Generalized low-rank nonrigid motion-corrected reconstruction for MR fingerprinting

Gastao Cruz¹ | Haikun Qi¹ | Olivier Jaubert¹ | Thomas Kuestner¹ | Torben Schneider² | Rene Michael Botnar¹,³ | Claudia Prieto¹,³

¹School of Biomedical Engineering and Imaging Sciences, King's College London, London, United Kingdom
²Philips Healthcare, Guilford, United Kingdom
³Escuela de Ingeniería, Pontificia Universidad Católica de Chile, Santiago, Chile

Purpose: Develop a novel low-rank motion-corrected (LRMC) reconstruction for nonrigid motion-corrected MR fingerprinting (MRF).

Methods: Generalized motion-corrected (MC) reconstructions have been developed for steady-state imaging. Here we extend this framework to enable non-rigid MC for transient imaging applications with varying contrast, such as MRF. This is achieved by integrating low-rank dictionary-based compression into the generalized MC model to reconstruct MC singular images, reducing motion artifacts in the resulting parametric maps. The proposed LRMC reconstruction was applied for cardiac motion correction in 2D myocardial MRF (T₁ and T₂) with extended cardiac acquisition window (~450 ms) and for respiratory MC in free-breathing 3D myocardial and 3D liver MRF. Experiments were performed in phantom and 22 healthy subjects. The proposed approach was compared with reference spin echo (phantom) and with 2D electrocardiogram-triggered/breath-hold MOLLI and T₂ gradient-and–spin echo conventional maps (in vivo 2D and 3D myocardial MRF).

Results: Phantom results were in general agreement with reference spin-echo measurements, presenting relative errors of approximately 5.4% and 5.5% for T₁ and short T₂ (<100 ms), respectively. The proposed LRMC MRF reduced residual blurring artifacts with respect to no MC for cardiac or respiratory motion in all cases (2D and 3D myocardial, 3D abdominal). In 2D myocardial MRF, left-ventricle T₁ values were 1150 ± 41 ms for LRMC MRF and 1010 ± 56 ms for MOLLI; T₂ values were 43.8 ± 2.3 ms for LRMC MRF and 49.5 ± 4.5 ms for T₂ gradient and spin echo. Corresponding measurements for 3D myocardial MRF were 1085 ± 30 ms and 1062 ± 29 ms for T₁, and 43.5 ± 1.9 ms and 51.7 ± 1.7 ms for T₂. For 3D liver, LRMC MRF measured liver T₁ at 565 ± 44 ms and liver T₂ at 35.4 ± 2.4 ms.
**INTRODUCTION**

Quantitative MRI is a powerful tool for tissue characterization. Measuring the physical properties of tissue (such as $T_1$ and $T_2$) can objectively identify underlying disease, offering the potential for standardized clinical diagnosis. In cardiac MR, $T_1$ is used to characterize a wide range of disease including myocardial infarction, amyloidosis, or fibrosis; similarly, $T_2$ is sensitive to several diseases including myocardial edema, myocarditis, or transplant rejection. In liver MR, $T_1$ and $T_2$ perform similar roles in the detection of fibrosis or inflammation.

Physiological motion is a common source of error in both of these applications; therefore respiratory and/or cardiac motion must be addressed to achieve good-quality parametric maps. Conventional $T_1$ and $T_2$ cardiac parametric mapping is achieved with electrocardiogram (ECG)-triggered acquisitions performed under breath-hold to minimize cardiac and respiratory motion, respectively. Consequently, scan time is limited to the achievable breath-hold (usually less than 20 seconds), resulting in limited 2D coverage. Furthermore, the acquisition window at each heartbeat is limited to the cardiac resting period (~200 ms), further limiting scan efficiency. Similarly, conventional $T_1$ and $T_2$ liver parametric mapping is performed in 2D under multiple breath-holds to achieve sufficient coverage. Free-breathing acquisitions have enabled the acquisition of 3D parametric maps, for both cardiac and liver applications. However, these approaches usually rely on diaphragmatic respiratory-gating approaches that lead to long and unpredictable scan times. Respiratory translational motion-compensation methods have been recently introduced to reduce the scan times of 3D $T_1$ and $T_2$ parametric mapping; however, these approaches do not account for nonrigid motion and are limited to steady-state encoding approaches.

Magnetic resonance fingerprinting (MRF) has been introduced for simultaneous, multi-parametric mapping. It continuously samples the transient state magnetization of an acquisition with varying sequence parameters to simultaneously encode multiple parameters, estimating these parameters through dictionary matching. This approach leads to co-registered, jointly modeled, multi-parametric maps from a single scan. Cardiac and liver applications, relying on cardiac triggering and breath-holds to minimize motion, have been proposed. Rigid motion correction (MC) has been proposed for brain MRF. More recently, translation MC has enabled free-breathing 3D myocardial MRF with predictable scan time, although only facilitating translational MC, whereas nonrigid cardiac or respiratory MC reconstruction with MRF has not been yet investigated. Image-based nonrigid cardiac motion alignment has been considered in 2D $T_1$ mapping by applying cardiac motion fields to dynamic contrast images, and a similar strategy has also been considered for 2D cardiac MRF; however these approaches simply average the registered images and do not incorporate the nonrigid motion in the iterative reconstruction process.

Global low-rank models derived from the MRF dictionaries have been proposed for undersampled MRF reconstruction, enabling reconstruction of transient magnetization from highly undersampled data. However, these low-rank matrix formulations inherently share information among all acquired k-space data, assuming a static image. In the presence of motion, these reconstructions produce motion artifacts in the resulting parametric maps, determined by the underlying motion, k-space trajectory, and parametric encoding (ie, pulse sequence).

In this work, we propose a novel low-rank nonrigid motion-corrected (LRMC) reconstruction for transient imaging applications with varying contrast, such as MRF. The proposed framework integrates low-rank matrix dictionary-based subspace models into a generalized MC reconstruction to produce a time series of motion-compensated varying contrast images from highly undersampled data. Moreover, the reconstruction is regularized with a patch-based low-rank tensor approximation (HD-PROST) for improved performance. To evaluate this LRMC reconstruction (regularized with HD-PROST), we considered three applications: (1) cardiac motion in 2D extended cardiac window 2D myocardial MRF (where increased scan efficiency can improve SNR); (2) respiratory motion in 3D free-breathing myocardial MRF; and (3) respiratory motion in 3D free-breathing liver MRF.
(enabling free breathing without gating); and (3) respiratory motion in 3D free-breathing liver MRF (demonstrating free-breathing 3D liver MRF without gating). The proposed approach was evaluated in a total of 22 healthy subjects, in comparison with conventional methods. To the best of our knowledge, this corresponds to the first demonstration of the feasibility of free-breathing 3D liver MRF.

2 | METHODS

The proposed framework is summarized in Figure 1. The framework can be divided into three steps: (1) data acquisition and dictionary generation to estimate low-rank dictionary-based compression, (2) auxiliary motion-resolved reconstruction to enable nonrigid motion estimation (via image registration), and (3) LRMC reconstruction. The low-rank compression operator is derived from the MRF dictionary (step 1), whereas the motion operator is obtained from the actual acquired data (step 2). Nonrigid motion estimation is performed in a bin-to-bin basis using image registration and relying on a 1D motion surrogate to bin the data in different motion states. Electrocardiogram and respiratory bellow signals are used as surrogate for cardiac and respiratory motion, respectively. In the case of respiratory motion, the relative bellows signal is used to drive a localized autofocus algorithm, producing an absolute measurement of 1D respiratory translational motion. Auxiliary motion-resolved reconstructions (ie, at different cardiac or respiratory motion states) are used to estimate dense, nonrigid motion fields that are then incorporated (together with low-rank compression) into the proposed LRMC reconstruction.

