vivo metabolic processes that are complementary to the information derived from water imaging. However, obtaining clinically beneficial MRSI data is challenged by the much lower metabolite concentration compared to water, which in turn often imposes long scanning times. Here we present brain MRSI data from a volunteer and from a brain tumor patient acquired at 3T with the standard 12-channel Head Matrix coil and with the recently released 32-channel phased-array head coil. The SNR of the

32-channel coil can be flexibly traded for reduced scan time and/or increased resolution of MRSI, or it can be used to improve the accuracy and reliability of spectral data.

Methods

All measurements were conducted on the same whole body 3T scanner (MAGNETOM Trio, A Tim System, Siemens, Erlangen) using similar MRSI/MRI protocols. One healthy volunteer and one high-grade glioma patient were studied after informed consent approved by IRB protocols at our institution. For both subjects, one high resolution 3D structural scan (0.9 mm isotropic) was acquired with a multi-echo MPRAGE (MEMPRAGE) [5, 6] sequence (TR = 2.5 s, TI = 1.2 s, TE1/TE2/TE3/TE4 = 1.59/3.19/4.79/6.39 ms, FA = 7°, NA = 1) to identify the anatomy and position the volume of interest (VOI) for the CSI measurement. In the case of the glioma patient a single dose of Gd-DTPA (Magnevist) was bolus injected before the MEMPRAGE. In order to ensure the same CSI slice prescription when changing coils, an AutoAlign scout [7] was run

prior to the spectroscopy measurement. The 2D CSI slice position was approximately at the same level from the skull base for the volunteer and for the patient. In both subjects the VOI could be positioned roughly in the center of the reference MEMPRAGE image. The hybrid 2D CSI protocol consisted of a PRESS [8] pre-selected VOI surrounded by 8 outer volume saturation bands (OVS) [9, 10] placed around the skull for lipid suppression. The WET (Water Enhanced Through T1 Effects) [1] method was used for water suppression (30 Hz bandwidth). Acquisition parameters were common for both subjects: FOV of 200 x 200 x 15 mm³, VOI 80 x 80 x 15 mm³, weighted elliptical phase encoding (16 x 16 matrix size yielding a voxel size of 12.5 x 12.5 x 15 mm³), TR = 1.5 s, TE = 30 ms, NA = 4 (TA = 7:12 min), 1250 Hz bandwidth and 1024 points for the time-domain. In addition, in the volunteer case two more 2D CSI data sets were recorded when using the 32-channel coil: 1) having the same resolution (16 x 16 matrix) and NA = 2 (TA = 4:45 min), and 2) increasing the resolution (20 x 20 matrix) and NA = 1 (TA = 6:26 min).

During data processing the data were interpolated to a 32 x 32 matrix. Processing included: k-space Fourier transformation and a spatial 50 Hz Hanning filter, subtraction of the residual water signal, time domain 1 Hz exponential apodization, zero filling to 2048 points, Fourier transformation of the time domain signals, frequency shift correction as well as phase correction and baseline correction. Data were quantified either with the fitting routine from *syngo* MR B15A or exported to jMRUI3.0 software [12] and fitted using the AMARES algorithm with soft constraints [13]. Shimming of the unsuppressed water signal from the VOI was performed with each coil until similar $T2^*$ (45 ms) and linewidth (9 Hz) values were obtained for both subjects to achieve similar spectral quality. However, in the case of the 32-channel coil convergence towards the optimal shim was slower and the shim values were higher suggesting either a slight increase in the distortion of the main magnetic field homogeneity (possibly due to the more compact geometry compared to the 12-channel coil), or an influence on the shimming algorithm of the less uniform receive profile for the 32-channel coil.

