Full Paper

Isotropic 3D Cartesian single breath-hold CINE MRI with multi-bin patch-based low-rank reconstruction

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Abstract

Purpose: To develop a novel acquisition and reconstruction framework for isotropic 3D Cartesian cardiac CINE within a single breath-hold for left ventricle (LV) and whole-heart coverage.

Methods: A variable-density Cartesian acquisition with spiral profile ordering, outwards sampling, and acquisition-adaptive alternating tiny golden/golden angle increment between spiral arms is proposed to provide incoherent and nonredundant sampling within and among cardiac phases. A novel multi-bin patch-based low-rank reconstruction, named MB-PROST, is proposed to exploit redundant information on a local (within a patch), nonlocal (similar patches within a spatial neighborhood), and temporal (among all cardiac phases) scale with an implicit motion alignment among patches. The proposed multi-bin patch-based low-rank reconstruction is compared against compressed sensing reconstruction, whereas LV function parameters derived from the proposed 3D CINE framework are compared against those estimated from conventional multislice 2D CINE imaging in 10 healthy subjects and 15 patients.

Results: The proposed framework provides 3D cardiac CINE images with high spatial (1.9 mm³) and temporal resolution (~50 ms) in a single breath-hold of ~20 s for LV and ~26 s for whole-heart coverage in healthy subjects. Shorter breath-hold durations of ~13 to 15 s are feasible for LV coverage with slightly anisotropic resolution (1.9 × 1.9 × 2.5 mm) in patients. LV function parameters derived from 3D CINE were in good agreement with 2D CINE, with a bias of −0.1 mL/0.1 mL, −0.9 mL/−1.0 mL, −0.1%/−0.8%; and confidence intervals of ±1.7 mL/±3.7 mL, ±1.2 mL/±2.6 mL, and ±1.2%/±3.6% (10 healthy subjects/15 patients) for end-systolic volume, end-diastolic volume, and ejection fraction, respectively.

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1 | INTRODUCTION

Cardiac CINE MRI is the gold standard for the assessment of cardiac morphology and function.¹ Conventionally, multislice 2D CINE is performed under multiple breath-holds to achieve left ventricular (LV) coverage. This requires patient cooperation and can result in slice misalignments due to different or imperfect breath-hold positions. For fast LV coverage, only a few short-axis 2D slices with anisotropic resolution in the slice direction are acquired. The slice misalignments and anisotropy may lead to staircase artifacts and erroneous assessment of ventricular volume.² Moreover, due to the anisotropic resolution, reformats to arbitrary orientations are not feasible, requiring multiple acquisitions to be performed in several geometric views and thereby increasing overall planning and scan time.

Recent developments in pulse design enabled simultaneous multi-slice for cardiac imaging.³-⁵ However, simultaneous multi-slice CINE is still limited to multiple breath-holds with anisotropic resolution and specific absorption rate restricting the attainable flip angle. 2D free-breathing CINE imaging with retrospective motion correction⁶-⁸ as well as single breath-hold 2D real-time acquisitions⁹,¹⁰ have been proposed to minimize slice misalignment and improve patient comfort. Recent techniques have been proposed to reconstruct pseudo 3D cardiac CINE datasets from multiple multi-slice anisotropic 2D volumes through the use of motion-corrected super-resolution frameworks.¹¹-¹³ Whereas those techniques have shown potential in recovering 3D CINE images in free-breathing, they require long acquisitions of several low-resolution volumes (~10 minutes), and performance heavily depends on the accuracy of slice-to-slice/volume registrations.

The above-mentioned shortcomings and inaccuracies can be mitigated by employing a single breath-hold 3D cardiac CINE. Accelerated acquisitions with parallel imaging and compressed sensing (CS) reconstruction¹⁴-¹⁸ have enabled this application. The approaches differ in the applied sampling trajectory, such as stack of spirals¹⁴,¹⁶ stack of stars,¹⁷ Cartesian with parallel imaging acceleration,¹⁸ or Cartesian with CS acceleration.¹⁵ As a tradeoff between spatial and temporal resolution for LV coverage, all of these methods provide anisotropic slice resolution in the range of 2.5 to 10 mm within a breath-hold of 10 to 27 seconds. Hence, in most cases reformats to arbitrary orientations are only possible at the cost of reduced image quality.