2.1 | Acquisition

The LRMC approach was validated in three MRF applications in which nonrigid cardiac or respiratory motion pose a challenge: (1) 2D myocardial MRF with higher cardiac acquisition efficiency using prolonged acquisition window (cardiac MC); (2) 3D myocardial MRF; and (3) 3D liver MRF with free-breathing, nongated acquisitions (respiratory MC). The acquisition protocol for each of these applications is described subsequently.
2.1.1 | Two-dimensional myocardial MRF

Two-dimensional ECG-triggered myocardial MRF scans were performed under breath-hold with an extended acquisition window of 450 ms (at each heartbeat), capturing midsystole through diastole using a constant density spiral trajectory. This strategy minimizes through-plane motion in 2D; however considerable in-plane motion is expected to remain. The acquisition protocol used three inversion-recovery (IR) preparation pulses and six \( T_2 \) preparation (\( T_2p \)) pulses over the course of 16 heartbeats (Figure 2A) to encode \( T_1 \) and \( T_2 \), respectively. The preparation pulse scheme was based on previous work on 3D myocardial MRF\(^{22} \); for each heartbeat, a different preparation pulse (or no pulse) was applied: IR15, 3× NP, IR130, 3× NP, IR260, NP, 3× \( T_2p20 \), and 3× \( T_2p50 \). In this notation, NP denotes “no preparation pulse”; the numerical values after IR or \( T_2p \) denote the corresponding TI or \( T_2p \) TE, respectively; and “\( N \times \)” represents the number of times the same preparation is applied consecutively. Spectral presaturation with inversion recovery\(^{36} \) was used in heartbeats with a TI superior to 300 ms to suppress fat signal, and gradient and RF-spoiled readouts (FLASH) were used to reduce sensitivity to field inhomogeneities. Fixed TE, TR, and flip angle (FA) were used.

2.1.2 | Three-dimensional myocardial MRF

Three-dimensional ECG-triggered free-breathing myocardial MRF data were acquired with a stack of variable density spirals using the acquisition scheme described in previous work\(^{22} \) (Figure 2B): IR15, 4× NP, IR150, 4× NP, IR300, NP, \( T_2p20 \), \( T_2p30 \), \( T_2p40 \), \( T_2p50 \), \( T_2p60 \), and \( T_2p80 \). Gradient spoiled readouts, fixed TE/TR, and sinusoidally varying FA in the range of 5° to 10° were used. The preparation pulse scheme described previously was used for each slice encoding of the stack of spirals with a minimum of 4-second waiting period between slice encodings to allow for longitudinal magnetization (Mz) to recover.

**Figure 2** Sequence diagrams of the MRF protocols used in this study. A, The sequence for 2D myocardial MRF used inversion-recovery (IR) pulses, \( T_2 \) preparation pulses (of 20 ms and 50 ms), fat suppression (spectral presaturation with inversion recovery [SPIR]) pulses, and cardiac-triggered readouts, in 18 heartbeats. B, The sequence for 3D myocardial MRF used a similar encoding scheme to the 2D myocardial MRF, except \( T_2 \) preparation pulses were varied over 20 ms to 80 ms. This sequence of 18 heartbeats was repeated for each slice encoding. C, The sequence for 3D liver MRF was acquired continuously, with preparation pulses interrupting an otherwise “free-running” acquisition. The sequence depicted was repeated for each slice encoding.
2.1.3 | Three-dimensional liver MRF

Three-dimensional free-breathing liver MRF was acquired with a similar encoding strategy to previous work in 2D liver MRF.\textsuperscript{18} Data were acquired with a free-running acquisition, otherwise interrupted by IR, T2p, and spectral presaturation with inversion-recovery pulses. Additionally, pause sections of 400 ms (no RF pulses applied) were used after T2p shots to improve SNR. Data were acquired with stack of spirals. The preparation scheme used for each slice encoding was as follows (Figure 2C): IR15, 19× NP, IR15, 9× NP, 15× T2p20, and 15× T2p50 for a total of 60 preparation blocks. Radiofrequency-spoiled readouts were used with fixed TE/TR and sinusoidally varying flip angle in the range of 8° to 12°. After each set of 38 preparation blocks, a pause of 3 seconds was introduced to allow for Mz to recover similar to 3D myocardial MRF; the same preparation scheme was repeated for the following slice encoding.

2.2 | Motion estimation

Auxiliary motion-resolved reconstructions (to estimate nonrigid motion fields) were obtained using soft-weighted low-rank inversion (LRI)\textsuperscript{37} with HD-PROST regularization (LRI-HD-PROST)\textsuperscript{38}:

\[
\left(\hat{S}_n, \hat{\mathcal{F}}_b^n\right) = \arg\min_{\mathbf{y}_n, \mathcal{F}_b^n} \frac{1}{2} \left\| \mathbf{W}_n \left( \mathbf{A}_n \mathbf{U}_r, F C_\alpha \mathbf{y}_n - \mathbf{k}_n' \right) \right\|_2^2 + \lambda \sum_b \left\| \mathcal{F}_b^n \right\|_1, \quad \text{s.t.} \quad \mathcal{F}_b^n = Q_b(\mathbf{y}_n) \tag{1}
\]

where \(\mathbf{y}_n\) are the reconstructed singular images for the \(N\)th motion state (or bin); \(\mathbf{W}_n\) are soft weights; \(\mathbf{A}_n\) corresponds to k-space sampling; \(\mathbf{U}_r\) is the dictionary-based low-rank compression (to rank \(r\)) operator derived from the MRF dictionary; \(F\) is the nonuniform Fourier transform; \(C\) are the coil sensitivities; \(\mathbf{k}_n'\) is the k-space or the translationally corrected k-space for the \(N\)th bin (if intrabin MC is considered as described subsequently); and \(Q_b\) generates a 3D tensor \(\mathcal{F}_b^n\) of voxels associated with the \(b\)th voxel (and \(N\)th bin) by concatenating local voxels (within a local patch) along the first dimension, nonlocal voxels (from patches that exhibit structural similarity with the patch around \(b\)), and contrast voxels (along the compressed singular value domain). Image registration based on free-form deformations\textsuperscript{39,40} is then applied to the auxiliary motion-resolved reconstructions to retrieve the corresponding nonrigid motion fields. When a binning approach is considered to define motion states; this reconstruction is relatively well-posed, leading to images with sufficient quality for motion estimation. However, the underlying parametric encoding may vary between motion states and may not be optimal for immediate template matching without incorporating information from all motion states (through LRMC).

2.2.1 | Two-dimensional myocardial MRF

Auxiliary cardiac motion-resolved images were obtained by binning the data within each 450-ms acquisition window into six cardiac bins of equal size. To improve contrast in this reconstruction (Equation 1) and facilitate image registration, only data with T2p (ie, the last six heartbeats) with good blood/myocardium contrast was used for this purpose. In this case, because a single contrast is used for the reconstruction, \(\mathbf{U}_r = \mathbf{I}\) in Equation 1 and a 2D matrix (instead of 3D tensor) is used for the HD-PROST regularization.

2.2.2 | Three-dimensional myocardial MRF

Auxiliary respiratory-resolved images were obtained by binning the acquired data according to the signal provided by the respiratory bellows. Before this binning, a localized autofocus algorithm was used to determine an absolute estimate of the translational motion of the heart due to breathing, as described in previous work.\textsuperscript{22} Briefly, the respiratory motion in each dimension is assumed to be proportional to the acquired bellows signal, \(r(t)\). A set of translationally corrected images \(\mathbf{x}_a\) are obtained by reconstructing the data with different motion signals \(ar(t)\). The optimal scaling \(\hat{a}\) (that determines an absolute estimate of respiratory motion) is obtained by minimization of the localized gradient entropy \(H(\mathbf{x}_a) = -\sum_{i} h_a(\mathbf{x}_a(i)) \log_2 h_a(\mathbf{x}_a(i))\), where \(h_a\) is the normalized spatial gradient. After estimating respiratory translational motion, data are grouped into three equally sized respiratory bins; for each bin, k-space is translationally corrected toward the bin center (intrabin MC) to minimize remaining intrabin motion. Similar to the case of 2D myocardial MRF, only the T2p data are selected for the auxiliary motion-resolved reconstruction.