Volunteer results

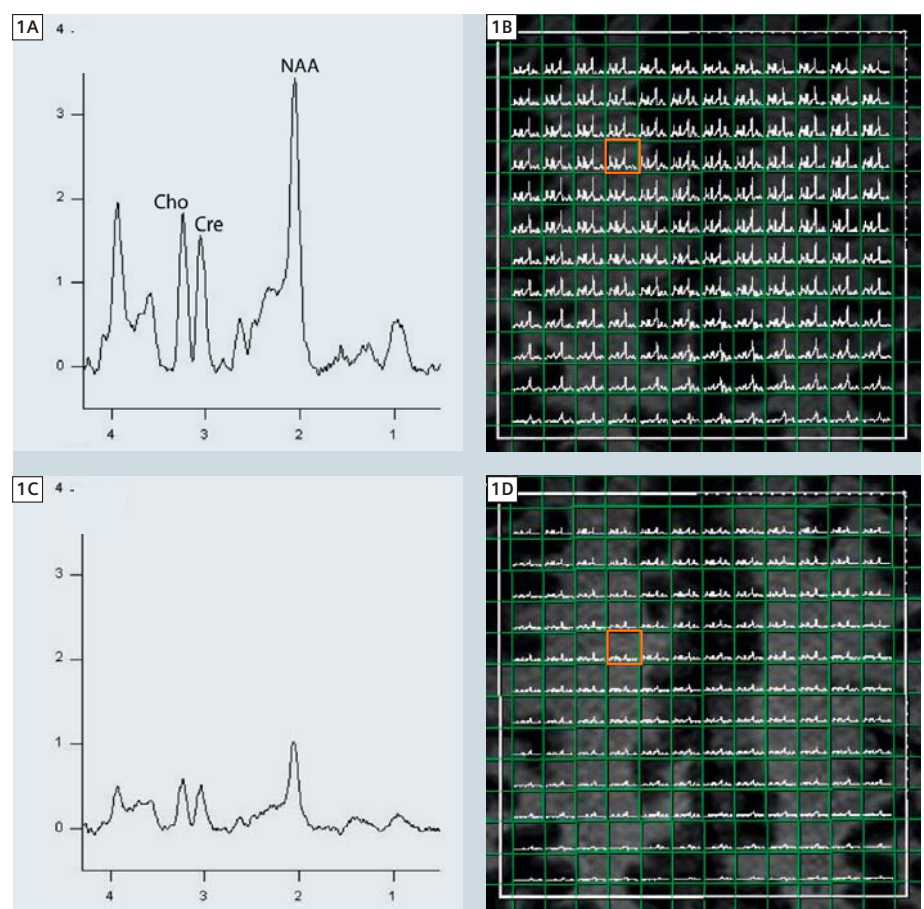
The need for averaging due to low SNR represents one of the main reasons for the increased scanning times in MRSI. To demonstrate the SNR gain of the 32-channel coil and how this can be traded to reduce scan time and/or increase resolution, three different protocols were compared on an adult (40 years age) healthy volunteer:

- 1) NA = 4, 16 x 16 weighted elliptical phase encode matrix (TA = 7:12 min);
- 2) NA = 2, 16 x 16 weighted elliptical phase encode matrix (TA = 4:45 min);
- 3) NA = 1, 20 x 20 weighted elliptical

phase encode matrix (TA = 6:26 min). The reference scan obtained with the 12-channel Head Matrix coil was acquired using the first protocol and the parameters described in the methods section.

Protocols 1 and 2 produced similar SNR with the 32-channel coil, resulting in an average gain of 3–4 times across all voxels and all metabolites compared to the reference CSI acquired with the 12-channel Head Matrix coil. A some-

what smaller SNR gain of 2–3 folds was obtained with protocol 3. Figure 1 shows examples of the 2D-CSI data acquired following protocol 2 for 32-channel coils (Figs. 1A, B) and protocol 1 for 12-channel Head Matrix (Figs. 1C, D). The spectra are scaled to the same intensity (vertical axis) and frequency (horizontal axis) ranges. Enlarged views of spectra from the voxels highlighted in red are shown in figures 1A and 1C.