In contrast, free-breathing 3D methods¹⁹-²₆ provide high isotropic resolution with whole-heart (WH) coverage to enable visualization of small structures (such as coronaries) as well as functional assessment; however, this is achieved at the expense of a prolonged scan time in the order of several minutes. Moreover, these approaches usually require long reconstruction times associated to CS-based algorithms that exploit redundancies in both temporal dimensions (respiration and cardiac movement) and to additional regridding operations in the case of commonly used acquisitions with non-Cartesian trajectories.

We recently proposed a 3D patch-based low-rank reconstruction (PROST)²⁷ that outperforms CS approaches by exploiting redundancies in 3D static images. PROST reconstruction enables operator splitting and is thus scalable to several threads on central processing unit or graphical processing unit, which allows a fast reconstruction. However, PROST does not exploit temporal redundancies. An extension of PROST to high-dimensionality PROST (HD-PROST)²⁸ was introduced to additionally exploit patch-based redundancies among different contrasts in 3D static images but does not again consider temporal redundancies.

In this work, we sought to combine a novel 3D Cartesian acquisition (continuous variable-density Cartesian trajectory with spiral-like order [VD-CASPR] sampling), accompanied with a novel multi-bin PROST (MB-PROST) reconstruction for the application of single breath-hold 3D cardiac CINE. We propose a 3D Cartesian CINE acquisition with isotropic resolution of 1.9 mm³ and novel MB-PROST reconstruction to provide high spatial and temporal resolution images in a single breath-hold of ~20 seconds for LV and ~26 seconds for WH coverage. Shorter breath-hold durations of ~13 to 15 seconds with a slightly anisotropic resolution (1.9 × 1.9 × 2.5 mm) have been studied as well. In the acquisition, we extend our previously proposed VD-CASPR²⁷,²⁹ for continuous sampling. This is achieved with an acquisition-adaptive alternating tiny golden/golden angle increment between spiral arms and a spiral out-inward sampling to provide incoherent and nonredundant sampling within and among cardiac phases while ensuring robustness to eddy current artifacts for high acceleration factors. The sampling is thereby adaptive to the subject’s heart rate and desired spatial and temporal resolution.
temporal resolution. PROST reconstruction is extended to additionally exploit redundancies in the temporal domain with implicit motion alignment among patches at different cardiac phases. The proposed method was evaluated in 10 healthy subjects and 15 patients and compared against previously proposed approaches with CS-based reconstruction. LV functional assessment with the proposed 3D cardiac CINE is compared against measurements from clinical gold standard multi-slice 2D CINE imaging.

2 | METHODS

The proposed method consists of 2 main contributions: a continuous 3D VD-CASPR sampling for CINE acquisition (Figure 1) and a novel multi-bin patch-based low-rank reconstruction (MB-PROST) (Figure 2) to enable 3D cardiac CINE with isotropic resolution in a single breath-hold.

2.1 | Acquisition

The overall temporal course of the acquisition is depicted in Figure 1. VD-CASPR trajectory for 3D cardiac CINE subsamples the phase-encoding plane \( k_y - k_z \) of size \( N_y \times N_z \), where \( N_y \) is the number of \( k_y \) and \( N_z \) is the number of \( k_z \) sampling points. The trajectory samples along a spiral-like arm with alternating angle increment between spiral arms and out-inward sampling. One spiral arm is acquired per cardiac phase at every heartbeat using prospective electrocardiogram triggering. The number of sampling points along each spiral arm \( R \) is computed before the acquisition based on the repetition time \( TR \) and the desired temporal resolution to match the subject’s heart rate (RR interval) with a predefined number of cardiac phases \( N_C \).

\[
R \cdot TR \cdot N_C \leq RR. \tag{1}
\]

The desired acquisition time \( TA \) (limited by breath-hold duration) determines the number of spiral arms \( SP \) to be acquired.

\[
SP = \left\lfloor \frac{TA}{L \cdot R \cdot TR \cdot N_C} \right\rfloor, \tag{2}
\]

where \( L \) is the undersampling factor for each cardiac phase. Samples along the spiral arm are spread out according to a variable-density sampling with an inverse-square relationship. In order to ensure good initialization for the reconstruction and to allow coil sensitivity map estimation from the data, a fully sampled region in the low-frequency range is acquired for each cardiac phase. The size of the fully sampled center is set as percentage \( c \) of all sampling points.

