Pilot tone–based motion correction for prospective respiratory compensated cardiac cine MRI

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Purpose: To evaluate prospective motion correction using the pilot tone (PT) as a quantitative respiratory motion signal with high temporal resolution for cardiac cine images during free breathing.

Methods: Before cine data acquisition, a short prescan was performed, calibrating the PT to the respiratory-induced heart motion using respiratory-resolved real-time images. The calibrated PT was then applied for nearly real-time prospective motion correction of cine MRI through slice tracking (ie, updating the slice position before every readout). Additionally, in-plane motion correction was performed retrospectively also based on the calibrated PT data. The proposed method was evaluated in a moving phantom and 10 healthy volunteers.

Results: The PT showed very good correlation to the phantom motion. In volunteer studies using a long-term scan over 7.96 ± 1.40 min, the mean absolute error between registered and predicted motion from the PT was 1.44 ± 0.46 mm in head-feet and 0.46 ± 0.07 mm in anterior-posterior direction. Irregular breathing could also be corrected well with the PT. The PT motion correction leads to a significant improvement of contrast-to-noise ratio by 68% ($P \leq .01$) between blood pool and myocardium and sharpness of endocardium by 24% ($P = .04$) in comparison to uncorrected data. The image score, which refers to the cine image quality, has improved with the utilization of the proposed PT motion correction.

Conclusion: The proposed approach provides respiratory motion-corrected cine images of the heart with improved image quality and a high scan efficiency using the PT. The PT is independent of the MR acquisition, making this a very flexible motion-correction approach.

KEYWORDS
cardiovascular MR, cine imaging, pilot tone, prospective respiratory motion correction
1 | INTRODUCTION

Cardiac cine MRI is a commonly used tool to assess cardiac function in clinical practice. The excellent soft-tissue contrast allows for the evaluation of cardiac function and calculation of diagnostic parameters such as ejection fraction. One major challenge for cine MRI is to minimize artifacts due to respiratory motion, which corrupt image quality, and therefore diagnostic accuracy. The preferred imaging strategy is to avoid respiratory motion with a breath-hold, in which often only a single 2D cine slice can be acquired. However, for a full cardiac examination, this has to be repeated multiple times. This procedure requires dedicated operator involvement and high patient cooperation, which is often not feasible for patients who have difficulty holding their breath. Real-time imaging has been proposed to allow for cine imaging without breath-holding, but often requires dedicated sequences and advanced reconstruction schemes and limits the achievable temporal resolution. In addition, respiratory-motion artifacts might still be present.

Another method to minimize image artifacts is respiratory gating. Here, an image-based navigator is used to determine the respiratory-motion state of the heart, and only data acquired in a predefined breathing state are used for image reconstruction. This can also be combined with prospective motion correction (slice tracking), in which navigator data are used to update the slice position before every spin excitation. Commonly, to avoid disturbing the magnetization in the heart region, the navigator monitors the position of the liver rather than the heart itself. This reduces the accuracy of slice tracking. The disadvantage of both gating and slice tracking is that the acquisition of diagnostic image information has to be interrupted for the acquisition of the navigator, and not all data are used for the final image reconstruction, leading to lower scan efficiencies and hence long scan times. Retrospective motion correction for 2D cine MRI has also been proposed. Nevertheless, this approach can only correct for in-plane motion and cannot recover artifacts caused by through-plane motion requiring also additional gating.

To overcome the problem of interrupting the MR data acquisition, methods have been proposed that use external navigators, such as thermal noise variances, respiratory bellows, or the pilot tone (PT). So far, these techniques have only been used qualitatively for gating, but not quantitatively for prospective motion correction.

Here, we present a novel respiratory motion-correction approach (ie, PT-MOCO) that uses the PT to perform prospective slice tracking for cine MRI followed by in-plane k-space-based motion correction. A short (≈ 60-second) calibration scan is used before the cine acquisition to calibrate the PT to the actual respiratory-induced heart motion. The obtained patient-specific motion model is then used during the cine acquisition for motion correction. Our new technique allows for continuous data acquisition and an effective use of scan time, yielding cine images without the need of breath-holding. A feasibility study of the method was conducted in a motion phantom and in 10 healthy volunteers.

