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Reverse Retrospective Motion Correction

Benjamin Zahneisen^{#, \$}, Brian Keating[#], Aditya Singh[#], Michael Herbst^{#, @}, and Thomas Ernst[#]

[#]University of Hawaii, Department of Medicine, John A. Burns School of Medicine, Honolulu

[@]Department of Radiology, University Medical Center Freiburg, Germany

^{\$}Stanford University, Department of Radiology, Stanford, CA

Abstract

Purpose—One potential barrier for using Prospective Motion Correction (PMC) in the clinic is the unpredictable nature of a scan because of the direct interference with the imaging sequence. We demonstrate that a second set of “de-corrected” images can be reconstructed from a scan with PMC that show how images would have appeared without PMC enabled.

Theory and Methods—For 3D scans, the effects of PMC can be undone by performing a retrospective reconstruction based on the inverse of the transformation matrix used for real time gradient feedback. Retrospective reconstruction is performed using a generalized SENSE approach with continuous head motion monitored using a single-marker optical camera system.

Results—Reverse retrospective reconstruction is demonstrated for phantom and in vivo scans using an magnetization-prepared rapid gradient echo (MPRAGE) sequence including parallel and Partial Fourier acceleration.

Conclusion—Reverse retrospective reconstruction can almost perfectly undo the effects of prospective feedback, and thereby provide a second image data set with the effects of motion correction removed. In case of correct feedback, one can directly compare the quality of the corrected with that of the uncorrected scan. Additionally, since erroneous feedback during PMC may introduce artifacts, it is possible to eliminate artifacts in a corrupted scan by reversing the false gradient updates.

Introduction

Motion correction in MRI is becoming increasingly important, in part since recent technical advances enhanced the spatial resolution achievable in routine applications into the sub-millimeter range. However, the effective spatial resolution in this regime may be limited by involuntary, physiologic head motion (e.g. due to breathing) even for cooperative patients. Motion correction can overcome this limitation, as well as reduce or eliminate motion artifacts in case of severe patient. Therefore, motion correction can ultimately save scan time by diminishing the need for repeated scans.

A range of motion correction strategies has been developed over the past decade. One class of methods make use of the actual MR signal to provide motion information (MR navigators (1)), whereas a second class of techniques relies on information from an external tracking device (e.g. optical (2–4), infrared, ultra-sound (5), active MR probes (6)) that typically does not interfere with the MRI measurement process. Motion correction can also be based on intrinsic image properties, i.e. motion estimation is based on minimizing a suitable image metric (autofocus strategies (7)). Furthermore, one has to distinguish between prospective and retrospective motion correction (PMC and RMC, respectively). Prospective techniques aim at providing real time motion feedback that can be used to update acquisition parameters, either once per TR or continuously (8), during the encoding process (4, 9, 10). Conversely, retrospective methods do not update acquisition parameters in real-time. These methods gather motion information during the scan, which is then incorporated into the reconstruction process by updating the k-space trajectory to achieve motion correction (11–13). Combining prospective and retrospective motion correction was proposed to de-noise the outcome of PMC (14), and to minimize the effects of imperfections in the camera cross-calibration (15). One advantage of PMC is that it can be applied to 2D multi-slice techniques, which require real time updates of the slice position and orientation. Furthermore, PMC results in k-space data on a Cartesian grid, which dramatically facilitates image reconstruction because standard methods can be employed.

Conversely, RMC requires non-Cartesian reconstruction and implementations are generally limited to sequences with volume excitation (i.e. 3D acquisitions). However, RMC also has several advantages over PMC. First, because the complete motion trajectory of the object of interest is known at the time of reconstruction, it is possible to filter and pre-process the motion data before the actual trajectory updates are applied. For instance, tracking data can be de-noised, and time periods with unreliable or erroneous tracking data may be excluded from the reconstruction process.

In fact, inaccurate tracking data may degrade the effectiveness of PMC, or even induce artifacts that would not have been present without PMC (for instance, if subject motion was minimal). The introduction of motion correction techniques into clinical routine clearly requires a minimal level of false positives; i.e. it has to be ensured that motion correction does not worsen quality of images compared to an uncorrected scan.

A related problem is the difficulty in quantifying the performance of newly developed PMC methods since the control measurement, i.e. the uncorrected scan, is not available. Herbst et al. proposed a method where the tracking information of a patient scan (with motion) is used to rotate the gradients for a second scan without subject motion in order to re-generate the expected motion artifacts (16). While this technique can provide important information in a research setting, it requires a second scan and therefore doubles scan time.

