Purpose: Segmented Cartesian acquisition in breath hold represents the current gold standard for cardiac functional MRI. However, it is also associated with long imaging times and severe restrictions in arrhythmic or dyspneic patients. Therefore, we introduce a real-time imaging technique based on a spoiled gradient-echo sequence with undersampled spiral k-space trajectories corrected by a gradient pre-emphasis.

Methods: A fully automatic gradient waveform pre-emphasis based on the gradient system transfer function was implemented to compensate for gradient inaccuracies, to optimize fast double-oblique spiral MRI. The framework was tested in a phantom study and subsequently transferred to compressed sensing–accelerated cardiac functional MRI in real time. Spiral acquisitions during breath hold and free breathing were compared with this reference method for healthy subjects (N = 7) as well as patients (N = 2) diagnosed with heart failure and arrhythmia. Left-ventricular volumes and ejection fractions were determined and analyzed using a Wilcoxon signed-rank test.

Results: The pre-emphasis successfully reduced typical artifacts caused by k-space misregistrations. Dynamic cardiac imaging was possible in real time (temporal resolution < 50 ms) with high spatial resolution (1.34 × 1.34 mm²), resulting in a total scan time of less than 50 seconds for whole heart coverage. Comparable image quality, as well as similar left-ventricular volumes and ejection fractions, were observed for the accelerated and the reference method.

Conclusion: The proposed technique enables high-resolution real-time cardiac MRI with no need for breath holds and electrocardiogram gating, shortening the duration of an entire functional cardiac exam to less than 1 minute.

Keywords: CMR, GSTF, pre-emphasis, real time, spiral
1 | INTRODUCTION

Cardiovascular MRI (CMR) in clinical routine typically relies on Cartesian data sampling. Nevertheless, non-Cartesian trajectories undoubtedly exhibit several advantages like more efficient k-space coverage, shorter acquisition times, or a favorable use of undersampling strategies. Spiral imaging, for example, inherently offers a broad spectrum of adjustment possibilities and can therefore be applied very flexibly: It can be implemented as a single-shot\textsuperscript{1-3} or interleaved multi-shot\textsuperscript{4,5} sequence. Moreover, the k-space trajectory can be designed Archimedean\textsuperscript{6} or with a variable density.\textsuperscript{7,8}

However, spiral MRI pushes hardware technology to its limits by applying rapidly oscillating magnetic field gradients. Imperfections of the gradient dynamics, such as residual eddy currents, amplifier inaccuracies, timing errors or the coupling of acoustic and main magnetic field fluctuations,\textsuperscript{9-13} result in nonideal transfer characteristics during an MRI exam. This leads to k-space misregistrations, because the true gradient output deviates from the nominal input as defined in the pulse sequence. Ultimately, these k-space trajectory errors appear as artifacts in the reconstructed image,\textsuperscript{12,14,15} potentially impairing diagnostic capabilities.

Magnetic resonance gradient systems can be modeled as linear and time-invariant in good approximation.\textsuperscript{16} This means that the transfer behavior is comprehensively characterized by the gradient system transfer function (GSTF). A field camera\textsuperscript{16,17} or phantom-based methods\textsuperscript{12,18-22} can be used to determine a GSTF, and thus for efficiently correcting deviations in the k-space trajectory associated with gradient inaccuracies. These corrections can be applied in postprocessing by convolving the spiral gradient waveforms with the GSTF within the reconstruction pipeline.\textsuperscript{17,20} Alternatively, the GSTF can be used as a pre-emphasis. This approach convolves the nominal gradients with the inverse GSTF before the measurement, as proposed previously in a prototype version.\textsuperscript{12}

A number of promising applications of spiral k-space sampling in CMR have been proposed in the past.\textsuperscript{23-25} Gradient imprecisions appears to be an important factor hindering clinical implementation for those spiral applications so far.

In this paper, we present a real-time acquisition technique with offline compressed sensing (CS) reconstruction for cardiac functional MRI. It is based on a spoiled gradient-echo (GRE) sequence with undersampled spiral k-space acquisitions (CRISPI, cardiac real-time imaging using a spiral k-space trajectory). Gradient waveforms were corrected by an integrated pre-emphasis, incorporating the information of a uniquely predetermined GSTF. This correction is fully automatic and works for arbitrary k-space trajectories and any slice orientation. Image reconstruction was realized by a CS method based on a low-rank plus sparse decomposition.\textsuperscript{26}

The aim of this work is to establish a real-time acquisition in free breathing that achieves comparable image quality as the current gold standard of electrocardiogram-triggered and segmented acquisitions in breath hold. As data are not sampled over several heartbeats, arrhythmia no longer represents a relevant problem in cardiac functional MRI. The CRISPI technique furthermore not only avoids motion artifacts, but also substantially increases patient comfort, as the entire functional exam can be performed in less than 1 minute.

