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Prospective Motion Correction of Segmented Diffusion Weighted EPI

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Abstract

Purpose—Recently, a new algorithm was introduced to combine segments of under-sampled diffusion weighted data using multiplexed sensitivity encoding. While the algorithm provides good results in cooperative volunteers, motion during the data acquisition is not accounted for. In this work the continuous prospective motion correction of a segmented diffusion weighted acquisition is combined with multiplexed sensitivity encoding.

Methods—Simulations investigate the influence of motion on the reconstruction. Additionally, the change in coil sensitivities due to patient motion is taken into consideration. Finally, in vivo experiments display the effects of motion and its prospective correction on high resolution diffusion weighted imaging.

Results—Inconsistencies of the imaging plane lead to artifacts and blurring in the reconstructed dataset. Additionally, motion during the diffusion weighting period can lead to substantial image artifacts and signal dropouts. The change in coil sensitivities shows minor effect for the simulated range of motion (5 degrees). Prospective motion correction is shown to improve image quality in the case of large motion (5 degrees) and to reliably correct for small motion (1 degree).

Conclusion—The combination of prospective motion correction and multiplexed sensitivity encoding allows for high resolution diffusion weighted imaging even in the presence of substantial head motion.

Keywords

prospective motion correction; diffusion; segmented EPI; MUSE; real-time

Introduction

Diffusion weighted imaging (DWI) is one of the most popular techniques in MRI research, due to its sensitivity to molecular motion. However, the desired sensitivity to molecular motion also increases the likelihood of artifacts from bulk motion [1]. Additionally, DWI and diffusion tensor imaging (DTI) in particular may suffer from long examinations. To

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reduce measurement time, a single-shot approach is desirable (commonly echo planar imaging, EPI [2]), which however can only provide lower spatial resolution. Segmentation of the EPI readout allows for higher resolution [3], but at the cost of even longer examination times which augment the problems resulting from patient's motion.

Even without subject motion, the reconstruction of images from a segmented diffusion weighted dataset can be demanding, because of phase differences between segments due to blood flow or brain pulsation. For single shot techniques, these phase errors can usually be neglected as consistency of each *k*-space dataset is maintained. However, when a higher resolution is desired, a segmentation of the readout is required due to the signal decay during long readout trains since gradient slew rates are limited. When the segments are combined to one dataset without further processing, the described phase differences lead to substantial image artifacts.

Some solutions to the problem of reconstructing segmented diffusion weighted scans involve the use of phase navigators [3–8]. The measured phase difference between segments can then be included to the reconstruction. However, navigators either increase the minimum echo time (TE) when introduced before the data readout, or the minimum repetition time (and thus measurement time) when introduced after the readout. Alternatively, it has been shown that an iterative approach can be used to estimate the phase errors by minimizing background ghosting [9]. Other techniques like diffusion weighted PROPELLER [10] allow the combination of *k*-space segments by reconstructing low resolution images from the acquired data. Since every segment acquired covers the center of *k*-space, low-resolution navigator images can be reconstructed that contain information on the phase error as well as the subject's position. However, this oversampling of the *k*-space center comes at the cost of longer examination times. Additionally, while these approaches work in cooperative patients with minor movements, they fail in the case of through plane movements or fast motion during segments.

Recently, a new algorithm was introduced to combine segments of under-sampled diffusion weighted EPI data using multiplexed sensitivity encoding (MUSE, [11]). This approach uses sensitivity encoding (SENSE [12]) to reconstruct each under-sampled segment of *k*-space. These images are then used to determine phase differences between segments, which are utilized in a further SENSE reconstruction at full resolution (using all segments), yielding the final image.

It has been shown that prospective motion correction can be used to ensure consistency of the imaging plane during the measurement [13, 14] and to prevent signal dropouts due to motion during the diffusion weighting, either by continuously adjusting the strong gradients to the patient's position [15] or by dynamically applying short gradient blips to rebalance gradient moments [16, 17]. However, phase changes due to non-rigid body motion, e.g., due to flow or brain pulsation, cannot be corrected using these methods. In summary, a combination of reliable motion correction and sophisticated image reconstructed is required.

For this work, continuous prospective motion correction was incorporated to a segmented diffusion weighted EPI sequence to prevent motion artifacts in high resolution DWI. The

acquired data were reconstructed using the MUSE algorithm. In this paper the motion sensitivity of the sequence and the reconstruction are investigated and the successful implementation of a combined approach is presented.

Methods

Hardware

All experiments were conducted on a 3T Trio a Tim scanner (Siemens Healthcare, Erlangen, Germany). For motion tracking, an in-bore camera system was used (Metria Innovation, Milwaukee, USA [18]). This camera was mounted on top of the scanner bore above the RF coil and monitored a single marker fixed to the forehead (12 channel coil) or to a custom made mouthpiece (32 channel coil).

Sequence

The diffusion weighted segmented EPI sequence uses a double spin echo approach [19] and was modified to enable prospective position updates between excitations and during the diffusion weighting period [15].

Image Reconstruction

The MUSE algorithm uses a two-step approach to reconstruct segmented EPI scans. First, each segment is reconstructed individually using SENSE [12]. These images are then used to calculate the phase map of each segment in image space. However, instead of using these phase maps to perform a phase-correction for single segments, the phase data from individual segments are combined with the coil sensitivities to perform a second-level SENSE reconstruction of the full data set.

