In Vivo Correction of 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 sequence most commonly used to acquire diffusion-weighted (DW) images. Unfortunately it is very sensitive to field inhomogeneities and the images suffer from spatial distortions and signal loss particularly at tissue-air interfaces. Single shot fast spin echo (ssFSE) does not present these limitations, but in order to retain the full signal throughout the whole series of echoes, precise control of the signal phase at the start of the refocusing pulse train is required (CPMG condition). This is particularly challenging to achieve when DW is applied. Rigid-body motion during the application of the diffusion gradients can lead to spatial phase gradients [1], while pulsatile brain motion [2] and scanner vibrations [3] produce non-linear phase patterns. One option is to modify the sequence so that the signal becomes phase insensitive. This can be achieved using the method introduced by Alsop, but at a cost of 50% signal reduction [4]. An alternative is to modulate the phase of the refocusing pulses as suggested by Le Roux [5], yet residual signal fluctuations can result in image artifacts. A different strategy altogether consists in correcting for the motion-induced phase patterns ensuring that the CPMG condition is satisfied after diffusion sensitization. This work demonstrates that it is possible to achieve such a correction *in vivo*.

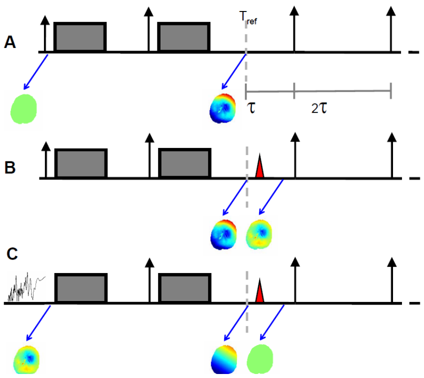

Figure 1- A) DW-ssFSE sequence showing the phase patterns produced at the first spin echo (Tref) as a result of motion; B) Gradient blips can be used to correct for rigid-body motion effects measured in real time; C) Full correction is possible by using a tailored RF pulse which pre-compensates the non-linear phase component.

Theory and Methods: To obtain diffusion weighting, a SE diffusion module is placed before the ssFSE readout. The time delay (τ) introduced between the SE obtained from the diffusion module (at time T_{ref}) and the first pulse of the FSE readout sets the echo train spacing (ETS). To fulfill the CPMG condition, $\phi(r,T_{ref})$, the phase over the whole object at the time of the first spin echo T_{ref}, should be flat (offset of $\pi/2$ relative to the phase of refocusing pulses). Unfortunately this is not true in the presence of motion (Fig 1A). Using a real time measurement of the phase at T_{ref} and introducing a gradient blip before the next RF pulse allows correction for linear phase gradients produced by rigid body motion [6], but even if cardiac triggering is performed, there is still a residual non-linear phase pattern (Fig. 1B). Recently, the idea of using a tailored RF excitation pulse to correct for these residual non-linear patterns was shown to be effective to image phantoms [7]. To measure $\phi(r,T_{ref})$, a 1D navigator was used with phase encoding performed over multiple shots. As this approach cannot be used *in vivo*, in this work a 2D navigator was introduced between the end of the diffusion gradients and the next refocusing pulse. Ideally, the navigator center of k-space should be read at T_{ref}. A spiral in trajectory would be suitable, but non-uniform data requires processing that is challenging to achieve in real time, making an EPI trajectory more appealing. To avoid prolonging the ETS, the center of the navigator preceded T_{ref} by a time interval ΔT. This meant that the phase measured by the navigator, $\phi^m=\phi(r,T_{ref}-\Delta T)$, differed from $\phi(r,T_{ref})$ due the phase evolution which occurs during that period. To calibrate this phase, two sequence variants were implemented: the one just described (i) and another one with the navigator centered on T_{ref} (ii). The phase evolution due to B0 during ΔT is given by $\phi^{B0} = \phi^{no\text{-}DW}(r,T_{ref}) - \phi^{no\text{-}DW}(r,T_{ref}-\Delta T)$. Application of DW causes a phase shift due to eddy currents, which was pre-calibrated by scanning an oil phantom using the same scan geometry: $\phi^{eddies} = [\phi^{DW}(r,T_{ref})-\phi^{DW}(r,T_{ref}-\Delta T)]-[\phi^{no\text{-}DW}(r,T_{ref})-\phi^{no\text{-}DW}(r,T_{ref}-\Delta T)]$. The non-linear phase contribution due to pulsatile motion and vibrations ($\phi^{Non\text{-}Lin}$) was estimated using sequence (ii) with DW and subtracting off any constant and linear phase terms. The total phase to be corrected was therefore: $\phi^{Tot}(r,T_{ref})=\phi^m+\phi^{B0}+\phi^{Eddies}+\phi^{Non\text{-}lin}$. To compensate for non-linear phase patterns, a small tip angle spatial pulse design method was used [8] employing a 2D spiral gradient trajectory and including B_0 correction. The resulting excitation pulse (Fig 1C) is not slice selective; this is achieved by the refocusing pulses. To correct for the linear component, gradient blips were applied. A healthy volunteer was scanned having been instructed to remain as still as possible. Imaging was performed on a Philips 3T Achieva magnet using a 6 channel head coil. An axial slice was imaged with: FOV 230×230 mm^2, resolution 3×3 mm^2, slice thickness 4 mm, TE 60 ms, half Fourier encoding (60%), low-high phase encode ordering, maximum b value of 500 s/mm^2, ETS 2.7 ms. Cardiac triggering was used to ensure the diffusion module was applied during diastole resulting in a TR of approximately 6 s. B_0 maps were obtained using the difference of two 2D field echo acquisitions with a ΔTE of 2.3 ms. The RF excitation pulse details were: duration 4.3 msec, k-space coverage ~200 radm^{-1} using variable velocity spiral to limit power deposition; peak B1 5.4 µT, 60^{O} flip angle. The pulses were designed using Matlab. For the calibration measurements, a flat phase excitation pulse was used.