2 | METHODS

2.1 | Gradient system transfer function

The determination of the system-specific GSTF follows the approach presented previously.\textsuperscript{12,19,20} The main aspects are summarized in this section. Twelve triangular input gradients with constant slew rate but of different amplitudes and durations were played out, and the phase evolutions $\phi_1(t)$ and $\phi_2(t)$ were measured in two parallel slices of a homogeneous spherical phantom. In addition, reference scans $\phi_{1,\text{ref}}(t)$ and $\phi_{2,\text{ref}}(t)$ were performed in each slice with a triangle gradient amplitude of zero and subsequently subtracted from $\phi_1(t)$ and $\phi_2(t)$, respectively.\textsuperscript{12} Thereby, the true gradient output $g_{\text{out}}(t)$ can be calculated as

$$g_{\text{out}}(t) = \frac{1}{d} \cdot \frac{d}{dt} \left( \left[ \phi_1(t) - \phi_{1,\text{ref}}(t) \right] - \left[ \phi_2(t) - \phi_{2,\text{ref}}(t) \right] \right),$$

(1)

where $\gamma$ is the gyromagnetic ratio of $^1$H, and $d$ is the slice distance. By means of the Fourier-transformed, idealized gradient input $\hat{G}_{\text{in}}(f) = \mathcal{F} \left[ g_{\text{in}}(t) \right]$ and the deduced gradient output $\hat{G}_{\text{out}}(f) = \mathcal{F} \left[ g_{\text{out}}(t) \right]$, the frequency-dependent system response GSTF $(f)$ can be calculated as

$$\text{GSTF}_{k,l}(f) = \frac{\sum_k G_{\text{in}}^{k,l}(f) \cdot G_{\text{out}}^{k,l}(f)}{\sum_i \left| G_{\text{in}}^{i,k}(f) \right|^2}.$$

(2)

where $k$ and $l$ represent the input and output channels, and index $i$ counts the number of triangular waveforms used. In this study, only the GSTF self-terms (ie, $k = l$) were calculated and used for trajectory correction: $\text{GSTF}_{x,x}(f)$, $\text{GSTF}_{y,y}(f)$, and $\text{GSTF}_{z,z}(f)$. In other words, the output signal is acquired in the same direction as the triangular input gradient is applied.

The GSTF measurements were performed on a clinical 3T MR scanner (MAGNETOM Prisma\textsuperscript{61}, Siemens Healthcare, Erlangen, Germany), which was subsequently used for the cardiac exams. The applied test sequence was based on an imaging prototype, which already incorporated an eddy current compensation. The triangular input gradients covered a duration of 100-320 µs at a slew rate of...
180 mT/m/ms. Measurement parameters were set to dwell time = 8.7 µs, TR = 1.0 seconds, slice thickness = 3.0 mm, slice distance = 33 mm, flip angle = 90°, and 100 signal averages. The GSTF measurement for all axes took about 4 hours in total.

2.2 Automatic pre-emphasis

To realize the automatic pre-emphasis for double-oblique imaging on the scanner, the inverted GSTFs were embedded in the sequence code. Numerical inversion without filtering was done for f < 12.2 kHz. For frequencies larger than 12.2 kHz, a decaying function given by curve fitting of the original GSTFs was used. The nominal spiral readout gradients are initially projected onto the physical axis k ∈ {x, y, z} by means of a rotation matrix corresponding to the planned orientation. In a second step, the resulting gradient waveform g_k(t) is Fourier-transformed and multiplied with the inverse GSTF of the respective spatial direction in frequency domain. Note that different dwell times in the GSTF acquisition and the spiral measurements lead to an additional global delay, which is identical for all three axes. An inverse Fourier transform subsequently yields the pre-emphasized gradient input as follows:

\[ g^\text{pre}_k(t) = \mathcal{F}^{-1} \left\{ \mathcal{F} [g_k(t)] \cdot \text{GSTF}_{k,k}^{-1}(f) \right\}. \] (3)

Consequently, the scanner outputs the actually intended spiral gradient waveforms during the measurement, and image reconstruction can be performed using the nominal spiral gradients.