Here, we suggest using the framework of retrospective motion correction to reverse the effects of prospective motion correction (“reverse retrospective correction”) for 3D brain scans. Therefore, each scan yields two sets of images, one reconstructed from the original data with PMC enabled, and a second “reference” set that mimics a scan with PMC disabled. This allows for a direct evaluation of the effects of PMC on image quality when motion is

present, and provides a means to restore an uncorrected image set in case of erroneous corrections. Retrospectively reversing effects of PMC is limited to conditions where PMC and retrospective correction perform equally well and are therefore able to mutually invert the effects of each other. From our data we conclude that this is the case for a restricted range of motion (ca. 10 degrees) with a motion trace that does not lead to large gaps in k-space due to a single large rotation of the head. In combination with 3D sequences, where subject motion and the choice of the k-space trajectory do not interfere with the steady-state magnetization, this allows for a reversal of PMC effects. This is particularly true for GRE sequences and the MPRAGE (magnetization-prepared rapid gradient echo) sequence with non-selective RF excitation. We validate this by comparing the outcome of a prospectively and a retrospectively corrected scan. For the retrospective correction we used a 3D multi-channel iterative augmented SENSE (11) reconstruction approach.

Theory

Reverse retrospective correction

Motion correction can be formulated in terms of a change of reference frames. We assume that head motion at time point t relative to an initial time point t_0 is given as the 4×4 homogeneous transform matrix $\mathbf{A}(t)$, which describes the position and orientation (pose) change relative to the initial logical coordinate system. The logical coordinate frame is defined by three orthogonal gradient directions (e.g. phase, read, and slice direction).

Using an augmented 4d k-space vector $\mathbf{k} = (k_x, k_y, k_z, \phi_0)^T$ where the 4th component represents an initial phase, the MRI signal equation including motion can be written in a compact form (17):

$$s(t) = \int \rho(\mathbf{r}_0) e^{i\mathbf{r}_0^T \mathbf{A}^T \mathbf{k}_0} dV \quad [1]$$

where $\mathbf{k}_0(t)$ represents the undistorted, i.e. Cartesian, k-space trajectory, which is transformed by $\mathbf{A}^T$ into $\mathbf{k}_{\text{motion}}(t) = \mathbf{A}^T \mathbf{k}_0(t)$. Retrospective motion correction is performed by implicitly or explicitly using the actual k-space trajectory $\mathbf{k}_{\text{retro}}$ when performing the transformation from signal space back to image space. In general, augmented 4D pose vectors are transformed by matrix multiplication with $\mathbf{A}$ and augmented 4D k-space vectors are transformed with $\mathbf{A}^T$ (17).

Prospective motion correction compensates for the motion-induced k-space transformation by dynamically transforming the initial k-space into $\mathbf{k}_{\text{PMC}}(t) = \mathbf{A}_{\text{PMC}}^T(t) \mathbf{k}_0(t)$ so that $\mathbf{A}^T(t) \mathbf{k}_{\text{PMC}}(t) = \mathbf{k}_0(t)$. From this requirement it follows that the prospective transformation must be the inverse of the head motion transformation: $\mathbf{A}_{\text{PMC}}^T(t) = \mathbf{A}^{-T}(t)$. K-space data are thus acquired on a Cartesian grid and can be reconstructed using $s_{\text{PMC}} = \int \rho(\mathbf{r}_0) e^{i\mathbf{r}_0^T \mathbf{k}_0} dV$ as a signal model.

As noted above, the goal of this work is to reverse the effects of prospective motion correction. This can be achieved by inverting the applied 4D k-space transformation.

Therefore, the k-space trajectory for the inverse reconstruction of data acquired with PMC enabled is $\mathbf{k}_{\text{PMC}}(t) = \mathbf{A}^{-T}(t) \mathbf{k}_0(t)$.

