Prospective Motion Correction for Diffusion Imaging Using FID Navigators

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Introduction Diffusion imaging is extremely susceptible to macroscopic motion. This is particularly critical in (i) long acquisitions schemes, as they appear in diffusion tensor, spectrum and q-ball imaging, and (ii) clinical settings with young or uncooperative patients. The application of registration-based correction methods is problematic; the changing contrast between image volumes due to the varying diffusion encoding directions and weightings introduces a high uncertainty, in particular for diffusion weightings of b>500 s/mm^2. Moreover, corrections a posteriori often do not consider the corresponding changes in the b-matrix, which may have a significant impact on the results [1]. Free induction decay (FID) navigators have been shown to provide information about motion with no or negligible time penalty [2-7]. In the present work, we combine this navigator technique and a well-established registration method with the aim to prospectively or retrospectively correct motion in diffusion images of the head.

$$rd(n) = 100 \bullet \left\langle \max_5 \left| \forall s \in UsedSlices : median\left[\mathrm{Re}\left(\frac{nav_n(s,c) - nav_{n-1}(s,c)}{nav_{n-1}(s,c)} \right) \right] \right| \right\rangle - rd(2) \qquad [eq1]$$

Material and Methods A FID readout, sampling 168 points in ~1ms after the slice rewinder (similar to [3]), was added to a conventional SE diffusion EPI sequence. The increase of the lower echo time limit (i.e. the time penalty due to the insertion of the FID readout) amounted to less than 1ms. The navigator data was volume-wise evaluated in real-time using [eq1], *rd(n)* being the relative difference (%) to the navigator of the volume before, $nav_n(s,c)$ the complex FID value of coil element *c* acquired in slice *s* of volume #*n* and *UsedSlices* a subset of slices where the navigator exceeded a minimum energy threshold to remove empty slices that would only contribute noise. If *rd(n)* exceeded an empirical threshold of 1%, a volume without diffusion encoding (termed "extra-b0" scan here) was inserted into the acquisition. Subsequently, the scan continued by repeating the motion-corrupted volume. The extra-b0 volume was registered in real time using the pre-existing scanner framework described in [8]. The motion parameters determined online can be stored (for later correction, *retrospective mode*) or directly be used to update the gradient coordinate system online (*prospective mode*).

A clinical diffusion scan protocol (TE=102 ms, TR=4800 ms, TD=500 ms, matrix size 84x84, FOV 212x212 mm^2, 32x3mm slices, bandwidth 1384Hz/pixel, 12 directions, bipolar gradient scheme) with 5 averages (5*13 volumes yielding TA=5:28 min) was set up. Eight healthy, young volunteers (4 male, 4 fem.), providing informed written consent, were scanned with a 32-channel head coil on a 3T MR system (Magnetom TIM Trio, Siemens Healthcare Sector, Germany). They were instructed to perform varying, realistic, small free head movements upon an oral command from the investigator. Each experiment comprised a *rest* (no movement command) and a *motion* acquisition, where 8-10 movement commands were issued during the scan and their times recorded. Three sets were acquired in retrospective mode with b-values of 500, 1000 and 3000 s/mm^2. In addition, a seventh scan (b=1000) was conducted in prospective mode. To investigate the stability of the navigator signal, three phantom scans with the above protocol were performed (10 instead of 5 averages, resulting TA=10:40 min). Standard deviation (SD) maps and mean images over the 5 averages were calculated of each direction from all time series. Sensitivity and specificity were determined using the documented timings of the motion commands.