For comparison, a retrospective application of the GSTF to the nominal waveforms was used within a postcorrection of the gradients before image reconstruction, as follows:

\[ g^\text{post}_k(t) = \mathcal{F}^{-1} \left\{ \mathcal{F} [g_k(t)] \cdot \text{GSTF}_{k,k}(f) \right\}. \] (4)

Additionally, reconstructions were performed with global delays only, identical for all axes. The delays were added between the timing of the ADC and the timing of the nominal k-space trajectory.

2.3 Spiral trajectory

The spiral readout gradients were designed using a freely available MATLAB (The MathWorks, Natick, MA) toolbox initiated by Brian Hargreaves.\(^8\) The maximum values for gradient slew rate and amplitude were set to \(S_{\text{max}} = 149.5\) mT/m/ms and \(G_{\text{max}} = 36\) mT/m. The in-plane spatial resolution was \(\Delta r = 1.34\) mm, and the trajectory was equipped with a variable density (FOV decreases linearly from 482 mm to 161 mm).

For cardiac real-time imaging, 10 equidistant (angle increment \(\varphi_{\text{inter}} = (2 \cdot \pi) / 10\) and consecutively acquired spiral arms formed one undersampled k-space (ie, the raw data for one real-time frame) (Figure 1B). 37 spiral arms would be required for Nyquist sampling. The maximum gap in k-space corresponded to a minimum FOV of 44 mm. With a TR of 4.96 ms, the temporal resolution per image was less than 50 ms. To cover k-space successively across several frames, the interference angle increment was set to \(\varphi_{\text{inter}} = (2 \cdot \pi GA) / 10\), with the golden angle \(GA = (2 \cdot \pi) / \left(\sqrt{5} + 1\right) \approx 111.25°\). Twice the golden angle was divided by 10 to efficiently fill gaps of adjacent frames each consisting of 10 equidistant spiral arms. This spiral sampling pattern was used in both phantom and in vivo studies. A sequence diagram of the applied 2D spoiled GRE sequence with the nominal spiral gradient waveforms is shown in Figure 1A. Further spiral imaging parameters were \(TE = 0.84\) ms, dwell time = 2.2 µs, number of sampling points per spiral arm = 1408, readout duration = 3.10 ms, flip angle = 15°, and slice thickness = 8 mm.

2.4 Image reconstruction

All spiral measurements were reconstructed offline in MATLAB. Initially, raw data were transferred to a Cartesian grid using the nominal k-space trajectory and GRAPPA operator gridding.\(^28\) To calibrate GRAPPA operators, a Cartesian k-space obeying the Nyquist-sampling criterion was determined by applying convolution gridding to a temporally averaged data set. In a second step, these operators were then applied to shift any sampling point within the undersampled frames to the closest position on the chosen Cartesian grid.

The low-rank plus sparse model presented by Otazo et al\(^26\) was exploited to obtain fully sampled data for each real-time frame:

\[ \min_{L,S} \frac{1}{2} \|E(L + S) - d\|^2_2 + \lambda_1 \|L\|_S + \lambda_2 \|TS\|_1. \] (5)

where \(L\) and \(S\) represent the low-rank and sparse component of the undersampled dynamic MRI data \(d\), respectively. Fourier-transformation to the temporal frequency domain serves as an additional sparsifying transform \(T\). \(E\) is the encoding operator, also incorporating coil sensitivity information. The minimization problem is solved by iterative soft thresholding with the empirically chosen regularization parameters \(\lambda_1 = 0.01\) and \(\lambda_2\), which is set to 5% of the maximum value in the sparse domain.
The ratio of the maximum k-space gap in the undersampled spiral acquisition and Δk of the chosen Cartesian grid during reconstruction yielded a maximum acceleration factor of 15.6.

2.5 Phantom measurements

To demonstrate the success of the proposed automatic pre-emphasis technique, the actual output of the gradient system was determined in a separate experiment measuring the signal’s phase in slices defined by one-dimensional phase encoding in a spherical phantom.