Methods

Image acquisition and prospective feedback

Experiments were performed on a 3T Tim Trio (Siemens Healthcare, Erlangen). Data were acquired in N=4 healthy subjects after obtaining informed consent using an institutionally approved protocol. Motion tracking was performed with a prototype optical motion tracking system based on Moiré-Phase-Tracking (MPT) of a single marker (15mm) (18). The tracking system was operated at 85 frames per second, and has an absolute precision of <0.1mm and <0.1° over the entire measurement range (Metria Innovation, Milwaukee, WI, (18)). PMC was performed using a magnetization prepared gradient echo sequence (MPRAGE) capable of adapting the scan geometry according to object motion in real time (2). The communication interface between the sequence and the camera was provided by the XPACE library (2). The camera cross-calibration was determined and refined using a method based on feedback from residual tracking errors (19). Scan parameters of the MPRAGE sequence used in combination with a 12-channel head coil were: TR/TE/TI = 2000/5/1100, readout bandwidth 130 Hz/px, base resolution 192×192×160, voxel size 1.1 mm isotropic, partial Fourier 6/8 in phase direction and 7/8 in slice direction. In addition, we acquired an MPRAGE sequence employing parallel imaging and a 32-channel head coil array. The scan parameters were: TR/TE/TI = 2000/5/1100, readout bandwidth 130 Hz/px, base resolution 128×128×112, voxel size 1.8 mm isotropic, partial Fourier 6/8 in phase direction and 7/8 in slice direction, twofold GRAPPA acceleration along phase direction, 24 reference lines. Protocol parameters for the scan with selective squint removal (Figure 6) were: 12-channel head coil, TR/TE/TI = 2000/5/1100, readout bandwidth 130 Hz/px, base resolution 256×256×160, voxel size 1 mm isotropic, partial Fourier 6/8 in phase direction and 7/8 in slice direction, 2× GRAPPA acceleration with 24 reference lines.

Retrospective motion correction using iterative augmented SENSE

Based on Eq.[1] retrospective motion correction is implemented within the framework of generalized iterative reconstruction (11, 20) by transforming the trajectory used to define the forward operation. Therefore, compensating for arbitrary head rotations during image reconstruction (retrospective MoCo) effectively leads to non-Cartesian k-space sampling. Reconstruction involved the following steps.

1. Synchronize the sequence timing of each line in k-space with the frame times of the camera system, based on the real-time communication interface between the scanner and the camera. As a result, each line in k-space (index n) then has a corresponding 4×4 homogeneous matrix $\mathbf{B}_n$ encoding the head pose relative to the initial pose as seen by the camera.
2. Transform the synchronized camera motion into the logical coordinate frame (phase, read, slice) by using the logical cross-calibration $\mathbf{Y} = \mathbf{S}_0\mathbf{X}$ that combines the effects of camera position relative to the fixed gradient frame (cross-calibration $\mathbf{X}$)

and the initial slice/slab orientation $\mathbf{S}_0$ relative to the gradient frame (see Figure 1 and (17)). Head motion in the logical frame is then $\mathbf{A}_n = \mathbf{Y}\mathbf{B}_n\mathbf{Y}^{-1}$

3. Transform the uncorrected augmented k-space data vectors $\mathbf{k}_n^0$ according to

$$\mathbf{k}_n^{\text{motion}} = \begin{cases} \mathbf{A}_n^T \mathbf{k}_n^0 & \text{RMC} \\ \mathbf{k}_n^0 & \text{forno correction} \\ \mathbf{A}_n^{-T} \mathbf{k}_n^0 & \text{reverse PMC} \end{cases}$$

4. Generate coil sensitivities by reconstructing a fully encoded centered subset of k-space data (e.g. 24 reference lines in case of GRAPPA (21) type acceleration) using the motion corrected trajectory $\mathbf{k}_n^{\text{motion}}$ from steps 1–3. The coil sensitivities are assumed to be stationary, i.e. effects of motion in the inhomogeneous receiver coil field during the acquisition are neglected.
5. Perform retrospective, iterative SENSE reconstruction using $\mathbf{k}_n^{\text{motion}}$ from all acquired lines (including the reference lines in GRAPPA accelerated scans). Because of the iterative CG reconstruction, no density compensation is needed for the variable density k-space. Computational effective calculation of the forward model and its transpose was performed using the nuFFT gridding algorithm (available at <http://www.eecs.umich.edu/~fessler/irt/irt>). Image quality typically did not improve after 8–12 conjugate gradient iterations (judged by visual inspection). Reconstructions were performed on a quad-core computer with 16GB memory; a full reconstruction for a $196 \times 196 \times 148$ dataset acquired with a 12-channel head coil took approximately 20min.

No attempt was made to re-generate missing data due to Partial Fourier acceleration, which corresponds to zero-filling in case of standard FT reconstruction.

