Strategies to improve intratrain prospective motion correction for turbo spin-echo sequences with constant flip angles

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Purpose: To investigate the effects of prospective motion correction on turbo spin echo sequences and optimize motion correction approaches, mitigating signal dropout artifacts caused by the imperfections of motion tracking data.

Methods: Signal dropout artifacts caused by undesired phase deviations introduced by tracking errors are analyzed theoretically. To reduce the adverse effect of such deviations, two approaches are proposed: (1) freezing the correction for example, for even-numbered or higher number of echoes and (2) shifting the correction event prior to the left crusher gradient preceding the refocusing pulse. A comprehensive analysis is presented, including both signal simulations and experimental verifications in phantoms and in vivo. Performance of the proposed approach is validated in two healthy volunteers imaged under two types of motion conditions simulating inadvertent fast motions associated with discomfort and continuous large motions.

Results: The results show that the proposed optimization is able to efficiently correct for the motion artifacts and at the same time avoid signal dropout artifacts. Specifically, performing correction every 4th echo prior to the left crusher gradient was shown to improve image quality.

Conclusion: An optimization approach is proposed to exploit the potential of external tracking for intra-echo-train motion artifact correction for turbo spin echo sequences.

KEYWORDS
coherence history, motion tracking system, phase graph, prospective motion correction, turbo spin echo

1 | INTRODUCTION

Turbo spin echo (TSE) also known as rapid acquisition with relaxation enhancement (RARE)1 or fast spin echo (FSE) is one of the MR sequences most frequently used in clinical routine procedures. In the TSE sequence, a train of spin echoes is formed by a series of refocusing pulses following the spin excitation. TSE is well known for its reduced sensitivity to B₀ inhomogeneities and susceptibility artifacts in comparison with gradient refocused echo (GRE) sequences.2

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In brain MRI, TSE provides excellent image quality throughout the brain, also in the areas where GRE suffers from signal dropout artifacts. However, scan time for TSE varies between 2 and 5 min rendering it hard for subjects to remain sufficiently motionless because of physiological motion, discomfort, or pathology. Typical multislice clinical protocols employ TSE trains of 10—35 echoes with no dead times; therefore, subject motion is more likely to take place during an echo train. This is expected to disturb the signal consistency in k-space, which typically results in motion artifacts after image reconstruction.

In recent years, prospective motion correction (PMC) has achieved many promising results. PMC relies on external motion tracking to update parameters of the running sequence in real time. Although external tracking systems are able to reach an impressive tracking precision in 6 degrees of freedom (dfs), residual tracking noise cannot be neglected in some situations. In GRE imaging, the tracking noise may be sufficiently low, not to show an evident image-quality reduction. Also, for TSE with short echo trains, where a conventional correction prior to the spin excitation is sufficient, tracking noise of the state-of-the-art MRI-compatible motion-tracking devices does not introduce artifacts. Performing intratrain PMC may increase the correction rate from approximately 5 Hz (1/pulse repetition time [TR] per slice) up to the camera frame rate of approximately 60–85 Hz, which may be important for faster subject motion. However, our pilot experiments have shown that in a naive and straightforward intratrain PMC implementation, tracking errors may lead to visible signal drop and ghosting artifacts.

Further inconsistencies, such as the temporal delay caused by a system latency and systematic deviations caused by an insufficiently rigid marker fixation may affect tracking data integrity. In contrast to the tracking noise, these tracking errors do not affect the running sequence regularly, but only occur sporadically. Nonetheless, as shown for TSE below, a single spike-like error in the beginning of the echo train may disturb the rest of the k-space trajectory encoded by the train.

