Robustness of Phase Sensitive Reconstruction in Diffusion Spectrum Imaging

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Target Audience: This abstract is targeted to those interested in diffusion spectrum imaging and diffusional kurtosis.

Introduction:
The Cartesian acquisition of q-space, termed diffusion spectrum imaging (DSI)[1], is one of the most comprehensive acquisition schemes in MR diffusion weighted (DWI) imaging, allowing for resolving crossing fiber tracts in the brain or characterizing diffusion with advanced metrics like diffusional kurtosis[2]. As diffusive signal decay is a magnitude effect, common data processing in DSI is based on the magnitude, discarding any phase information. The observed non-zero net phase however has a variety of contributing sources (B0 inhomogeneity, eddy currents, motion, etc.) which are difficult to disentangle. Separating these phase contributions can be advantageous for several reasons, mainly because the phase contains information on coherent motion (i.e. brain pulsatility[3]); and because processing of DWI to obtain parametric quantities like DTI and Kurtosis benefits from taking real valued data, as magnitude processing introduces Rician bias[4]. This work examines the robustness of phase sensitive reconstruction applied to full DSI data in phantoms and in vivo human brain, but may also be applied to further other diffusion acquisition schemes in particular compressed-sensing-accelerated DSI[5].

Purpose: To test the robustness of phase sensitive reconstruction in diffusion spectrum imaging.

Methods: A non-zero phase φ in DWI can be attributed to different components as follows, with φ_0 – RF pulse phase, φ_x static magnetic field inhomogenieties and eddy currents of imaging gradients, φ_{eddy} – eddy currents induced by diffusion-(q-space) gradients, and φ_{B0} phase induced by motion:

$$\varphi_{total} = \varphi_0 + \varphi_X + \varphi_{eddy} + \varphi_{motion} \qquad (1)$$

Through a series of consecutive DWI acquisition, the first two terms can safely be regarded as static, as imaging gradients repeat, rf phase profiles stay constant and a potential drift of B_0 can be accounted for (i.e. through interleaved b=0 acquisitions without diffusion encoding gradients). Throughout such a set of DWI experiments, the two other major contributions to phase, namely by eddy currents induced by the strong diffusion gradients (which vary in strength and duration with every acquisition) and by motion (bulk motion due to subject movement but also pulsatile brain motion) vary from acquisition to acquisition, however.

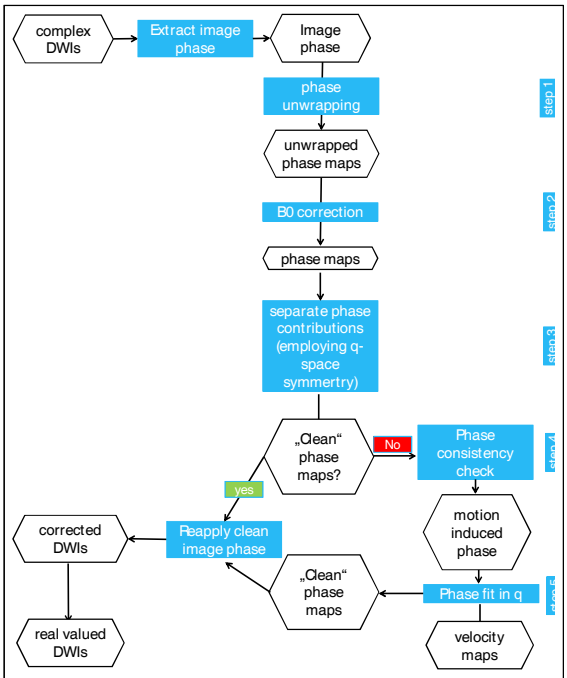

Fig 1: Phase sensitive reconstruction workflow

Echo-planar diffusion weighted imaging experiments were performed on phantoms and healthy volunteer (peripherally gated, trigger delay = 14/50/100/150/200 ms) using a 3T GE MR750W MR scanner (GE Healthcare, Waukesha, WI, USA) (TE=80.7ms, TR=2s, 96x96, FOV=24 cm, slice=2.5 mm, ASSET factor 2, bmax=2,000 s/mm²) using fully (no of samples = 531 and only cardinal-axes sampled (no of samples =94) acquisition schemes with interleaved b=0 acquisitions.