Results

Phantom validation

To validate the mathematical framework of our approach, three scans were acquired with position lock functionality; this ensured that the pose of the first scan is stored to define a pose reference (Figure 2a). Prior to the second scan (Figure 2b), the phantom was moved. The second scan then was acquired with PMC switched on, using the reference pose from the first scan. The resulting images show (b) the same internal phantom structure as the reference images. However, the air bubble (red circle) disappears in **b** because it is not subject to rigid body motion. Figure 2c shows difference images between **a** and **b**. Figure 2e shows that retrospective de-correction of the scan in **b** restores the phantom orientation to its actual position in the same way the scan with PMC switched off in Figure 2d does. Sensitivity maps from scan **d** were taken to perform the reverse reconstruction in **e**. This was necessary because we performed inter-scan correction and the position of the phantom in the reconstructed frame changed, while the support of the coil sensitivities is limited to the region the scanned object supports. However, for a typical scenario with intra-scan corrections and no apparent change of object pose (after correction), this issue does not

occur. In both cases, the phantom is acquired (PMC off) or appears (reverse-corrected PMC) in its transformed orientation. The air bubble remains stationary between (d) and (e) because no actual motion took place between the two scans. In Figure 2f the residual difference between d and e is shown.

Retrospective de/correction of MPRAGE scans

Figure 3a displays three orthogonal slices of a “baseline” MPRAGE sequence acquired without intentional motion and without prospective feedback (upper motion trace from the right column in Figure 3). A second scan with PMC enabled (Figure 3b) results in virtually identical image quality despite severe head motion, with rotations of over $\pm 5^\circ$ (2nd motion trace).

During a third scan; tracking data were recorded continuously but PMC feedback was disabled. Head motion for this and the prior scan were almost identical (bottom motion graph). Reconstructing the third scan without retrospective correction results in images that have substantial motion artifacts, and thus would be non-diagnostic and could not be used for gray/white matter segmentation (Figure 3c). When images are reconstructed using retrospective correction (Figure 3d), image quality is similar to that of the scans in Figures 3a or 3b. Figure 3e was reconstructed by reversing the effects of PMC in the second scan. The resulting images show motion artifacts that resemble those from the uncorrected scan with motion and PMC off (Figure 3c).

Figure 4 shows the results for a scan acquired with the 32-channel head coil and including two-fold GRAPPA acceleration along the phase direction. For Figure 4a the subject kept his head still (upper motion trace) and the scan was acquired without motion correction. Both prospective (b) and retrospective motion correction (c) result in an image that is comparable in quality as that in (a), despite the substantial head motion during the scans (greater than ± 5 degrees, middle and lower motion traces). Reverse correction of the scan acquired with PMC in (d) results in image artifacts comparable to the uncorrected reconstruction in (e). Also, the pronounced ringing artifact in the sagittal view is present in scans a–c, from which we conclude that it is not related to motion-correction but rather due to a suboptimal choice of imaging parameters.

Artifact removal in case of erroneous prospective feedback

The ability to de-correct scans acquired with PMC also makes it possible to remove artifacts that were introduced by inaccurate motion feedback. For instance, Figure 5a shows a reference MPRAGE scan acquired without intentional head motion and without PMC. Next, a scan was acquired with PMC enabled, and the subject was asked to keep his head still but to squint and yawn periodically. Since the marker was attached to the forehead above the eyebrows, the squinting and yawning events were associated with false marker movements that induced spikes in rotational and translational components of tracking data (Figure 5d). Because of these erroneous tracking data, the PMC enabled images exhibit motion artifacts (Figure 5b). Reverse-correction of this PMC enabled scan using retrospectively corrected reconstruction improves image quality (Figure 5c versus Figure 5b) and results in images that are of the same quality as those of the baseline scan (Figure 5c versus Figure 5a).

Selective removal of erroneous prospective feedback

Figure 6 displays how a severely corrupted scan with PMC can be recovered if false motion updates can be identified reliably. The raw motion trajectory $\mathbf{A}_{full}(t)$ as seen by the camera is displayed in Figure 6d. Because of the nature of this motion trajectory, false marker movement $\mathbf{S}(t)$ (squints) can be readily identified as sharp peaks between 100s and 200s. This false marker motion causes prospective motion correction to fail (Figure 6a). Reversing all PMC updates in Figure 6b somewhat improves the image quality but still suffers from degradation because the actual head oscillations at the beginning and the end of the scan are also removed. After the squints were identified, an estimate $\mathbf{A}_{true}(t)$ of the head motion trajectory was derived by replacing false poses during the squint events with the last pose assumed to reflect the true head motion. The squint-only motion trace in **e** is then given by dividing the raw motion trace by the estimated head motion trajectory $(t)=\mathbf{A}_{full}(t)\mathbf{A}_{true}^{-1}(t)$. For selective removal, **S** is supplied to step 3 of the retrospective reconstruction in the “reverse PMC” mode. Figure 6c shows the result from the retrospectively de-corrected scan when only the squint motion (Figure 6e) is removed. In doing so, the image quality of an uncorrupted PMC scan can be restored.