2.2.3 | Three-dimensional liver MRF

Auxiliary respiratory-resolved images were obtained in a similar fashion to the case of 3D cardiac MRF previously. Respiratory bellows were used to drive an autofocus algorithm and estimate translational motion, which was used to both bin the data into three respiratory-motion states and translationally correct each bin k-space toward the center of its bin. Instead of selecting the subset of T2p data, all data were considered for the auxiliary respiratory-resolved reconstruction (as it provided better contrast between
abdominal organs), and nonrigid motion estimation was performed from the reconstructed first singular image.

2.3 | Low-rank MC reconstruction

The generalized matrix formulism for MC introduced by Batchelor et al.\(^\text{28}\) is characterized by the following problem:

\[
\hat{x} = \arg\min_{x} \frac{1}{2} \left\| \sum_{n} A_{n} FCM_{n} x - k \right\|_{2}^{2}
\]  

where \(x\) is the MC image; \(M_{n}\) is a sparse matrix that encodes the motion transformation for the \(n\)th motion state; and \(k\) is the acquired (MC) k-space. This formulation incorporates MC directly into the reconstruction process and is not limited to affine MC unlike k-space-based corrections. Initially proposed for single-contrast imaging, this formulism for the LRI formulism) within the generalized MC matrix reconstruction:

\[
\hat{y}, \hat{\mathcal{B}} = \arg\min_{y, \mathcal{B}} \frac{1}{2} \left\| \sum_{n} A_{n} U_{n} FCM_{n} y - k' \right\|_{2}^{2} + \lambda \sum_{b} \left\| \mathcal{B}_{b} \right\|_{*} \text{ s.t. } \mathcal{B}_{b} = Q_{b}(y)
\]

where \(y\) are MC singular images, and \(k'\) is the sampled k-space (or the translationally corrected k-space in case intrabrin correction is considered). All remaining operators are defined in Equations 1 and 2.

2.4 | Simulations

Digital phantoms based on realistic nonrigid motion and parametric maps were used to investigate the performance of the proposed LRMC approach. Three digital phantom simulations were carried out: (1) myocardial MRF with cardiac motion, (2) myocardial MRF with respiratory motion, and (3) liver MRF with respiratory motion. Ground truth for the digital phantoms (\(T_{1}, T_{2}, M_{0}\), coil sensitivities, and motion fields) was initially obtained from in vivo MRF experiments. For each simulated sequence, associated Bloch simulations were performed producing the corresponding k-space from a simulated MRF acquisition (considering k-space undersampling, motion, and coil sensitivities). In all cases, a single 2D slice was simulated. Data acquisitions were simulated in each case using the parameters described in section 2.5.2. Reconstructions were performed using the same parameters as described in section 2.6. In each case, simulated data were reconstructed with (1) no MC using LRI-HD-PROST (Equation 3 with \(M_{n} = I\)), (2) the proposed LRMC using nonrigid motion estimated using image registration from auxiliary motion-resolved images, (3) LRMC using the known (simulated) motion, and (4) without motion corruption (reconstructed using LRI-HD-PROST). For the simulation, coil sensitivity maps and motion fields were assumed to be known. To investigate the behavior of the reconstruction in the presence of large motion errors, an additional simulation was performed using a cardiac digital phantom containing respiratory and cardiac motion, where errors were introduced to the motion model by scaling it to 105%, 110%, 115%, 120%, 125%, and 130%.

2.5 | Experiments

The proposed LRMC was evaluated in a standardized phantom (static) and in a total of 22 healthy subjects (12 males, age 31 ± 3 years) on a 1.5T Ingenia MR system (Philips, Best, The Netherlands) using a 28-channel cardiac coil. Eight subjects were scanned for 2D myocardial MRF; 8 subjects were scanned for 3D myocardial MRF; and 6 subjects were scanned for 3D liver MRF. The study was approved by the institutional review board, and written informed consent was obtained from all subjects according to institutional guidelines.

2.5.1 | Phantom acquisitions

A standardized \(T_{1}/T_{2}\) phantom (TIMES)\(^\text{41}\) was validated with (1) 2D myocardial MRF, (2) 3D myocardial MRF, and (3) 3D liver MRF sequences. In general, MRF sequences used spiral readouts, constant TE/TR, and low flip angles. Specific acquisition parameters are provided in Supporting Information Table S1.
All MRF phantom scans were compared with 2D reference IR spin echo and spin echo for \( T_1 \) and \( T_2 \), respectively. Key parameters for the IR spin echo included TE/TR = 15/15 000 ms and 15 TIs in the range of 50 to 5000 ms; key parameters for spin echo included 8 TEs in the range of 10 to 640 ms.

2.5.2 | In vivo acquisitions

**Two-dimensional myocardial MRF**

Eight healthy subjects were scanned in mid-short axis with the ECG-triggered 2D myocardial MRF protocol with an extended cardiac acquisition window of 450 ms as previously described. Additionally, a conventional 2D cardiac MRF\(^{35} \) protocol was acquired for comparison. The same parameters were used, except a shorter cardiac window of 150 ms was used, leading to the acquisition of only 336 time points (Supporting Information Table S1). Additionally, MOLLI\(^{15} \) and \( T_2 \)-GRASE\(^{8} \) were acquired for comparison using the same FOV and resolution. The MOLLI (5[3]3 variant) key parameters included FOV = 256 × 256 mm\(^2 \); resolution = 1.6 × 1.6 mm\(^2 \); TE/TR = 1.4/2.8 ms; FA = 35\(^\circ \); SENSE factor = 2; and acquisition window = 224 ms. The \( T_2 \)-GRASE (with black-blood preparation) key parameters included nine TEs equally spaced from 9.3 ms to 83.7 ms, FA = 90\(^\circ \); EPI factor = 7, SENSE factor = 3, and acquisition window = 84 ms.

**Three-dimensional myocardial MRF**

Eight healthy subjects were scanned in short-axis orientation with the same 3D myocardial MRF protocol as in the phantom experiments. Electrocardiogram and respiratory bellows were used to monitor cardiac and respiratory motion. For comparison, conventional 2D MOLLI and 2D \( T_2 \)-GRASE acquisitions were performed in basal, mid, and apical slices. The MOLLI (5[3]3 variant) was acquired with the following parameters: FOV = 300 × 300 mm\(^2 \); resolution = 2 × 2 mm\(^2 \); TE/TR = 1.19/2.40 ms, FA = 35\(^\circ \); SENSE factor = 1.8, and acquisition window = 244 ms. The \( T_2 \)-GRASE (black-blood suppression) used the following parameters: nine TEs from 8.3 ms to 74.7 ms, EPI factor = 7, SENSE factor = 2.8, FA = 90\(^\circ \), and acquisition window = 75 ms.

**Three-dimensional liver MRF**

Six healthy subjects were scanned with the same 3D liver MRF protocol as in the phantom experiments. Respiratory bellows were used to monitor respiratory motion.

2.6 | Magnetic resonance fingerprinting reconstruction

Auxiliary motion-resolved reconstructions to enable nonrigid motion estimation through image registration (Equation 1) and the proposed LRMC (Equation 3) reconstruction were solved with the alternating direction method of multipliers (ADMM)\(^{42} \); conjugate gradient (CG) was used to solve the internal least-square problem. For comparison, data were also reconstructed with no MC using LRI-HD-PROST (Equation 3 with \( M_n = I \)), using otherwise corresponding parameters to the ones used for the proposed LRMC. Coil sensitivities were derived from ESPIRiT\(^{43} \) and density-compensation functions estimated using voronoi diagrams. NiftyReg\(^{39} \) was used for image registration to estimate nonrigid motion fields, using local normalized cross correlation for similarity metric. All dense motion fields were cast as sparse matrices (\( M_n \)), considering linear interpolation.

