31P NMR Spatial Localization in Heart with Inhomogeneous Surface Spoiling Gradient

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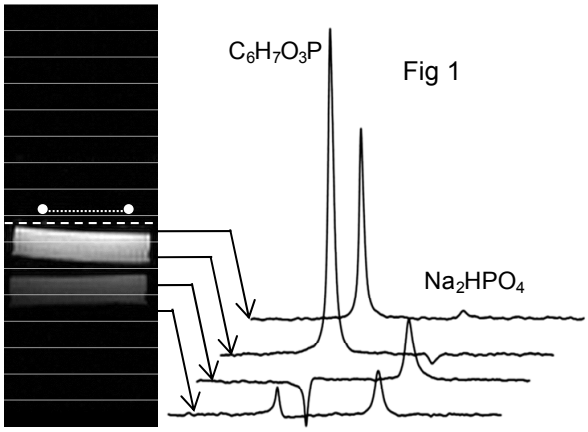
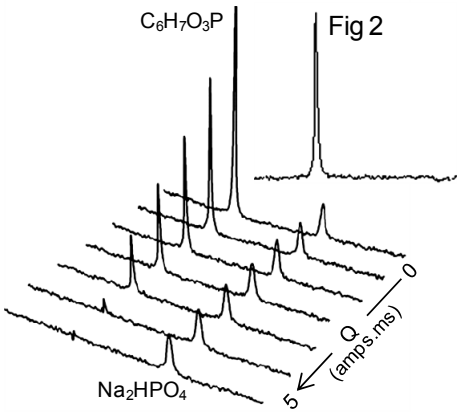
Introduction: Phosphorous (^{31}P) magnetic resonance spectroscopy (MRS) is the only tool to non-invasively measure high energy phosphate metabolism *in vivo*. It is regularly used to measure phosphocreatine (PCr) and adenosine triphosphate (ATP) levels as an estimate of cardiac energy status in a variety of disease processes [1]. *In vivo* localization of ^{31}P signal from heart requires the removal of signal contribution from the chest muscle. 1D-CSI in combination with surface coil reception is the most widely used method for cardiac ^{31}P-MRS localization. However, the discrete Fourier transform used to spatially construct low resolution phase encoded CSI data can lead to Fourier ringing and contaminate signal from heart tissue. Furthermore, the respiratory motion of the chest muscle can make signal contamination worse in the phase encoded experiment [2]. We have previously demonstrated the use of a superficial and a highly non-linear magnetic field gradient to eliminate the signal contribution from surface lying tissue [3]. Spins located within the inhomogeneous gradient rapidly loose coherence and, thus, the net magnetization produced by the spins is zero. The magnetic field gradient only exists within the surface lying regions, therefore, it does not affect the signal from deep lying heart tissue. Here we demonstrate the applicability of this technique for cardiac ^{31}P spectroscopy and compare it with 1D-CSI spatial localization.

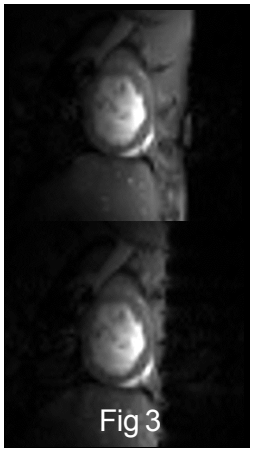
Methods: All experiments were done on 4.7-T Agilent/Varian scanner. A two compartment phantom containing 0.5 M sodium phosphate (Na_2HPO_4) and phenylphosphonic acid ($C_6H_7O_3P$) was constructed to give distinct ^{31}P resonances. The depth of each compartment was ~5-6 mm. A surface-spoiling gradient coil (SSGC) was built with linear anti-parallel current elements spaced 5 mm apart and running perpendicular to the magnetic field. The SSGC coil was placed on top of the phantom and a 2.5 cm diameter ^{31}P surface coil was used to acquire data. A TTL pulse from spectrometer was used to trigger the SSGC gradient driver circuit.

A 1D-CSI experiment was performed with FOV = 64 mm, matrix size = 16, TR = 2 sec and TE = 1ms. A pulse and acquire sequence was used for SSGC experiments. Acquisition was delayed after the excitation pulse to accommodate the inhomogeneous spoiling gradient of 1ms duration. The charge (Q) through the SGCC was varied from 0 – 5 (amps.ms). Since it is not possible to distinguish between the ^{31}P signal from chest and heart, to test the performance of the SGCC in animal experiments we acquired high resolution ^{1}H images of the heart with and without application of the inhomogeneous spoiling gradient.

Results: Figure 1 shows the image of the phantom and CSI grid. The schematic layout of the SSGC (heavy dashed line) and RF coil is also displayed. The spectrum from selected voxels of 1D-CSI experiment demonstrates inter-voxel signal leakage, e.g., phenylphosphonic acid ^{31}P signal (left peak) is present in voxels that only contain sodium phosphate in the CSI experiment. The spectrum from the SSGC experiment as a function of charge through the coil is shown in figure 2. The signal from the regions of the phantom closer to the gradient coil disappears as the charge is increased with almost no affect on the signal from deep lying regions of the phantom (inset shows the difference spectrum between Q=0 and 5 (amp.ms)). The rat heart image without and with charge (Q = 4 amps.ms) is shown in figure 3 demonstrating elimination of signal from the chest tissue.







Conclusions: The 1D-CSI can lead to severe inter-voxel signal contamination. This contamination is made worse when a surface coil is used for acquisition as the chest muscle is positioned in the high sensitivity region of the coil. Using the surface spoiling gradient coil approach, the signal can be quickly and effectively localized to the deep lying tissue with little or no contamination. The difference spectrum (fig 2) demonstrates that there is almost negligible effect on the signal from deep regions of the phantom. Another benefit of SSGC coil is that it will move with the chest and can inherently compensate for breathing motion. If further localization is needed, the SGCC approach can be easily combined with other spatial localization techniques.

References: [1] Beadle R, Frenneaux M; *Expert Rev Cardiovasc Ther*; 2010. [2] Bashir A, Yablonskiy DA; *Magn Reson Med;* 2006. [3] Chen W, Ackerman JJ*; NMR biomed*; 1990.