2 | METHODS

In this paper, we propose a respiratory motion-correction method for free-breathing cardiac cine MRI that uses the PT (ie, a continuous-wave RF signal as a surrogate motion signal). A conceptual overview of our method is shown in Figure 1. In a calibration phase, 2D images of the moving heart are simultaneously acquired with the PT, and a motion model, linking the intensity changes of PT to the motion of the heart, is derived. In a correction phase, through-plane motion is compensated for by performing nearly real-time prospective slice tracking during the running MR cine sequence. In-plane motion is corrected retrospectively by applying corresponding phase shifts to k-space data before image reconstruction. Our approach was compared with the breath-hold technique as a ground truth. Examples of the intermediate steps of our method are shown in Figure 2.

2.1 | Pilot tone

The PT is a continuous RF signal that is sent into the bore of the MR scanner. The frequency of the PT is set, such that it can be recorded in the two-fold oversampled region of the FOV. Therefore, the PT is obtained simultaneously with each readout line. To minimize the potential influence of the pilot tone on the image data, a modeled PT is subtracted from the k-space data. Figure 2A illustrates the pilot tone detection and subtraction from the k-space data. In Figure 2B, three exemplary PT signals of an in vivo scan are shown.

Patient motion (ie, due to breathing or the heartbeat) leads to a coil-dependent variation of the signal amplitude. The underlying principle is based on the different wave impedances due to different loading conditions of the receiver coils. Motion causes changes in local coil load and in the coupling between transmitter and receiver. The PT signal amplitude is approximately proportional to the motion amplitude, yet unitless; therefore, the PT is a useful tool to monitor patient motion.

2.2 | Calibration

In a first step, a calibration scan is performed to determine a linear motion model that describes the correlation of the PT and the heart motion. Previous studies have shown that respiratory motion amplitudes are strongest in head–feet (HF)
and anterior–posterior (AP) directions. Therefore, a series of sagittal images is acquired together with the PT, capturing several breathing cycles. The region of interest, covering the heart, is chosen manually before the calibration scan. Image registration with a normalized 2D cross-correlation function is used to determine the translation of the heart in HF (ΔHF_reg) and AP (ΔAP_reg) directions. An exemplary cardiac shift in HF is shown in Figure 2C. Linear motion models are then derived between ΔHF_reg, ΔAP_reg, and the PT for each coil separately:

\[
\Delta HF_{\text{reg}} = a \times PT + b \quad (1) \\
\Delta AP_{\text{reg}} = m \times PT + n \quad (2)
\]

For the motion correction, the coil that yields the motion model with the highest coefficient of determination, \( R^2 \), for \( \Delta HF_{\text{reg}} \) is used. As an example, Figure 2D shows motion models for different coils resulting from a calibration scan. The model parameters \( a, b, m, n \), and the coil number are stored for the following scans.

### 2.3 Prospective through-plane correction

The second step is the motion correction during the cine scan. Using the model parameters and the PT, the respiratory motion of the heart (ΔHF_pred, ΔAP_pred) is predicted for every readout line. This motion is separated into through-plane and in-plane components, as illustrated in Figure 2E. Through-plane motion correction is then applied prospectively during measurements by adapting the frequency of the RF pulse, to ensure that the excited slice follows the motion of the heart (ie, slice tracking). The change in slice position \( \Delta \vec{SL} \), which is applied during data acquisition, can be described as the orthogonal projection of the predicted shifts \( \Delta HF_{\text{pred}} \) and \( \Delta AP_{\text{pred}} \) onto the slice normal \( \vec{SN} \) of the scan orientation:

\[
\Delta \vec{SL} = P_{\vec{SN}} \left( \vec{M} \right),
\]

where \( \vec{M} \) is the motion vector:

\[
\begin{bmatrix}
0 \\
\Delta AP_{\text{pred}} \\
\Delta HF_{\text{pred}}
\end{bmatrix}
\]

The value of \( \Delta \vec{SL} \) is then used during the scan by adapting the RF pulse, accordingly. This ensures that the excitation follows the respiratory motion of the heart for each TR.