The following key parameters were used for the auxiliary motion-resolved reconstruction (Equation 1): CG iterations = 4, ADMM iterations = 2, number of similar patches = 20, patch search window size = 41 pixels, \( \lambda = 5 \times 10^{-2} \), LRI rank = 2 (3D liver MRF only), linear soft weighting with bin’s width effectively increased by 50% for cardiac motion and 25% for respiratory motion. For respiratory-motion estimation, the unit-normalized respiratory bellows signal was scaled by \( \alpha = [0:0.1:1] \beta \), where \( \beta \) was the maximum amplitude observed over all subjects for the autofocusing step. Data binning along cardiac (for 2D myocardial MRF) and respiratory (for 3D myocardial and liver MRF) allowed for successful auxiliary motion-resolved reconstructions in a modest number of motion states. Six motion states were considered for cardiac motion (effective resolution of ~75 ms), and three motion states were considered for respiratory motion (in both 3D myocardial MRF and 3D liver MRF).

The following key parameters were used for the proposed LRMC reconstruction (Equation 3): CG iterations = 3, ADMM iterations = 3, number of similar patches = 25 (2D) and 20 (3D), patch search window size = 51 (2D) and 41 (3D) pixels, and \( \lambda = 4 \times 10^{-3} \). All reconstruction parameters were determined experimentally in two representative cases.

Reconstructions were performed in a Linux workstation with 12 Intel Xeon X5675 (3.07 GHz) and 200 GB RAM. Acquired raw data were about 0.1 GB for 2D MRF and about 1-2 GB for 3D MRF; corresponding memory burden of the full pipeline was about 10 GB for 2D MRF and about 100 GB for 3D MRF. We estimate the computational cost of the LRMC reconstruction as \( \frac{\sigma}{r} \left( \{ [aM + bMlogM]2rcnN_g + (psr)^2M \} N_{ADMM} \right) \), where \( a \) and \( b \) are gridding parameters; \( M \) is the total number of data points; \( r \) is the rank; \( c \) is the number of coils; \( n \) is the number of motion states; \( N_g \) is the number of conjugate gradient iterations; \( p \) is the number of self-similar patches; \( s \) is the patch size; and \( N_{ADMM} \) is the number of ADMM iterations. The reconstruction times for 2D myocardial MRF were 1 hour and 40 minutes, with about 10
minutes for motion-resolved reconstruction/image registration (step 1) and about 1.5 hours for LRMC (step 2). Corresponding results for 3D myocardial MRF were 6 hours and 50 minutes, with a corresponding split of about 50 minutes (step 1) and about 6 hours (step 2); corresponding results for 3D liver MRF were 11 hours and 50 minutes with a corresponding split of approximately 90 minutes (step 1) and approximately 10 hours (step 1). High-dimensionality PROST (denoising step), NiftyReg (image registration step), ESPiRiT, and the nonuniform fast Fourier transform ran as C++ compiled code; the remaining code was implemented in MATLAB (R2018b; The MathWorks, Natick, MA).

2.7 Magnetic resonance fingerprinting dictionaries

The MRF dictionaries were computed using extended phase graphs (EPGs). For 2D myocardial MRF, the following values were considered: \( T_1 = [50:20:700, 700:10:900, 900:5:1300, 1300:20:1400, 1400:50:2000, 2000:100:4000] \) ms and \( T_2 = [5:5:20, 20:0.5:60, 60:2:100, 100:10:500] \) ms. For 3D myocardial MRF, the following values were considered: \( T_1 = [200:30:900, 900:20:1200, 1200:30:1400, 1400:50:2000] \) ms and \( T_2 = [20:1:60, 60:5:100, 100:20:300] \) ms. For 3D liver MRF, the following values were considered: \( T_1 = [10:10:100, 100:5:900, 900:20:1000, 1000:30:1300, 1300:50:2000, 2000:100:5000] \) ms and \( T_2 = [10:1:80, 80:2:100, 100:10:200, 200:20:300, 200:50:800] \) ms. The extended phase graph model did not consider \( B_0/B_1 \) imperfections, slice profile, diffusion, or magnetization transfer effects. The MRF sequences used here employed gradient spoiling (and RF spoiling) with low FAs, which are relatively insensitive to \( B_0, B_1 \), and slice profile errors as show in previous works. Mean values in different myocardial segments (or liver) were used as surrogates for accuracy; SDs of these measurements were used as surrogates for precision.

3 RESULTS

3.1 Simulations

Simulated nonrigid cardiac and respiratory motion resulted primarily in blurring artifacts in the parametric maps. The proposed LRMC led to a substantial reduction in motion artifacts as seen in Supporting Information Figures S1-S3 for cardiac motion in myocardial MRF, respiratory motion in myocardial MRF, and respiratory motion in liver MRF, respectively. Large deviations in parametric error maps and elevated mean square errors (MSEs) measured in regions of interest around the heart and liver (\(-115 \text{ ms for } T_1, -27 \text{ ms for } T_2\)) were observed without MC and subsequently reduced with the proposed (estimated motion) LRMC (\(-40 \text{ ms for } T_1, -9 \text{ ms for } T_2\)), similar to LRMC using known motion (\(-20 \text{ ms for } T_1, -5 \text{ ms for } T_2\)) (Supporting Information Figures S1-S3). When evaluating the behavior of LRMC in the presence of large motion errors, we can observe that errors in the motion model propagate into the parameter maps (Supporting Information Figures S7-S10). In the presence of artificial errors in the motion model, we can observe some errors in the maps (error 46-77 ms for \( T_1 \) and 13.0-15.7 seconds for \( T_2 \), although much smaller than with no MC. In contrast, LRMC from accurate image registration (error \(-45 \text{ ms for } T_1 \text{ and } 13.0 \text{ ms for } T_2\)) achieves similar quality to LRMC using the true motion (error \(-35 \text{ ms for } T_1 \text{ and } 11.7 \text{ ms for } T_2\)), and both are of similar quality to the ground truth (where no motion exists).

3.2 Phantom acquisitions

Phantom measurements performed for 2D myocardial MRF (150-ms and 450-ms windows), 3D myocardial MRF, and 3D liver MRF were in general agreement with conventional spin-echo IR and spin-echo values; however, a slight underestimation of \( T_2 \) was generally observed (Supporting Information Figures S4-S6, respectively). Larger \( T_2 \) errors were observed for long \( T_2 \) values, particularly for the 3D myocardial MRF sequence that used gradient spoiled readouts. For the 2D myocardial MRF with 150-ms acquisition window, normalized RMS errors for \( T_1 \), short \( T_2 \), and long \( T_2 \) were 4.9%, 5.4% and 17.2%, respectively; for 2D myocardial MRF with 450-ms acquisition window, the corresponding values were 6.4%, 6.3%, and 17.3%. For 3D myocardial MRF, the corresponding values were 3.7%, 5.3% and 36.3%, whereas for 3D liver MRF, they were 7.1%, 4.8%, and 12.8%.

3.3 In vivo 2D myocardial MRF

Singular images from 2D MRF (450-ms window) presented blurring artifacts around the myocardium and papillary muscles when cardiac motion was not accounted for (no MC); these artifacts were considerably reduced with the proposed LRMC (representative subject A; Figure 3). These motion artifacts propagated into the parametric maps and were again considerably reduced with LRMC (representative subject A; Figure 4). Conventional MRF (150-ms window) showed slightly lower precision than LRMC MRF (450-ms window), likely due to only acquiring one third of the data (representative subject A; Figure 4).
The MOLLI technique showed lower $T_1$ values and higher noise amplification in the lateral wall than LRMC MRF. A myocardial segment analysis of mean and SD of $T_1$ and $T_2$ revealed higher $T_1$ values and higher apparent precision in the septal wall (compared with the lateral wall) for all methods; for $T_2$, values were slightly higher in the septal wall (compared with the lateral wall) for the MRF approaches (Figure 5, showing aggregated results over the subject cohort). No significant differences in $T_1$ and $T_2$ values were observed with/without MC for MRF with a 450-ms acquisition window. Two-dimensional myocardial MRF $T_1$ values were significantly elevated relative to MOLLI (more so for LRM MRF with 450-ms window); $T_2$ values were significantly lower than $T_2$-GRASE. $T_1$ precision was significantly improved with LRMC MRF (450-ms window), and $T_2$ precision for LRMC MRF (450-ms window) and MRF (150-ms window) was significantly higher than $T_2$-GRASE. These values are presented in Supporting Information Table S2, aggregated over the respective subject cohorts.
Myocardial segmental analysis for 2D myocardial MRF. Higher $T_1$ values and lower SDs (precision surrogate) were observed for MRF (particularly with 450 ms window) when compared with MOLLI. $T_1$ values and apparent precision were generally higher in the septal wall than in the lateral wall (for all methods). For $T_2$, a difference between septal and lateral values were observed with MRF (not with $T_2$-GRASE), and lower precision was obtained (particularly with 150-ms window myocardial MRF). Lower $T_2$ values were obtained with myocardial MRF compared with $T_2$-GRASE.