In addition, a phantom with structures of different sizes was used to further validate the spiral pre-emphasis. A 16-channel head coil array was used to measure transversal and double-oblique slice orientations with and without pre-emphasis. Fully sampled acquisitions were reconstructed using the gridding procedures described previously.

2.6 In vivo study

The study was approved by the local ethics committee, and written informed consent was obtained from each participant. An 18-channel body coil array in combination with a 16-channel spine coil array was used for the cardiac acquisitions. Seven healthy participants and 2 patients with cardiac disease were examined according to the following protocol:

1. A breath-held acquisition in midventricular short-axis (SAX) slice orientation with spiral pre-emphasis mode switched on and off (duration of each breath hold = 3.5 seconds).
2. Left-ventricular SAX slices from base to apex (N = 10-14) in free breathing (acquisition time = 35-49 seconds) as well as in breath hold (3.5 seconds per slice). The spiral pre-emphasis was used during these acquisitions.
3. Reference technique: electrocardiogram-gated, Cartesian, spoiled gradient echo in breath hold (the slice positions matched those of the spiral acquisitions in 2).
Shimming of the examined volume was performed with a vendor-provided cardiac shim and kept constant for all acquisitions in 1 volunteer/patient. The reference method was applied using the following protocol (adjusted individually depending on the subject size): TE = 2.29-2.41 ms, TR = 4.57-4.80 ms, FOV = 280-360 × 229-294 mm², in-plane spatial resolution = 1.35-1.73 × 1.41-1.80 mm², temporal resolution = 27.42-50.82 ms, dwell time = 3.4 µs, image matrix = 208 × 163, readout duration = 1.41 ms, flip angle = 15°, slice thickness = 8 mm, and T-GRAPPA factor = 2.

2.7 | Quantification of functional parameters

Quantitative cardiac functional parameters were derived from the images obtained by the different methods (parts 2 and 3 from the protocol). In particular, end-diastolic volume (EDV), end-systolic volume (ESV), stroke volume (SV), and ejection fraction (EF) of the left ventricle were determined by a trained operator (4 years of experience in cardiac imaging) using a dedicated cardiovascular imaging software (Circle Cardiovascular Imaging, Calgary, Canada). In addition, a Wilcoxon signed-rank test was performed to test for significant differences between the parameters of the CRISPI sequence and the Cartesian reference.

3 | RESULTS

3.1 | Gradient system transfer function

The magnitude and phase transmission of the measured GSTF self-terms for all three physical axes x, y, and z are depicted in Figure 2A,B, respectively. Similar transfer characteristics can be observed for the x-axis and y-axis, whereas those of the z-axis differ clearly. The MRI scanner exhibits low-pass behavior (ie, the magnitude decreases for higher frequencies). Interestingly, the transmission ratio of the z-axis exceeds that of the x-axis and y-axis for all frequencies. Scanner-specific resonances at approximately 1.7 kHz and 4 kHz are highlighted as an inlay in Figure 2A. The GSTF phase transmission of the x-axis and y-axis approximately exhibit a linear shape, whereas the course for the phase of the z-axis is flatter.

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**FIGURE 2**  Magnitude (A) and phase (B) of the measured gradient system transfer function (GSTF) self-terms of the x-axis, y-axis, and z-axis are shown in red, blue and green, respectively. Scanner-specific resonances are highlighted as an inlay in (A). C, Magnitude of the representation of one of the spiral readout gradients in the frequency domain. D, Inverted GSTFs used for the automatic pre-emphasis.
and oscillating for the observed frequency range. Figure 2C shows the magnitude of the representation of one of the spiral readout gradients in the frequency domain. Figure 2D depicts the inverted GSTFs used for the automatic pre-emphasis as described in section 2.2.

Figure 1C shows a detailed view of the region with maximum amplitude for one spiral readout gradient. The pre-emphasized, nominal, and measured waveforms (with and without pre-emphasis) are depicted, respectively. By playing out the pre-emphasized gradient (blue curve), the actually intended waveform was obtained (green curve matches dashed black curve with high precision). If no pre-emphasis was performed, the system transferred the waveform toward a falsified gradient output (orange curve).