Minor changes to the algorithm described in [11] were implemented. The MUSE algorithm uses images with zero diffusion weighting to calculate the coil sensitivities for the SENSE reconstruction. However, in our experience, these images might suffer from signal fluctuations due to flow or brain pulsation. Since this step is crucial for all following reconstructions, a 20 second pre-scan (gradient echo sequence, TR = 200 ms, TE = 4.9 ms, matrix = 80 × 80) was used to acquire reliable coil sensitivity information. This scan was prospectively corrected to account for motion. Additionally, the tracking data was used to guarantee the consistency between pre-scan and imaging scan (position lock). Furthermore, the SENSE reconstruction proposed by Chen et al. [11] was replaced by an iterative SENSE [20] approach to provide the flexibility for different readout strategies in the future. Finally, instead of the phase cycling method used in the original MUSE approach, one dimensional phase correction using 1D navigators (3 non-encoded readout lines before diffusion encoding) was implemented. This substantially reduced reconstruction times and was found to provide sufficient accuracy for the scanner system used.

To explore the abilities of the reconstruction method in closer detail additional experiments were performed. These were conducted without the use of prospective motion correction, instead the subject’s head was stabilized using the maximum of acceptable padding. Sequence parameters were: TR = 3000 ms, TE = 108 ms, b = 0, and 1000 s/mm^2, 7

directions, matrix 256×256, field of view (FoV) 220×220 mm^2, and 15 slices of 3 mm. A 32 channel head-coil was used and a range of different segmentation factors (4–7) were applied.

Simulations

All simulations were performed in MATLAB, using an artifact free image together with the measured coil sensitivities. The data were acquired with the new segmented diffusion weighted EPI sequence and reconstructed using the MUSE algorithm (Sequence parameters: TR = 2000 ms, TE = 100 ms, 2 segments, b-value = 500 s/mm^2). The final image was then transformed into two *k*-space segments. These two segments were modified according to the simulation parameters and reconstructed without further modifications to the algorithm.

To investigate the motion sensitivity of the imaging sequence and the reconstruction algorithm, a number of simulations was performed. First, the effect of motion between segments was simulated by introducing in-plane pose changes. These simulations were performed separately for rotations and translations, with a range of motion of 0.1–3.0 degrees and 0.1–3.0 mm. The change in coil sensitivities (12 channel coil) due to the simulated motion was taken into account. In a second experiment, only a change in coil sensitivities was simulated, using the same motion parameters as above. This represents an experiment with prospective motion correction enabled, i.e. the object retains a constant position in the field of view, but the coil sensitivities change according to the actual motion. These simulations were performed for a 12 channel and a 32 channel coil.

Additionally, the effect of motion during the diffusion weighting period was simulated. In these simulations, the first segment was assumed to be the reference scan; therefore, the second segment reflected simulated effects of phase and position. To separate effects from different motion patterns, the position during excitation and readout was assumed to be constant. Two situations need to be taken into consideration: a) through plane translation which leads to signal loss due to misalignment of the excited slice and the refocusing pulses and b) signal dephasing due to rotations relative to the gradient direction. Translations relative to the diffusion weighting gradients were not simulated since the resulting global phase offset is corrected by the MUSE reconstruction. Effects of rotations and translations during diffusion encoding were simulated using MATLAB (0.1–3.0 degrees and 0.1–3.0 mm over the duration of the diffusion weighting), assuming a slice thickness of 3 mm, a diffusion weighting gradient of 20 mT/m, a voxel size of 1.7 mm and a TE of 120 ms, which corresponds to a b-value of about 1000 s/mm^2.

In vivo experiments

Three healthy human subjects were enrolled for the study (all male, 30, 34, and 32 years). All subjects provided verbal and written consent, using a protocol approved by our IRB.

In a first experiment, the effects of intentional head motion were investigated. A reference scan without motion was performed first, followed by two scans with comparable motion (absolute rotation of about 4 degrees). One of the ‘motion’ scans was corrected prospectively, using the tracking data from a marker fixed to the forehead of the volunteer.

A 12 channel head-coil was used. Sequence parameters were: TR = 2000 ms, TE = 93 ms, b = 0, 500, and 1000 s/mm^2, 6 directions, matrix 128×128, field of view (FoV) 220×220 mm^2, 2 EPI segments, and 9 slices of 3 mm.

In the second experiment, the second volunteer was instructed to lay as still as possible during two measurements (‘no intentional motion’). No additional padding was used to stabilize the head. One of the measurements was prospectively corrected to account for involuntary head motion. For this experiment, a 32 channel head-coil was used and the marker was fixed to a custom made mouthpiece to provide optimal stability and to enable a line of sight from the camera to the marker. Sequence parameters were: TR = 3700 ms, TE = 114 ms, b = 0, 1000 and 1500 s/mm^2, 20 directions, matrix 256×256, FoV 220×220 mm^2, 4 EPI segments, and 9 slices of 3 mm. The acquisition time for each measurement was approximately 10 min.

Results

Simulations

The image used for the simulations is shown in fig. 1. First (a), the image reconstructed using the adaptive (conventional) combination of all coils without further phase correction between segments is shown. Figure 1 b shows an image reconstructed from one undersampled segment. The third image (c) shows the same data reconstructed with the MUSE approach which results in a higher signal to noise ratio when compared to the undersampled single shot data (b). The third image was then used for the following simulations.