Three-dimensional myocardial MRF singular images for subject B, reconstructed with LRI-HD-PROST without MC and with the proposed LRMC. Motion artifacts can be observed with no MC and subsequently reduced with LRMC (arrows).
3.4 | In vivo 3D myocardial MRF

Singular images for free-breathing 3D myocardial MRF with (LRMC) and without (no MC) respiratory MC were compared, revealing minor blurring artifacts with no MC, which were reduced after LRMC (representative subject B; Figure 6). Corresponding effects were observed in the parametric maps: residual blurring in T1 and T2 was observed with no MC and considerably reduced with LRMC, achieving parametric quality more in line with conventional MOLLI and T2-GRASE (representative subject B; Figure 7). A myocardial segmental analysis (Figure 8, showing aggregated results over the subject cohort) revealed slightly higher septal values (compared with lateral) for T1 in all methods; slightly higher septal values were also observed with MRF for T2. The T1 values with MRF were higher than MOLLI (significantly so for no MC); both no-MC MRF and LRMC MRF presented significantly improved precision relative to MOLLI. The T2 values with MRF were significantly lower than T2-GRASE, and a significant difference in T2 values was observed between no MC and LRMC. These results are compiled in Supporting Information Table S2.

FIGURE 7 T1 and T2 maps obtained with MOLLI (and T2-GRASE for T2) and with 3D myocardial MRF with no MC and with the proposed LRMC, for subject B. Residual respiratory motion artifacts are visible with no MC, although these are reduced with LRMC (arrows), achieving more comparable quality to conventional methods
Free-breathing 3D liver MRF was also reconstructed without (no MC) and with (LRMC) respiratory MC, leading to considerable differences in residual motion artifacts in the singular images (representative subject C; Figure 9). Similar to the two previous cases, considerable motion artifacts were observed in the parametric maps with no MC; however, a considerable reduction of these artifacts was achieved with LRMC (representative subject C; Figure 10). Mean and SD of T₁ and T₂ measured in a region of interest in the liver presented similar values with and without MC. T₁ values were in agreement with those reported in literature; however, T₂ values were considerably

3.5 | In vivo 3D liver MRF

Myocardial segmental analysis for 3D myocardial MRF. Higher T₁ values and lower SDs (precision surrogate) were observed for MRF when compared with MOLLI. For T₂, a difference between septal and lateral values was observed with MRF (not with T₂-GRASE), and lower precision was obtained (particularly with 150-ms window myocardial MRF). Slightly higher values in the septal wall (compared with the lateral wall) were observed for all T₁ methods, as well as for T₂ with myocardial MRF. Lower T₂ values were obtained with myocardial MRF compared with T₂-GRASE

FIGURE 8 Myocardial segmental analysis for 3D myocardial MRF. Higher T₁ values and lower SDs (precision surrogate) were observed for MRF when compared with MOLLI. For T₂, a difference between septal and lateral values was observed with MRF (not with T₂-GRASE), and lower precision was obtained (particularly with 150-ms window myocardial MRF). Slightly higher values in the septal wall (compared with the lateral wall) were observed for all T₁ methods, as well as for T₂ with myocardial MRF. Lower T₂ values were obtained with myocardial MRF compared with T₂-GRASE.

FIGURE 9 Three-dimensional liver MRF singular images for subject C, reconstructed with LRI-HD-PROST without MC and with the proposed LRMC. Motion artifacts can be observed with no MC, and subsequently reduced with LRMC (arrows).

Singular images
lower than literature values. These results are included in Supporting Information Table S2.

4 | DISCUSSION

A reconstruction method was introduced to enable generalized nonrigid MC of MRF imaging with varying contrast. This approach incorporates low-rank (dictionary-based) and nonrigid MC (LRMC) models into a regularized iterative reconstruction. The feasibility of this approach was investigated for cardiac and respiratory MC in three different applications: 2D myocardial MRF, 3D myocardial MRF, and 3D liver MRF. The average relative errors in phantom over the three applications were about 5% for $T_1$ and about 6% for $T_2$ (for $T_2$ values less than 100 ms).

The study on 2D myocardial MRF demonstrated that the proposed approach is capable of correcting for nonrigid in-plane cardiac motion. Cardiac parametric mapping often relies on ECG triggering to minimize cardiac motion, at the expense of scan efficiency. This limitation can impose longer scans and/or limit spatial resolution. Extending the cardiac acquisition window beyond the middiastolic period (~150 ms) increases the amount of data collected at each heartbeat at the expense of motion artifacts. Here, the proposed LRMC was used to increase the scan efficiency by extending the acquisition to about 450 ms, while correcting for cardiac motion. This approach achieved higher apparent precision and parametric map quality than MOLLI and myocardial MRF with middiastolic acquisition window. The $T_2$-GRASE technique achieved the highest apparent precision, and middiastolic myocardial MRF presented the lowest apparent precision for $T_2$. $T_1$ measured with the proposed MC 2D myocardial MRF were higher than those obtained with middiastolic myocardial MRF, particularly in blood, potentially due to flow effects. The LRMC MRF myocardium $T_1$ was generally higher than MOLLI (bias of +140 ms), and $T_2$ was generally lower than $T_2$-GRASE (bias of −5.7 ms), in line with previous studies in cardiac MRF. These biases are likely associated with magnetization-transfer effects and diffusion, in addition to flow, all of which are not considered in the current model. When comparing parametric maps without (no MC) and with (LRMC) cardiac MC, a considerable reduction of motion artifacts could be observed.

The study on free-breathing 3D myocardial MRF demonstrated the feasibility of the proposed approach to correct for 3D nonrigid respiratory motion. Three-dimensional imaging is often impossible to perform within a breath-hold, requiring diaphragmatic respiratory gating or MC solutions. For MRF, however, respiratory gating can affect the underlying parametric encoding in addition to increased and unpredictable scan times. Free-breathing 3D myocardial MRF with translational respiratory MCs...
has been recently proposed\textsuperscript{22}; however that approach does not enable nonrigid motion compensation. Here, LRMC was used to enable MC of remaining nonrigid motion components in free-breathing 3D myocardial MRF, resulting in comparable parametric maps to 2D breath-held conventional methods. A slight positive bias relative to MOLLI (+23 ms) and negative bias relative to T\textsubscript{2}-GRASE (−8.2 ms) was observed. Apparent precision for LRMC 3D cardiac MRF (61 ms for T\textsubscript{1} and 4.7 ms for T\textsubscript{2}) was better than MOLLI (77 ms) and comparable to T\textsubscript{2}-GRASE (4.9 ms). When comparing the results before and after MC, an increase in sharpness and delineation of myocardium wall and papillary muscles can be appreciated.