TSE is more sensitive to PMC imperfections because of the intrinsic properties of the sequence. According to the partitioning effect, a splitting of coherence pathways at every radiofrequency (RF) pulse leads to an exponential increase of the number of pathways with the number of RF pulses. TSE relies on the so-called Carr-Purcell-Meiboom-Gill (CPMG) condition, where all coherences add-up constructively along the main SE pathway. The symmetry of TSE is also exploited by the efficient signal calculation algorithms such as the extended-phase graph (EPG) to reduce memory requirements to a linear function of the numbers of echoes. If motion takes place during the echo train, it is likely to disturb the coherence pathways, resulting in artifacts. It appears reasonable to apply intraecho-train PMC to potentially reduce the signal deterioration. However, if motion correction is applied within the echo train, any inaccuracy of the tracking data may also influence the coherence of the subsequent echoes. Noise in rotational dfs in the motion-tracking data propagates in the form of undesired gradient moment deviations along the echo train, which affects the signal coherence, potentially leading to a signal intensity drop, even if imaging a stationary object. Additionally, for slice-selective RF pulses further artifacts might be brought about by the noise in the translational dfs in the through-plane direction because of the spin history effects. Based on the lever effect, the apparent shift induced by rotational tracking noise also increases with the distance to the tracking marker. Although the artifacts originating from the in-plane tracking noise in the translational dfs for TSE imaging can be expected to be similar to those in gradient echo imaging analyzed in our previous work, artifacts brought about by rotational noise have not been characterized yet.

In this work, we propose a sequence optimization approach to increase the motion correction efficiency and mitigate the signal drop artifacts caused by tracking errors in the process of applying PMC within the echo train. We mainly focus on the random tracking noise, and concisely limit ourselves to TSE with constant refocusing flip angles (FAs) to focus on the underlying spin physics and to simplify the analysis. For clarity reasons, only imaging results acquired with two-dimensional (2D) TSE are presented; however, similar effects were also found in our experiments within 3D TSE with constant refocusing FAs.

We first show the optimization approach based on simulated tracking noise in a stationary object. A comprehensive analysis is carried out, including both signal simulations and experimental verification in phantom and in vivo. Finally, the Moire phase tracking (MPT) system is employed for phantom and in vivo imaging. Two in vivo experiments with a deliberate fast small motion and a continuous large motion are performed to show the performance of the proposed PMC approach under various motion conditions.

2 | THEORY

In PMC the position and orientation of the imaged object is tracked in real time and the MR pulse sequence is updated to achieve the desired spatial encoding in the coordinate frame moving together with the object. However, tracking errors in rotational dfs, taking a Cartesian acquisition as an example, may destroy the parallel structure of the k-space trajectory by incorrectly rotating the readout gradient. If such an inappropriate rotation of the coordinate system takes place when the current encoding position along the k-space trajectory is far from the k-space center, a rotation even to a small angle effectively generates a substantial additional gradient moment. Although this has a negligible effect on the
frequency encoding direction, the accumulated phase in the phase-encoding direction for 2D TSE (and the through-slice direction for 3D TSE) might destroy the CPMG condition and cause the magnetization to depart from the desired coherence pathway in the subsequent echoes (see Figure 1). In the case of tracking noise, random gradient moments generated in each refocusing period clearly contradict the CPMG condition. We refer to such destructive pathways caused by tracking imperfections as coherence history effects.

2.1 | Coherence history effects

In a naive implementation the intratrain correction is applied immediately prior to each refocusing pulse (Figure 2, dashed line). The k-space position at this moment is far beyond the nominal k-space extent as typically a substantial crusher in the readout direction is applied prior to the next refocusing pulse. Any small errors in rotation in this case prevent a correct rephasing in the next echo interval, leading to the accumulated phase deviation to be represented as:

\[
\varphi (n) \approx \pm \varphi (n-1) + \delta \varphi (n),
\]

\[
\delta \varphi (n) = \theta (n) \times K \cdot P.
\]  

Here, \( \varphi (n) \) is the phase deviation at the \( n \)th echo, which is a sum of the previous phase deviation \( \varphi (n-1) \) and the new deviation \( \delta \varphi (n) \) generated; \( \varphi (1) = 0 \) per convention; \( n = 2, 3 \ldots N \) is the echo number. Equation (2) defines the new phase deviation \( \delta \varphi (n) \) caused by the tracking error in rotation \( \theta (n) \) at the \( n \)th echo for the voxel located at the position \( P; K \) is the k-space position when correction is applied.