Figure 2 shows the comparison of DWI data with different segmentation factors (4–7) of the EPI readout. Each dataset was reconstructed using standard Cartesian (a–d) and MUSE (i–l) reconstruction. To compare with accelerated single shot EPI acquisitions, an additional SENSE reconstruction was performed using only one EPI segment (e–h). The images displaying standard Cartesian reconstruction all show the well known ghosting dependent of the number of segments (a–d showing 4–7 segments). Using a 32 channel coil acceleration factors up to 4 can be achieved with single shot EPI (e), but at a cost in the signal to noise ratio (SNR). While for higher acceleration factors SNR of the single shot image is further reduced (f–h), it is still possible to reconstruct the image phases used for the MUSE reconstruction (i–l). Note, that larger voxel sizes will increase SNR and therefore allow for higher acceleration factors for single shot sequences as well as for higher segmentation factors for MUSE reconstruction.

Figure 3 shows the effects of position differences between segments. The effect of motion in the range of 1 mm (a) and 1 degree (e) on segmented imaging is minimal and mostly results in slight blurring. However, movements greater than 3 mm (b,c) or 3 degrees (f,g) cause a substantial decrease in image quality. Note, that these artifacts would be corrected by prospectively adjusting the imaging volume before each excitation pulse. Figure 3d and h display the effect of changes in coil sensitivities after motion was prospectively corrected. For the 12 channel head-coil used, this effect seems to be negligible even for corrected motion of 5 mm or 5 degrees. Figure 4 shows the same simulation performed for a 32

channel coil. Figure 4a shows the unmodified image for comparison. Figure 4b and c show the effects of a 5 degree rotation and a 5 mm translation. Additionally, the differences relative to the image without coil motion (fig. 4a) are shown in fig. 4d and e.

Figure 5 displays the effect of motion during the diffusion encoding period. These effects lead to the well known signal dropouts which can be observed in DWI acquisitions. Since these effects are already well documented for single shot acquisitions [1, 15], our simulations only evaluate motion during one of the two acquired segments. In fig. 5 a–c, through plane motion between signal excitation and refocusing pulses is simulated. Such misalignment leads to signal loss in the affected segment and therefore results in ghosting in the final image.

Figure 5 d–f demonstrates that even very small rotations (e.g. f: 0.1 degree) over the duration of the strong gradients substantially degrade image quality. Since such rotations cause a shift of the signal in *k*-space, a sudden loss (compare e: 0.09 degree and f: 0.1 degree) of signal intensity can be observed when the *k*-space center is shifted outside of the field of view [15]. Note that, the artifacts shown in fig. 5 cannot be corrected using prospective slice-by-slice updates, but require a correction of intrascan motion.

In vivo experiments

Figure 6 shows data from the first experiment, with and without motion. Each row displays one exemplary slice and 3 out of 6 weighting directions for both b-values. In fig. 6 a and b (two top rows), the results of the scan with no intentional motion are displayed for comparison. The images in fig. 6 a were reconstructed using a conventional Cartesian reconstruction while the images in fig. 6 b were reconstructed using MUSE. Figure 6 c (third row) demonstrates substantial degradation of image quality when motion was not corrected; the same was true for all slices acquired. However, motion correction dramatically improved image quality (fig. 6 d); in fact, motion corrected images were comparable to those acquired without motion. In this experiment, the subject performed quasi-periodic head rotations of up to 4 degrees (tracking data of the motion corrected scan, fig. 6 e, bottom trace). The motion trajectory was similar for the scan without motion correction (trajectory not shown). The improvement in quality due to motion correction is also visible in the mean diffusivity maps (fig. 6, last column), reconstructed for a diffusion weighting of b = 500 s/mm^2.

Figure 7 shows results of the second in vivo experiment (no intentional motion, a: no correction, b: prospective motion correction), as well as the tracking data showing the total rotation for both scans. The three rows show diffusivity images (MD) for the two diffusion weightings (1000 and 1500 s/mm^2, left and center) and the fractional anisotropy (FA) maps (right column). Both scans show comparable image quality, probably due to the relatively small amplitude of the motion as well as the reliability of the reconstruction algorithm. However, without motion correction (top row), small changes in head position during the long acquisition (about 10 min) lead to some blurring in the MD images and decreased quality of the FA maps (top row). These artifacts were substantially reduced when prospective motion correction was enabled (bottom row).

Discussion

This work shows the successful implementation of a prospectively corrected segmented DWI-EPI sequence using multiplexed sensitivity encoding [11] for reconstruction. In combination these techniques allow for robust segmented DWI acquisitions even in the presence of substantial head movements.

While the modified MUSE reconstruction algorithm provides good results in cooperative volunteers, motion during the data acquisition can substantially reduce image quality. The effects of different motion paradigms were simulated and the results suggested the use of a motion correction technique. Since movements between individual segments as well as within diffusion encoding gradients (intra-scan) have to be taken into account, a prospective motion correction approach (with a fast tracking system) is favorable.

Prospective motion correction during DWI substantially improved image quality in the case of subject motion. In fact, our approach resulted in better image quality even without intentional subject motion. Of note, our experiments do not only demonstrate that it is possible to correct for such small motions, but also that prospective motion correction does not increase the artifact level (possibly introduced by erroneous tracking data).

A potential problem also investigated is that prospective adjustments of the imaging volume to the head's position dynamically alters coil sensitivities during segmented acquisitions. These changes might lead to image artifacts since these sensitivities are assumed to be constant during reconstruction [21]. Our simulation demonstrates that a change in coil sensitivities can be neglected for movements as large as 5 degrees. However, this issue may become more problematic for head-coils with a higher number of coil elements.

To enable the sequence to perform prospective updates during diffusion encoding, the correction scheme presented in [15] was implemented. It should be noted, that the continuous correction of diffusion weighting gradients is highly platform specific. Other approaches to correct for intrascan rigid-body motion (using external hardware [17] or navigator data [16]) or reacquisition methods [14] might provide more flexibility and lead to comparable results involving some increase in scan time.