In the third study, LRMC was used to enable for the first time (to the best of our knowledge) free-breathing 3D liver MRF T\textsubscript{1} and T\textsubscript{2} mapping with 100% respiratory scan efficiency (ie, no respiratory gating). Due to the complex nature of motion, simpler affine models may not fully capture the elastic respiratory induced motion of the liver.\textsuperscript{50} Our experiments showed that in the absence of MC, blurring artifacts are present in both T\textsubscript{1} and T\textsubscript{2} 3D-MRF maps, obscuring vessels in the liver and other small structures. The LRMC technique led to a considerable reduction in motion artifacts, allowing the visualization of these structures and removing biases in regions of large motion (eg, diaphragm dome) on the T\textsubscript{1} and T\textsubscript{2} maps. No significant differences for T\textsubscript{1} and T\textsubscript{2} (in terms of mean values and SDs) were observed before/after MC in homogenous regions of interest in the liver. T\textsubscript{1} values were in agreement with literature values (bias of −9 ms); however, T\textsubscript{2} values were underestimated relative to literature (bias of −10.6 ms).

All acquisition sequences used in this work were heuristic, guided by previous preliminary works on sequence design for MRF. For cardiac MRF, a choice of small FAs\textsuperscript{46} (and FLASH readouts for 2D cardiac MRF and 3D liver MRF) were considered to minimize sensitivity to B\textsubscript{1} and slice profile. Previous work on liver MRF noted considerable sensitivities to field errors\textsuperscript{17} and the need for B\textsubscript{1} correction. In an effort to minimize this problem, here we opted to generate all parametric encoding through preparation pulses with adiabatic elements\textsuperscript{18} and low-FA FLASH readouts. Additionally, fat-suppression modules were used to reduce potential aliasing signal originating from fat. Further investigation on optimal sequence design in the presence of field inhomogeneities for specific cardiac and liver applications is warranted, similarly to previous work done on brain MRF.\textsuperscript{51–53}

Motion estimation is a key challenge for the proposed framework. Similarly to previous work on nonrigid MC steady-state imaging,\textsuperscript{25,54} dense motion fields were obtained through image registration of auxiliary motion resolved images. Here we use a soft-weighted LRI-HD-PROST reconstruction to produce these images; however, other motion-resolved approaches like XD-GRASP\textsuperscript{55} or multitasking\textsuperscript{56} could be considered for similar purposes. Elastic motion fields have also been produced using localized autofocus ideas\textsuperscript{57}; however, that approach has not yet been validated in dynamic contrast applications, like MRF. These images are used here for motion estimation and thus can suffer from insufficient diagnostic image quality. Poor contrast, low SNR, and residual artifacts in the auxiliary images can all affect the accuracy of image registration. Because non-Cartesian trajectories are used, it is possible to improve apparent SNR/residual aliasing by reconstructing lower resolution for the purposes of motion estimation, as motion is generally smooth. Preliminary experiments (not shown) indicated better motion estimation using T2p data in cardiac; however, this was not the case for the liver. Simulations demonstrated that accurate motion estimation was obtained from the auxiliary motion-resolved images with the proposed framework. In the presence of errors in the motion model, residual blurring is present in the resulting maps; however, map quality is still considerably higher than no MC. Additional simulations showed that larger errors in the motion model (up to 30%) can introduce errors in the parameter maps (increase in MSE from 35 ms to 77 ms for T\textsubscript{1}, increase in MSE from 11.7 ms to 15.7 ms in T\textsubscript{2}), although these errors remain considerably lower than the case in which motion is not corrected (MSE of 171 ms for T\textsubscript{1}, MSE of 37.0 ms for T\textsubscript{2}). Finally, the binning step before reconstructing motion-resolved images assumes periodic motion. Irregular motion like arrhythmia may have to be discarded, reducing scan efficiency and potentially affecting map quality.

Alternatively, motion could be resolved instead of corrected. Magnetic resonance multitasking\textsuperscript{56} has recently been proposed to resolve motion in quantitative MRI. This is a promising method, although with considerable computational complexity. Multitasking uses self-navigation signals to estimate cardiac and respiratory-motion states, casting the image reconstruction of the dynamic motion and contrast data as a low-rank tensor. The estimation of the associated motion subspaces for multitasking is expected to require less information than for dense motion fields in LRMC; however, the reconstruction costs are also considerably different. We note that although the LRMC produces a single MC state, multiple LRMC reconstructions (toward different motion states) could be used to achieve motion-resolved imaging (through multiple MC reconstructions), albeit at increased computational cost. Comparisons between motion-resolved and MC approaches would be interesting in future work.

This work features several limitations. Each study included a small number of subjects to evaluate feasibility.
More in-depth studies with larger cohorts should be considered in the future to further characterize the performance of the proposed LRMC in specific applications. The in vivo study for liver application focused on demonstrating the feasibility of free-breathing 3D liver MRF and in comparing LRMC against no motion compensation. The accuracy of this approach was demonstrated in phantom experiments; however, further validation of $T_1$ and $T_2$ accuracy in vivo will be considered in dedicated studies in the future. Cardiac MC was only evaluated within an approximate 450-ms acquisition window (in 2D) to minimize through-plane motion, as it can affect $T_1$ and (especially) $T_2$ values. Nonetheless, the increased acquisition window in 2D cardiac MRF provides increased scan efficiency (3 times) that could be exploited to achieve higher spatial resolution, shorter scan time, or potentially map additional parameters. Investigating the feasibility of LRMC for 3D cardiac imaging would be of interest in future studies. Future works should also validate this approach in specific patient populations.

### 5 CONCLUSIONS

A novel low-rank (dictionary-based) MC reconstruction was proposed to enable generalized nonrigid MC of MRF imaging with varying contrast. This approach enabled 2D myocardial MRF with higher cardiac scan efficiency with cardiac MC and free-breathing 3D myocardial and demonstrated for the first time free-breathing 3D liver MRF with respiratory MC.

### CONFLICT OF INTEREST

Torben Schneider was Philips employed during the development of this research.

### ORCID

Gastao Cruz [https://orcid.org/0000-0002-7397-9104](https://orcid.org/0000-0002-7397-9104)

Olivier Jaubert [https://orcid.org/0000-0002-7854-4150](https://orcid.org/0000-0002-7854-4150)

Thomas Kuestner [https://orcid.org/0000-0002-0353-4898](https://orcid.org/0000-0002-0353-4898)

### TWITTER

Gastao Cruz [@GastaoCruz](https://twitter.com/GastaoCruz)

### REFERENCES

1. Moon JC, Messroghli DR, Kellman P, et al. Myocardial T1 mapping and extracellular volume quantification: a Society for Cardiovascular Magnetic Resonance (SCMR) and CMR Working Group of the European Society of Cardiology consensus statement. *J Cardiovasc Magn Reson.* 2013;15:1-12.

2. Messroghli DR, Moon JC, Ferreira VM, et al. Clinical recommendations for cardiovascular magnetic resonance mapping of $T_1$, $T_2$, $T_2$ and extracellular volume: a consensus statement by the Society for Cardiovascular Magnetic Resonance (SCMR) endorsed by the European Association for Cardiovascular Imaging. *J Cardiovasc Magn Reson.* 2017;19:1-24.

3. Haimerl M, Verloh N, Zeman F, et al. Assessment of clinical signs of liver cirrhosis using $T_1$ mapping on Gd-EOB-DTPA-enhanced 3T MRI. *PloS One.* 2013;8:1-8.

4. Luetkens JA, Klein S, Träber F, et al. Quantification of liver fibrosis at $T_1$ and $T_2$ mapping with extracellular volume fraction MRF: preclinical results. *Radiology.* 2018;288:748-754.

5. Messroghli DR, Radjenovic A, Kozerke S, Higgins DM, Sivananthan MU, Ridgway JP. Modified look-locker inversion recovery (MOLL) for high-resolution $T_1$ mapping of the heart. *Magn Reson Med.* 2004;52:141-146.

6. Chow K, Flewitt JA, Green JD, Pagano JJ, Friedrich MG, Thompson RB. Saturation recovery single-shot acquisition (SASHA) for myocardial $T_1$ mapping. *Magn Reson Med.* 2014;71:2082-2095.