The direction of the vector rotation \( \theta (n) \) corresponds to the rotation axis with its magnitude expressing the rotation angle in radians. All three parameters are 3D vectors with components along the readout, phase-encoding, and through-slice directions, respectively. When no slice crusher gradient is involved, as for example in 3D TSE, the cross-product symbol \( \times \) reveals the deviation \( \delta \varphi (n) \) generated in the phase-encoding or through-slice directions by a deflection of a large gradient moment accumulated in the readout direction to the moment of the coordinate update. Phase-encoding gradients do not contribute to \( K \) because they are balanced prior to each RF pulse. The \( \pm \) sign before \( \varphi (n-1) \) reflects that for some coherence pathways a positive sign is taken, whereas for refocused coherence pathways previously accumulated phase is inverted. The sign \( \approx \) indicates that the proposed treatment is provided as a qualitative illustration only. To arrive at quantitative results, a spatially resolved phase-graph analysis is required (see below). For TSE, for example, as shown in Figure 2, a pair of slice-crusher gradients is applied on both sides of each slice-selective refocusing RF pulse. These gradients, known as left and right crushers, serve to reduce the effects of slice-profile imperfections and suppress undesired free induction decay (FID) signals. The slice-crusher gradient moment contributes to the \( K \) vector when the echo train correction is applied (Figure 2, dashed line), and correspondingly also to the phase deviation \( \varphi (n) \).

The above-mentioned phase contributions violate the CPMG condition, causing the longitudinal and transverse magnetization to depart from the pseudo-steady state as the echo train proceeds. This loss of coherence typically manifests itself in a signal reduction. Because the phase deviation is proportional to the voxel position, signal intensity becomes spatially dependent. For small rotations, the phase deviation
\( \delta \varphi (n) \) is directly proportional to the rotational tracking error \( \theta (n) \). Because of the accumulation of phase errors, the additional signal attenuation becomes echo dependent. To mitigate the above artifacts, we aimed to suppress accumulation of the phase deviation \( \varphi (n) \) along the echo train. We optimized it using a combination of two strategies, as indicated below.

2.2 | Frame sharing

The first approach mainly targets reducing deteriorative effects of the tracking noise on the signal evolution. It is based on the assumption that if within the echo train more echoes maintain their relative phase relations, the coherence pathways remain partly preserved. Freezing correction for some cycles causes the MR data acquired during these echoes to share the same frame of reference. As the phase relations within the train temporary fulfill the CPMG condition the signal intensity is likely to be partly preserved. We term the approach, where subsequent echoes are acquired sharing the same reference frame position, frame sharing (FS). Especially for high-constant refocusing FAs (\( \sim 180^\circ \)), the low-order states (\( F_0, F_1, F_1 \) and \( Z_0, Z_{-1}, Z_1 \) in the EPG terminology) are populated mostly. Focusing on the pure spin-echo pathway, the phase deviation can be described as:

\[
\varphi (n) = - \varphi (n - 1) + \delta \varphi (n).
\]

Let us assume an example of erroneous tracking data consisting of a stationary pose stream with a single spike in the tracking data, where the original correct pose is restored immediately after the spike. As seen from the recurrent relation in Equation (3), the phase error introduced by the spike will be undone only for every second echo. All subsequent odd echoes will still be affected by this single spike. By freezing the correction for even-numbered echoes, the phase error will be perfectly restored for all subsequent echoes for this hypothetical example of a single spike.

For random tracking noise following a normal distribution with a zero mean and a given standard deviation (SD) \( \sigma \), according to Equation (1) the SD of the phase error accumulated at \( n \)th echo is proportional to \( \sigma \sqrt{n} \), meaning that the phase deviation grows along the echo train. Freezing the correction for even-numbered echoes enforces \( \delta \varphi (2m) = \delta \varphi (2m - 1) \) for any integer \( m \), which according to Equation (3) for the perfect refocusing pathway will cancel any introduced phase error in the following step.

Although FS effectively suppresses accumulation of phase errors along the echo train in case of random tracking errors, it also reduces our ability to correct for fast motions. Depending on the frame rate of the tracking system and the echo spacing time (ESP), for high FS factors the position update rate will be defined entirely by the FS, leading to an increased latency and consequently to a reduced correction efficiency.