This work used an in-bore camera to track the head position [18]. This approach involves the use of additional hardware such as a marker, but has the advantage of a continuous position tracking without time penalty. The mouthpiece used for high resolution DWI enables an undistracted line of sight, and is extremely stable and comfortable to wear. However, since it is custom made, it might not be acceptable for a broader patient population. Approaches like navigators could also be used to ensure the consistency of the imaging plane during acquisition, however, the additional time for the acquisition of the navigator signal needs to be taken into account. Additionally, continuous correction of motion during the diffusion encoding periods will not be possible with standard navigator approaches.

While the prospective correction of rigid-body motion ensures the consistency of the imaging plane during the measurement and prevents signal dropouts caused by motion

during the diffusion encoding period, the MUSE algorithm corrects for non-rigid motion. Previous work on the correction of phase errors resulting from such motion was based on data from phase navigators [3, 5, 6] or on oversampling of the center of *k*-space [10]. In comparison to these methods the MUSE reconstruction has the advantage of providing stable results without additional time penalty. However, the algorithm relies on the reconstruction of images from undersampled *k*-space data. Therefore the number of *k*-space segments is limited by the number of coils [11].

In conclusion, the combination of prospective motion correction and multiplexed sensitivity encoding allows for high resolution DWI even in the presence of substantial head motion, and might enable the use of multi-shot DWI in patients that cannot hold still.

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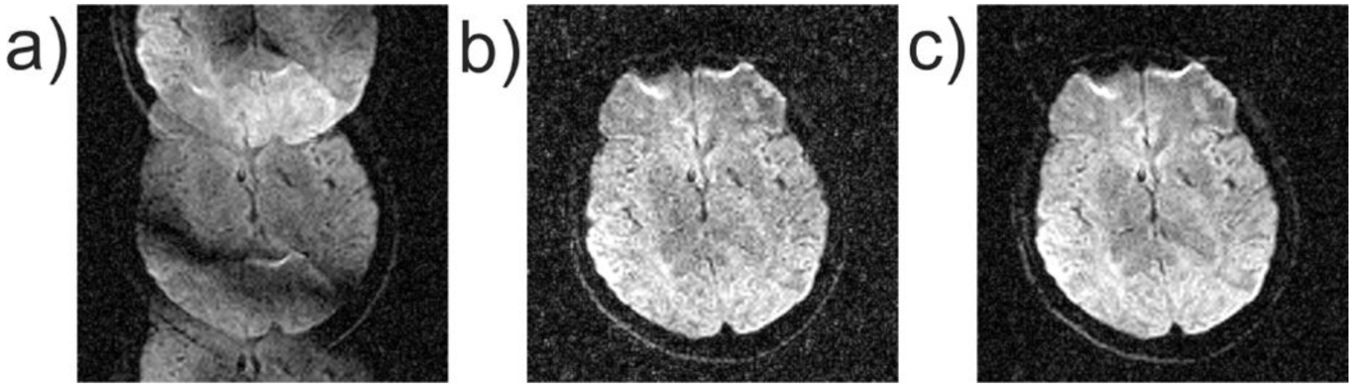

Figure 1.

Diffusion weighted image (b = 500 s/mm^2) acquired with the developed sequence using 2 EPI segments. In a) standard Cartesian reconstruction was used, in b) SENSE was used to reconstruct an image out of one segment (corresponding to a twofold accelerated dataset), in c) the result of the reconstruction with multiplexed sensitivity encoding is displayed.

