Prospective Correction of Spatially Non-Linear Phase Patterns for Diffusion-Weighted FSE Imaging Using Tailored RF Excitation Pulses

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Introduction: Echo planar imaging (EPI) is the workhorse for *in vivo* diffusion-weighted (DW) imaging; however it suffers from spatial distortions and signal loss particularly at tissue-air or bone interfaces. Single shot fast spin echo (ssFSE) does not suffer these limitations, but requires precise control of the signal phase at the start of the refocusing pulse train to avoid signal losses (CPMG condition). The application of the diffusion sensitizing gradients results in spatial phase gradients and global phase shifts caused by subject motion [1] as well as non-linear phase variations arising from brain pulsations throughout the cardiac cycle [2] , and as a result of scanner vibration [3]. Alsop developed a phase insensitive modification to the ssFSE that can provide robust performance at the cost of 50% signal reduction [4]. To stabilize performance for arbitrary phases, Le Roux proposed an elegant method of phase modulation of the refocusing pulses [5], but this approach often leads to residual signal fluctuations between echoes that can cause image artifacts. Taking a different approach, Norris demonstrated the use of real time correction of linear phase gradients, with cardiac gating used to minimize non-linear phase variations [6]. Here we explore the uses of tailored RF excitation pulses combined with additional gradient pulses to prospectively correct phase structures associated to scanner vibrations and ensure that the CPMG condition is satisfied after diffusion sensitization.

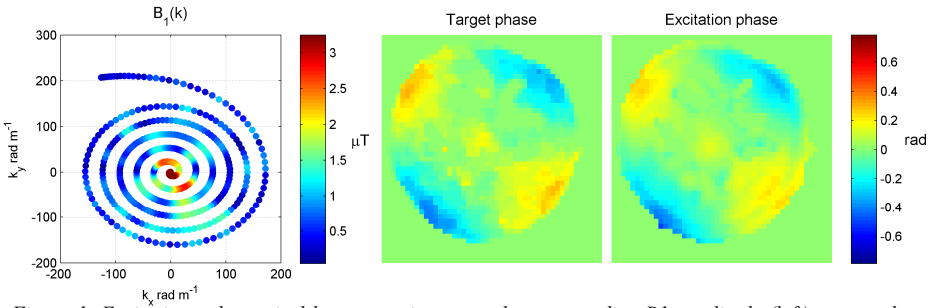

Figure 1: Excitation pulse: spiral k-space trajectory and corresponding B1 amplitude (left), target phase pattern (middle) and predicted excitation phase (right).

Theory and methods: In pilot experiments we have verified that non-linear phase patterns following diffusion sensitizing gradients are strikingly consistent across normal adult subjects when cardiac synchronization is used, which would indicate reproducible scanner vibrations [7]. We propose removing these non-linear phase patterns by designing excitation pulses that produce a compensating reversed spatial phase structure. We used a small tip angle spatial pulse design method [8] employing a spiral gradient trajectory and including B_0 correction to achieve the required non-linear phase modulation. To correct for the linear component, gradient blips were used. The excitation was followed by a standard pulse gradient spin echo with a slice selective refocusing pulse and a ssFSE pulse train with minimal inter echo spacing (2.9ms). The tailored excitation pulse method was tested on Jelly Phantoms and melons scanned using a Philips 3T Achieva scanner with an 8 channel head coil. An axial slice was imaged with: FOV 230×230 mm^2 , resolution 2×2 mm^2, slice thickness 10 mm, SENSE factor of 2 and half Fourier encoding (60%) with maximum b value of 1000 s/mm^2, TE/TR 90ms/10s. For full prospective phase correction a navigator would have been acquired between the end of the diffusion gradients and the ssFSE train as in the Norris method. In these first tests we have not yet implemented this component, and measured the phase patterns using a separate ssEPI scan instead. For this scan a flat phase excitation pulse was used and k-space fully encoded. B_0 maps were obtained using the difference of two 3D field echo acquisitions with an echo time difference of 1 ms. Pulse details were as follows: duration 4.3 msec, k-space coverage ± 200 radm^{-1} using variable velocity spiral to limit power deposition; peak B1 7.5 μT, 60° flip angle. The pulses were designed using Matlab and took 0.5sec for optimization. For comparison EPI diffusion data were also acquired (same FOV and resolution, 60% half Fourier and SENSE factor of 2, TE/TR 65ms/10s).