7. Giri S, Chung Y-C, Merchant A, et al. $T_2$ quantification for improved detection of myocardial edema. *J Cardiovasc Magn Reson.* 2009;11:1-13.

8. Sprinkart AM, Luetkens JA, Träber F, et al. Gradient Spin Echo (GraSE) imaging for fast myocardial $T_2$ mapping. *J Cardiovasc Magn Reson.* 2015;17:1-9.

9. Nordio G, Henningsson M, Chiribiri A, Villa ADM, Schneider T, Botnar RM. 3D myocardial $T_1$ mapping using saturation recovery. *J Magn Reson Imaging.* 2017;46:218-227.

10. Guo R, Chen Z, Herzka DA, Luo J, Ding H. A three-dimensional free-breathing sequence for simultaneous myocardial $T_1$ and $T_2$ mapping. *Magn Reson Med.* 2019;81:1031-1043.

11. Qi H, Jaubert O, Bustin A, et al. Free-running 3D whole heart myocardial $T_1$ mapping with isotropic spatial resolution. *Magn Reson Med.* 2019;82:1331-1342.

12. Qi H, Bustin A, Cruz G, et al. Free-running simultaneous myocardial $T_1$/$T_2$ mapping and cine imaging with 3D whole-heart coverage and isotropic spatial resolution. *Magn Reson Imaging.* 2019;63:159-169.

13. Milotta G, Ginami G, Bustin A, Neji R, Prieto C, Botnar RM. 3D whole-heart free-breathing qBOOST-$T_2$ mapping. *Magn Reson Med.* 2020;83:1673-1687.

14. Ma D, Gulani V, Seiberlich N, et al. Magnetic resonance fingerprinting. *Nature.* 2013;495:187-192.

15. Hamilton JJ, Jiang Y, Chen Y, et al. MR fingerprinting for rapid quantification of myocardial $T_1$, $T_2$, and proton spin density. *Magn Reson Med.* 2017;77:1446-1458.

16. Jaubert O, Cruz G, Bustin A, Schneider T, Botnar RM, Prieto C. Dixon cMRF: cardiac tissue characterization using three-point Dixon MR fingerprinting. In: Proceedings of the 27th Annual Meeting of ISMRM, Montréal, Canada, 2019. Abstract #1100.

17. Chen Y, Jiang Y, Pahlwa S, et al. MR fingerprinting for rapid quantitative abdominal imaging. *Radiology.* 2016;279:278-286.

18. Jaubert O, Arrieta C, Cruz G, et al. Multi-parametric liver tissue characterization using MR fingerprinting: simultaneous $T_1$, $T_2$, $T_2^*$, and fat fraction mapping. *Magn Reson Med.* 2020;84:2625-2635.

19. Cruz G, Jaubert O, Schneider T, Botnar RM, Prieto C. Rigid motion-corrected magnetic resonance fingerprinting. *Magn Reson Med.* 2018;81:947-961.

20. Bipin B, Dan M, Eric M, Pierre Y, Coppo S, Alan M. Magnetic resonance in medicine image reconstruction algorithm for
motion insensitive MR fingerprinting (MRF): MORF. *Magn Reson Med*. 2018;80:2485-2500.

21. Yu Z, Zhao T, Cloos MA, Assländer J, Lattanzi R, Sodickson DK. Exploring the sensitivity of magnetic resonance fingerprinting to motion. *Magn Reson Imaging*. 2018;54:241-248.

22. Cruz G, Jaubert O, Qi H, et al. 3D free-breathing cardiac magnetic resonance fingerprinting. *NMR Biomed*. 2020;33:1-16.

23. Becker KM, Blaszczyk E, Funk S, et al. Fast myocardial T1 mapping using cardiac motion correction. *Magn Reson Med*. 2020;83:438-451.

24. Hamilton JI, Jiang Y, Eck B, Griswold M, Seiberlich N. Cardiac cine magnetic resonance fingerprinting for combined ejection fraction, T1 and T2 quantification. *NMR Biomed*. 2020;33:e4323.

25. Cruz G, Atkinson D, Buerger C, Schaeffter T, Prieto C. Accelerated motion corrected three-dimensional abdominal MRI using total variation regularized SENSE reconstruction. *Magn Reson Med*. 2016;75:1484-1498.

26. Schaeffter T, Kolbitsch C, Dössel O, Paschke NK, Prieto C. Comparison of image-based and reconstruction-based respiratory motion correction for golden radial phase encoding coronary MR angiography. *J Magn Reson Imaging*. 2015;42:964-971.

27. Zhu X, Chan M, Lustig M, Johnson KM, Larson PEZ. Iterative motion-compensation reconstruction ultra-short TE (iMoCo UTE) for high-resolution free-breathing pulmonary MRI. *Magn Reson Med*. 2020;83:1208-1221.

28. Batchelor PG, Atkinson D, Irarrazaval P, Hill DLG, Hajnal J, Larkman D. Matrix description of general motion correction applied to multishot images. *Magn Reson Med*. 2005;54:1273-1280.

29. McGivney DF, Pierre E, Ma D, et al. SVD compression for magnetic resonance fingerprinting in the time domain. *IEEE Trans Med Imaging*. 2014;33:2311-2322.

30. Zhao BO, Setsompop K, Adalsteinsson E, et al. Improved magnetic resonance fingerprinting reconstruction with low-rank and subspace modeling. *Magn Reson Med*. 2018;79:933-942.

31. Assländer J, Cloos MA, Knoll F, Sodickson DK, Hennig J, Lattanzi R. Low rank alternating direction method of multipliers reconstruction for MR fingerprinting. *Magn Reson Med*. 2018;79:83-96.

32. Bustin A, Cruz G, Jaubert O, Lopez K, Botnar RM, Prieto C. High-dimensionality undersampled patch-based reconstruction (HD-PROST) for accelerated multi-contrast MRI. *Magn Reson Med*. 2019;81:3705-3719.

33. Atkinson D, Hill DLG, Styole PNR, Summers PE, Keesvl SF. Automatic correction of motion artifacts in magnetic resonance images using an entropy focus criterion. *IEEE Trans Med Imaging*. 1997;16:903-910.

34. Cheng JY, Alley MT, Cunningham CH, Vasanawala SS, Pauly JM, Lustig M. Nonrigid motion correction in 3D using autofocus with localized linear translations. *Magn Reson Med*. 2012;68:1785-1797.

35. Ingle RR, Wu HH, Addy NO, et al. Nonrigid autofocus motion correction for coronary MR angiography with a 3D cones trajectory. *Magn Reson Med*. 2014;72:347-361.

36. Oh C, Hilal S, Cho Z. Selective partial inversion recovery (SPIR) in steady state for selective saturation magnetic resonance imaging (MRI). In: Proceedings of the Seventh Annual Meeting of the ISMRM, Vol. 2, San Francisco, California, 1988. p 1042.

37. Jaubert O, Cruz G, Bustin A, et al. Free-running cardiac magnetic resonance fingerprinting: joint T1/T2 map and cine imaging. *Magn Reson Imaging*. 2020;68:173-182.

38. Bustin A, Cruz G, Jaubert O, Lopez K, Botnar RM, Prieto C. High-dimensionality undersampled patch-based reconstruction (HD-PROST) for accelerated multi-contrast MRI. *Magn Reson Med*. 2019;81:3705-3719.

39. Modat M, Ridgway GR, Taylor ZA, et al. Fast free-form deformation using graphics processing units. *Comput Methods Programs Biomed*. 2009;89:278-284.

40. Rueckert D, Sonoda LI, Hayes C, Hill DLG, Leach MO, Hawkes DJ. Nonrigid registration using free-form deformations: application to breast MR images. *IEEE Trans Med Imaging*. 1999;18:712-721.

41. Captur G, Gatehouse P, Kellman P, et al. A T1 and ECV phantom for global T1 mapping quality assurance: the T1 mapping and ECV standardisation in CMR (TIMES) program. *J Cardiovasc Magn Reson*. 2016;18:4-6.