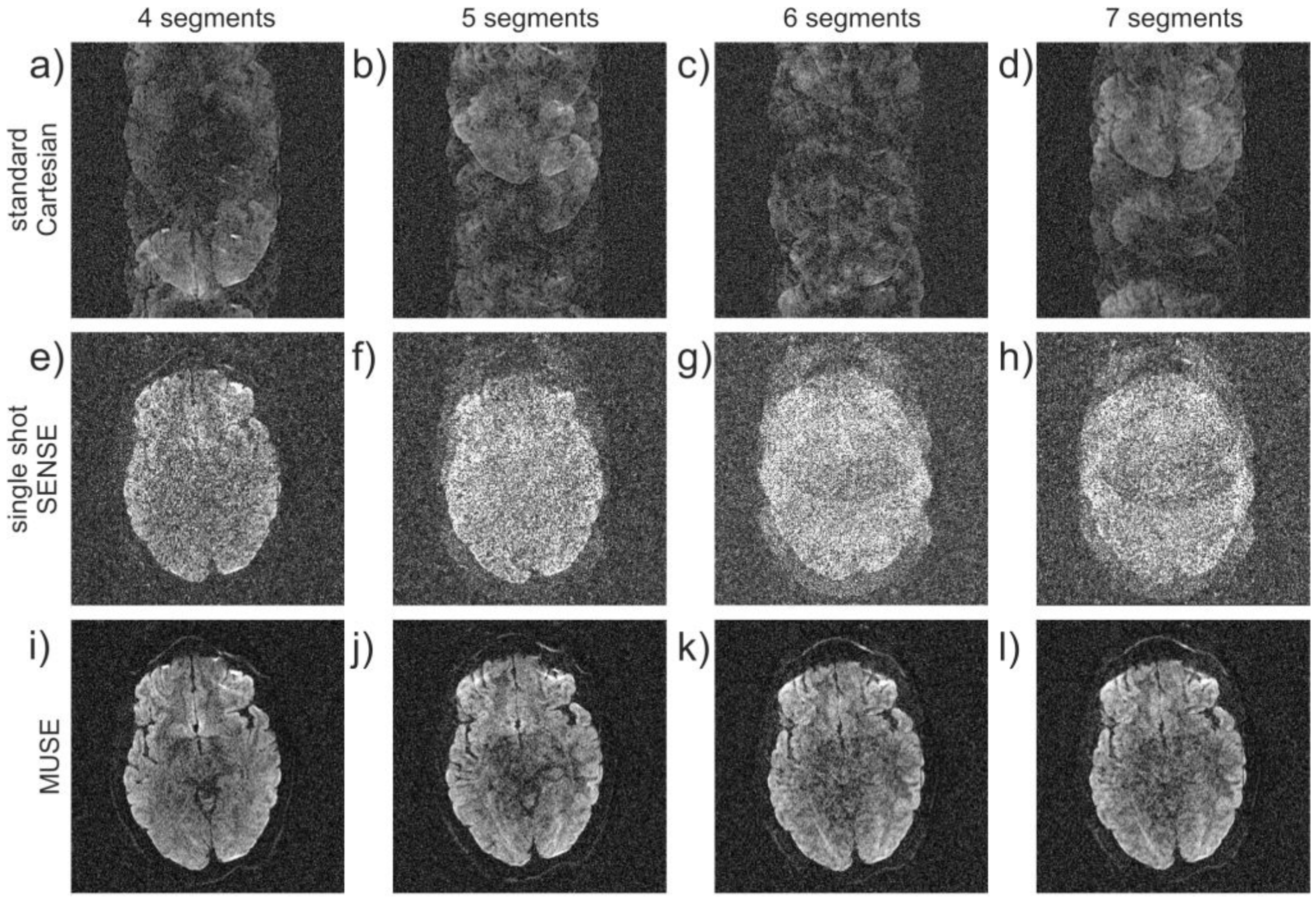

Figure 2.

DW Images acquired using a 32 channel head-coil with different number of segments (4, left to 7, right). All datasets were reconstructed using a standard Cartesian approach (top row) and MUSE (bottom row). Additionally, one segment of each dataset was reconstructed using SENSE (middle row); this corresponds to an accelerated single shot scan with an undersampling factor represented by the number of segments.

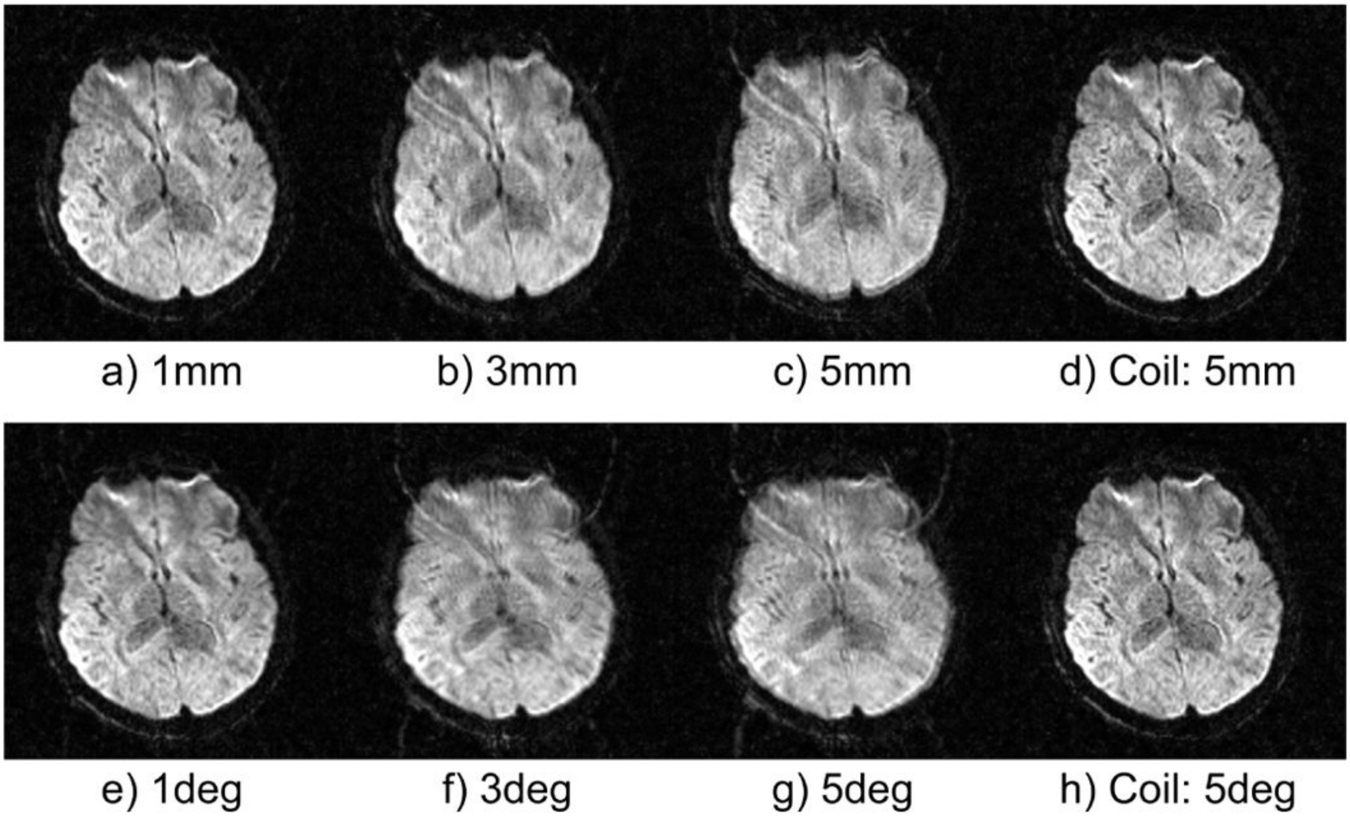

Figure 3.