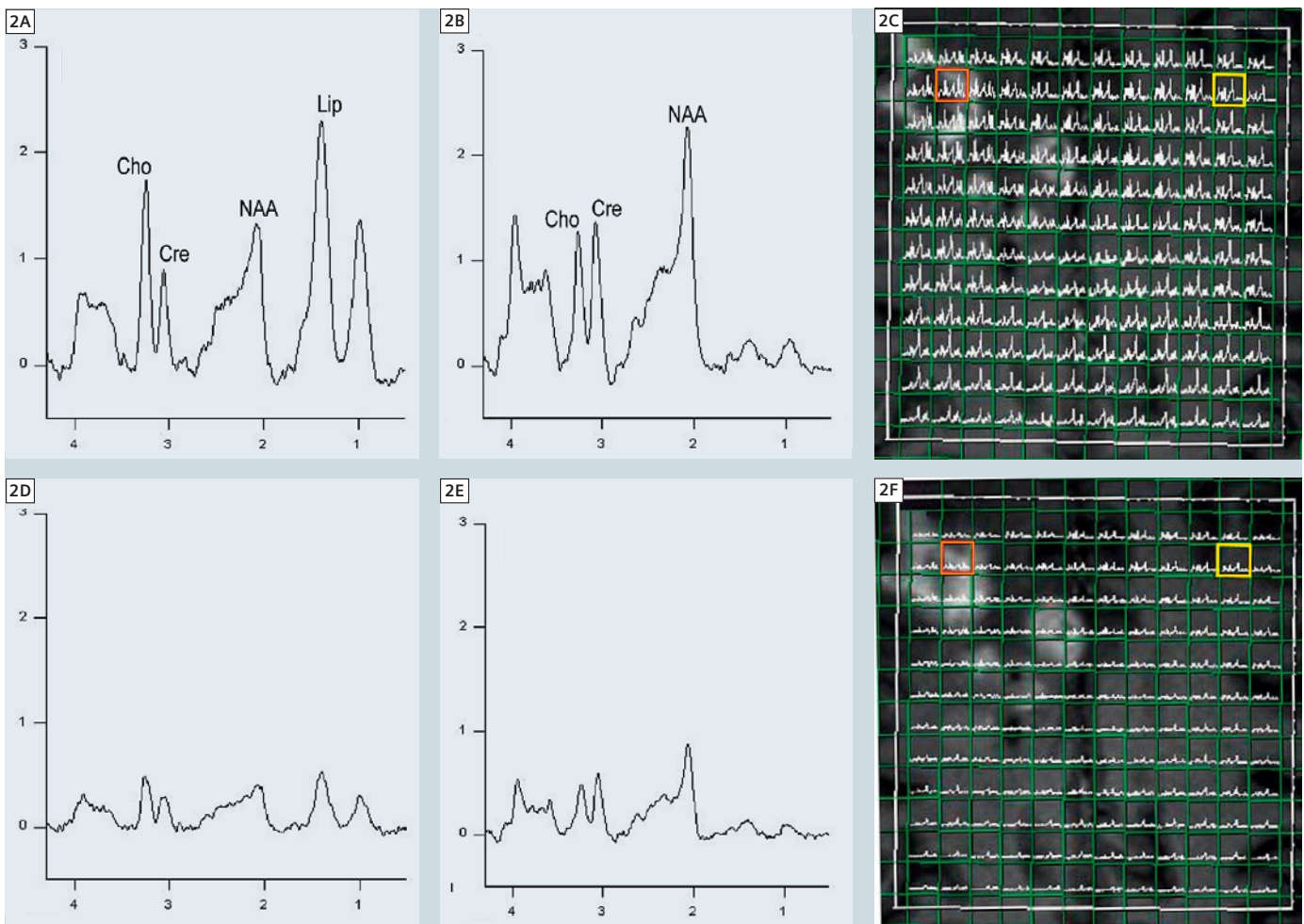
1 CSI data sets from a healthy volunteer acquired with the 32-channel phased-array head coil (upper row, NA = 2, TA = 4:45 min) and with the 12-channel Head Matrix coil (lower row, NA = 4, TA = 7:12 min). The spectra shown in 1A and 1C originate from the voxels highlighted in red in the CSI matrix (1B and 1D, respectively).

Patient results

One young patient (24 years age) with a recurrent high-grade glioma enrolled in a phase II study of a new antiangiogenic drug was imaged with structural, perfusion and diffusion-weighted imaging [14] which were followed by 2D CSI. All data were obtained within the same visit after 1 day of treatment. All the necessary data were first collected using the 32-channel coil. After switching to the 12-channel coil only the AAscout,

MEMPRAGE and the CSI protocols where repeated, adding in total an extra 15 minutes to the scan time, including the time needed for changing coils. No extra Gd-DTPA dose was injected. For consistency with the previous visits, the 2D CSI measurement with the 32-channel coil was performed with the protocol 1 (NA= 4, TA = 7:12 min). The same protocol was also employed with the 12-channel Head Matrix coil.