### 3.2 Structural phantom study

Figure 3 shows spiral images of the structural phantom, representatively for transversal (Figure 3A) and double-oblique (Figure 3B) slice orientations. For the measurement of the double-oblique slice orientation, the phantom was rotated in the experimental setup, and the slice position was chosen perpendicular to the phantom. All images were reconstructed with the nominal k-space trajectory. Signal pile-ups and voids together with ghosting artifacts were present when no trajectory correction was applied (Figure 3A1,B1), which were greatly reduced when switching on the pre-emphasis (Figure 3A2,B2). The supporting line profiles of Figure 3A1,B1 reveal signal dropouts at the edges of phantom structures as well as corrupted signal intensities within phantom structures. This was successfully suppressed by the pre-emphasis.

In Figure 4, the image quality of the double-oblique slice is additionally compared with reconstructions using global delays but no pre-emphasis. Figure 4A,D corresponds to the measurements without and with pre-emphasis, and both were reconstructed with the nominal trajectory. In Figure 4B,C, the nominal trajectory was modified by implementing a global time delay, identical for all axes. The values of 7.70 µs and 11.3 µs were chosen to minimize the effect of the delay at the x-axis and z-gradient axis, respectively. Some artifacts could be mitigated by the global delay compensation. However, residual artifacts, highlighted by orange arrows, are present in both cases, which could be further reduced by the automatic pre-emphasis.

### 3.3 In vivo study: healthy participants

Figure 5A demonstrates the influence of the embedded pre-emphasis on an exemplary midventricular slice of 1 healthy participant in breath hold. To highlight artifacts that arise from trajectory errors, 704 consecutive spiral arms of the measurements both without (Figure 5A1) and with (Figure 5A2) pre-emphasis were reconstructed with the nominal k-space trajectory. These temporal averages are free of undersampling artifacts.
artifacts, which could corrupt the evaluation of the GSTF performance. Similar artifacts that appeared in the phantom study were present without trajectory correction (Figure 5A1), which could be clearly reduced by applying the gradient pre-emphasis (Figure 5A2). Figure 5B emphasizes the effect of the CS framework on an end-diastolic real-time frame of the CRISPI acquisition. Apparently, incoherent aliasing and blurring impaired the image quality when reconstructing the undersampled k-space without CS (Figure 5B1). These artifacts were eliminated by the CS algorithm, as shown in Figure 5B2.

Figure 6 illustrates that the quality of the trajectory correction works well for different image orientations. Spiral, end-diastolic real-time frames of a healthy volunteer in free breathing are shown in four-chamber view (Figure 6A) and SAX view (Figure 6B). All real-time images were reconstructed with the CS framework as presented in section 2.4. Images in Figure 6A1,A2,B1,B2 were reconstructed with the nominal trajectory. For comparison, the data acquired without pre-emphasis were reconstructed with retrospective application of the GSTF to the nominal waveforms during image reconstruction (postcorrection). The images without trajectory correction (Figure 6A1,B1) exhibit apparent artifacts, which could be mitigated equivalently by the pre-emphasis (Figure 6A2,B2) and the postcorrection (Figure 6A3,B3). Note that the pre-emphasized end-diastolic real-time frames (Figure 6A2,B2) and those acquired without pre-emphasis (Figure 6A1,B1,A3,B3) correspond to two separate measurements acquired in free breathing. The real-time CRISPI image series of a midventricular SAX slice of 1 healthy volunteer in free breathing is shown in Figure 7. Twenty time frames represent approximately one cardiac cycle from end-diastole 1 (top left) toward end-diastole 2 (bottom right). To highlight breathing motion,
the diaphragm–lung interfaces within frame 1 and frame 20 are denoted by a green and orange line, respectively. The end-systolic cardiac phase is marked by a dashed blue box.

Figure 8 compares the gated Cartesian reference (left column) with the real-time CRISPI sequence in breath hold (middle column) and free breathing (right column). End-systolic and end-diastolic cardiac phases are shown for three exemplary slices in SAX orientation. For slice 2, the temporal course is presented for the pixels marked by the orange line (x-t). Forty-five consecutive frames are shown, whereas for the gated reference, the R-R interval was duplicated along the temporal dimension. In general, reference and CRISPI acquisitions reveal similar image quality. Solely, regions with fat tissue exhibit blurring artifacts due to off-resonance when using the spiral sequence. The x-t diagrams support the observation that temporal fidelity is preserved by the real-time CRISPI framework, although the gated reference shows a slightly better contrast between myocardium and blood pool.