Results: Figure 2 shows images obtained using the flat phase spiral for low and high b-values (A and B), the corrected DW image (C) and the corresponding ADC map (D). The images E and F show the non-linear phase patterns measured before and after correction. At this stage no real time linear phase correction was in place, but the motion induced phase could be confirmed and using multiple repeats it was found that when the measured ϕ^m was zero, signal loss was mitigated.

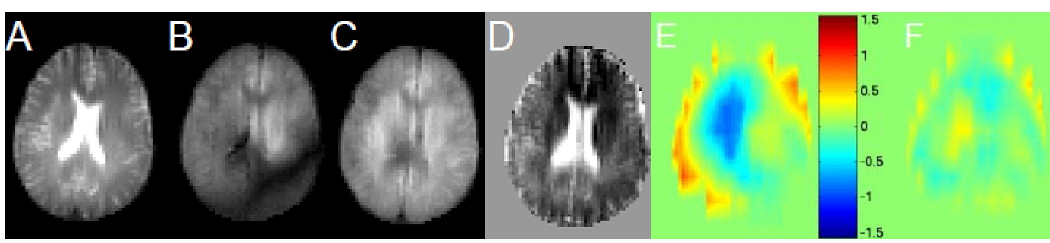

Figure 2- A) non-DW image; B) uncorrected DW image; C) corrected DW image; D) corrected ADC map. Original (E) and corrected (F) non-linear phase components (radians).

Discussion and Conclusions: It was demonstrated that a tailored RF excitation pulse can successfully compensate for non-linear phase patterns *in vivo* and that these patterns are sufficiently repeatable to allow correction based on a prior measurement. Calibrations scans were performed to account for other linear phase errors not related to motion (eddy currents and B0 field effects). This acquisition was performed on a cooperative volunteer and therefore rigid body motion could be minimized. The final step in this project will be to implement true real time correction of the motion related phase ramps as suggested by Norris [6]. Other developments will include further exploration of spiral navigators with real time processing and use of parallel transmit techniques [9] to shorten the duration of the excitation RF pulse.

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] Nunes RG et al, ISMRM 2011, 172; [8] Yip C et al, MRM 2005; 54:908; [9] Zhu Y, MRM 2004; 51:775.