Natural Linewidth Chemical Shift Imaging (NL-CSI)

A. Bashir[1], D. A. Yablonskiy[1]

[1]Mallinckrodt Institute of Radiology, Washington University in St. Louis, St. Louis, MO, United States

INTRODUCTION: Due to the low density of metabolites and scan time restrictions spectroscopic imaging is typically done at low spatial resolution, i.e. only 16 or 32 phase encoding steps are usually acquired. When Discrete Fourier transform is used for spatial reconstruction there is substantial inter-voxel signal contamination due to the sinc-like point spread function (Gibbs ringing) inherent to Fourier transform (1). Few methods have been designed to reduce/eliminate this inter-voxel spectral leakage (2-4). In this study we show that the presence of magnetic field inhomogeneities creates additional to Gibbs-ringing mechanism of inter-voxel spectral leakage. We also demonstrate a new method (Natural Linewidth CSI (NL-CSI)) that allows complete removal of field inhomogeneity effects and Gibbs ringing from the acquired MR signal so that the signal decay is determined only by the internal "natural" R2 relaxation rate constant, hence the acronym – Natural Linewidth CSI.

THEORY: MR signal acquired during a 1D phase encoded spectroscopic experiment is given by

$$s(k_n, t) = \int_{-\infty}^{\infty} \int_{\Re} \rho(\Re, f) e^{-i2\pi(k_n x - ft)} e^{-i\phi(\Re, t)} d\Re df \qquad (1)$$

where $\rho(\Re, f)$ is the spin density function representing the spatial distribution of the spectral information of the excited region $\Re$ and k_n is the k-space value corresponding to the *n*-th phase encoding step. If excited region $\Re$ is segmented into *M* homogeneous compartments with spin densities $\rho_m(t)$, then Eq. 1 can be written as a set of linear equations

$$s_n(t) = \sum_m g_{nm} \rho_m(t); \quad where \quad g_{nm}(t) = \int_{\Re_m} e^{-i2\pi k_n x} e^{-i\phi(\Re, t)} d\Re \quad and \quad \rho_m(t) = \int_{-\infty}^{\infty} \rho(\Re, f) e^{i2\pi ft} df \qquad (2)$$

Compartmental FID signal $\rho_m(t)$ can now be obtained by solving this set of linear equations. The important difference from earlier proposed SLIM technique (4) is that the encoding matrix takes into account existence of magnetic field inhomogeneities becoming a function of time.

METHODS: 1D simulations were performed to show the effect of field inhomogeneities on CSI experiment. Simulation parameters: FOV = 256 mm, phase encoding (PE) steps = 16, vector size of 1024 and SW = 2kHz. PE gradient duration was 750 μsec. A background field gradient of 1% of the maximum PE gradient strength was used to mimic field inhomogeneity for simulation studies. The simulated object occupied 96 mm (grey area in Fig 1) in the field of view. The simulated data was analyzed by Fourier, SLIM and NL-CSI reconstruction algorithms. Phantom and *in vivo* human 1D CSI experiments were done on Siemens 3T Magnetom Allegra system with the following parameters; Fourier voxel size = 1.6 x 1.6 x 1.6 cm^3, TR/TE = 1500/40 ms, SW = 2kHz, and 8 averages. High resolution 3D images were acquired with isotropic 1mm^3 resolution at echo times of 3 and 20 ms. The magnitude images were used to segment the object into homogeneous compartments and the phase difference at the two echo times was used to calculate the field inhomogeneities. For phantom studies, the compartments were chosen to align with the Fourier voxels. For *in vivo* experiments the skull and scalp were chosen as independent compartments, while the brain compartments were chosen to align with Fourier voxels. Spatial reconstruction was done using Fourier, SLIM and NL-CSI reconstruction algorithms.

RESULTS AND DISCUSSIONS: Fig 1 shows simulation results (natural T2 relaxation is ignored to amplify effects of field inhomogeneities). One would expect that Fourier reconstruction results in monotonous signal decay due to phase dispersion in the presence of field gradient in voxel (a) occupied by the object, and much smaller decaying signal in voxel (b) neighboring to the object. In fact, the signal in voxel (a) oscillates for short times and then decays rapidly at around 70ms. The signal in voxel (b) initially increases with time. Such behavior is present for both Fourier and SLIM reconstructions and can only be explained by substantial cross-contamination of signals from adjacent voxels. Only NL-CSI reconstruction algorithm provides "pure" signal free of cross-voxel contamination and inter-voxel phase dispersion. Figure 2 shows data from a human subject. Semi-log plots demonstrate highly nonlinear signal decay with Fourier and SLIM reconstruction, whereas the NL-CSI reconstructed signal displays practically pure T2* decay. Peak fitting of signal also showed a decrease of 23% in proton and 52% in NAA signal line-width. NL-CSI was also very effective in removing the extra-cranial lipid signal leakage as compared to Fourier reconstruction (signal oscillations are present only in Fourier reconstructed data). Phantom data (not shown) also demonstrate similar results with significantly improved line shapes.

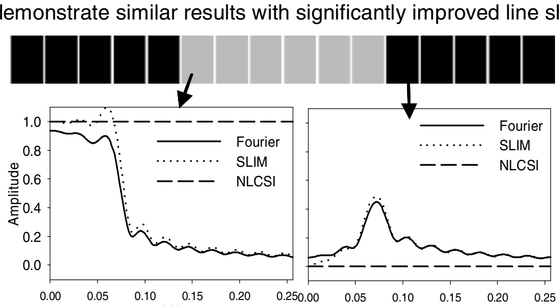

Figure 1: Image (upper panel) shows the simulation object (gray area) and Fourier voxels. The plots are reconstructed FIDs from selected voxels.

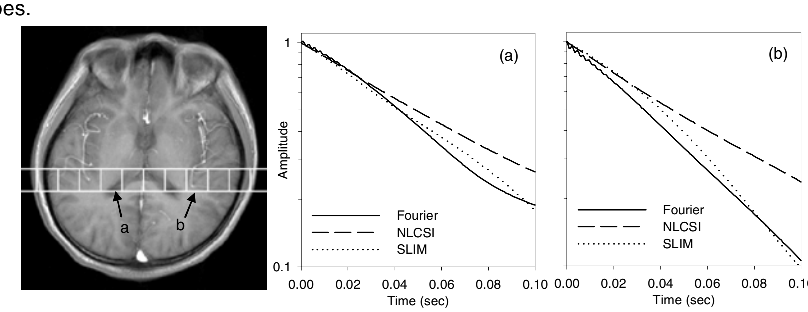

Figure 2: Image showing location of 1D spectroscopy region along with location of Fourier voxels. The plots (semi-log scale) show reconstructed FID signals from two selected voxels.

CONCLUSIONS: We have demonstrated that even small magnetic field inhomogeneities present during any spectroscopic imaging experiment can lead to severe contamination of spectroscopic data and can severely compromise the results of spectroscopic imaging if discrete Fourier transform is used for spatial reconstruction. We have also demonstrated a new non-Fourier approach – NL-CSI – for spatial reconstruction of CSI data. Our approach takes into account magnetic field inhomogeneities and produces voxel spectra which are free of inter-voxel contamination and intra-voxel phase dispersion that significantly improves signal lineshape.

REFERENCES: 1. Mathews J. et. al. Mathematical Methods of Physics (1970); 2. Plevritis S. K. et. al. *Magnetic Resonance in Medicine* (1995); 3. Kienlin M. V. et. al. *Journal of Magnetic Resonance* (1991); 4. Hu X. et. al. *Magnetic Resonance in Medicine* (1998).