Motion-Induced Phase Error Correction in 3D Diffusion-Weighted Imaging

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INTRODUCTION – Owing to its capability to achieve sufficient signal-to-noise ratio (SNR) at high isotropic resolution, 3D diffusion-weighted imaging (DWI) has recently become a focus in diffusion acquisition studies [1-4]. Since 3D DWI is usually acquired with multishot techniques, one of its most challenging problems is the motion-induced phase errors, which result from object motion during diffusion encoding periods. A few studies have attempted to resolve this phase error problem; however limited success has been achieved [1-5]. In the present study, we propose a new motion-induced phase error estimation and correction algorithm in which parameters of the phase errors are estimated by nonlinear fitting of navigator images to a motion-induced phase error model and then used to correct the *k*-space data. The proposed estimator is unbiased and gives mean square error that approaches the Cramer-Rao lower bound. Regardless of the *k*-space trajectory used, the correction method in the present study is robust and time-efficient. Both simulation and *in vivo* results were obtained to demonstrate the performance of the proposed method.

METHOD

<u>*Phase Error Parameterization and Estimation*</u>: Under the assumption of rigid body motion, it has been proved that the non-diffusion-weighted image ($I_0(\mathbf{r})$) and the phase-corrupted diffusion-weighted image ($I_b(\mathbf{r})$) can be modeled as [6]:

$$I_0(\mathbf{r}) = \left|I_0(\mathbf{r})\right| e^{j\varphi_0(\mathbf{r})}; \qquad I_b(\mathbf{r}) = \left|I_b(\mathbf{r})\right| e^{j\left[\varphi_0(\mathbf{r}) + (\mathbf{a}\bullet\mathbf{r} + a_0)\right]}; \qquad (1)$$

where $\varphi_0(\mathbf{r})$ is common to both non-diffusion-weighted and diffusion-weighted images and is the combination of receiver coils' phases and magnetic-susceptibility-induced phases, **a** and a_0 are the slope and the offset of the linear motion-induced phase error, respectively. If the imaging noise is Gaussian, with reasonable SNR, the maximum likelihood estimation of {**a**, a_0} is obtained by minimizing the following cost function:

$$R(\mathbf{a}, a_0) = \sum_{\mathbf{r}} \left| I_b(\mathbf{r}) - \left|I_b(\mathbf{r})\right| e^{j\left[\angle I_0(\mathbf{r}) + (\mathbf{a}\bullet\mathbf{r} + a_0)\right]} \right|^2; \qquad (2)$$

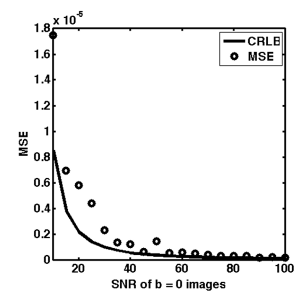

Fig. 1: Cramer-Rao lower bound.

where $\angle\bullet$ is the phase operator. With non-diffusion-weighted and diffusion-weighted navigator images for each shot of the acquisition, shot-dependent phase error parameters can be estimated using Equation (2). However, minimizing Equation (2) is a nonlinear least squares optimization problem, which does not generally have a closed-form solution. Therefore, the minimization has to be solved iteratively with a descent-based algorithm. To make matters worse, the cost function $R(\mathbf{a}, a_0)$ is also nonconvex with multiple local minima that might prevent descent-based algorithms from converging to the global minimum unless a proper initialization is provided. From the Fourier transform relationship between a linear phase in image space and a frequency shift and constant phase offset in *k*-space and with the notice that *k*-space data reach the maximum magnitude value at the *k*-space origin, we propose the following initialization:

$$\mathbf{k}_0^* = \arg\max_{\mathbf{k}} \left|S_0(\mathbf{k})\right|; \quad \mathbf{k}_b^* = \arg\max_{\mathbf{k}} \left|S_b(\mathbf{k})\right|$$
$$\mathbf{a}^{init} = 2\pi\left(\mathbf{k}_b^* - \mathbf{k}_0^*\right); \quad a_0^{init} = \angle S_b(\mathbf{k}_b^*) - \angle S_0(\mathbf{k}_0^*); \qquad (3)$$

where $S_0(\mathbf{k})$ and $S_b(\mathbf{k})$ are navigator non-diffusion-weighted and diffusion-weighted *k*-space data, respectively.