2.3 | Minimal-moment minimal-moment motion correction

From Equation (2), we notice that \( \delta \varphi \) is proportional to the distance to the k-space center when correction is applied. In the naive implementation, intra-train PMC is executed before each refocusing pulse (Figure 2 dashed line). To reduce \( \delta \varphi \), another time point of correction needs to
be selected, minimizing the distance between the current k-space position and the k-space center. By placing the correction point before the left crusher gradient, $|K|$ would be reduced to only correspond to a half of the readout gradient moment (Figure 2, dotted line). In this case, the phase deviation generated by the readout crusher gradient is in the same frame of reference with the corresponding refocusing pair prior to the next readout. Additionally, for 2D TSE the $K_z$ component of the k-space vector at this time point still equals 0. In theory, placing the correction point after the right crusher gradient has the same effect, but would not correct for the slice position of the current refocusing pulse leading to a prolonged latency for through-slice translations.

A potential disadvantage is, however, that now also the phase-encoding moments in phase (and slice for 3D TSE) directions will contribute to the accumulated phase errors. Nonetheless, bearing in mind that for the center of k-space, which contributes the majority of signal and contrast, the phase-encoding moments are low, and an overall reduction of signal dropout is expected.

We term this approach a minimal-moment motion correction (MM-MOCO). Dotted lines in Figure 2 indicate the corresponding optimal positions for motion correction.

## 3 | METHODS

To prove the concept, we first simulate the k-space trajectory under different motion and tracking noise conditions. Thereafter, magnetization simulations and corresponding MR experiments in a small phantom without phase encoding are presented. Next, thorough MR experiments are performed in spherical phantom and in vivo, both with a simulated tracking noise and with a tracking system.

For all simulations and experiments, different motion correction strategies with FS1 through FS4, with and without MM-MOCO are compared; motion correction per excitation is taken as a reference. All MR measurements were performed on 3T Magnetom Prisma (Siemens Healthineers) with modified 3D/2D TSE sequences. The XPACE$^7$ platform was used to both inject pseudo-random tracking noise and perform real-time corrections. A root mean square error (RMSE) was used for assessing both the deviation of the sampling trajectory and image artifacts. All volunteers provided written informed consent according to the experimental protocol approved by the ethics committee of the University Medical Center Freiburg.

### 3.1 Proof of concept: K-space trajectory simulation

To visualize our theoretical assumptions, an open source framework pulse sequence programming environment$^{18}$ (https://pulseq.github.io/) was employed. Several k-space trajectories were calculated assuming the dominance of the spin-echo pathway and a perfect spoiling/dephasing of undesired pathways for a train of 15 sequentially encoded echoes. Initially, a single spike-like in-plane rotation (0.1°) was assumed (simulating subject motion followed by a slightly delayed correction), only affecting a single (the second) line in k-space. For FS mode (FS2) each pair of k-space lines was forced to share the tracking information (frame of reference), leading to two lines becoming affected by a single spike in tracking. Consequently, random in-plane rotations (SD, 0.1°) were simulated for imitating tracking noise with and without FS and with and without MM-MOCO.

### 3.2 Proof of concept: magnetization simulations and a small phantom MR experiment

For a thorough systematic analysis, simulations and MR measurements were performed with simulated Gaussian rotational noise (SD, 0.02° for each rotational df) with 3D TSE sequences with nonselective constant refocusing FAs, as shown in Table 1 (protocol A and B). The 3D TSE sequence was chosen for isolating the coherence history effect, while avoiding spin history effects caused by inhomogeneities of the slice profile.

| Protocol                  | FA        | FOV (mm) | Matrix | Position (mm) | ETL | PE |
|---------------------------|-----------|----------|--------|---------------|-----|----|
| 3D Simulation (A)         | 178°/150°/120° | 192      | 192    | 15-24         | 61  | No |
| 3D Small phantom (B)      | 178°/150°/120° | 192      | 192    | 15-24         | 61  | No |
| 2D Spherical phantom (C)  | 150°      | 220      | 320    | 16            | Yes |    |
| 2D In vivo (D)            | 150°      | 220      | 320    | 16            | Yes |    |
| 2D Spherical phantom (E)  | 180°      | 192      | 512    | 20            | Yes |    |
| 2D In vivo (F)            | 180°      | 192/220  | 448    | 20            | Yes |    |

**Table 1** Protocols for simulation and MR experiments for three-dimensional (3D) and two-dimensional (2D) turbo spin echo imaging.
As the conventional EPG is unable to consider magnetization dephasing by random gradients in 3D, a custom 3D-SRPG simulator was built based on the extension of the original work of Kiselev.\textsuperscript{19} The details of the 3D-SRPG simulator are presented in a separate article.\textsuperscript{20} The following parameters were assumed in the simulations: $T_1/T_2 = 400/200$ ms; position = 20 mm off-center; nominal refocusing FA = 178°/150°/120°. The perfect 180° refocusing would only affect the signal phase instead of the intensity. Therefore, FA = 178° was used for the high refocusing FA case. To accelerate 3D-SRPG calculations, a k-vector merging was used with the merging grid step of 1 rad/m.