### 2.4 Retrospective in-plane correction

In-plane motion is corrected for by adapting the phase of the acquired k-space data before image reconstruction. The phase correction is applied by multiplication of the correction values with the k-space data:

\[
\Delta PE = P_{PE} \left( \vec{M} \right) \quad (4)
\]

\[
\Delta RO = P_{RO} \left( \vec{M} \right), \quad (5)
\]
(A) Pilot tone detection and subtraction

![Diagram showing pilot tone detection and subtraction](image)

(B) Extracted pilot tone from 3 coils (out of 32)

![Graphs showing extracted pilot tone from 3 coils](image)

(C) Obtained respiratory induced heart motion in HF direction

![Graph showing respiratory induced heart motion](image)

(D) Derived motion models from 3 coils (out of 32)

![Graphs showing derived motion models from 3 coils](image)

(E) Calculation of real-time and retrospective correction shifts

![Graphs showing calculation of real-time and retrospective correction shifts](image)
where \( \overrightarrow{PE} \) and \( \overrightarrow{RO} \) are the slice orientation vectors of the readout and phase-encoding directions.

### 2.5 | Experiments

All experiments were carried out on a Siemens 3T scanner (MAGNETOM Verio; Siemens Healthcare, Erlangen, Germany). The local ethics board approved the in vivo experiments, which were performed on 12 volunteers (7 male, 5 female, age = 38 ± 15 years, weight = 72 ± 16 kg). Experiments on long-term stability and motion correction were each carried out on 10 of the subjects from the cohort. The parts of image reconstruction and evaluation that were required for the application of the prospective correction were implemented on the scanner, customizing the reconstruction chain (software syngo.MR B17). Application of correction values to k-space data and the image visualization were carried out using MATLAB 2017a (The MathWorks, Natick, MA).

#### 2.5.1 | Pilot tone generation

The PT was produced by a prototype setup similar to previous applications for respiratory gating and cardiac triggering. A commercial RF synthesizer (ESG 1000A; Hewlett Packard, Palo Alto, CA) was connected to a nonresonant 7-cm surface coil. Due to the vibrations of the scanner cover, it was practical to place the emitting coil on an external table within 20 cm distance of the head end of the bore. The frequency of the continuous-wave RF signal was set 70 kHz higher than the center frequency of the scanner. With an image bandwidth of 86.2 kHz, corresponding to a FOV of 32 cm, the PT was placed 25.8 cm away from the image center in the head direction, such that it was received in the two-fold oversampled readout signal of the MR data and outside the frequency band of the desired FOV.

#### 2.5.2 | Pilot tone acquisition

The pilot tone is recorded in the oversampled region of the readout of the Fourier-transformed k-space data, as depicted in Figures 2A and 3. For each k-space line and coil, a complex signal is obtained at the readout position that corresponds to the PT frequency. Before image reconstruction, the PT is removed from the data by fitting the signal to a model \( A \ast \exp(i2\pi ft) \) and then subtracting it from the k-space data, as depicted in Figure 2A. The complex amplitude \( A \) is logged as the PT, as described by Speier et al. To ensure greater temporal stability, the PT was median-filtered in steps of 100 data points with a sliding window.

#### 2.5.3 | Phantom setup and experiments

For the phantom experiments, a 1.2-L agarose phantom containing cubic structures with side lengths of 0.7 cm was placed onto a wagon. To imitate free breathing, the wagon performed a translational sinusoidal-like movement along the HF direction of the scanner with an amplitude of 3.2 cm and a frequency of 6 repetitions per minute (0.1 Hz). An electrocardiogram signal was simulated within the scanner software with 1 Hz and was used for triggering. The phantom setup is displayed in Supporting Information Figure S1.

The calibration scans and the dynamic scans (ie, scans with one image per RR cycle) as well as the assessment of the temporal stability were performed on the phantom with the same scan parameters as for the in vivo scans.
2.5.4 | Calibration scan

For calibration, 2D electrocardiogram-triggered dynamic data were acquired over 60 cardiac cycles in sagittal view with FOV = 320 × 320 mm², voxel size = 2.1 × 1.7 × 8 mm³, TE/TR = 3.22/5.7 ms, pixel bandwidth = 449 Hz/pixel, and flip angle = 12º during free-breathing using an in-house-modified gradient-echo sequence and two-fold parallel imaging acceleration with 24 reference lines. Each single dynamic image was acquired in an acquisition window of 507 ms in end diastole.

2.5.5 | Dynamic scan

To assess the performance of the proposed method, dynamic measurements with the same parameters as for the calibration scan were acquired. Twenty images were captured, with motion correction and without motion correction.

