SNR and Spectral Line-Width Improvements between 1.5 and 7 T in 1H Echo-Planar Spectroscopic Imaging

R. Otazo[1], T. Li[1], A. Caprihan[1], B. Mueller[2], K. O. Lim[3], K. Ugurbil[4], L. L. Wald[5], B. G. Jenkins[6], B. R. Rosen[5], C. Zuo[7], P. Renshaw[7], J. R. Alger[8], S. Posse[9]

[1]MIND Institute, University of New Mexico, Albuquerque, New Mexico, United States, [2]Psychiatry, University of Minnesota - Center for Magnetic Resonance Research, Minneapolis, MN, United States, [3]Psychiatry, Center for Magnetic Resonance, Minneapolis, MN, United States, [4]Radiology, Center for Magnetic Resonance, Minneapolis, MN, United States, [5]Radiology, Massachusetts General Hospital - Athinoula A. Martinos Center for Biomedical Imaging, Charlestown, MA, United States, [6]Massachusetts General Hospital - Athinoula A. Martinos Center for Biomedical Imaging, Charlestown, MA, United States, [7]Brain Imaging Center, McLean Hospital, Belmont, MA, United States, [8]Radiology, University of California, Los Angeles, California, United States, [9]Psychiatry, University of New Mexico, Albuquerque, NM, United States

Introduction.

Proton NMR spectroscopy benefits from increasing the field strength due to increases in sensitivity and spectral resolution [1-2]. Signal-to-noise ratio (SNR) gains may be a linear or nonlinear function of field strength, depending on several competing factors [2]. The aim of this study was to compare SNR and spectral line width at 1.5, 3, 4 and 7 Tesla in the human brain using Proton-Echo-Planar-Spectroscopic-Imaging (PEPSI) [3] on scanners sharing similar software and hardware platform.

Methods.

Ten measurements were performed on healthy volunteers using clinical 1.5 Tesla and 3 T scanners (Siemens Sonata and Trio), and research scanners at 4 Tesla (Bruker MedSpec) and 7 Tesla equipped with Siemens console and gradients. Quadrature head coils and 8-channel surface array coils were employed. Nine short-TE (30 ms) PEPSI measurements were performed with 8-slice outer volume suppression. Encoded spectral width increased with field strength. Even- and odd-echo data were reconstructed separately using a water reference scan as described previously [3]. One measurement at 4 T was performed with conventional phase encoding and PRESS volume localization at TE: 30 ms. Sensitivity was compared using a normalized SNR defined in equation (1) [4-5], where T_{adq} is the total measurement time and V_x is the voxel volume. To measure SNR_{meas} in a homogeneous white matter region, we used either the NAA peak amplitude S_{NAA} (eq. 2), or the NAA peak integral C_{NAA} (eq. 3), and the standard deviation of the noise measured outside of the brain in a corner voxel, assuming that thermal noise is independent of the spatial location [5]. An exponential matched-filter (using the NAA line width at each field strength) was used to maximize the SNR of the NAA peak. For line width comparisons, we used the full-width at half-maximum (FWHM) of the NAA peak.

$$SNR_{v/t} = \frac{SNR_{meas}}{\sqrt{T_{adq}} \cdot V_X} \quad (1) \qquad SNRs = \frac{S_{NAA}}{\sigma} \quad (2) \qquad SNRc = \frac{C_{NAA}}{\sigma} \quad (3)$$

Results.

Table 1 summarizes pulse sequence parameters and results. Figure 1 shows examples of the spectra obtained at different field strengths. As expected, the $SNR_{v/t}$ is independent of the spectral width (SW) [5] and the $SNR_{v/t}$ of PEPSI is similar to conventional phase encoding [3]. The field strength dependence of NAA SNR is shown in figure 2, for the CP coil and in figure 3 for the phased-array coil. A power law explains the field strength dependence better than a linear or quadratic function (as evidenced by a larger R^2 value). For the phased-array coils we measure better SNR than with the CP coils at the edges of the FOV, as expected, whereas the sensitivity is similar in the center. The NAA line width in Hz increases with field strength, but on the ppm scale (figure 4) it decreases approximately linearly with field strength.

