Joint estimation of microstructural and biomechanical features of the brain using a phase sensitive reconstruction of DWIs

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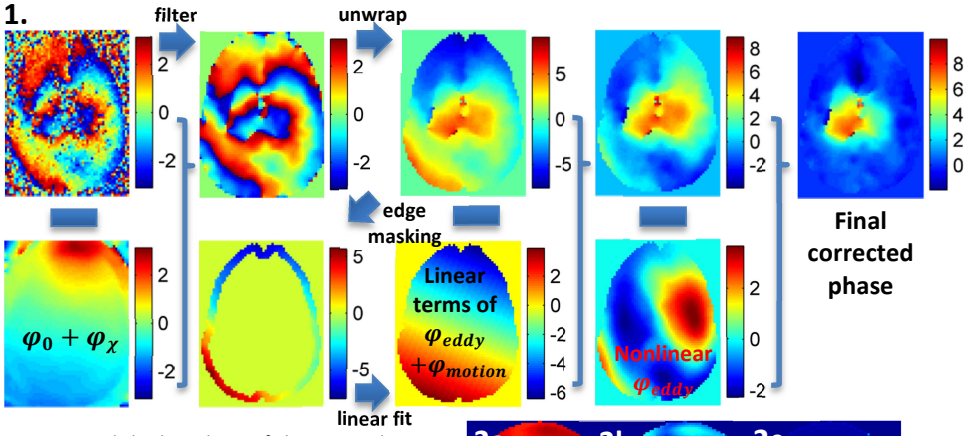
Introduction: Diffusion weighted magnetic resonance imaging (DWI) allows for non-invasively measuring microstructural features of the human brain [1]. DWI is used to investigate a wide range of brain disorders, e.g. neurodegenerative disorders like multiples sclerosis and Alzheimer or Traumatic Brain Injury (TBI). Usually all data processing in DWI is based on the magnitude of the complex MR signal, and the inherent phase of the signal is discarded as it is considered to be spoiled by different sources. However one of these sources for a non-zero phase in DWI signals is the pulsation of the brain itself [3], which is encoded using the diffusion sensitizing gradients. The technique of velocity encoding via the signal phase is well-known in the field of MRI [4], however it is usually used for the measurement of blood flow featuring a much higher velocity than brain pulsation. Changes in the biomechanical properties of the brain however are of potential importance for disorders such as hydrocephalus (HC), brain tumor or traumatic brain injury [5]. We developed a method to process the complex DWI signals in order to extract a meaningful phase, which allows for inferring information of the pulsation of the brain. This novel approach enables joint examination of microstructural as well as biomechanical features of the brain without increasing scan time as compared to regular DWI.

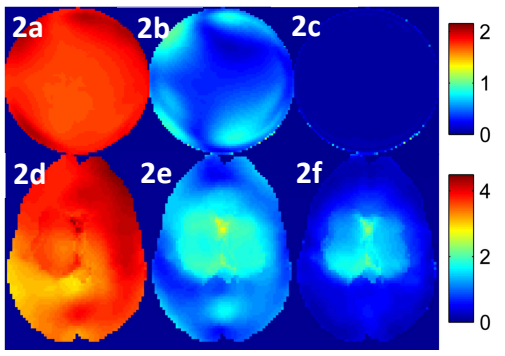
Theory: Phase of the complex signal of a DWI can be described by equation (1)

$$\varphi_{total} = \varphi_0 + \varphi_\chi + \varphi_{eddy} + \varphi_{motion} \quad (1)$$

where φ_0 is the offset of the RF-pulse, φ_χ is the susceptibility induced phase, φ_{eddy} is the eddy current induced phase and φ_{motion} the motion induced phase. Note that the phase term due to the concomitant electromagnetic field cancels out for a Stejskal-Tanner diffusion preparation. The terms φ_{eddy} and φ_{motion} consist of linear and nonlinear components where the linear component of the φ_{motion} is caused by bulk motion and the nonlinear by the brain pulsation. If the DWI acquisition is not cardiac gated, the nonlinear φ_{motion} can be considered as pseudo random because there is no permanent coherence between the sequence timing and the heartbeat of the scanned person. Hence the average phase over a sufficiently large number of acquisitions (i.e. over the whole DWI exam) cancels out while the mean of it absolute value is directly proportional to the motion.