42. Boyd S. Distributed optimization and statistical learning via the alternating direction method of multipliers. *Found Trends Mach Learn*. 2010;3:1-122.

43. Uecker M, Lai P, Murphy MJ, et al. ESPIRiT—an eigenvalue approach to autocalibrating parallel MRI: where SENSE meets GRAPPA. *Magn Reson Med*. 2014;71:990-1001.

44. Weigel M. Extended phase graphs: dephasing, RF pulses, and echoes—pure and simple. *J Magn Reson Imaging*. 2015;41:266-295.

45. Jaynes E. Matrix treatment of nuclear induction. *Phys Rev*. 1955;98:1099-1105.

46. Hamilton JI, Jiang Y, Ma D, et al. Investigating and reducing the effects of confounding factors for robust T1 and T2 mapping with cardiac MR fingerprinting. *Magn Reson Imaging*. 2018;53:40-51.

47. Stanisz GJ, Odrobina EE, Pun J, et al. T1, T2 relaxation and magnetization transfer in tissue at 3T. *Magn Reson Med*. 2005;54:507-512.

48. Hilbert T, Xia D, Block KT, et al. Magnetization transfer in magnetic resonance fingerprinting. *Magn Reson Imaging*. 2020;84:128–141. https://doi.org/10.1002/mrm.28096.

49. Kobayashi Y, Terada Y. Diffusion-weighting caused by spoiler gradients in the fast imaging with steady-state precession sequence may lead to inaccurate T2 measurements in MR fingerprinting. *Magn Reson Imaging*. 2018;53:40-51.

50. McClelland JR, Hawkes DJ, Schaeffter T, King AP. Respiratory motion models: a review. *Med Image Anal*. 2013;17:19-42.

51. Zhao BO, Haldar JP, Liao C, et al. Optimal experiment design for global T1 mapping quality assurance: the T1 mapping and ECV standardisation in CMR (TIMES) program. *J Cardiovasc Magn Reson*. 2014;68:384-861.

52. Assländer J, Novikov DS, Lattanzi R, Sodickson DK, Cloos MA. Hybrid-state free precession in nuclear magnetic resonance. *Commun Phys*. 2019;2:1-73.

53. Leitao D, Hajnal JV, Teixeira RP, Malik S. Parameter encoding efficiency in transient and steady-state quantitative MRI methods. In: Proceedings of the 27th Annual Meeting of ISMRM, Montréal, Canada, 2019. p 813.

54. Cruz G, Atkinson D, Henningsson M, Botnar RM, Prieto C. Highly efficient nonrigid motion-corrected 3D whole-heart coronary vessel wall imaging. *Magn Reson Med*. 2017;77:1894-1908.

55. Feng L, Axel L, Chandarana H, Block KT, Sodickson DK, Otazo R. XD-GRASP: golden-angle radial MRI with reconstruction of extra motion-state dimensions using compressed sensing. *Magn Reson Med*. 2016;75:775-788.
56. Christodoulou AG, Shaw JL, Nguyen C, et al. Magnetic resonance multitasking for motion-resolved quantitative cardiovascular imaging. *Nat Biomed Eng*. 2018;2:215-226.

57. Luo J, Addy NO, Ingle RR, et al. Nonrigid motion correction with 3D image-based navigators for coronary MR angiography. *Magn Reson Med*. 2017;77:1884-1893.

**SUPPORTING INFORMATION**

Additional Supporting Information may be found in the online version of the article at the publisher’s website.

**FIGURE S1** Simulated 2D myocardial MRF with cardiac motion occurring in a 450 ms window, reconstructed with LRI-HDPROST (no MC) and with the proposed Low Rank Motion Corrected (LRMC), in comparison to the simulated case without motion (no motion). Cardiac motion introduces blurring artefacts in T1 and T2, particularly in the borders of the myocardium (as visible in the error maps) which are removed with LRMC. A considerable reduction in mean squared error (MSE, computed in a region of interest around the heart) is also observed after LRMC.

**FIGURE S2** Simulated 2D myocardial MRF with respiratory motion reconstructed with LRI-HDPROST (no MC) and with the proposed Low Rank Motion Corrected (LRMC), in comparison to the simulated case without motion (no motion). Respiratory motion introduces minor blurring artefacts in T1 and T2, particularly in the borders of the myocardium (as visible in the error maps) which are removed with LRMC. A considerable reduction in mean squared error (MSE, computed in a region of interest around the heart) is also observed after LRMC.

**FIGURE S3** Simulated 2D liver MRF with respiratory motion reconstructed with LRI-HDPROST (no MC) and with the proposed Low Rank Motion Corrected (LRMC), in comparison to the simulated case without motion (no motion). Respiratory motion introduces considerable blurring artefacts in T1 and T2, obscuring vessels and other image structures (as visible in the error maps) which are removed with LRMC. A considerable reduction in mean squared error (MSE, computed in a region of interest around the liver) is also observed after LRMC.

**FIGURE S4** 2D myocardial MRF measurements in a parametric phantom in comparison to reference inversion recovery spin echo and spin echo for T1 and T2, respectively. T1 measurements are in general agreement with the reference (with a slight overestimation at high values), however a slight underestimation of T2 is observed (larger for high T2 values).

**FIGURE S5** 3D myocardial MRF measurements in a parametric phantom in comparison to reference inversion recovery spin echo and spin echo for T1 and T2, respectively. T1 measurements are in general agreement with the reference (with a slight underestimation at high values), however a slight underestimation of T2 is observed (larger for high T2 values).

**FIGURE S6** 3D liver MRF measurements in a parametric phantom in comparison to reference inversion recovery spin echo and spin echo for T1 and T2, respectively. T1 measurements are in general agreement with the reference (with a slight underestimation at high values), however a slight underestimation of T2 is observed (larger for high T2 values).

**FIGURE S7** T1 maps obtained from a realistic MRF simulation including respiratory and cardiac motion. Different degrees of error in the motion fields were added to the estimated motion (ranging from 5-30%); the proposed LRMC was used to reconstruct each map. Motion artefacts are present in the case with no motion correction and minor blurring appears in some cases where the artificial motion errors are high. LRMC using the real estimation motion produces similar quality to LRMC using true motion and to the ground truth. Corresponding errors are shown in Figure B.

**FIGURE S8** T1 errors obtained from a realistic MRF simulation including respiratory and cardiac motion corresponding to the maps shown in Figure A. Considerable errors are present in the case of no motion correction (>170 ms). Artificial motion errors (range of 5-30%) produce errors in the range of 46-77 ms, considerably lower than no motion correction, but still higher than LRMC with estimated motion (45 ms) and LRMC with true motion (35 ms).

**FIGURE S9** T2 maps obtained from a realistic MRF simulation including respiratory and cardiac motion. Different degrees of error in the motion fields were added to the estimated motion (ranging from 5-30%); the proposed LRMC was used to reconstruct each map. Motion artefacts are present in the case with no motion correction and minor blurring appears in some cases where the artificial motion errors are high. LRMC using the real estimation motion produces similar quality to LRMC using true motion and to the ground truth. Corresponding errors are shown in Figure C.

**FIGURE S10** T2 errors obtained from a realistic MRF simulation including respiratory and cardiac motion corresponding to the maps shown in Figure A. Considerable errors are present in the case of no motion correction (>37 ms). Artificial motion errors (range of 5-30%) produce errors in the range of 13.1-15.7 ms, considerably lower than no motion correction, but still higher than LRMC with estimated motion (130.0 ms) and LRMC with true motion (11.7 ms).
TABLE S1 Summary of the MRF sequence parameters employed in each study

TABLE S2 Summary of the T1 and T2 measurements observed for in-vivo 2D myocardial MRF, 3D myocardial MRF and 3D liver MRF. (*) denotes statistical difference to the MOLLI-T2GraSE methods; (‡) denotes statistical difference to the 150 ms myocardial MRF; (†) denotes statistical difference between no MC and LRMC.