2.5.6 | Cine scan

For the 2D cine scans, the slice orientation was changed to a short-axis view. Depending on the volunteer’s heart rate, 28-30 cardiac phases were captured within 15-25 seconds using the same FOV, voxel size, flip angle, and TE as for the calibration scan and a TR of 5.9 ms. Data were acquired with retrospective electrocardiogram gating and two-fold parallel imaging acceleration with 24 reference lines. In 1 volunteer, a full stack of 10 short-axis-view images and additional long axis, 4-chamber view and transversal cine scans were acquired.

For the evaluation of the performance of our proposed approach, the cine scans were acquired during free-breathing, with and without slice tracking, and for reference purposes during one breath-hold per slice. Cine images were reconstructed using SENSE without temporal or spatial filtering of the data.

2.5.7 | Temporal stability

The temporal stability was assessed in 10 healthy volunteers who were asked to perform normal continuous breathing. Additionally, 1 volunteer was asked to breathe irregularly. A dynamic scan was performed with the same parameters as the calibration scan but with 513 RR cycles.

To evaluate the temporal stability, the first 60 acquisitions were used for calibration (ie, the derivation of the motion model). With the model and the PT, the heart motion was then predicted (ΔHF\textsubscript{pred}) and compared with the registered heart motion. The motion model was not updated after initial calibration.

The mean absolute error \( (MAE) \) was calculated by subtracting the predicted shift from the registered heart shift:

\[
MAE = |\Delta HF_{\text{pred}} - \Delta HF_{\text{reg}}|.
\]

For analysis, the values were averaged over 10 scan repetitions. For the in vivo data, the same approach was also used with \( |\Delta AP_{\text{pred}} - \Delta AP_{\text{reg}}| \).

2.5.8 | Analysis

Analyses were carried out regarding the contrast-to-noise ratio (CNR) between the myocardium and the blood pool. For comparison, images from diastole without motion correction, with PT-MOCO, and of a breath-hold scan were used. To determine the CNR with regard to the motion artifacts, areas of the septum and the blood pool were selected manually. In addition, the sharpness of the endocardium along the septum was also determined similar to previous work for coronary arteries and abdominal imaging. The highest sharpness value is 1, describing the edge of a heavy side step function.

To assess cine image quality, the images were reviewed and evaluated by two independent observers with more than 5 years of experience in cardiac MRI in a randomized blinded reading session. For this purpose, the overall image quality was evaluated on a scale of 0 to 3 for each slice. The following scale was used for blinded reading: 0, images with poor and nondiagnostic quality due to motion-induced blurring; 1, image quality impaired by motion that may lead to misdiagnosis; 2, good image quality, motion artifacts hard to recognize; and 3, excellent image quality, no motion artifacts observed.

3 | RESULTS

3.1 | Phantom experiments

3.1.1 | Evaluation of temporal stability

Due to the setup of the phantom, only motion along HF was present; hence, the analysis was limited to that direction. Figure 4A shows the shift found by image registration (ΔHF\textsubscript{reg}) compared with the predicted shift (ΔHF\textsubscript{pred}) estimated from the model and the PT. The linear regression correlation coefficient was \( R^2 = 0.95 \). The amplitude of the registered shift was 31.7 mm, and for the predicted shift 31.7 ± 3.7 mm. Figure 4B shows the evolution of the MAE. The average MAE over the complete measurement was 2.5 ± 0.7 mm.
3.1.2 | Motion correction

Dynamic images of the moving phantom, with and without slice tracking, are shown in Figure 5 for the different scan orientations, sagittal and transversal. To visualize the motion of the structures during the entire measurement, all 20 images were summed up. Without motion correction, the image content changes during data acquisition due to the motion of the phantom (Figure 5, top row). With motion correction, the acquired slice follows the motion of the phantom, leading to the same image content during measurement. Motion correction improves the depiction of the same underlying structure compared with the uncorrected scan, leading to reduced blurring in the summed images.
3.2 | In vivo experiment

3.2.1 | Evaluation of temporal stability

The respiratory-induced heart motion for regular breathing was evaluated for 10 volunteers in HF and AP directions from sagittal images, and the corresponding MAEs are shown in Figure 6. The mean $R^2$ of the registered and the predicted shift for the first 60 RR cycles were $0.94 \pm 0.04$ in HF and $0.67 \pm 0.19$ in AP direction. The measurement duration of 513 RR cycles varied between 6.5 and 11.1 minutes. The average peak-to-peak difference along HF over all volunteers was $5.7 \pm 3.0$ mm, with a maximum heart displacement of 28.3 mm. The average of the MAE along HF was $1.4 \pm 0.5$ mm. For AP, the average peak-to-peak difference was $1.8 \pm 0.5$ mm for all measurements, with a maximum heart shift of 6.7 mm. The average MAE along AP was $0.5 \pm 0.1$ mm.