In order to evenly distribute the spiral arms among the cardiac phases while fulfilling (1) interleaved acquisition (i.e., one spiral arm per cardiac phase), (2) incoherent sampling for sampling masks of each cardiac phase, and (3) minimal gradient switching for high-frequency samples

FIGURE 1  ECG-triggered 3D CINE variable-density Cartesian acquisition with spiral profile ordering (VD-CASPR) sampling. A continuous acquisition with out-inward sampling and acquisition-adaptive alternating tiny golden \( \Psi_M \)/golden angle \( \Psi_1 \) increment throughout the cardiac cycle ensures incoherent aliasing within and between cardiac phases with minimal influence of eddy currents. The order \( M \) of the tiny golden angle ensures angle increment to equally space all acquired spiral arms among the cardiac phases depending on the chosen acceleration factor and heart rate. After \( H \) heartbeats (bottom row), each cardiac phase is sampled with a prescribed acceleration factor \( L \) while also providing a fully sampled center region. ECG, electrocardiogram; VD-CASPR, variable-density Cartesian trajectory with spiral-like order
where \( x, y, z \times V \), we seek to reconstruct the 3D multi-cardiac phase complex image \( X \in \mathbb{C}^{N_x \times N_y \times N_z \times N_C} \), where \( N_x, N_y, N_z \) represent the number of voxels in the spatial directions \( x, y, z \), \( N_C \) being the cardiac phases, and \( C \) being the number of receiver channels.

To enhance sampling density, a spatiotemporal soft-weighting between neighboring cardiac phases is applied under the assumption that homogeneous regions, that is, low-frequency components, vary slowly.\(^{31}\) For this, the sampling region in the \( k_y - k_z \) plane is divided into \( K \) concentric areas with varying radius \( r_k, k \in [1, K] \). Each area gets assigned a different view-sharing factor \( V_k \in \left[ 0, \frac{N_C}{2} \right] \) indicating how far in the temporal direction the neighboring cardiac phases can contribute samples to the current cardiac phase, as illustrated in Figure 2. For each position \( k_y - k_z \), the soft-weight \( W \) is given by a Gaussian function centered at the current cardiac phase, with SD given by \( V_k \). Spatially coinciding view-sharing samples are averaged.

Coil sensitivity maps \( S \in \mathbb{C}^{N_x \times N_y \times N_z \times C \times \mathbb{R}} \) are obtained via ESPRIT\(^{32}\) from the time-averaged (across all cardiac phases) central fully sampled k-space data.

MB-PROST assumes that 3D cardiac CINE imaging contains a rich amount of redundancy on a local (intensities within a patch), nonlocal (between patches within a spatial neighborhood), and temporal (between cardiac phases or bins) scale. This information can be exploited through patch-based matrix representation and singular value decomposition to regularize the overall MR reconstruction problem, yielding.

\[
\arg \min_{x,t} \frac{1}{2} \| AWFSX - Y \|_F^2 + \lambda_p \sum_p \| T_p \|_s, s.t. T_p = P_{p,t}(X), t \in [1, N_C]
\]

\( \lambda_p \) balances the data fidelity and smoothness terms.

2.2 | Reconstruction

From the acquired subsampled k-space \( Y \in \mathbb{C}^{N_x \times N_y \times N_z \times N_C \times C} \), we seek to reconstruct the 3D multi-cardiac phase complex image \( X \in \mathbb{C}^{N_x \times N_y \times N_z \times N_C} \), where \( N_x, N_y, N_z \) represent the number of voxels in the spatial directions \( x, y, z \), \( N_C \) being the cardiac phases, and \( C \) being the number of receiver channels.

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\]

\( \lambda_p \) balances the data fidelity and smoothness terms.
where $A$ describes the subsampling operator; $F$ denotes the Fourier transform; $\| \cdot \|_F$ is the Frobenius norm; $\lambda_P > 0$ is the sparsity-promoting regularization parameter for patch $P$; and $\| \cdot \|_*$ is the nuclear norm enforcing low rank on a multi-bin patch scale. In a first step (3D block matching + patch search in Figure 2), the patch selector $P_{P,i}(\cdot)$ extracts 3D patches similar to a reference patch centered at voxel $P$, within a given spatiotemporal search window (yellow shaded rectangle in Figure 2), among all cardiac phases or bins $t \in [1, N_C]$. The reference patch is selected from the first cardiac phase (here $t = 1$). Assuming that similar structures are existing in different cardiac phases, but at different spatial locations, the search for similar patches in the spatiotemporal window enables (1) exploiting temporal redundancies with (2) an implicit motion-aligned content. This is different to PROST, which searches for similar 3D patches only in a spatial window for each cardiac phase independently. Similar patches are chosen based on minimal mean-squared error. The size of the search window and the amount of patches to be selected can be adjusted, which influences the achievable low rank as well as the reconstruction time. This process is repeated for all voxels $P$.