<u>*Phase Error Correction:*</u> For trajectory-independent robustness, the motion-induced phase errors are corrected in *k*-space with the previously estimated parameters as follows:

$$\mathbf{k}_{cor} = \mathbf{k} - \frac{\mathbf{a}^*}{2\pi}; \qquad S_{cor}(\mathbf{k}_{cor}) = S(\mathbf{k}_{cor}) e^{-ja_0^*}; \qquad (4)$$

where $\mathbf{k}$ is the designed *k*-space trajectory, $\mathbf{k}_{cor}$ is the actual trajectory under the effect of the motion-induced phase error, $S(.)$ is the uncorrected *k*-space, $S_{cor}(.)$ is the corrected *k*-space data, and $\mathbf{a}^*$ and a_0^* are the optimal slope and offset minimizing Equation (2), respectively.

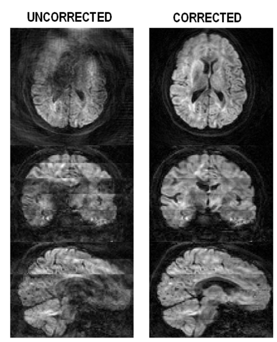

Fig. 2: Diffusion-weighted images.

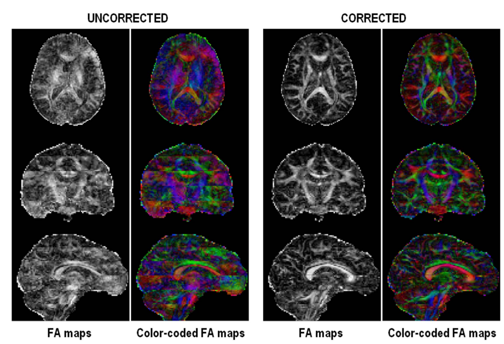

Fig. 3: FA and color-coded FA maps. Red: left-right, green: anterior-posterior, blue: inferior-superior.

<u>*Data Acquisition:*</u> A previously proposed 3D multislab stack of spirals with 3D navigator acquisition strategy was used [4]. *In vivo* data were acquired using Siemens 3 T Trio scanner with a 12-channel head coil on a healthy subject in accordance with the institutional review board. The obtained resolution was 1.88 x 1.88 x 2 mm^3 with 24 x 24x15cm^3 field-of-view. Other imaging parameters were: TE1 = 64 ms (for data), TE2 = 105 ms (for navigator), navigator matrix size = 15 x 15 x 10, and b = 1000 s/mm^2. The acquisition was cardiac-gated for enhancement of the rigid body motion assumption resulting in an effective TR of two R-R intervals (approximately 2 s).

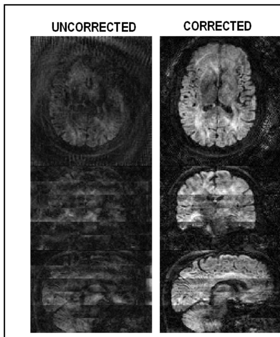

Fig. 4: Non-cardiac-gated, diffusion encoding direction [1 0 1].

RESULTS – <u>*Simulation:*</u> Fig. 1 shows the comparison between the empirical mean square errors (MSE) and the theoretical Cramer-Rao bound (CRLB, a lower bound on the variance of any unbiased estimator [7]) for the estimation of the first element of the slope (**a**(1)). Different SNR in the simulation were achieved by adding complex Gaussian noise with a range of variances to the simulated image. From Figure 1, the proposed estimation algorithm gives MSE approaching the CRLB with increased SNR, which implies the efficient property of the algorithm.

<u>*In vivo*</u>: Fig. 2 shows the obtained diffusion-weighted image with and without motion-induced phase error correction. Artifacts were removed and signal loss was restored in the corrected images. Without phase error correction, both the fractional anisotropy (FA) maps and the color-coded FA maps are heavily corrupted as shown in the left panel of Fig. 3. Removing the motion-induced phase errors produces FA maps and color-coded FA maps with well-resolved fiber structure and direction as shown in the right panel of Fig. 3. When non-cardiac-gated acquisition is used, the proposed algorithm performs the first order correction of the nonlinear phase error, yielding reasonable images with minor signal loss due to pulsation (Fig. 4).

CONCLUSION – We proposed and successfully tested a robust and time efficient algorithm for 3D motion-induced phase error correction in *in vivo* cardiac-gated diffusion-weighted images. The presented algorithm makes feasible *in vivo* 3D DWI with high isotropic resolution and sufficient SNR.

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