Results and Discussion: While the spatial patterns of eddy currents induced by so-called donor-gradients (a diffusion encoding gradient pair can be regarded a donor gradient) and their influence on the image magnitude is well understood [6], their phase-contribution has had limited attention so far. Exploiting symmetry relations of the phase variation in q-space, we present a robust scheme to separate the different contributions, yielding a significantly reduced phase, which allows for real-value based processing of the data (DTI, Kurtosis). In Fig. 1 the phase-sensitive reconstruction workflow is presented. Step 3, the q-space symmetry step takes advantage of the pointsymmetry of a DSI acquisition and the scaling properties of donor gradient induced eddycurrents/phase. Consequently, through adding/substracting of corresponding DWIs, both point- and axisymmetric phase contributions can be isolated from each other. This approach allows to then selectively correct the original DWI phase by these terms, leaving only non-scalable phase contributions, which can be attributed to pulsatile brain motion. The consistency of these brain pulsatility motion phase contributions were found to highly depend on the trigger reproducibility and quality. Fig. 2 depicts exemplary phase images before and after the series of corrections (steps 2-4) and corresponding real/magnitude images. The total observed image phase could be reduced by a factor of ~10, resulting in minimal residual phase contributions, which demonstrated to not significantly affect fitting of derived metrics like mean and orthogonal kurtosis (Fig. 3).

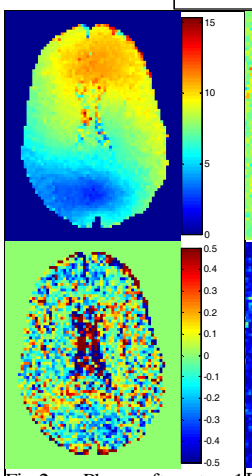

Fig.2a : Phase after step 1 (upper) and after step 4 (lower)

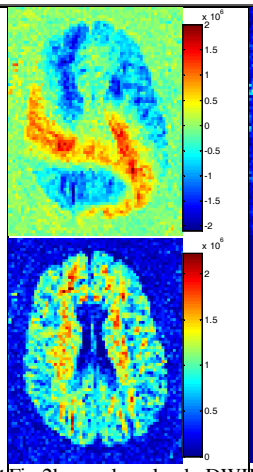

Fig.2b: real valued DWI before (upper) and after step 4 (lower)

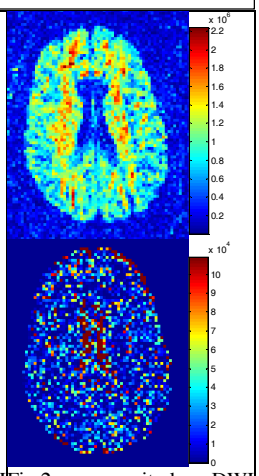

Fig.2c: magnitude DWI (upper) and difference Mag-Real (lower). Note the different colorbar scalings

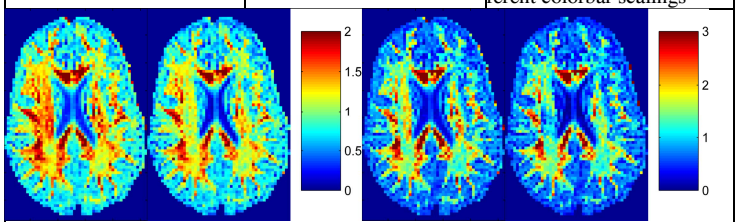

Fig. 3: Kurtosis Fit comparison magnitude (left) vs. real-valued (right) data processing for mean kurtosis (left) and orthogonal kurtosis (right)

Conclusion: Phase sensitive reconstruction was successfully applied to PG triggered DSI data, demonstrating a substantial reduction of phase, except for ventricular areas, which exhibit a totally different motion regime. Using the resulting real-valued DWI enables bias-free fitting of derived diffusion metrics, like DTI, FA and Kurtosis.

References:

[1] Wedeen et al., MRM 2005; 54(6), [2] Jensen et al., MRM 2005; 53(6), [3] Soellinger et al., MRM 2009; 61(1), [4] Cárdenas-Blanco et al., Concepts Magn. Reson. Part A 2009, [5] Menzel et al., MRM 2011; 66(5), [6] Xu D, et al MRMed. 2013 70(5)