The patient results of Figure 2 demonstrate a 3–4 fold increase in SNR with the 32-channel coil, which is similar to the increase in SNR observed in the volunteer data. The spectra are scaled to the same intensity and frequency range. The data in the upper (lower) row are obtained with the 32(12)-channel coil. Examples of tumor (Figs. 2A, D) and healthy (Figs. 2B, E) brain spectra are shown from the highlighted voxels (red

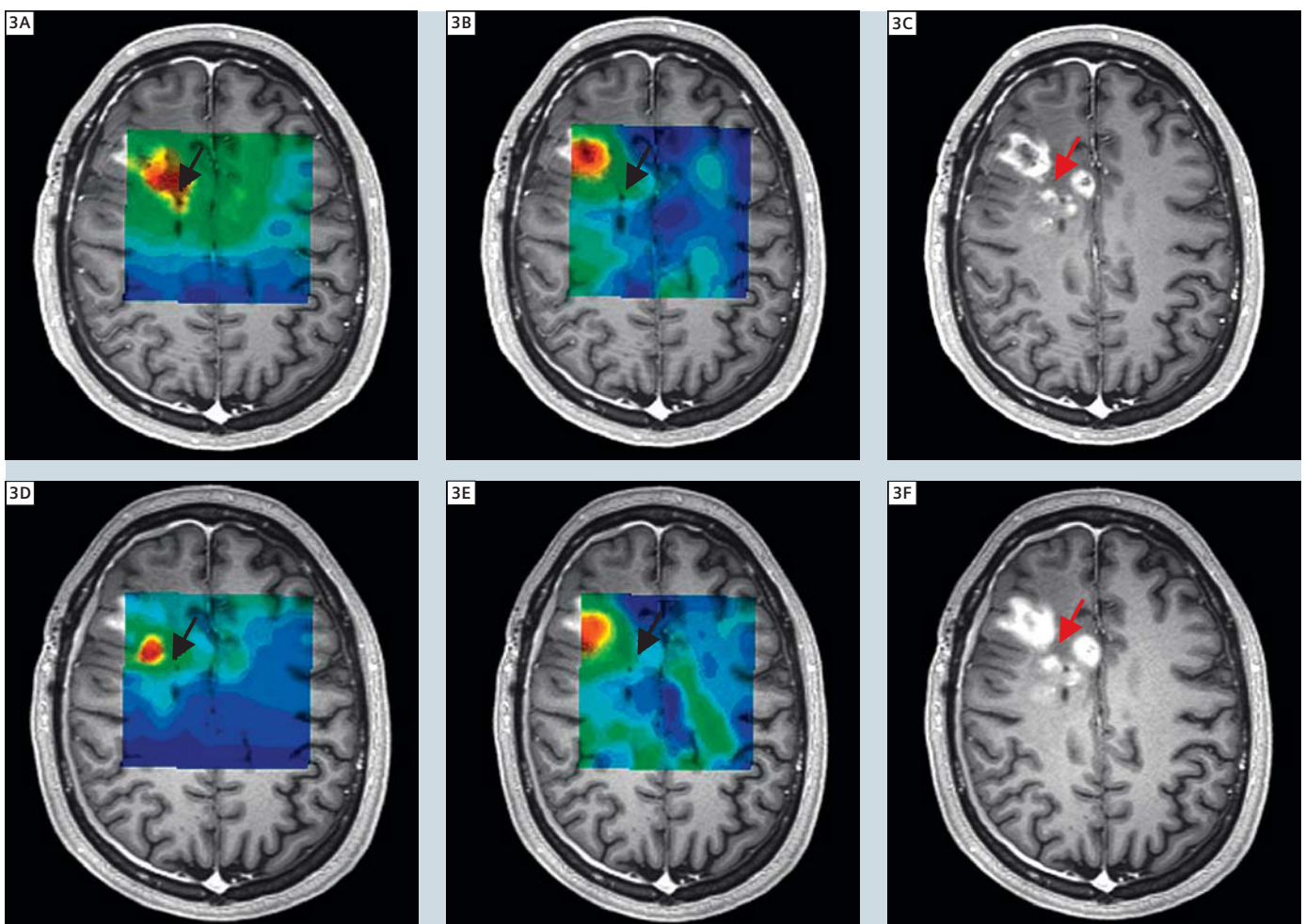


2 CSI data sets from a high-grade glioma patient acquired with the 32-channel phased-array head coil (2A–C) and with the 12-channel Head Matrix coil (2D–F). The same protocol with TR = 1.5 s, TE = 30 ms, NA = 4, TA = 7:12 min was employed. Tumor spectra are shown in 2A and 2D, originating from the red highlighted voxels (2C and 2F, respectively). Spectra of the healthy contra-lateral side are presented in 2B and 2E (yellow voxels in 2C and 2F, respectively).

and yellow, respectively) in the corresponding 2D CSI data matrix (Figs. 2C, F). A large contribution from lipid signal is found in the tumor. The particular position of the tumor in the anterior part of the VOI reduces the possibility of contamination with lipid signal from the skull due to the direction of the chemical shift error displacement from posterior to anterior (i.e. posterior voxels are more likely to be contaminated with lipid

signal from skull). A close inspection of the voxels inside and outside the VOI reveals reduced lipid signals outside the VOI. The highest lipid signals are within the lesion; it was concluded that a lipid metabolite map can be reliably computed as demonstrated in figures 3B and 3E. As evident from spectra of Figs. 2A and D the tumor is characterized also by decreased NAA and increased Cho levels, confirming together with the

lipids findings a high-grade glioma [15, 16]. A Cho/NAA map was calculated and overlaid on the Gd-DTPA enhanced image. Figure 3 shows the corresponding Cho/NAA maps (Figs. 3A, D), lipid maps (Figs. 3B, E) and post-Gd MEMPRAGE reference image (Figs. 3C, F), where the upper (lower) row data are obtained with the 32(12)-channel coil.



3 Metabolite maps and Gd-DTPA enhanced images for a high-grade glioma patient obtained using the 32-channel phased-array head coil (upper row) and the 12-channel Head Matrix coil (lower row). Cho/NAA maps are computed in 3A, D and lipid maps in 3B, E. The black arrows (3A, B, D, E) indicate areas where the 32-channel coil detects changes more reliable than the 12-channel coil. The red arrows (3C, F) point to the region of decreased intensity among the three main Gd enhancing lesions.