Analysis of the relative error per peak showed that there is no correlation between respiratory amplitude and the deviation between prediction and true motion.

Figure 7A shows the respiration curve in HF direction of a volunteer who was asked to breathe irregularly for 6.9 minutes. An approximately 10-second breath-hold after end-exhalation was also included in the calibration phase. The MAE of the predicted shift ($\Delta HF_{pred}$) compared with the registered shift ($\Delta HF_{reg}$) was less than 1 pixel for almost the entire measurement. Only very deep breathing led to an increase of the MAE, because the deep breathing was not part of the calibration, as shown in Figure 7B (arrows). After the deep-breathing phase, the error decreases again.

3.2.2 | Motion correction

Dynamic MR images, acquired with the electrocardiogram triggering, without motion correction, and with PT-MOCO, are depicted in Figure 8. Images were selected from the image stream, such that the depicted motion alternates between end-inhalation and end-expiration, identified based on the PT. Anatomical features such as the papillary muscles (red arrows), the pericardium (green arrows), and the liver (yellow arrows) change during the measurement, as the tissue moves through the slice. In-plane displacement of the left ventricle can also be clearly seen (blue arrows) and was up to 11 mm. The PT-MOCO accurately corrects for these motion displacements.

3.2.3 | Cine

Motion estimation succeeded for all volunteers. Figure 9 shows the scale-free pilot tone and the motion shifts calculated with the motion model along the slice direction (Figure 9B) and in plane (Figure 9C).

Our motion-correction approach resulted in improved cine image quality for all volunteers relative to
Figure 7  A. Registered cardiac shift and predicted shift of a volunteer with irregular breathing for 513 RR cycles. B. Temporal stability of the motion model. The first 60 RR cycles were used to calibrate the PT to the respiratory-induced heart motion. The heart motion was then predicted and compared with the registered heart motion. The MAEs in steps of 10 RR cycles are displayed.

Figure 8  Free-breathing, 2D dynamic scans with 20 repetitions without (A) and with (B) PT-MOCO. The MR images from five different respiratory motion states are depicted, with numbers indicating the repetition number. The images were chosen to alternately show end-inspiration (in) and end-expiration (ex). A, Changes of the anatomy due to respiratory through-plane motion are indicated with arrows. A reference for in-plane motion is given by the horizontal blue line. The motion of the heart (through-plane and in-plane) was minimized with the slice-tracking method, as can be seen by comparing the regions where the arrows point at and the heart position with reference to the blue line in (A) and (B).

The uncorrected images. Figure 10 shows end-systole and mid-diastole of 4 subjects acquired during free-breathing without correction, and with correction using PT-MOCO. For comparison, data were also obtained during a single breath-hold. The corresponding cine images are shown in Supporting Information Video S1. It is important to note that all three scans were acquired as separate scans, and can therefore be at slightly different scan positions. In addition, Supporting Information Video S2 displays different scan orientations (long axis, four-chamber view, short axis, and transversal), and Supporting Information Video S3 shows a stack of 10 slices in short-axis view on 1 volunteer.
Supporting Information Figure S2 shows the calibration curves for all 10 volunteers, also distinguishing between inspiration and expiration.

The CNR was found to be $7.6 \pm 2.7$ with regard to motion artifacts for the uncorrected diastolic images, $12.8 \pm 4.0$ for PT-MOCO, and $13.8 \pm 4.9$ for the breath-held data. Significant improvements in CNR by 68% ($P \leq .01$) were found between the uncorrected data and PT-MOCO, and by 80% ($P \leq .01$) between the uncorrected data and the breath-held data. The difference in PT-MOCO compared with the breath-held was not significant ($P = .9$).