B_o [T]	SW [Hz]	V_X [cc]	T_{ADQ} [min]	$SNRs_{v/t}$ [$min^{-1/2}cc^{-1}$]	$SNRc_{v/t}$	LW [ppm]
1.5	770	2	8	13.54	3.12	0.0675
1.5	770	2	8	15.32	2.82	0.0650
3	770	2	1	22.29	4.11	0.0597
3	770	2	4	21.93	3.91	0.0589
3	926	2	16	23.05	3.76	0.0556
4	770	2	4	29.39	5.17	0.0595
4	926	1.9	8.5	31.15	4.96	0.0579
4*	2000	1.9	8.5	31.7	5.26	0.0571
7	1380	2	1	39.54	6.12	0.0451
7	1380	2	8	38.3	6.15	0.0447

Table 1: In vivo experiments parameters (*=PRESS-CSI).

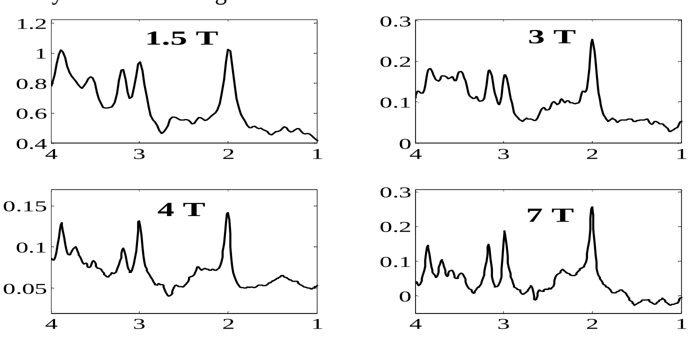

Figure 1: Spectra at different Bo's (2 cc)

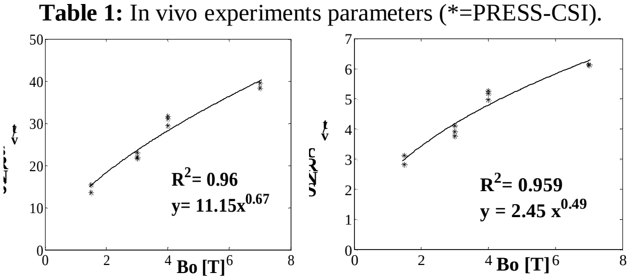

Figure 2: SNR vs. B_o (CP Coil).

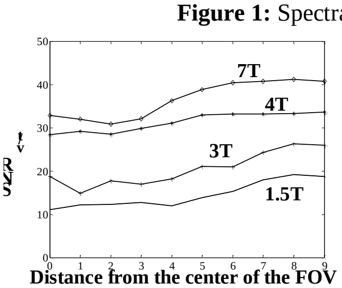

Figure 3: SNR vs. B_o (PA Coil).

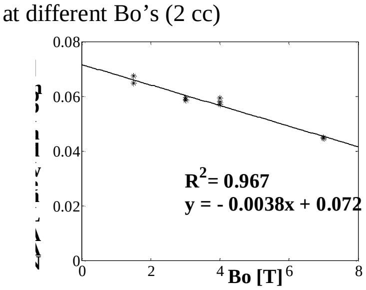

Figure 4: Line width vs. B_o.

Discussion.

This study demonstrates feasibility of high speed spectroscopic imaging over a wide range of field strength and considerable gains in sensitivity and spectral line width with field strength. These gains suggest that very high-spatial resolution metabolite mapping is feasible at high field strength. The decrease in NAA line width on the ppm scale may in part be due to the less than linear decrease in metabolite T2 values with field strength [6].

References.

[1] R. Gruetter *et al*, J. Magn. Reson. **135**, 260 (1998). **[2]** K. Ugurbil *et al*, Magn. Reson. Imag. **21**, 1263 (2003). **[3]** S. Posse *et al*, Magn. Reson. Med. **33**, 34 (1995). **[4]** R. Pohmann *et al*, J. Magn. Reson. **129**, 145 (1997). **[5]** A. Macovski, Magn. Reson. Med. **36**, 494 (1996). **[6]** S. Posse *et al*, Magn. Reson. Med. **33**, 246, 1995. *Supported by NIDA 1 R01 DA14178-0,* NIH RR008079 *the MIND Institute and the Keck Foundation.*