Simulated in plane motion between excitations is shown for translations (a–c) and rotations (e–g). Additionally, the influence of the change in coil sensitivities after prospective correction was simulated for a translation of 5 mm (d) and a rotation of 5 degrees (h). The reference image can be found in fig. 1c.

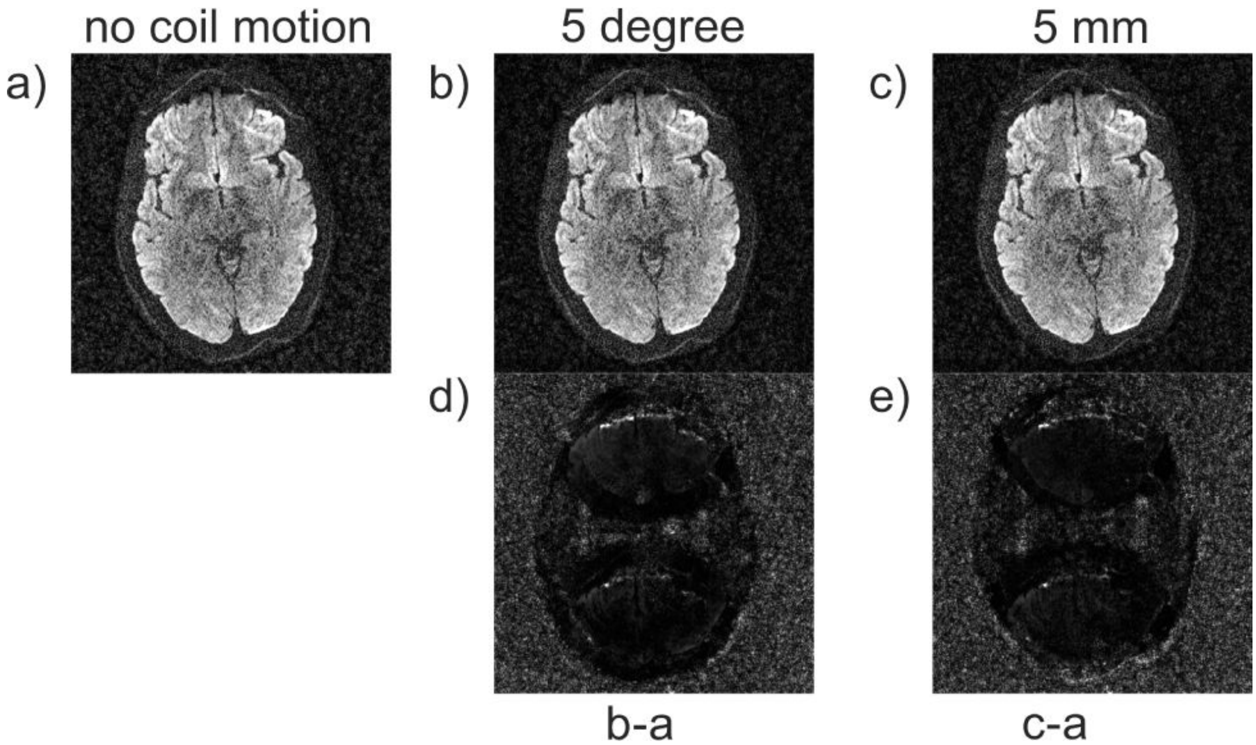

Figure 4.

The effect of a change in coil sensitivities after prospective correction was simulated for a 32 channel coil. a) Shows the unmodified image for comparison, b) the simulation of a 5 degree rotation and c) the effect of a 5 mm translation. Since no substantial effect can be seen in the magnitude image, the differences to the unmodified image is shown in d (b-a) and e (c-a).

