Double Spin-Echo Spiral Chemical Shift Imaging for Rapid Metabolic Imaging of Hyperpolarized [1-^{13}C]-Pyruvate

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Introduction

Rapid chemical shift imaging (CSI) techniques are of considerable interest for hyperpolarized ^{13}C metabolic imaging [1,2]. Undersampled spiral CSI (spCSI) using a free induction decay (FID) acquisition mode has been applied to real-time metabolic imaging with acquisition times of less than 1 s per slice [3]. The FID acquisition necessitates phase correction during reconstruction for accurate spectral quantitation as the metabolite signals cannot be separated in magnitude mode. Phase correction can be challenging, especially with contributions from aliased out-of-phase peaks. However, when the full spin echo is acquired, the magnitude operation does not increase the linewidth, hence eliminating the need for phase correction. This work extends the spiral CSI sequence to incorporate a double spin-echo (DSE) [4] using a pair of adiabatic refocusing pulses.

Methods

All measurements were performed on a GE 3T MR scanner equipped with self-shielded gradients (40 mT/m, 150 mT/m/ms). A custom-built dual-tuned (^{1}H/^{13}C) quadrature coil (dia=80 mm, length=90 mm) was used for both RF excitation and signal reception. Healthy male Wistar rats were anesthetized with 1-3% isoflurane in oxygen (~1.5 l/min). The rats were injected in a tail vein with 2.5-3 ml of 80-mM solution of [1-^{13}C] pyruvate that was hyperpolarized using HyperSense DNP (15-20% liquid state polarization). 2D ^{13}C data were acquired using the DSE-spCSI sequence with slice-selective excitation followed by 2 adiabatic refocusing pulses for a double spin-echo (pulses described in [4]) and a spiral readout trajectory that encoded the spectral and both spatial dimensions. The spiral waveforms were designed for FOV=80x80 mm^2 and nominal 5x5 mm^2 in-plane resolution, with 3 spatial interleaves and spectral width of 276.2 Hz. The TE was 151 ms for DSE-spCSI and 3 ms for FID-spCSI.

The DSE acquisition was compared to FID in a single time-point measurement of a 10 mm axial slice through the kidneys. Imaging was started 20 s after the injection of hyperpolarized pyruvate. A variable-flip-angle scheme [5] of 35º, 45º, and 90º was used for the three spiral interleaves to account for depletion of the longitudinal magnetization with multiple excitations and excite the same amount of transverse magnetization for each acquisition. Longitudinal relaxation and metabolic turnover can be neglected for the short time duration between the excitations (125 ms for FID and 220 ms for DSE). The data were apodized with 15 Hz Gaussian line broadening in the time dimension and reconstructed as described in [3]. Metabolic maps for pyruvate (Pyr), lactate (Lac), alanine (Ala) and bicarbonate (Bic) were calculated by integrating the signal within 20 Hz (10 Hz for Bic) around each peak in absorption mode for FID and magnitude mode for DSE.

The DSE-spCSI sequence was also used in dynamic imaging experiments and compared to FID-spCSI, for hyperpolarized Pyr solution in vitro and in vivo. A constant 5º flip angle excitation was used and images were acquired every 3 s, starting at the time of injection. Another DSE acquisition was done with the acquisition starting at the end of injection (15 s delay).

Results and Discussion

Figure 1 shows the metabolic images of Pyr, Lac, Ala and Bic superimposed onto a ^{1}H single-shot FSE image (2-mm slice) for anatomical reference. The spectra (magnitude spectra for DSE and absorption spectra for FID) from a voxel in the kidney are also shown**.** The DSE-spCSI exhibits high quality spectra, with the magnitude mode linewidth being comparable to absorption mode FID-spCSI spectra. The Bic signal in particular is obscured in the FID acquisition due to the contribution from the out-of-phase aliased Pyr peak, but can easily be distinguished in the DSE magnitude spectrum. Automatic phase-correction of the FID spectra failed for Bic due to its proximity to the Pyr peak, and the spectrum from the kidney voxel was manually phase-corrected to obtain the Bic peak shown here. An added benefit of the DSE sequence is that it suppresses the signal in blood vessels, as crusher gradients around the refocusing RF pulses attenuate the magnetization from flowing spins. In the FID, the high Pyr signal in the vessels can contaminate the signal from the kidney at the resolution used here.

To verify that the DSE-spCSI preserves signal over multiple excitations, the T_1 of hyperpolarized Pyr solution in a syringe was measured. The T_1 of 59 sec observed with DSE-spCSI was similar to the 60 sec measured with slice-selective FID acquisition.

The time courses of the mean Pyr and Lac signals (corrected for polarization level) from an ROI in the right kidney are shown in Fig. 2. The DSE showed lower signal and decayed slightly faster than the FID, as the refocusing pulses spoil any magnetization passing through the transition bands/fringe field at the edges of the RF coil. This is a significant loss when there is Pyr flowing through anatomy lying near the coil edges. Delaying the acquisition until the end of injection may provide some gain in this case.

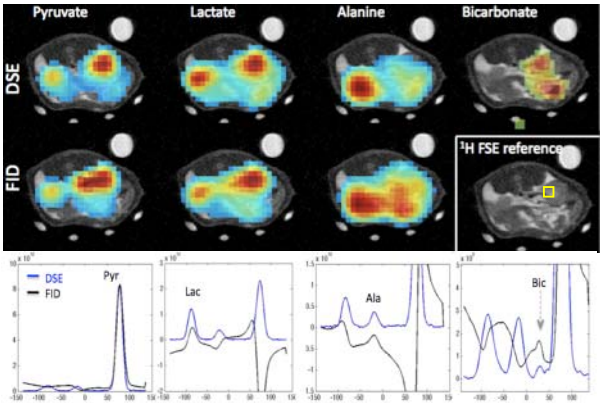

Figure 1: Metabolic images of Pyr, Lac, Ala, and Bic along with spectra from a voxel in the rat right kidney (voxel location shown in ^{1}H anatomical image). DSE-spCSI magnitude mode yields similar linewidth as FID-spCSI absorption mode spectra, eliminating the need for phase correction. Bic image is not shown for FID as it was contaminated by out-of-phase Pyr signal.

Conclusion

This work demonstrates the use of DSE-spCSI for fast metabolic imaging, attaining high quality spectra without the need for phase correction along with the advantage of attenuating signal in the vessels. Care must be taken for dynamic imaging to ensure that the spins remain within the RF coil volume.

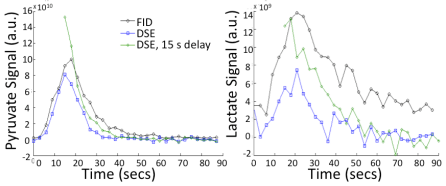

Figure 2: Time course of mean signal in right kidney. The DSE sequence allows time-resolved metabolic measurements, but loses some signal compared to FID acquisition as the refocusing pulses crush magnetization passing through RF coil edges.

References : [1] Golman K *et al* [2006], Acad Radiol 13:932-942, [2] Kohler S *et al* [2007], MRM 58:65-69, [3] Mayer D *et al* [2009], MRM 62:557:564, [4] Cunningham C *et al* [2007], JMRI 187:357:362, [5] Zhao L et al [1996], JMR 113:179-183

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