Because the magnetization evolution is position-dependent, the phantom was placed about 20 mm off-center in line with the simulation settings. To compare simulations with MR experiments, a small phantom (height = 10 mm, radius = 2.5 cm) was scanned without phase encoding. A careful manual FA calibration was performed using the FID sequence.

### 3.3 MR measurement with simulated tracking noise: spherical phantom and in vivo

Spherical phantom and in vivo measurements (without motion) were performed with the same rotational noise realization as in protocols A and B with 2D TSE. During in vivo measurements, subjects were asked to keep as still as possible. The details of the protocol are given in Table 1 (protocols C and D).

### 3.4 MR measurements with position noise generated by the tracking system

As a source of the real tracking data affected by noise, we used a MPT system (Metria Innovation Inc). To compare the influence of different frame update frequencies of the tracking system, high-resolution spherical phantom imaging experiments (Table 1, protocol E) were performed in two tracking conditions: with the MPT system running at a 85-Hz and 60-Hz frame rate, respectively. The ESP was set to 13 ms (~77 Hz) to lie between the two frame rates.

### 3.5 In vivo MR measurements with intentional subject motion

Two healthy volunteers were imaged with high-resolution 2D TSE with the protocol listed in Table 1 (protocol F). The MPT system running at 85 Hz was used for motion tracking and ESP was extended to 15 ms to guarantee that new tracking information is available for each echo. Volunteers were asked to perform two types of motion: (1) to replicate inadvertent fast motions associated with discomfort by raising their leg abruptly periodically and (2) continuous large-amplitude motions by shaking their heads left to right. The position-locking feature of the XPACE library (intersequence motion correction) was used to align images from the subsequent runs to the same frame of reference.

### 4 RESULTS

#### 4.1 Proof of concept: k-space trajectory simulation

Figure 3 depicts the spin-echo k-space trajectories affected by the simulated motion and tracking noise for a single echo train, with the phase encoding ordered from bottom to top. The sampling trajectory for each neighboring readout line is equidistant, and all spoiling/dephasing moments are fully balanced when no motion occurs (Figure 3A, REF). Figure 3B shows the sequence affected by a single spike-like rotation right before the second echo, with the position returning to the original for the third and following echoes. As seen, a single rotation spike affects the entire k-space vastly, resulting in phase-encoding steps overlapping each other. In contrast to the previous case, Figure 3C presents the situation when the third echo shares the same motion-tracking information as the second echo (FS2). In this case, only these two echoes are influenced (the second and third lines from the bottom), whereas the following echoes are not affected. In Figure 3D and 3E, the effect of MM-MOCO is shown. As seen, almost no influence on the trajectory can be noticed when the same spike-like rotation is applied.

For the random noise-like rotations the periphery of k-space experiences massive phase deviations (see Figure 3F). Such deviations are somewhat reduced when applying the FS method (Figure 3G), and almost fully disappear when the MM-MOCO method is applied (Figure 3H). After combining these two methods, the spin-echo trajectory in Figure 3I manifests an even better resistance against random rotations in comparison with MM-MOCO only (RMSE = 3E-5 vs 6E-4).

#### 4.2 Proof of concept: magnetization simulations and small phantom MR experiments

Figure 4 shows a signal drop in a small phantom based on simulations (Figure 4A–F) and MR measurements (Figure 4G–L) with different FAs of the refocusing RF pulse using simulated rotation noise as position input. Notable signal drop mitigation is visible for high FS modes (FS2, FS3, and FS4) in comparison to FS1 in all cases. Signal intensity gain
of the MM-MOCO method is clearly visible in both simulations (Figure 4D-F in comparison with Figure 4A-C) and experiments (Figure 4J-L in comparison with Figure 4G-I).