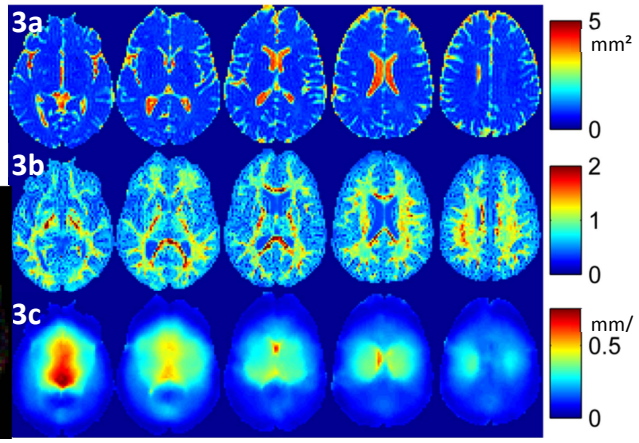


Methods: Echo-planar DWI experiments were performed on phantoms and a healthy volunteers (N=2) using a 3T GE MR750 MR scanner (GE Healthcare, Waukesha, WI, USA), equipped with a 32 channel head coil (TE=80.7ms, TR=2s, 96x96, FOV=4 cm, slice=2.5 mm, ASSET factor 2). Q-space acquisition scheme comprises 3 shells with 25, 40, 75 directions and b-values of b=750, 1070, 3000 s/mm². Fig. 1 schematically describes the filtering algorithm to remove all phase contributions from the complex MR signal except for the motion induced phase of the brain. In a first step, φ_0 and φ_χ are removed using the spatially filtered phase of b_0-images and the remaining maps are unwrapped. In a second step the linear terms of φ_{eddy} and φ_{motion} are removed by a linear fit based on the phase at the surface (i.e. along a 3 pixels thick ribbon) of the brain only where no brain pulsation is expected. In a last step nonlinear $\varphi_{eddy,NL}$ is removed using the assumption that the remaining φ_{motion} cancels out in average. A quadratic fit in q-space (see eq. (2)) is applied whereas each voxel had been filtered using a 5x5x5 Gaussian kernel (FWHM = 2 voxel) in image space to impose smoothness. $C_{1..9}$ are the coefficients of the polynom and q_x, q_y, q_z are the amplitudes of the 3 gradient axes normalized to the maximum gradient strength used during the acquisition. The mean velocity maps were calculated as the pixel wise mean over the absolute value of the remaining phase of all DWIs where each DWI was divided by the first gradient moment respectively. A correction factor of 2 was used to account for the random orientation of the velocity encoding gradient with respect to the motion. Mean diffusivity and mean kurtosis was calculated using an in-house weighted linear least squares (WLLS) algorithm.

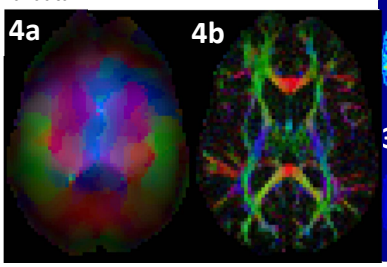


$$\varphi_{eddy,NL} = C_1 q_x + C_2 q_y + C_3 q_z + C_4 q_x q_y + C_5 q_x q_z + C_6 q_y q_z + C_7 q_x^2 + C_8 q_y^2 + C_9 q_z^2 \quad (2)$$

Results: Fig. 2 shows the standard deviation (SD) over all DWIs for different steps of the filtering process. The SD is considered to be a quality measure for the filtering process because bulk eddy currents and bulk motion change from one DWI to another. Fig. 2a,d show the SD of the unwrapped data. The phantom and the brain have a strong phase fluctuation and no anatomical details are visible. Fig. 2b,e show the SD after the linear filtering step. The phase of the phantom is already significantly cleaner and in the brain, the ventricles are visible. Fig. 2c,f show the SD after the nonlinear correction. The phantom exhibits almost no phase fluctuation meaning that φ_{eddy} could be fully corrected. The brain motion is concentrated around the ventricles, which is physiologically plausible. Fig. 3 demonstrates the feasibility of extracting both microstructural and biomechanical features from one dataset, showing mean diffusivity (a), mean diffusional kurtosis (b) as well as the phase-based-metric mean |velocity| (c). Fig. 4a shows directional maps of the brain motion and Fig. 4b features the fractional anisotropy of the diffusion in the brain.



Discussion: A new phase filtering technique has been introduced which may allow for the quantification of the brain pulsation using DWIs. It can be applied to a wide range of q-space acquisition schemes with no need to acquire additional data.



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