Subsequently (vectorization and low-rank decomposition in Figure 2) and similarly to PROST, patches are vectorized and concatenated to form a low-rank 2D matrix $T_P$. Equation (4) can be solved by reformulating the problem to an unconstrained Lagrangian cost function.28

$$\arg \min_X \frac{1}{2} \| EX - Y \|_F^2 + \lambda_P \sum_P \| T_P \|_* + \frac{\mu}{2} \sum_P \left\| T_P - P_{P,i}(X) - \frac{b_P}{\mu} \right\|_F^2,$$  

(5)

where $b_P$ is the Lagrangian multiplier for patch $P$; $\mu > 0$ is the penalty parameter; and $E = AWFS$ describes the encoding operator. Equation (5) can be efficiently solved through operator-splitting via alternating direction method of multipliers27 by iteratively alternating the minimization between (1) an $\ell_2$ norm data consistency for the multi-bin set of images $X$, and (2) a low-rank patch-based matrix denoising $T_P$, followed by (3) an update of the augmented multiplier $b_P$.

Afterward, the denoised patches are back-aggregated into their spatial positions in the image (multi-bin aggregation in Figure 2). The principal schematics of MB-PROST are depicted in Figure 2, and a description of the algorithm is provided in Supporting Information Table S1.

2.3 | From PROST over HD-PROST to MB-PROST

3D PROST reconstruction was initially proposed to reconstruct mid-diastolic 3D coronary images by exploiting 3D patches spatial redundancies, showing that it outperforms CS reconstruction.27 Applying PROST to 3D cardiac CINE images corresponds to reconstructing each cardiac phase independently without considering patch-based temporal redundancies or implicit motion alignment between patches. HD-PROST28 was introduced to additionally exploit patch-based redundancies among different contrasts through a high-order tensor decomposition, but it does not again consider patch-based temporal redundancies or implicit motion alignment between patches and thus can be only applied to static images. The proposed MB-PROST jointly reconstructs motion-resolved multi-bin 3D images by adding an implicit temporal constraint and thus taking advantage of the common features present in adjacent cardiac phases through multi-bin patch selection. Global and local low-rank approaches have been previously proposed to exploit redundancies in the temporal cardiac dimension.34-37 A comparison of the spatiotemporal compression achieved with global low-rank, local low-rank, PROST, and the proposed MB-PROST is examined in Supporting Information Figure S2, showing that higher compression can be achieved with MB-PROST.

2.4 | Implementation details

Reconstruction parameters for the proposed MB-PROST technique were empirically optimized on several data-sets (not reported here) and were maintained for all reconstructions. View-sharing factors among neighboring phases were set to $V_1 = 2, V_2 = 1.5, V_3 = 1.25, V_4 = 1$. The degree of redundant structural information within each patch is controlled by the size of the patches and was set to $5^3$ voxels. Similar 3D patches were searched in a $3D \times N_C$ window of size $60^3 \times N_C$. Twenty similar patches were selected within this spatiotemporal search window. The truncation parameter $\lambda_P$ and regularization factor $\mu$ were set to 0.1 and 0.3, respectively. A fixed number of 5 alternating direction method of multipliers iterations and 7 conjugate gradient iterations were performed. MB-PROST reconstruction was implemented in MatLab (v9.6, MathWorks, Natick, MA), with the multi-bin 3D patch-based denoising implemented in C (MEX) and distributed over 12 CPUs (2.3 GHz Intel Xeon E5-2697) to provide fast computational times.