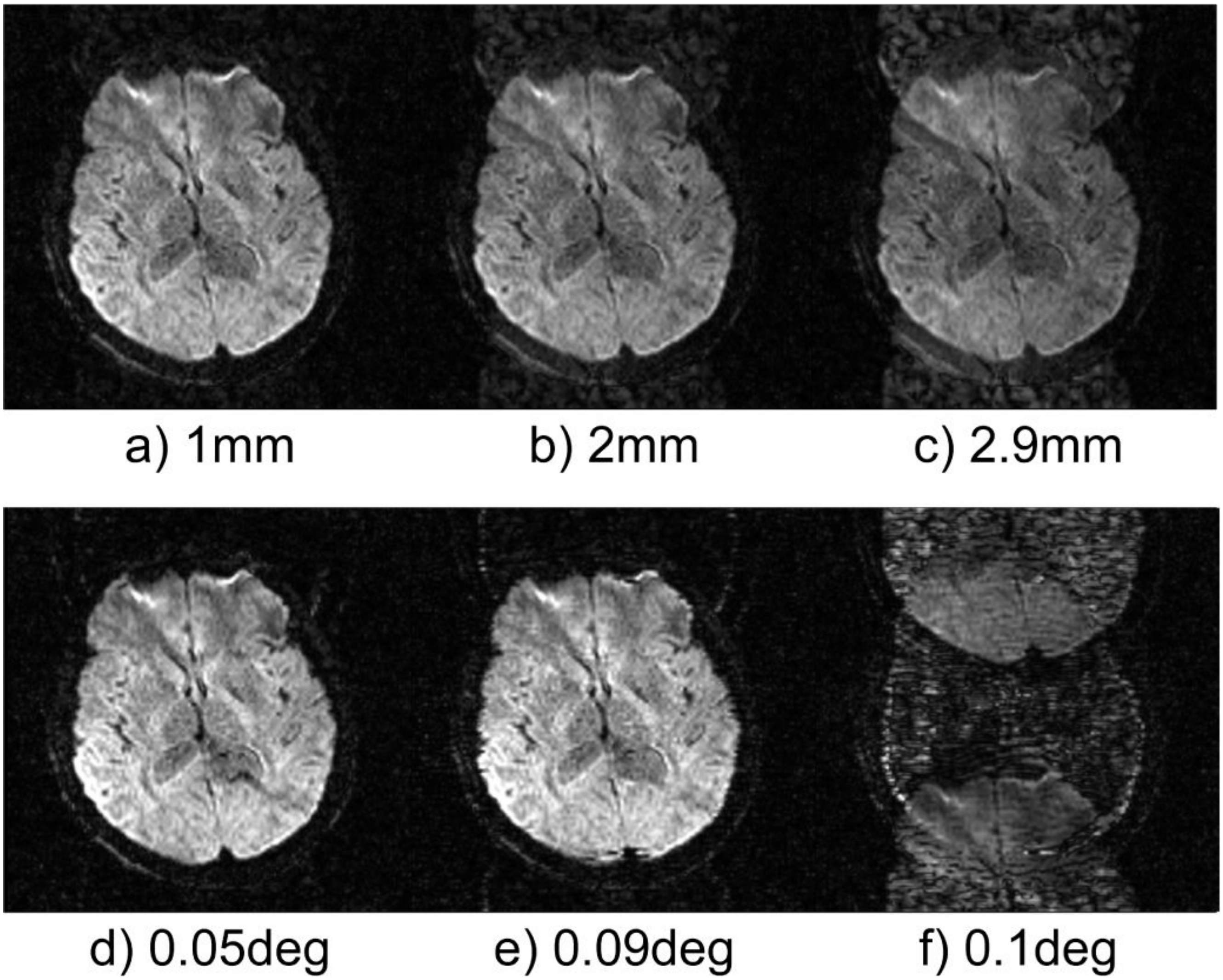

Figure 5.

Simulated motion during the diffusion weighting is shown for through plane translations (a–c) and in plane rotations with a diffusion weighting gradient orthogonal to the rotation axis (d–f).