Discussion

We demonstrate that forward and reverse, prospective and retrospective motion correction can approximate each other in selected 3D imaging sequences where the k-space trajectory does not interfere with the steady-state signal. This flexibility allows for reversing the effects of PMC retrospectively, either fully or only for selected time periods during the acquisition where PMC is assumed to be erroneous. One application of reversing PMC effects is testing of PMC methods, since one can directly compare the quality of a prospectively corrected scan with that of the corresponding “de-corrected” images, based on identical raw data and identical motion trajectory. This ability addresses an important problem in terms of validating PMC implementations in the clinical setting, since current approaches would require a patient to perform the same movement twice over the course of two scans (one with and one without PMC enabled), which is entirely unrealistic. From our results, which cover only a subset of potential settings and subject motion patterns, we conclude that under the specific but still representative circumstances presented in this work PMC and RMC perform equally well. A pragmatic conclusion could be to avoid implementing PMC in the MR sequence at all and entirely rely on retrospective correction. However, the intention of this work is to provide a link between both approaches, which ultimately allows one to combine the inherent advantages of both methods (Cartesian sampling for PMC and full motion trajectory available for RMC) and reduce the effect of their shortcomings. A possible scenario where RMC alone would perform not as well as PMC is given by a single large head rotation during a significant time of the k-space acquisition. This would result in a wedge-shaped region in k-space with zero sampling density. If this region is too large or if the coil array does not provide enough sensitivity variation, the missing k-space information can't be restored and will therefore affect image quality. In contrast, PMC maintains uniform sampling density throughout the entire acquisition.

A second application of reversing PMC effects lies in the removal of potential artifacts due to PMC tracking inaccuracies. For our tracking system, potential sources of tracking errors include noise in the tracking data (22), lag time of the tracking system (14), and non-rigid movements between the marker and the organ of interest (e.g. the brain) as exemplified above by squinting events. In the case of MR-based navigator scans, tracking errors may also be caused by orientation-dependent distortions of navigator images, or inaccuracies in registration algorithms. Therefore, while PMC may improve image quality in many instances, it also inherently has the potential to worsen image quality compared to a scan with no motion correction. Reconstructing a second data set with the effects of PMC reversed will resolve this dilemma, since at least one of the data sets will be equal to or better in quality than a scan without PMC.

A more sophisticated artifact removal approach relies on post-processing the motion trajectory data in order to decide whether a given line in k-space should be corrected, uncorrected or discarded (rejected) from the reconstruction process. In particular, since the complete marker trajectory is available during retrospective reconstruction, automatic or semi-automatic signal processing strategies can be applied to identify physiologically implausible marker motions. Criteria include extreme angular velocities and accelerations, marker slippage, or relative marker movement in case of multi sensor methods as shown in (23). The corrupt data can then be discarded from the reconstruction process which is expected to improve the validity of reconstruction (13). In Figure 6 we show that false marker movement that occurred during a scan with PMC can be selectively removed while leaving the correct PMC updates unaffected. For this initial demonstration with a focus on MR reconstruction, an example was chosen that allowed relatively simple and manual detection of false updates (squints as sharp peaks). The ability to partially reverse false updates crucially relies on sufficient k-space coverage throughout the entire scan in the same way RMC does. If marker slippage results in large rotations and k-space regions with very low sampling density, the information is lost and cannot be recovered. However, for false updates that are either limited in range or occurrence and moderate initial k-space acceleration, the parallel imaging reconstruction employed in this work will most likely be able to restore the missing k-space information.