For refocusing FAs of 178° and 150° (Figure 4A,B,D,E,G,H,J,K), signal intensity is better preserved in FS2 and FS4 in comparison with FS3. However, this superior performance is not observed for 120° refocusing FAs (Figure 4C,F,I,L). This indicates that for high FAs the spin-echo pathway dominates. For 120°, one expects substantial stimulated echo contributions, changing the parity of the coherence pathways. It may also be noted, that although the qualitative behavior of signal drop in simulations is comparable to measurement results, they do not match each other quantitatively (see Table 2). This is understandable because precise control of T1/T2 relaxation, B0/B1 inhomogeneity, phantom size, and shape and system imperfections would be required for a perfect match.

### 4.3 | MR measurements with simulated tracking noise: spherical phantom and in vivo

The results of 2D imaging in a spherical phantom and in vivo (protocol E and F) are shown in Figure 5. Additional artifacts presumably caused by the interference of the slice-selective RF pulses and the corresponding crusher gradients with the tracking noise have also been effectively suppressed by MM-MOCO (Figure 5F-I,P in comparison with Figure 5B-E,N), resulting in an RSME decrease by at least 40% for phantom images. Figure 5I and P corresponding to the FS4M

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**FIGURE 3** k-Space trajectories simulated with an open source framework pulse sequence programming environment: A, reference image. B–E, Single spike-like rotation with the corresponding respective methods. F–I, Random rotational motion with the respective correction methods. Presented correction methods are frame-share mode FS2, minimal-moment motion correction, and the combination of both methods. Zoomed-in graphs show the detail of the k-space periphery for each case to highlight the extent of phase deviations. Numbers in the bottom right corners indicate the root mean square error deviations of the analog-to-digital converter samples (red crosses) from their reference positions. TR, pulse repetition time; RF, radiofrequency.
Figure 4  A–F, Signal simulation (with protocol A in Table 1) and (G–L) small phantom experiments (with protocol B in Table 1). The simulation and experiments with simulated tracking noise are executed for different frame-share modes (FS), refocusing flip angles, and with and without minimal-moment motion correction (as a suffix M), respectively. The signal intensity of each condition is normalized to that under the motion-free condition. MM-MOCO, minimal-moment motion correction.
case manifest the best signal intensity and artifact-free images, both in vivo and in a phantom, which is reflected in an RSME reduction by 87.8% and 57%, respectively, in comparison with FS1. Arguably, for 150° refocusing FAs, FS3 outperforms FS2 or FS4 because of the increased influence of the stimulated echo pathway. Note that we were unable to accurately track the real movement of the subject based on a variety of reasons; nevertheless, the phantom and in vivo results show a good agreement.

4.4 MR measurements with tracking system noise: spherical phantom

The averaged SDs of noise in rotational dfs were 0.0064° and 0.0068° for MPT 85Hz and MPT 60Hz, respectively. Images of a spherical phantom acquired with the high-resolution 2D sequences (protocol E) are shown in Figure 6.

Artifacts in images acquire with MPT correction at 85 Hz (Figure 6B,C) are stronger than those at 60 Hz (Figure 6E,F, RSME improves by 65% and 57% in comparison with the corresponding reference (Figure 6A,D), respectively), although the tracking noise levels in these two cases are comparable. According to the results presented, FS mode FS4 with the minimum moment correction could effectively avoid artifacts caused by tracking noise (Figure 6C,F). The rate of the tracking information update is restricted by the camera frame rate, which for MPT 60 Hz (Figure 6F) is slower than the echo spacing in this experiment. The missing tracking data force the sequence to share the same tracking information occasionally and in this way form a FS mode spontaneously.

4.5 In vivo MR measurements with intentional subject motions

Figure 7 presents in vivo high-resolution imaging results in two subjects with the MPT tracking system running at 85 Hz. In the first experiment, the subjects were asked to move abruptly but periodically (Figure 7A-J), which resulted in a spike-like fast head motion (amplitude = ~10 mm and ~2°; speed up to ~80 mm/s and ~20°/s). In the second experiment, the subjects performed continuous large-amplitude head rotations (Figure 7K-T, amplitude = ~5 mm and ~15 to 20°; speed up to ~30 mm/s and 40°/s).