The maps obtained with the 32-channel phased-array head coil (Figs. 3A, B) have a better coverage of the extent of the lesion probed by the Gd image (Fig. 3C) as indicated by the black arrows. Interestingly the lipid map (Fig. 3B) overlaps well over the largest Gd enhancing lesion, while the increase of Cho/NAA ratio (Fig. 3A) seems to be more confined to the space among the three main Gd enhancing lesions, which corresponds to a region of reduced intensity in the MEMPRAGE image (Figs. 3C, F, red arrows) as compared to the contra-lateral normal side. However the extent of the Cho/NAA is better defined than the region of reduced intensity, possibly pointing to the most active part of the tumor that needs to be followed during the course of the treatment.

Conclusion

Data obtained from a healthy volunteer and from a patient with glioma indicate that CSI measurements with the 32-channel phased-array head coil can reduce by half the acquisition time (number of averages) while maintaining a 3–4 times gain in SNR compared to the 12-channel Head Matrix coil. SNR gain is also possible when simultaneously increasing the spatial resolution (with a larger phase encoding matrix) and reducing the scan time (number of averages).

Clinical benefits are demonstrated in the patient data, which show that possible hot spots of the tumors can be more reliably identified. Metabolite maps of the 32-channel coil conform better to the Gd enhancing lesions and the peritumoral regions.

The advantages reported here with the use of a 32-channel coil in the evaluation of tumors are also relevant to other brain spectroscopy applications including stroke, psychiatric or neurodegenerative diseases. The improved SNR and reliability, and the shorter scan times are likely to increase the clinical use of MR spectroscopy. They will also have considerable impact on the development of functional (dynamic) MR spectroscopy

[17] and its clinical applications. Besides reducing scan times, increasing resolution and improving accuracy, innovative ideas could utilize image encoding with the geometry of large phased arrays coils [1]; latter could then improve the chemical shift displacement error that still hampers MRSI signal localization using slice selective excitation and gradient encoding.

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