Implementation and Comparison of LASER- and Semi-LASER-based MRSI Pulse Sequences at 9.4T

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Introduction: Magnetic resonance spectroscopic imaging (MRSI) allows signal acquisition from multiple voxels in an imaging plane simultaneously. Due to its ability to non-invasively assess metabolic characteristics of the sample, MRSI has great potential for clinical applications and basic researches. In MRSI, the localization performance is the most important factor in order to obtain metabolite maps without spectral contaminations from unwanted signal such as peripheral lipid. Slab-selection in combination with outer volume suppression (OVS) is commonly used for spatial localization at low field (e.g., 1.5 and 3.0T). However, at high field, B_0 and B_1 inhomogeneity are of great concern for successful implementation of MRSI sequences. For this reason, a Point-REsolved SpectroScopy (PRESS)-based voxel localization method is often employed in MRSI. However, the resulting chemical shift displacement error (CSDE) can be substantial at high field. Given the excellent localization performance of semi-Localized by Adiabatic SElective Refocusing (sLASER) [1] and LASER [2] single voxel MRS pulse sequences, we have implemented sLASER- and LASER-based MRSI sequences (Fig.1) at 9.4T and compared their performance both in phantom and in vivo.

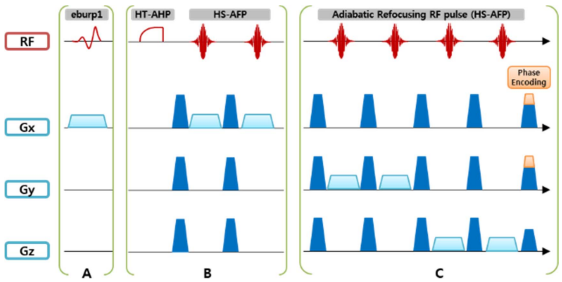

Figure 1: *A sLASER-based (A+C) and a LASER-based (B+C) MRSI pulse sequences. A total of 4 and 6 adiabatic pulses are used, respectively. In the sLASER-based sequence, a self-refocusing, excitation pulse (eburp1) was used to reduce TE.*

Methods: The animal study protocol was approved by IACUC. All experiments were performed on a 9.4T animal MR scanner (Agilent). A volume coil was used for both RF transmission and signal reception (Agilent). The first- and the second-order shimming were performed by using FASTMAP and 3Dgshim for phantom and in-vivo experiments, respectively. To evaluate the localization performance of the implemented MRSI pulse sequences, projection profiles of a voxel were obtained in the three orthogonal directions using a spherical water phantom (Fig.2), and the results were also compared to those obtained by using a conventional MRSI sequence with slab-selection and OVS. An eburp1 [3] (duration=500ms, bandwidth=10kHz) was used as an excitation RF pulse to minimize echo time (TE) for sLASER sequence. For LASER sequence, a hyper tangent adiabatic half passage (HT-AHP) [4] (duration=1000ms, bandwidth=10kHz) and a hyper secant adiabatic full passage (HS-AFP) [5] (duration=1400ms, bandwidth=10kHz) pulses were used for excitation and refocusing, respectively. The sequence parameters were: TR=2000ms, TE=12.26, 25.34, and 6.00ms for sLASER-based, LASER-based, and conventional MRSI sequences, respectively, and ROI=16x16x2mm^3. To further evaluate the localization performance of the implemented sequences against strong peripheral lipid signal, spectra were acquired without phase encoding from a spherical phantom (Fig.3) with two compartments (the inner compartment containing a metabolite solution (NAA, PCh, PCh, Gln, Tau, and m-Ins) and the outer compartment containing soy oil). The sequence parameters were: voxel size=8x8x2mm^3, 64 averages, OVS, and VAPOR water suppression. Finally, in-vivo data were acquired by using the sLASER MRSI sequence from the brain of a Sprague Dawley rat. The sequence parameters were: FOV=8x10x2mm^3, matrix size=4x8x1, pixel size=2x1.25x2mm^3, TR/TE = 2000/12.26ms, 64 averages, OVS, and VAPOR.