Executing motion correction per excitation shows a certain image-quality improvement (RSME reduction averaged over two subjects was 35% for small and 9% for large motion, respectively), in comparison with the experiment without correction. However, substantial motion artifacts prevail, especially for large continuous motions. Correcting each echo in the echo train (FS1), does not show a reliable image-quality increase compared with the per excitation case, both for sporadic and continuous motions. Although all correction approaches show residual artifacts for the fast motion case, presumably because of the speed of the performed of motion, the FS4M mode shows the clearest image-quality improvement judging by visual inspection. This is also in line with the RMSE values for all subjects and motion cases, except for subject 2 under the small motion condition. Interestingly, that despite of the RSME increase by 11% in comparison with FS1, the image in this case clearly shows the best subjective quality of correction, which underlines the limitations of RMSE as a quality measure. In this particular case, it could have failed for example, because of changes in the coil sensitivity distribution or residual through-slice motion despite of the position-locking feature.

5 DISCUSSION

Correcting for motion artifacts in echo-train sequences prospectively is particularly challenging because the tracking information is often affected by the residual tracking errors. Such tracking errors can arise for different reasons, such as intrinsic tracking system noise, out-of-date tracking information caused by latency, incorrect tracking information from insufficiently rigid marker fixation, and cross-calibration errors. Applied within echo-train, such tracking errors affect coherence pathways by inducing undesired phase deviations.

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**TABLE 2** The corresponding averaged signal intensity along echo train of protocol A under different conditions

| FA  | FS1  | FS2  | FS3  | FS4  | FS1M | FS2M | FS3M | FS4M |
|-----|------|------|------|------|------|------|------|------|
| Simulation | 178° | 0.933 | 0.999 | 0.976 | 0.999 | 0.979 | 1.000 | 0.992 | 1.000 |
|       | 150° | 0.929 | 0.994 | 0.981 | 0.993 | 0.979 | 0.998 | 0.994 | 0.997 |
|       | 120° | 0.950 | 0.985 | 0.991 | 0.984 | 0.986 | 0.996 | 0.997 | 0.995 |
| Experiment | 178° | 0.899 | 0.996 | 0.976 | 0.992 | 0.954 | 0.994 | 0.985 | 0.991 |
|       | 150° | 0.931 | 0.991 | 0.978 | 0.986 | 0.966 | 0.993 | 0.986 | 0.990 |
|       | 120° | 0.949 | 0.984 | 0.983 | 0.971 | 0.988 | 0.990 | 0.990 | 0.986 |

FS<n> represents different frame-share mode; FS<n>M indicates that minimal-moment motion correction is applied additionally.

Abbreviation: FA, flip angle.
FIGURE 5  A–I, Two-dimensional spherical phantom (with protocol C in Table 1) and (J–P) in vivo experiment (with protocol D in Table 1) with simulated tracking noise. B–E,K–N, Every single image indicates a different frame-share mode labeled as FS<\textit{n}>. F–I,O,P, The label FS<\textit{n}>M indicates that the minimal-moment motion correction was applied additionally. For each condition, the root mean square error with respect to the reference image is presented in the right bottom corner, and the red arrows point to the strongest artifacts.
because of additional gradient moments resulting in signal intensity fluctuations in the spatial domain.

To mitigate artifacts and signal dropouts caused by the abovementioned coherence history effect, an optimization approach is proposed for reducing phase error accumulation. According to the phase deviation equation for the high refocusing pulse TSE sequence (Equation 3), phase-error reduction can be achieved in two ways: (1) sharing the same tracking information with neighboring echoes (every second echo for perfect 180° refocusing pulses) to restore odd/even echo-refocusing mechanism, and (2) shifting the correction time point to a location in the sequence, where the correction is technically feasible and the current k-space position is characterized by the minimal distance to k-space center. The FS strategy is closely related to the even echo-rephrasing mechanism, well known for CPMG echo trains in the presence of flow.21 The presented combined optimization approach has been used comprehensively in both signal simulations and in experiments with phantoms and in vivo, with a simulated tracking noise and real tracking system, respectively. The efficiency of the proposed approach has been tested in vivo under motion conditions replicating inadvertent fast motions associated with discomfort and continuous large-amplitude motions.