### 3.4 In vivo study: cardiac patients

Figure 9 displays the results of the 2 patients with cardiac disease and a diagnosis of heart failure with reduced ejection fraction. Additionally, patient 1 suffered from ventricular arrhythmia with complex premature contractions. Again, the three methods are compared by showing end-systolic and end-diastolic frames. The results of patient 1 are presented in Figure 9A and those of patient 2 in Figure 9B. Supporting x-t diagrams that correspond to the orange line in the respective diastolic frame are included for all cases. Both x-t depictions for the CRISPI reconstructions in Figure 9B show 43 consecutive frames, whereas in the reference diagram, the R-R interval is repeated three times along the temporal dimension. In the case of the arrhythmic patient (Figure 9A), the x-t characteristic of a time window of $t = 2.2$ seconds is presented.

In Figure 9A, ventricular arrhythmia resulted in unreliable sorting of the acquired data into the different cardiac segments. Thus, no typical R-R interval could be observed in
the image series using the gated Cartesian reference. Instead, spatial as well as temporal blurring reduced the image quality. Ultimately, end-systolic and end-diastolic cardiac phases could not be identified unequivocally. The x-t diagrams of the spiral real-time measurements indicate the known arrhythmia in Figure 9A. Additional differences in between the two CRISPI diagrams are due to extensive breathing motion. With regard to Figure 9B, all three methods yielded similar results. To support the latter analysis, supporting videos of both patients are presented online (see Supporting Information Videos S1 and S2). When analyzing the real-time CRISPI sequence in a breath hold of patient 1 (central part of Supporting Information Video S1), residual respiratory motion can be traced, indicating that the patient had troubles holding his or her breath. This potentially contributes to the spatiotemporal blurring that is present in the Cartesian reference video (left side of Supporting Information Video S1).

3.5 | Quantitative analysis

Figure 10A-D summarizes the quantitative results of left-ventricular EDV, ESV, SV, and EF for the Cartesian reference (blue) and the CRISPI sequence in breath hold (red) and free breathing (yellow). The x-axis lists all 9 measured participants (7 healthy volunteers and 2 cardiac patients). Most importantly, the three methods yielded similar results, as only minor differences between the calculated values could be found. Only the SV for patient 2 shows an apparent difference between the reference and the spiral acquisitions. The Wilcoxon signed-rank test revealed a failure to reject the null hypothesis at the 5% significance level in the case of ESV and EF, but a rejection in the case of EDV and SV for both CRISPI breath hold (BH) and free breathing (FB). Patient 1 was excluded from the Wilcoxon signed-rank test due to the ventricular arrhythmia. The mean differences with SD when subtracting the Cartesian parameters from the CRISPI parameters yielded $E_{FH} = -1.81 \pm 3.02\%$, $E_{FB} = -2.05 \pm 2.58\%$, $SV_{BH} = -7.61 \pm 8.68 \text{ mL}$, $SV_{FB} = -9.29 \pm 11.38 \text{ mL}$, $EDV_{BH} = -9.71 \pm 9.03 \text{ mL}$, $EDV_{FB} = -8.29 \pm 9.51 \text{ mL}$, $ESV_{BH} = -1.94 \pm 6.21 \text{ mL}$, and $ESV_{FB} = 0.43 \pm 4.13 \text{ mL}$. In general, the results of the healthy volunteers are in the expectable range of typical left-ventricular data, referred to as healthy participants. Most of the EFs of 52.25%-64.79% for the healthy volunteers demonstrate normal cardiac pump function, whereas EFs of 21.02%-34.85% for the 2 patients express moderately up to severely abnormal cardiac pump function. The high values for EDV and ESV in the case of patient 2 correlate with the qualitative observation of a comparably large left ventricle, as can be seen in Figure 9B.
DISCUSSION

The developed CRISPI technique facilitates real-time cardiac MRI with high temporal and spatial resolution. It is based on an automatic correction of gradient imperfections by means of a pre-emphasis using the GSTF. Undersampled spiral k-space trajectories were exploited in conjunction with a CS reconstruction. The established framework was validated in volunteers and patients both during breath hold and free breathing.

Measuring the impulse response and using it for trajectory correction serves as a simple and robust method to mitigate k-space deviations when applying time-varying gradient waveforms. This is supported by the fact that the GSTF is determined by a unique measurement and shows only small deviations over time.\(^ {12,20}\) In addition, the phantom-based approach that is used in this and other studies\(^ {12,18-22}\) is feasible without great expense, as no supplementary hardware is needed. Moreover, the pre-emphasis is not limited to a certain k-space trajectory, as, in principle, any desired gradient waveform can be convolved with the inverse GSTF. Further optimization of the pre-emphasis could possibly be achieved by including cross-terms of the GSTF.

