Eddy Current Compensation for Double Wave Vector Diffusion MRI

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Target audience: Researchers interested in double wave vector diffusion MRI and eddy current compensation.

Purpose: Double wave vector (DWV) diffusion imaging is used to obtain information on microscopical tissue anisotropy, independent of the macroscopic configuration. It has recently been shown to be applicable *in-vivo* on a whole body MRI [1]. Such scanners have weaker gradients than animal scanners and therefore need longer gradient durations, which leads to increased eddy currents and image distortions. To get a rotationally invariant measure of microscopic anisotropy, images obtained with many different gradient orientations must be combined, which can lead to errors if eddy current induced image distortions are present. We propose to adapt the diffusion-weighting gradients of DWV-MRI such that they are intrinsically eddy current compensated by adopting the technique of the twice refocused spin echo diffusion weighted sequence [2].

Material and Methods: An in-house developed DWV echo planar sequence was used. Each of the two diffusion weightings could either be run with monopolar gradients (not eddy current compensated) or with twice refocused gradients (eddy current compensated). For the compensation, an exponential decay of the eddy currents of 70 ms was assumed. Measurements of a phantom consisting of plastic rods arranged in a quadratic grid in water were performed on a 3T system (Biograph, Siemens Healthcare, Erlangen, Germany) with the body coil.

In a first series of measurements, all 36 combinations of first and second gradient direction lying along either plus or minus read, phase, or slice direction were probed. Further parameters were: field of view 400x400 mm², nominal in plane resolution 4x4 mm², 10 slices of 5 mm thickness, 25 ms between the diffusion weightings. The duration of the uncompensated diffusion weighting was 42 ms (gradient duration of 18.99 ms), the compensated one was 46 ms long (gradient durations: 3.81, 15.17, 14.81, 4.17 ms). The four variations of no compensation, first or second weighting compensated, both compensated, were measured with echo times of 134, 140, 150 and 136 ms, respectively. The gradient amplitude was 33 mT/m without compensation and 36 mT/m with compensation leading to the same b-value of 450 s/mm².

A second series of measurements was performed using the isotropic direction combination of Jespersen *et al*. [3] with 72 direction combinations to determine the microscopic anisotropy in the pelvis of healthy volunteers. The nominal in plane resolution was 4.5x4.5 mm² with a slice thickness of 5 mm and a field of view 450x243 mm². Each diffusion weighting took 39 ms in the uncompensated and 48 ms in the compensated case. The time between the diffusion weightings was set to zero for the compensated and 15 ms for the uncompensated sequence, the echo time was 108 ms. The b-value was 450 s/mm² for each diffusion weighting, resulting in q-values of 170 mm^{-1}. Additionally, images without diffusion weighting were acquired to determine the fractional eccentricity (FE), a compartment size independent measure of microscopic anisotropy.

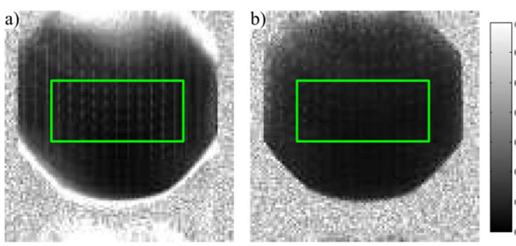

Fig. 1: Coefficient of variation over 36 diffusion direction combinations. There is a large area around the edges of the phantom where the signal variation is huge in the uncompensated images (a). These disappear mostly when using the eddy current compensated sequence (b).

Results: In Fig. 1, maps of the coefficient of variation of the signal intensities in the phantom for the uncompensated and the fully compensated weightings are shown. The edges of the phantom show a high signal variation in the uncompensated images, which can be seen as severe distortions in the individual images. The positions of the plastic rods vary over several pixels. These artifacts are reduced when using the eddy current compensation (Fig. 1b). In Fig. 2, the sums of the coefficients of variations of all pixels inside the green rectangular in Fig. 1 are listed for all slices for the uncompensated, only one weighting compensated and the fully compensated scheme. Compensating the second weighting is more effective than compensating the first one, which only shows little effect. The lowest signal variation is seen when compensating both weightings. The *in-vivo* results show high anisotropy in the bladder and around small structures for the uncompensated measurements (Fig. 3). The use of the eddy current compensation reduces these values to zero.

Discussion: The use of a bipolar eddy current-compensated gradient scheme for both diffusion blocks in a double wave vector experiment reduces the misregistration of structures as was demonstrated in a phantom as well as *in-vivo*. One effect of eddy currents is a change in actual pixel size and therefore signal intensity, which is assumed to have led to a high anisotropy value in the bladder of our volunteers. When performing measurements of small structures *in-vivo*, eddy current compensation, or a sophisticated postprocessing correction scheme [4], seems to be mandatory.

Conclusion: Eddy currents cause a problem in double diffusion weighted imaging on human whole-body scanners. We propose to compensate both diffusion weightings with respect to eddy currents.

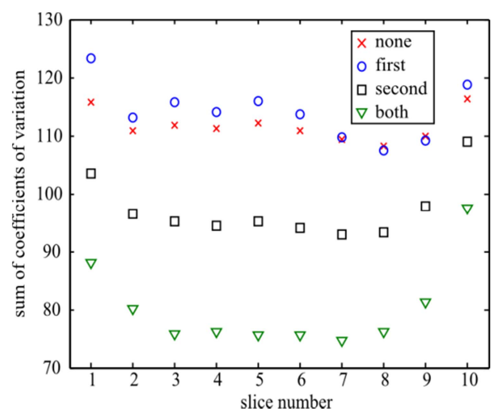

Fig. 2: Sums of the coefficients of variation over the region of interest (ROI) indicated in Fig. 1 for the different compensation schemes and slices. The variation reduces tremendously when compensating both weightings. The compensation of only one weighting proves to be more efficient if it is closer to the readout train. The size of the ROI was 1566 pixel.

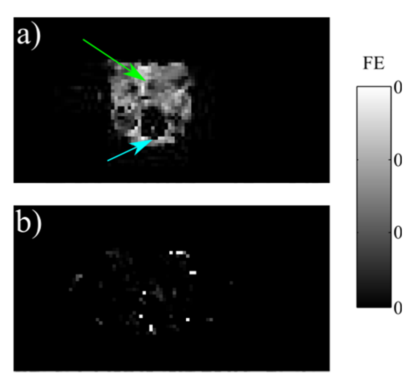

Fig. 3: Fractional eccentricity (FE) in the bladder. The images without eddy current compensation lead to a strong overestimation in the bladder (green arrow) and at small structures (blue arrow) (a). The values decrease when using the compensation (b).

References:
1. Lawrenz, Finsterbusch MRM 2013; 69:1072-1082, 2. Reese *et al*. MRM 2003; 49:177-182, 3. Jespersen *et al*. NIB 2013; 26:1647-1662
4. Bodammer *et al*. MRM 2004; 51:188-193