The sharpness of the endocardium was $0.13 \pm 0.04$ for the uncorrected images, $0.16 \pm 0.03$ for the PT-MOCO images, and $0.19 \pm 0.03$ for the breath-held images. The sharpness improved significantly by 24% ($P = .04$) between the uncorrected data and PT-MOCO. For the uncorrected data compared with the breath-held data, the sharpness improved significantly by 47% ($P \leq .01$), whereas for PT-MOCO

FIGURE 9  A. Median filtered PT acquired for each phase-encoding line (ie, each TR). Using the motion model from the calibration, correction values for the real-time through-plane correction (B) and the retrospective in-plane correction (along RO and PE) (C) can be calculated from the PT

FIGURE 10  Comparison of cine MRI acquired during free breathing without (uncorrected) and with PT-MOCO. A standard breath-hold cine scan is also shown for reference. For each volunteer, end-systolic and mid-diastolic phases are shown. It is important to note that each method is acquired in a separate scan that can also lead to small differences in the visualized anatomy
compared with the breath-held data, the differences were not significant at −19% (P = .054).

The image score for the uncorrected data was 0.3 ± 0.6, which was increased to 1.4 ± 0.7 using the proposed PT-MOCO (P ≤ .01). The difference in PT-MOCO compared with the breath-hold cines, with an image score of 2.6 ± 0.6, was also significant (P ≤ .01).

4 | DISCUSSION

This paper demonstrates the feasibility of PT-based prospective motion correction for free-breathing cardiac cine MRI. It was shown on a phantom and 10 healthy volunteers that our approach led to accurate motion correction. With the proposed approach, respiratory-motion artifacts could be reduced.

By analyzing the temporal stability, it was found that there is a slight increase of the MAE by approximately 1.6 mm in the HF direction over a period of 513 RR cycles, corresponding to 6% of the maximum motion amplitude and in the order of image resolution. A possible reason for this increase could be that the system warmed up slightly, leading to a change in coil loads, and therefore PT intensity. Despite this small increase, the temporal stability of our approach was still very good, allowing for one calibration to be used for multiple 2D free-breathing scans. Therefore, the additional scan time spent on calibration using PT-MOCO can be kept small compared with the total scan time. After 350 to 400 RR cycles, the MAE becomes higher than the in-plane resolution of the cine scans. This would require recalibration of the model to ensure accurate motion correction after that.

So far, during calibration a single coil was selected based on the linear regression correlation coefficient R². Bulk patient motion during the scan (ie, shifts of the entire body) could lead to a displacement of the receiver coils and amplitude changes of the PT, leading to errors in the motion prediction. The PT appeared very sensitive to a change in distance of the receiver coils and transmitter coil. Combining the information from all receiver coils using, for example, a principal component analysis, could be less sensitive to these influences and could lead to a more robust PT. In general, combining the PT signal from multiple receiver coils could improve the proposed slice tracking approach. Another option to overcome this problem could be to integrate the PT signal from multiple receiver coils could work even better for trajectories that sample the k-space center multiple times (ie, radial sequences), because these sequences are more robust against motion artifacts; therefore, wrongly corrected tissues would produce fewer artifacts.

For the approximation of the respiratory motion of the left ventricle, the patient-specific linear model proved to be useful. Supporting Information Figure S2 shows that the calibration worked well (lowest R² values is 0.8), and there are only small differences (ie, hysteresis effects) between inspiration and expiration. Nevertheless, errors occurred for motion that was not captured during the calibration phase (see deep breathing in Figure 7B). More advanced models such as affine motion models, or taking into account hysteresis effects between inspiration and expiration, could improve the prediction of respiratory motion and image quality.

Currently, we were using the same motion model for each slice position. The registration is carried out for the entire left ventricle, so the motion model fits very well for a midventricular region, but is less accurate for apical regions, which can be seen in Supporting Information Videos S2 and S3. The further the scan position deviates from the original calibration plane, the greater the errors of the correction, as the motion model no longer fits there. For example, in Supporting Information Video S2, for the long-axis orientation the motion correction in the apex works less accurate than for the midventricular region, for which the PT was calibrated. In addition, slices 1 and 10 in Supporting Information Video S3 displaying the apex and the base of the ventricle, respectively, show more motion artifacts than the slices covering the midventricular region. Affine motion models would allow to adapt the motion model better to the current slice position and overcome this problem.