While our current implementation does not address apparent changes in coil sensitivities due to rotation of k-space (“coil motion”), a GRAPPA like version of a 3D motion corrected reconstruction pipeline supporting partial Fourier acceleration and accounting for effective changes in coil sensitivity variation was recently shown in (24). The authors refined their retrospective correction by transforming the coil sensitivities for a group of k-space lines with similar orientation (binning into motion groups). However, due to the limited quality of the motion corrected scans in (24), it is difficult to decide whether a significant improvement in image quality is to be expected from such an approach. Specifically, the binning of motion corresponds to the multiplication of k-space with boxcar functions, and Fourier Transformation of these step functions during image reconstruction may reduce image quality since they correspond to a convolution in image space with a sinc-function, which leads to additional blurring as well. However, for the data presented in this paper we did not observe significant effects of coil motion when comparing datasets acquired under stationary conditions with datasets acquired under substantial head movement, neither for the 12-

channel nor for the 32-channel head coil with twofold GRAPPA acceleration. A finding that it is in agreement with recently published work (25) and is probably due the relative smoothness of B1-profiles. However, for higher acceleration factors and less favorable coil geometries, it is sensible that including apparent coil motion would further improve the reconstruction. In case of high acceleration factors and specific motion patterns that result in large regions of k-space with very low sampling density, the inherent parallel imaging reconstruction of our approach may fail to restore missing k-space information.

Retrospective motion correction ultimately requires a non-Cartesian reconstruction approach. We have chosen to use iterative CG-SENSE because of its accuracy, robustness and flexibility. The main drawback of CG-SENSE is certainly reconstruction time and hardware requirements. Full brain coverage with isotropic resolution of 1.2mm and acquired with a 12-channel head coil can be reconstructed in less than 30min on a regular quad-core computer. Other approaches like gridding are faster but were shown to be less accurate in the context of motion correction (14). Of note, the reconstruction time of the CG-SENSE implementation used here scales linearly with the number of individual receiver coils because they are treated as more or less independent sub-problems. This, in return, makes it an ideal candidate for a parallel computing environment.

Another limitation is the applicability of the proposed method to imaging sequences where spin history effects are important (such as many fMRI acquisitions or motion corrected TSE (26)) or where motion causes signal losses that cannot be recovered during the reconstruction, such as during DTI scans (8). The latter is particularly important for 2D multi-slice sequences in case of through-plane motion, since k-space based retrospective correction approach can only account for in-plane effects then.

In conclusion, the ability to perform forward and reverse prospective and retrospective motion correction provides a great deal of flexibility. Reverse corrections promise simplified testing and validation of PMC sequences in the research setting. However, the most important impact will most likely be in the clinical application of PMC, where our approach guarantees that a data set can be presented whose quality is at least as good as a scan acquired without PMC. This feature should eliminate one of the remaining barriers in introducing PMC into the clinic.

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Figure 2.

Phantom validation of reverse correction step. **a)** Initial pose of the phantom, defining the reference pose for the 2nd PMC enabled scan. **b)** Rotated/translated phantom acquired with PMC enabled shows the phantom in its initial orientation. **c)** Difference between **a** and **b**. **d)** PMC disabled reconstruction of the phantom in its 2nd pose. **e)** Reversing the effects of PMC in scan **b** reconstructs the phantom in a orientation that is aligned with reconstruction where PMC was disabled. **f)** Difference image between **d** and **e**.

Figure 3.

Retrospective de-correction of an MPRAGE scan. **a)** base line image: no motion, no feedback. **b)** head motion acquired with PMC using standard reconstruction. **c)** head motion acquired without PMC and no retrospective correction. **d)** retrospectively corrected reconstruction. **e)** De-corrected reconstruction of PMC. Right column: Motion traces from the three acquisitions.

Figure 4.

Retrospective de-correction using parallel imaging and a 32-channel head coil array. **a)** Base line image: no motion, no feedback; **b)** head motion acquired with PMC using standard reconstruction; **c)** retrospectively corrected reconstruction; **d)** de-corrected reconstruction of PMC; **e)** head motion acquired without PMC and no retrospective correction. Right column: Motion traces from the three acquisitions.

Figure 5.

Removal of artifacts introduced by erroneous prospective feedback using de-corrected retrospective reconstruction. **a)** Baseline image without motion and PMC disabled. **b)** Corrupted images due to intentional squinting during a scan with minimal motion but PMC enabled. **c)** De-correction of the erroneous PMC scan removes artifacts due to false tracking signals. **d)** motion traces with isolated, sharp peaks due to squinting.

Figure 6.

Partial de-correction of a prospectively corrected scan that was corrupted by erroneous updates. **a**) Direct reconstruction of corrupted PMC data; **b**) complete removal of all PMC updates based on the motion trace in **d**; **c**) partial removal of erroneous squint data based on the motion trace in **e**.