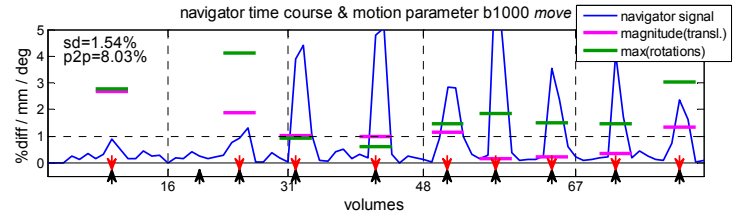

Fig. 1 Exemplary navigator time course of a retrospective motion scan (b1000). Arrows indicate when a motion command was given (black) and when the threshold was exceeded and thus an extra-b0 volume acquired (red). Note the undetected movement at vol. #20. To give an impression of the amplitude of the movements, the magnitude translation (magenta, [mm]) and the maximal rotation (green, [deg]), as revealed by the registration, is also shown.

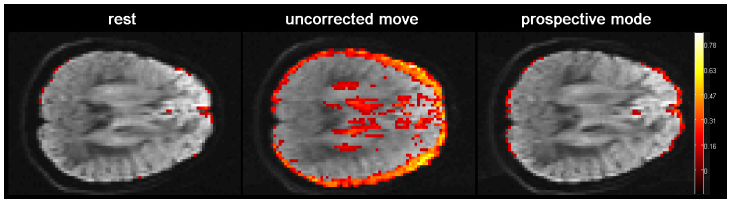

Fig 2 Mean over 5 averages of the same direction in three b1000 experiments (same subject) with comparable movements (amplitudes and number). Overlaid colour maps show the SD over the averages (normalised by the same factor). As it can be seen, the diffusion information is well preserved in the PACE, whereas high SD values in the move experiments and blurring show a strong effect of motion.

Results Phantom scans showed good temporal stability of the navigator signal for all measured b-values, albeit the noise level increased with the b-value (mean SD over all phantom scans for b500 / b1000 / b3000 = 0.14% / 0.19% / 0.23%). The same behaviour was observed in the human rest scans (mean SD=0.19% / 0.20% / 0.27%). The given motion commands and detected motions agreed with high sensitivity (SE) and specificity (SP). Over all retrospective mode experiments, they amounted to SE=92.0% / SP=99.8% (b500), SE=94.6%/ SP=99.6% (b1000) and SE=93.3% / SP=98.6% (b3000). An exemplary time course, together with the corresponding motion parameters as revealed by the co-registration, is shown in Fig. 1. Mean prospective mode images showed good preservation of the diffusion information, which was confirmed by only slight increases of the SD. In some cases, stronger ghosting and susceptibility artefacts in anterior brain regions were observed. Representative prospective mode as well as corresponding move and rest datasets are shown in Fig 2.

Discussion and Conclusions The FID navigator reliably detected movements of a large range. As an illustration, motion parameters for the smallest detected movement were trans=(0.31/-0.30/0.24)mm and rot=(-0.13/-0.14/-0.25)deg. The FID is sampled prior to the diffusion gradients and is hence (within limits) independent from the applied diffusion weighting. Currently, we see four main limitations in the presented implementation. Firstly, the method has not been tested with other coils and is likely to perform worse with non-helmet-shaped coils. Secondly, noise levels in the navigator rise with higher b-values, most likely to eddy current history effects. An adapted trigger threshold could help to prevent an increased false positive rate (by the cost of decreased sensitivity); the data suggests that, for lower b-values, a threshold of <1% is feasible (note the low noise level of the b1000 navigator dataset in Fig. 1). Thirdly, the volume-to-volume comparison scheme does not detect slow movements. Here, regularly triggered extra-b0 scans could be a straightforward remedy. Lastly, big movements (>8 deg, >5mm) may cause large local field variations, changing the local shim, which can yield image artefacts. It should also be noted that the total acquisition time increases slightly if motion occurred due to the extra-b0 and the repeated volume.

To conclude, we propose a robust and accurate method for prospective motion correction in diffusion imaging that maintains the diffusion directions consistency. It combines the advantageous characteristics of FID navigators, i.e. no impact on the imaging procedure with negligible time penalty, with a well-proven prospective registration method and should improve in particular long DSI/q-ball acquisitions as well as scans with uncooperative or paediatric patients.

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