A median filter with a width of 100 readout lines was applied to the PT to reduce noise in the signal. Therefore, the PT of the first 100 readout lines may not be as accurate as the following signal. However, this did not have any visible effect on the cine images, as, at most, seven readout lines per cardiac phase are affected. In addition, the filter leads to a temporal smoothing of the motion signal and hence a short delay between the estimated and true motion of the heart. Although this delay is short compared with the length of a respiratory cycle, other filters such as Kalman filters, as proposed in Spincemaille et al35 and already used with the PT for prospective cardiac triggering,36 could overcome this problem.

Depending on the subject and the slice orientation, the amplitude of the through-plane motion varies. The motion model was optimized for the left ventricle but is applied globally to the entire slice. The retrospective correction reduces motion artifacts for the region of interest, but the surrounding static tissue (eg, back and spine) or tissue moving differently to the heart (eg, liver) are wrongly corrected, leading to residual artifacts for segmented k-space acquisitions. A correction with the PT could work even better for trajectories that sample the k-space center multiple times (ie, radial sequences), because these sequences are more robust against motion artifacts; therefore, wrongly corrected tissues would produce fewer artifacts.
Based on the PT and the dynamic images used for calibration, end-expiration could be identified and used as reference for the motion model to ensure slice tracking is carried out to a more well-defined motion state.

The heart beat also leads to a modulation of the PT. Nevertheless, this contribution is much smaller than the contribution due to respiratory motion and is further attenuated by the applied median filter.

The proposed approach is independent of k-space sampling, and therefore could also be combined with other sampling trajectories. Motion information is available with high temporal resolution, taking also variations in the breathing cycles into account. In addition, in this prospective implementation, motion corruption is prevented during data acquisition; therefore, no additional reconstruction time for a retrospective motion correction is needed. This makes this approach easy to integrate into clinical practice.

The CNR and sharpness of the endocardium show a significant improvement using PT-MOCO compared with the uncorrected cine scan, but are still not as good as breath-hold acquisitions. The breath-hold scan was always performed in end-expiratory state. In contrast, the respiratory state that was used for slice tracking was arbitrary, and the motion-corrected image may therefore show a different short-axis view position and lead to errors in the quantitative comparison.

For healthy volunteers, a breath-hold acquisition will provide the best image quality, but for patients having difficulties holding their breath, our approach can improve the image quality significantly compared with an uncorrected free-breathing acquisition. Nevertheless, studies on patients are still required to assess the applicability of this approach in clinical routine. So far, our method was used for cine imaging, but other continuous acquisitions, like MR fingerprinting, could also benefit from the method.

5 CONCLUSIONS

In our study, we were able to show that the PT provides a suitable motion signal for prospective respiratory motion correction. The temporal stability of the PT was demonstrated in phantom and volunteer scans. The PT-MOCO showed a significant improvement in image quality compared with an uncorrected free-breathing acquisition for all volunteers. Further studies are required to assess the proposed approach for clinical application. The PT-MOCO is independent of the MR data acquisition, making it a very flexible motion-correction approach.

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**SUPPORTING INFORMATION**

Additional Supporting Information may be found online in the Supporting Information section.

**FIGURE S1** Phantom setup. The phantom is attached to a wagon that performs translational motion in the head–feet (HF) direction of the scanner. The rotating wheel is attached to a string going outside of the scanner room, which is pulled by a controlled step motor. The receiver coil is placed onto a fixed table above the phantom. The pilot tone (PT) is generated by a frequency generator outside the scanner room, and the emitting coil is placed 20 cm away of the bore of the scanner onto a holder, to avoid vibrational motion.

**FIGURE S2** Calibration data for 10 volunteers. The data from 60 dynamic acquisitions were fitted with linear regression curves and used for the motion models.

**VIDEO S1** Cine images in short-axis-view (SAX) orientation of 4 healthy subjects. The proposed PT motion correction (PT-MOCO) method improves the image quality for free-breathing cine acquisitions. For comparison, the standard breath-hold cine scan is also shown.
VIDEO S2 Cine images of four different orientations in 1 healthy volunteer. For comparison, the standard breath-hold cine scan is also shown

VIDEO S3 Short-axis cine images in 1 healthy volunteer. Ten slices (six displayed) were recorded, covering the complete left ventricle. The PT-MOCO method leads to a reduction of respiratory-motion artifacts

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