Many studies concerning spiral cardiovascular MRI do not consider trajectory errors due to gradient imperfections.\(^ {24,31-33}\) Others used an estimation model of gradient delays with eddy current compensation to account for these inaccuracies,\(^ {34}\) which was introduced by Tan and Meyer in brain imaging.\(^ {15}\) Previous studies\(^ {12,17,18,20}\) used the impulse response function to predict actual k-space trajectories that were used during image reconstruction. One aim of this work was the extension of a prototype pre-emphasis\(^ {12}\) toward a fully automatic procedure that enables double-oblique slice orientations. With that, the intended waveforms are played out so that no further preparations of the readout gradients and no efforts regarding trajectory correction during image reconstruction are necessary. In addition, the performance of the pre-emphasis was compared with an optimized global delay compensation, demonstrating that the latter approach could already substantially reduce the overall artifact power. However, the high image quality of using the GSTF pre-emphasis with its frequency-dependent and axis-dependent corrections could not be achieved.

The spiral sampling pattern was set up with a high spatial resolution of \(1.34 \times 1.34 \text{ mm}^2\) and a fixed temporal...
**FIGURE 9** Comparison of the gated Cartesian reference (left) with the spiral real-time CRISPI sequence in breath hold (BH, middle) and free breathing (FB, right) in 2 patients (A,B) with cardiac disease. End-systolic and end-diastolic cardiac phases are presented, supported by x-t diagrams that correspond to the marked orange line in the diastole frame, respectively.

**FIGURE 10** Summary of the quantitative analysis for the Cartesian reference (blue) and the CRISPI sequence in breath hold (red) and free breathing (yellow). A-D, End-diastolic volume (EDV), end-systolic volume (ESV), stroke volume (SV), and ejection fraction (EF). The 7 healthy participants are labeled as V1-V7, and the 2 cardiac patients are labeled as P1 and P2.
resolution of less than 50 ms (≥ 20 frames per second). Both parameters could be adjusted by using different spiral gradient waveforms or by changing the coverage of the undersampled k-space (eg, to further decrease the number of spiral interleaves in each frame). However, there is a trade-off between spatiotemporal resolution and image quality, as the cardiac kinetics are prone to temporal blurring in the case of overfitting the chosen model. Here, we focused on maintaining all of the cardiac kinetics at the cost of slight residual aliasing in some cases (Supporting Information Video S1). The current results resemble previous findings that CS presents itself as a powerful tool for accelerated MRI, and therefore can be used to optimize high-resolution cardiac imaging.\cite{31,33,35,36} Recent efforts of exploiting machine learning to optimize filter functions within CS-like reconstruction techniques\cite{37} are promising for even better suited models and thus improved image quality (eg, less temporal blurring).

The spatial resolution of the CRISPI protocol is superior to other spiral real-time CMR studies performed at 1.5 T\cite{24,33,34} or 3 T.\cite{32} Moreover, complete images were often acquired with acquisition times larger than 100 ms,\cite{34} and sliding window reconstructions were used to yield image series with more than 20 frames per second.\cite{24,32} In contrast, little or no undersampling was used, such that online reconstruction was feasible in many cases. This represents one of the limitations of our current framework and needs to be implemented before a routine clinical application of the method.

Steeden et al\cite{33} used a tiny golden-angle sampling pattern with an acceleration factor of 8 and CS during reconstruction to achieve a temporal resolution of 30 ms with a spatial resolution of 1.7 mm at 1.5 T. Furthermore, balanced SSFP sequences are used primarily in CMR, as they feature a superior image contrast\cite{24,33,34} compared with GRE sequences. On the downside, valuable time is needed for gradient balancing, which is critical especially in real-time CMR. To our knowledge, superior spatiotemporal resolution is only feasible in spiral gated cine imaging as, for example, presented in Zhou et al,\cite{5} in which undersampled GRE imaging was performed at 3 T. The used GRE sequence and the flip angle of 15° avoided specific absorption rate problems in our study at 3 T.