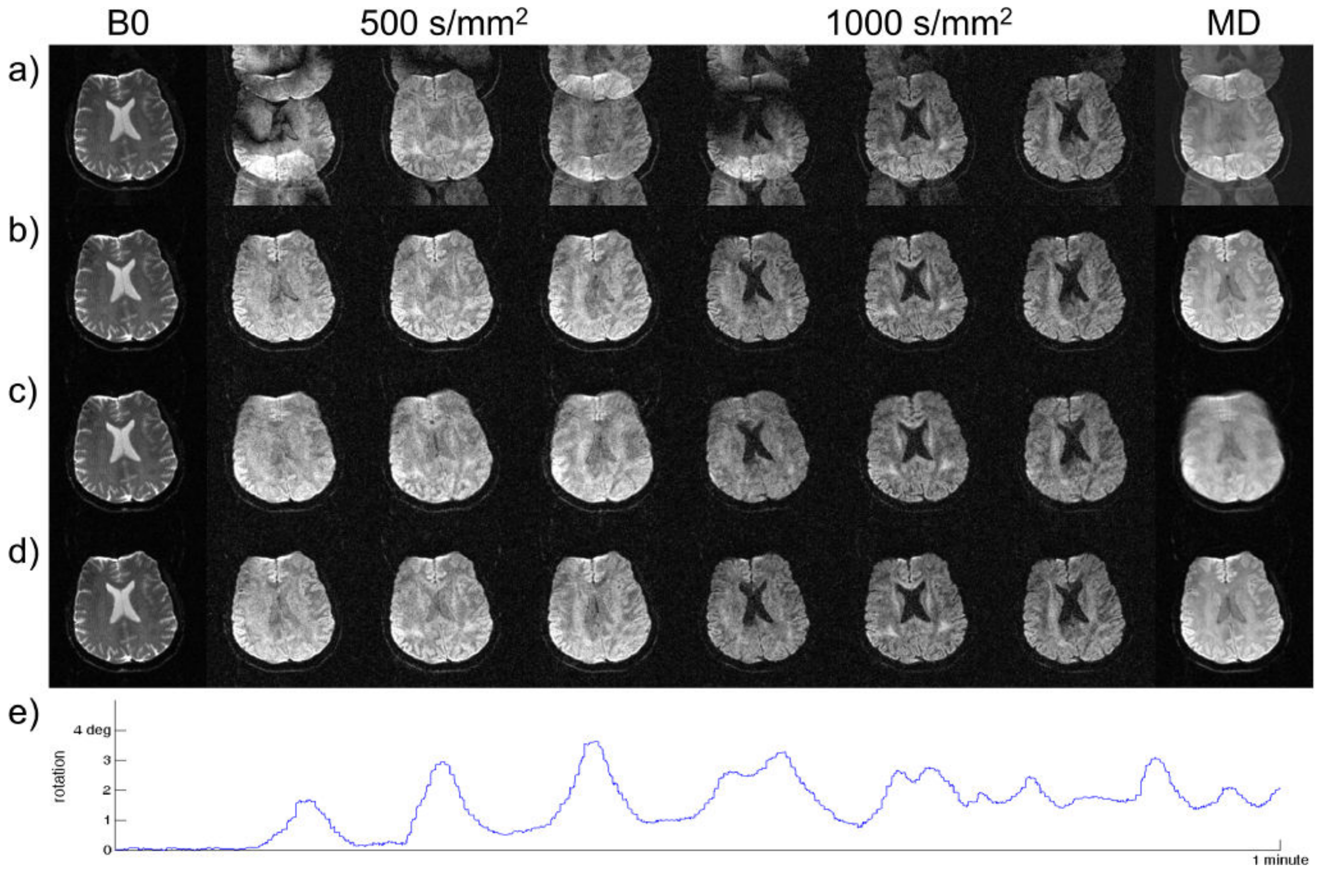

Figure 6.

The results of the first in vivo experiment are shown. For each measurement (b: no motion, c: motion, no correction, and d: motion plus correction) the unweighted B_0 image is shown, followed by the images from each weighting (500 and 1000 s/mm^2) and the MD image calculated from all six images with a weighting of 500 s/mm^2. In a: the images of the measurement without motion are shown, using a conventional reconstruction approach. Additionally, the tracking data from the third measurement are shown.

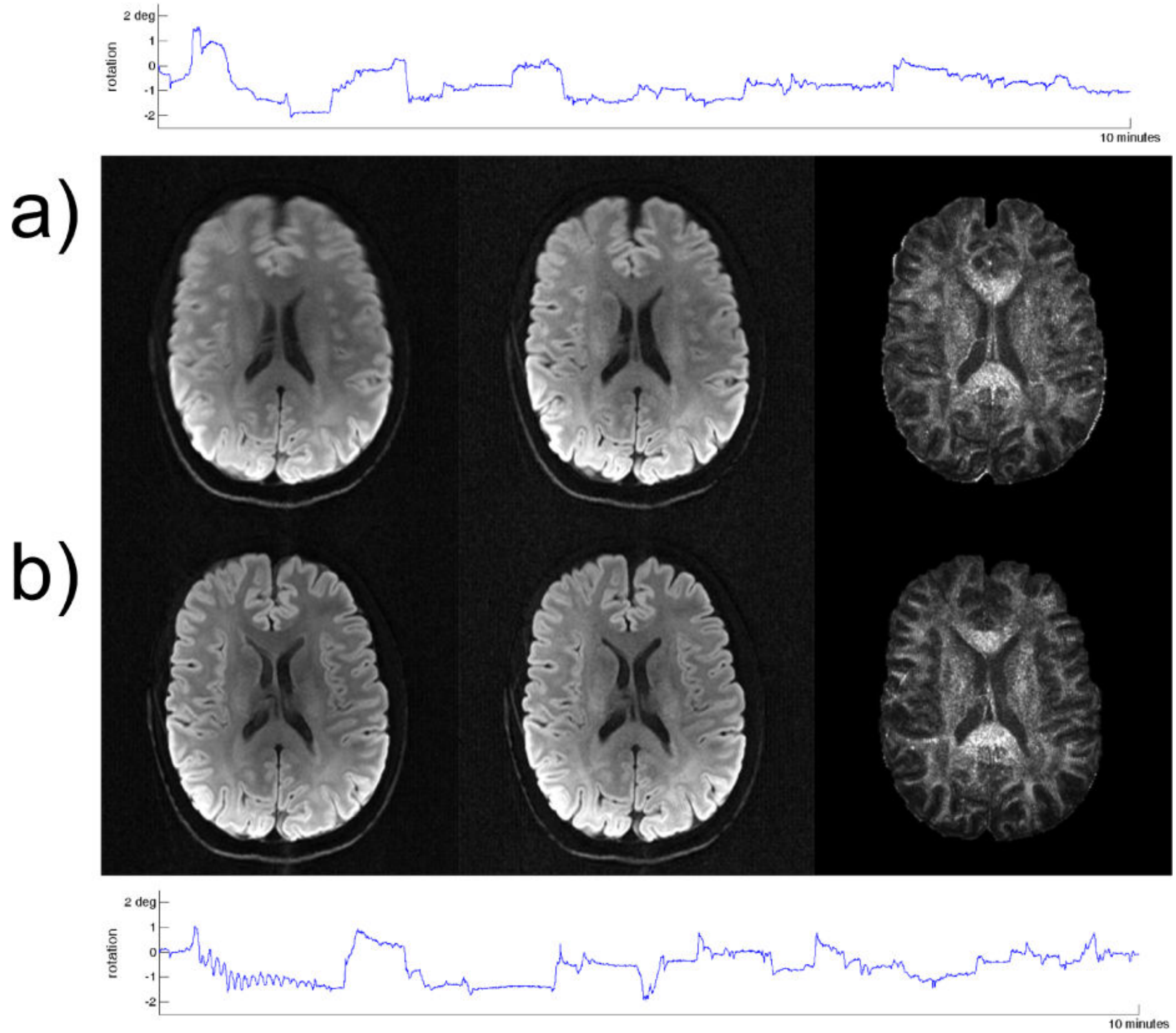

Figure 7.

The results of the second in vivo experiment without intentional motion are shown. Tracking data from both measurements are provided. Mean diffusivity images for b-values of 1000 and 1500 s/mm^2 and the FA maps combined from all measurements are displayed. In the first measurement (a) no correction was used, in the second one (b) prospective motion correction was enabled.