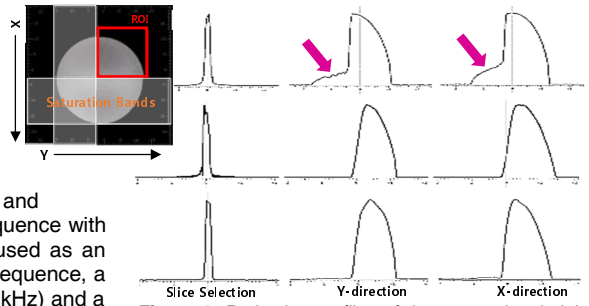

Figure 2: *Projection profiles of the conventional slab-selection MRSI sequence with OVS (top), LASER-based sequence without OVS (middle), and sLASER-based sequence with OVS (bottom). The definition of the ROI is also shown along with the 2 saturation bands.*

Results/Discussion: In Fig.2, the poor localization performance of the conventional sequence with slab-selection and OVS is problematic even in phantom. The localization performance of the sLASER-based MRSI sequence in combination with OVS is comparable to that of the LASER-based sequence without OVS. In Fig.3c, only negligible amount of lipid signal is observed in the spectrum obtained with the sLASER-based MRSI sequence with OVS despite the considerable amount of lipid in the outer compartment of the phantom. The poor localization performance of the conventional sequence is clearly demonstrated in Fig.3b where those metabolite signals are not visible due to the strong lipid signal in this spectrum scale. For the LASER-based MRSI sequence, excellent spatial localization was achieved without the need of an OVS module. However, the relatively long TE of 25.34ms with the sequence can result in unwanted signal loss due to T_2 relaxation and J-evolution of spin systems, which can be substantial in vivo. In Fig.4, in-vivo metabolite maps for NAA, tCr and tCho were successfully obtained with the sLASER-based MRSI sequence from a rat brain without spectral contamination from the peripheral lipid.

Conclusion: Given the comparable volume localization performance along with the shorter minimum TE attainable and less SAR with a sLASER-based MRSI sequence relative to a LASER-based MRSI sequence, a sLASER-based MRSI sequence may be the sequence of choice for in-vivo MRSI at high field.

References: [1] Scheenen TW et al, MRM. 2008;59(1):1-6. [2] Slotboom J et al, JMR (1969). 1991;95(2):396-404. [3] Nägele T et al, MRM. 1993;11(6):785-97. [4] Shen J. JMR, Series B. 1996;112(2):131-40. [5] Robin Bendall M et al, MRM. 1987;4(5):493-9.

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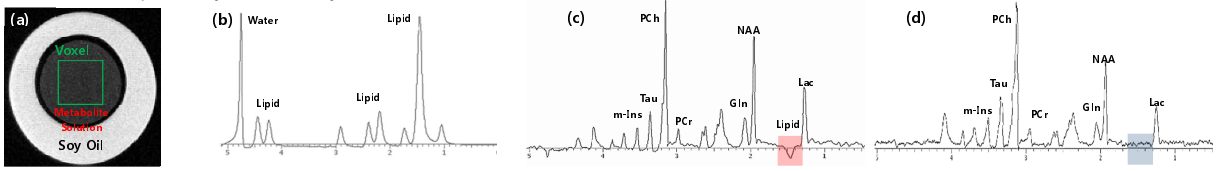

Figure 3: *(a) Voxel plan on the metabolite-oil phantom, (b) spectrum with the conventional MRSI sequence (c) spectrum with the sLASER-based MRSI sequence, and (d) spectrum with the LASER-based MRSI sequence.*

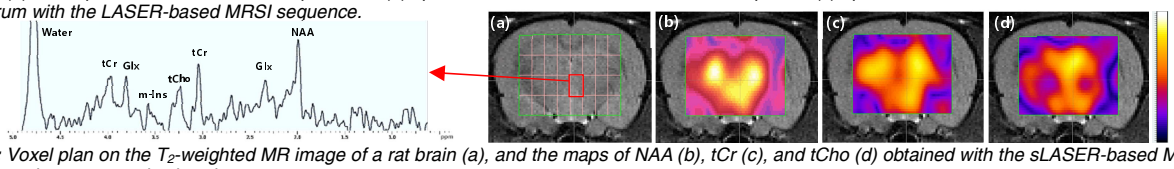

Figure 4: *Voxel plan on the T_2-weighted MR image of a rat brain (a), and the maps of NAA (b), tCr (c), and tCho (d) obtained with the sLASER-based MRSI sequence. A representative spectrum is also shown.*