Single-Shot Interleaved Gradient Compensation of Susceptibility Induced Spectral Line Broadening in Proton Spectroscopic Echo-Planar Imaging (PEPSI)

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Introduction Brain MR spectroscopic imaging (MRSI) suffers from spectral line broadening in orbito-frontal cortex and medial temporal regions (amygdala, hippocampus) caused by susceptibility induced line broadening. The severity of the line broadening can be reduced by using smaller voxel sizes and by acquisition of multiple data sets with different shim settings. In the fMRI literature single- and multi-shot gradient compensation schemes, such as Z-shimming[1] or multi-echo EPI[2] with interleaved gradient compensation have shown significant signal recovery in frontal and medial temporal regions. In this study we translate this advanced fMRI technology into the MRSI domain and present a novel single-shot spectroscopic imaging method that compensates local gradients in regions that suffer from susceptibility related spectral line broadening. The new method was integrated into a Proton Echo-Planar spectroscopic imaging (PEPSI) sequence [3] and uses a train of alternating gradient pulses of increasing strengths, interleaved into the spatial-spectral encoding scheme, to simultaneously collect an uncompensated and a compensated data set. Controlled spectral aliasing was used to overcome gradient slew rate and stimulation limitations. The NAA and the choline/creatine peaks now fall on the opposite sides of the water peak and can be deconvolved by suitable techniques.

Theory Compensating gradients along all three spatial axes are applied between the readout gradient lobes of a PEPSI sequence (Fig.1) to reverse the phase shift introduced by a local gradient G_l. The compensating gradients (Fig. 1) are alternating in sign and increase linearly in amplitude in order to simultaneously collect uncompensated and compensated data in a single shot. In order to compensate the kth even echo we require that the gradient moments observe $(2k-1)G_l T_d = G_c T_c$. The increasing compensation gradient amplitude will be limited by gradient slew rate and maximum amplitude. In order to achieve linear gradient moment increases during a long spectroscopic readout it is necessary to elongate the compensation gradients, which requires sacrificing spectral width. As a result the metabolite signals will alias into the encoded spectral range. In order to separate the spectral peaks with equal spacing the spectral width was optimally selected as a function of field strength using a minimization procedure. Fig. 2A displays the peaks in a conventionally acquired proton spectrum. With spectral aliasing due to reduced spectral width the sequence of peaks changes as shown in Fig. 2B. In this example the NAA and the choline and creatine peaks are shown on opposite sides of the water peak. Using spectral rotation to move the water peak to the left side a spectral pattern shown in Fig. 2C is obtained. A complication of this approach is increased first order phase shifts [3] due to the increased readout gradient spacing, which needs to be corrected in postprocessing. Furthermore, sensitivity per unit time decreases due to interleaving of compensation gradients.

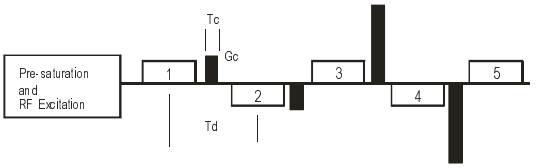
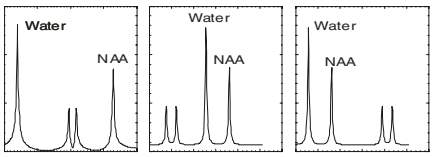


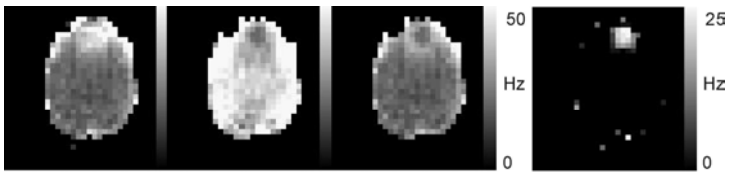
Figure 1



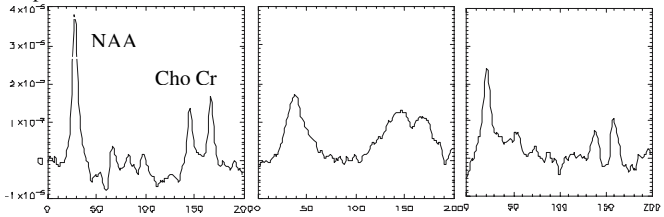
2A 2B 2C

Methods Brain scans on 5 healthy subjects were performed on a 4 Tesla Bruker MedSpec scanner equipped with a Siemens console and Sonata gradients. PEPSI data were collected with TR: 2 sec, TE: 50ms, spatial matrix: 32x32, pixel size: 10 x 10 mm^2, and slice thickness of 15 mm. Water-suppressed (8 averages) and non-water-suppressed (1 average) data were obtained, resulting in 10 minute scan time. The compensation gradients were calculated based on phase difference images obtained from a dual echo gradient echo sequence. For Td = 1.34 ms and Tc = 0.73 ms (see Fig.1), the maximum local gradient that can be compensate is 0.0426 mT/m. The maximum gradient amplitude (40 mT/m) with 256 readout gradients inversions is reached after 686 msec, which provides adequate spectral resolution at 4 T. The maximum local gradient strength that could be compensated with this approach was 0.0426 mT/m.

Results and Discussion Line width images of the water peak were calculated from the non-water suppressed data without gradient compensation (3A), with gradient compensation (3B) and a composite image was calculated by choosing the minimum line-width spectrum in each voxel of the two images (3C). The difference map in Figure 3D shows the region of improved line width. Figure 4A shows a spectrum from the uncompensated data set in a posterior brain region without susceptibility related line broadening. Fig. 4B shows a spectrum from the uncompensated image in a frontal region affected by susceptibility related line broadening. Fig.4C shows the same spectrum from the compensated data set, which displays considerably reduced spectral line width.



3A 3B 3C 3D



4A 4B 4C

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Fig 5 shows an NAA map obtained with this single-shot PEPSI method using spectral integration. The map is similar to maps obtained with conventional PEPSI scans, but more uniform in frontal cortex. In comparison to conventional PEPSI methods the SNR is reduced due to 2 mechanisms: Separation of compensated and uncompensated data reduces SNR by sqrt(2) as compared to conventional PEPSI reconstruction that averages even and odd echoes. Second, the interleaving of compensation gradients reduces sampling efficiency. We are developing methods to reduce this sampling inefficiency. The approach is most suitable for long echo times in its current implementation due to spectral overlap.

In conclusion, we have shown for the first time single shot compensation of local susceptibility related gradients in spectroscopic imaging. This methodology is particularly advantageous for 3D scanning, which does not allow collecting separately shimmed data sets due to time constraints. As with gradient compensated fMRI the methodology is limited to brain region where local gradients are relatively linear in space. In regions with spatially nonlinear local gradients it will be necessary to reduce the voxel to improve the efficiency of the compensation scheme.

References. 1) Deichman R., et al. Neuroimage 15(1) 120-135. 2) Posse S., et al. Neuroimage 18(2) 390-400. 3) Posse, S. et al. Magn Reson Med. 1995; 33:34

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