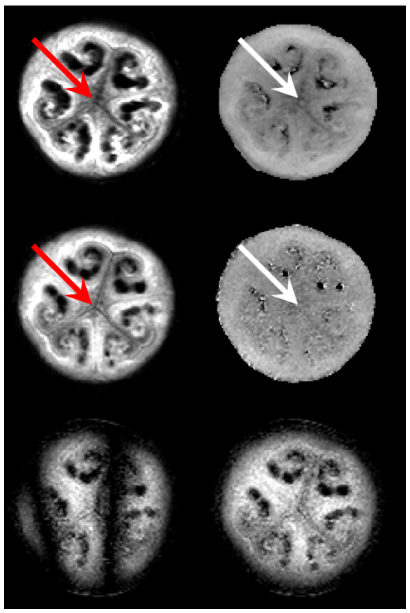

Figure 2: EPI and ssFSE images of a melon. Top row: non-DW EPI image and matching ADC map; middle row: non-DW ssFSE and corresponding ADC map; bottom row: uncorrected DW-ssFSE and prospectively corrected DW-ssFSE. The red arrows show an area of signal dropout in the EPI that is not damaged in the ssFSE image, leading to incorrect ADC estimation for EPI (white arrows). All images are displayed with independent window settings, except the ADC maps which are matched.

Results: Figure 1 shows the excitation pulse with its spiral trajectory color-coded by B1 amplitude, together with the target and predicted excitation phase structures. A flat excitation magnitude was predicted (within ± 2% of the nominal value) and thus a variable phase and constant magnitude excitation were achieved. Figure 2 shows images of a melon: non-DW EPI and corresponding apparent diffusion coefficient (ADC) map (top row), non-DW ssFSE and matching ADC map (middle), and un-corrected and prospectively corrected DW-ssFSE (bottom row). The correction has removed the regions of signal dropout, leading to undamaged DW images that are also free from susceptibility artifacts. The ssFSE acquisitions preserve signal in regions where there is dropout in the EPI images (red arrows) and so the ADC map obtained using ssFSE is flatter compared to EPI (white arrows).

Discussion and Conclusions: The concept of prospective correction of non-linear phase distributions using tailored RF excitation has been shown to be effective in phantoms, paving the way for a full *in vivo* test. Brain-pulsation effects can be avoided using cardiac gating, with the RF excitation pulse designed to cancel out the non-linear phase patterns produced due to scanner vibrations. Linear phase components including the effect of bulk patient motion could then be measured using a navigator and prospectively corrected using gradient pulses as suggested by Norris [6]. As the amplitude of the non-linear phase variations measured *in vivo* is larger compared to the phantom example shown here (± 1 rad versus ± 0.5 rad), we expect this correction to have an even more significant impact *in vivo*. Currently the excitation pulse is non-slice selective with slice selection being achieved through the refocusing pulses. Full brain coverage can still be achieved with one slice being acquired per heartbeat. Although in this implementation a long excitation pulse was required, its duration could be shortened using parallel transmit techniques [9]. The results indicate that undamaged ssFSE images can be obtained by these methods providing the benefits of FSE readout without the previous penalties of signal loss or echo fluctuation.

References: [1] Anderson AW and Gore JC, MRM 1994; 32:379; [2] Miller K and Pauly JM, MRM 2003; 50:343 [3] Gallichan D et al, HBM 2010; 31:193; [4] Alsop DC, MRM 1997; 38:527; [5] Le Roux P, JMR 2002; 155:278; [6] Norris DG and Driesel W, MRM 2001; 45:729; [7] Hiltunen J et al, NeuroImag 2006; 32:93; [8] Yip C et al, MRM 2005; 54:908; [9] Zhu Y, MRM 2004; 51:775.