With increasing $B_0$ field strength, off-resonance due to magnetic field inhomogeneity becomes a growing issue, which can result in blurring artifacts in the case of spiral sampling. Apart from slight blurring in fat regions, we have not observed severe off-resonance artifacts. We share the observation of Zhou et al,\cite{5} who found that readout durations of less than 5 ms produce acceptable results in CMR applications at 3 T without additional off-resonance correction. However, for readout durations longer than 8 ms, the spiral real-time study at 3 T presented in Nayak et al\cite{12} used an additional off-resonance correction requiring field maps.

Spiral pre-emphasis successfully reduced image artifacts resulting from k-space misregistrations. When compared with the Cartesian reference, equivalent image quality could be achieved in the heart using the real-time CRISPI sequence. The acquisition time for whole heart coverage could be decreased from several minutes (reference in breath hold) to less than 50 seconds (CRISPI in free breathing), substantially increasing not only time efficiency, but also patient comfort. This ultimately reduces the cost of cardiac MR exams and can contribute to the general accessibility of CMR. Moreover, the results on patient 1 showed that patients with cardiac arrhythmia greatly benefit from real-time CMR. In addition, a significant number of patients are not able to hold their breath for more than a few seconds.

The quantitative analysis with respect to volumetric and functional parameters agreed well with the Cartesian reference and CRISPI for all participants. The differences in the SV for patient 2 could be traced to those of the EDV. Due to the comparably large size of the left ventricle (EDV > 400 mL), the SV is greatly influenced by variations within the EDV. The Wilcoxon signed-rank test revealed a systematic deviation for the EDV and SV (ie, a small but significant underestimation of both parameters in the case of the spiral acquisition). However, the differences are rather small, and the test entails limited power due to a small sample size. For example, by additionally excluding patient 2, the test reveals no systematic deviation in the case of the EDV for the spiral acquisition in free breathing. A larger clinical study could provide a more reliable analysis of the discussed matter. In general, through-plane motion due to free breathing appears to have a minor impact on the quantitative results. This supports the robustness of the developed framework, not only from a technical, but also from a clinical point of view.

Recently, there has been a discussion\cite{38,39,40} about whether the term “real-time MRI” should be reserved for those applications that provide reconstructed images online with low latency (several hundred milliseconds). We share the opinion of Nayak et al\cite{40} and underline the main aspect of a definite time point that is represented by each frame. This study targeted high-quality CMR and not a minimization of the reconstruction time. All spiral measurements were reconstructed offline with a comparably high latency (several hours for whole heart coverage). Nevertheless, unrolled gradient descent schemes as, for example, presented in Hammernik et al,\cite{37} provide excellent possibilities for high-quality reconstructions in the shortest time and could therefore represent a remedy for the long reconstruction times of the technique presented in this paper.

It is noteworthy that the strategies introduced here are not only valuable for cardiac functional imaging but certainly
also for other investigations typically integrated in CMR exams (eg, late enhancement, myocardial perfusion).

5 | CONCLUSIONS

A real-time cardiac functional imaging technique dubbed CRISPI was presented and validated. A pre-emphasis based on the GSTF was embedded in a GRE sequence that automatically corrects spiral gradient waveforms before data acquisition. Artifacts that correspond with k-space trajectory errors could successfully be suppressed in a phantom and in vivo study. The CRISPI sequence benefits from high spatial resolution and concurrent short acquisition times to gather whole heart coverage. Typical issues of the current gold standard of electrocardiogram-gated acquisitions like corrupted data allocation due to arrhythmia, long breath holds, and low patient comfort are greatly improved or even eliminated by the proposed technique.

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CONFLICT OF INTEREST

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DATA AVAILABILITY STATEMENT

The code and data that support the findings of this study are openly available at https://github.com/expRad/CRISPI.

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**SUPPORTING INFORMATION**

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

**VIDEO S1** Comparison of the dynamic image series for patient 1, who suffered from cardiac disease with ventricular arrhythmia. The corrupted R-R interval of the gated Cartesian reference is shown on the left, and the videos in the middle and on the right depict 44 consecutive frames of the CRISPI (cardiac real-time imaging using a spiral k-space trajectory) sequence in breath hold and free breathing, respectively

**VIDEO S2** Comparison of the dynamic image series for patient 2, who suffered from cardiac disease. The gated Cartesian reference is shown on the left and the videos in the middle and on the right depict 49 consecutive frames of the CRISPI sequence in breath hold and free breathing, respectively

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