Upon a decrease of the refocusing FA, the balance between the low- and high-order dephasing states in the phase graph shifts towards the higher orders, which progressively wipes the differences between the odd and the even FS modes (see Figure 4). Although a dramatic recovery of coherence as with 180° refocusing FA is unachievable, FS seems to introduce temporary pseudo-steady-state–like conditions. At the same time, it allows for the reduction of the coherence history effects leading to the propagation of the phase errors along the entire echo train. Both these phenomena contribute to a partial recovery of the magnetization. However, an optimized FS factor is difficult to find experimentally for short ESPs caused by the phenomenon of incidental echo sharing because of the absence of fresh motion-tracking data at subsequent refocusing periods. For a slower frame-rate tracking, more echoes incidentally share the same tracking information; therefore, the situation realized in the experiment does not match the specified FS factor. This phenomenon explains a better image quality in absence of true motion when slower tracking was used (see Figure 6). Instead of modifying the MR sequence, one could use the undersampling option of the MPT system to reduce the effective frame rate and count on the incidental FS. However, in this case the increased effective latency.
FIGURE 7  Images acquired in the two-dimensional high-resolution in vivo experiment (protocol F in Table 1) in two subjects (S1 and S2) while replicating the fast small motions (A–J) and continuous large motions (K–T). Each subject was measured five times: No motion no correction as the reference, motion but no correction labeled as NMC, motion with correction per echo train labeled as the Per Ext (per excitation), executing motion correction within the echo-train for each echo, labeled as FS1, and executing correction with the proposed method, labeled as FS4M. For each condition, the root mean square error value with respect to the reference image is presented on the right bottom corner.
of tracking would also affect the excitation pulse, potentially increasing spin-history effects. Also the experiment with the true large motion underlines our finding that performing motion correction of every echo does not result in a superior image quality compared with correction per excitation (Figure 7). As the tracking noise in this case is certainly weaker than the motion amplitude, it is likely that the coherence history effects are responsible for the poor performance of FS1 as suggested in Figure 3. We hypothesize that a combination of the MPT system latency and the fast motion changing directions produces spike-like residual motions in the object frame of reference that influences the CPMG condition and creates persistent artifacts. Additional investigations of these phenomena are needed under carefully controlled reproducible motion, for example, using an MR-compatible programmable motion phantom.

Depending on the selected reordering scheme (e.g., to achieve a desired contrast), the performance of the proposed method may vary. Although additional phase errors contributed from phase and partition-encoding gradients in MM-MOCO could be ignored in our experiments, they may play a role for certain MR protocols and motion types. Sharing more frames (above four, not shown) of tracking information would further reduce the signal drop caused by tracking noise, especially for complex coherence pathways (e.g., such that are created by refocusing FA variations along the echo train). However, it is also expected to reduce the sequence ability to adapt to fast motions. Given the complex interdependencies of FS and the TSE set-up (FA amplitude, variable FA schemes, reordering), because of its simplicity the MM-MOCO method should be employed as a standard solution in practice.

Alternatively, tracking data could be filtered prior to the application of the pulse-sequence updates to reduce the noise level at the cost of increasing the effective latency. A comprehensive comparison of the proposed method with tracking data filtering as an alternative is worth exploring in the future. However, full control and reproducibility of motion, as well as careful design of the experiment will be required to arrive at conclusive results. The influence of high FS settings on the image quality under realistic in vivo conditions is also a topic of future research. In general, we expect the FA variation during the TSE echo train to affect the sensitivity of the sequence to the residual tracking imperfections caused by the complex magnetization storage and recovery mechanisms. Prospective motion correction of echo train sequences with FA variation is a topic of ongoing studies in our lab.

In this study, we evaluate the motion-correction performance for a constant refocusing FA echo-train sequence taking tracking errors into account. The proposed method is, however, not restricted to the external tracking and could also be combined with other advanced PMC methods.

6 | CONCLUSION

An optimization approach is presented to reduce signal drop and image artifacts resulting from tracking errors in motion-corrected TSE sequence with the echo-train correction. The approach is verified exhaustively by simulations and in phantom and in vivo experiments.

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DATA AVAILABILITY STATEMENT

The code that supports the findings of this study is openly available in GitHub at https://github.com/xiang-G/3D-Spatially-Resolved-Phase-Graph.

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