2.5 | In vivo study

Imaging was performed on a 1.5 tesla MRI (MAGNETOM Aera, Siemens Healthcare Erlangen, Germany) equipped with 18-channel body and 32-channel spine coils. Written informed consent was obtained from all subjects, and the study was approved by the local ethics committee.
The proposed prospectively electrocardiogram-triggered 3D Cartesian balanced steady-state free precession (bSSFP) sequence with CINE VD-CASPR sampling was acquired in 10 healthy subjects for LV coverage (4 female, age = 29 ± 3 years) and in 4 healthy subjects for WH coverage (2 female, age = 27 ± 1 years). For each subject, 2 acquisitions were performed in short-axis (SA) orientation with isotropic and slightly anisotropic resolution (comparable to Ref. 15). Isotropic resolution was 1.9 mm³, covering a FOV of 300 × 270 × 100 mm³ (LV) and 300 × 270 × 130 mm³ (WH). Slightly anisotropic resolution was 1.9 × 1.9 × 2.5 mm, covering the same FOVs as before. In one subject (subject C), an additional isotropic 3D CINE WH coverage scan was performed in coronal orientation to investigate reformating agreement with the short-axis scans. Remaining imaging parameters included: TE = 1.3 ms, TR = 2.6 ms, flip angle α = 39°, bandwidth = 1042 Hz/px, phase oversampling = 15%, slice oversampling = 20%, and fully sampled k-space center c = 15%. In each acquisition, Nc = 16 cardiac phases were acquired, and the number of segments per spiral arm R = 16–28 was adapted to fit within the subject’s cardiac cycle (50–91 BPM, 71 ± 12 BPM), which determined the temporal resolution R·TR = 42 to 72 ms (50 ± 4 ms) according to Equation (1). Acquisition times for LV were 20 ± 2 seconds with an acceleration of L ≈ 10 (isotropic)/8 (anisotropic), and for WH were 26 ± 1 seconds with acceleration of L ≈ 14 (isotropic)/10 (anisotropic) per cardiac phase.

The proposed 3D Cartesian CINE was also acquired in 15 patients (5 female, age = 47 ± 14 years), with LV coverage and slightly anisotropic resolution of 1.9 × 1.9 × 2.5 mm in a single breath-hold of ~13 to 15 seconds (L ≈ 10). Temporal resolution of 28 to 47 ms (39 ± 6 ms) with Nc = 16 acquired cardiac phases (49–90 BPM, 69 ± 14 BPM) was slightly higher than for the healthy subject acquisitions. Remaining imaging parameters were the same.

A multi-slice SA 2D balanced steady-state free precession CINE acquisition with retrospective gating and 2x GRAPPA acceleration was performed for all subjects. 2D CINE was acquired in 8 breath-holds of 15 s duration (2 slices per breath-hold), each with 20 seconds pause in between, resulting in an acquisition time of 4 minutes 20 seconds. 2D CINE acquisition parameters included: resolution of 1.9 × 1.9 mm in-plane (acquired and reconstructed), slice thickness 8 mm, temporal resolution ~40 ms, 20 cardiac phases (reconstructed), TE = 1.06 ms, TR = 2.12 ms, flip angle α = 52°, bandwidth = 915 Hz/px, and similar FOV coverage as for the proposed 3D CINE.

## 2.6 | Evaluation

The proposed MB-PROST is compared against two different CS¹⁵,¹⁹ with spatial and temporal regularization, PROST²⁷, and HD-PROST²⁸ reconstructions for the LV isotropic 3D CINE scans in healthy subjects. The first CS reconstruction (in the following denoted as CS-W) uses a fast iterative shrinkage-thresholding algorithm optimization for nonlinear SENSE and ℓ₁-regularized spatial and temporal single-level Haar wavelets. Spatial and temporal regularization parameters were optimized and fixed to λs = 0.001 and λt = 0.005, coinciding with those proposed in Ref. 15. The second CS reconstruction (in the following denoted as CS-TV) employs a total variation (TV) regularization along the cardiac temporal direction, with λt = 0.005 and ℓ₁-regularized spatial wavelets with λs = 0.001. Regularization parameters were carefully optimized for the datasets and are comparable to those employed in previously proposed CS reconstructions.¹⁹ CS reconstruction parameters were the same for all reconstructions reported in this work. Reconstruction parameters for PROST and HD-PROST were chosen to be the same as for MB-PROST.

Image quality was quantitatively assessed by edge sharpness (ES) and contrast ratios (CR) in end-systolic and end-diastolic cardiac phases of the LV coverage 3D CINE scans (isotropic and slightly anisotropic). ES was measured by Nl = 30 line profiles between right ventricle blood pool and epicardium (15 lines), as well as between left ventricle blood pool and endocardium (15 lines). ES is then defined as the mean inverse distance between the positions p_l,20 and p_l,80 of the 20% and 80% maximum values along the line profile l, similar to Ref. 38

\[
ES = \frac{1}{N_l} \sum \frac{1}{|p_{l,80} - p_{l,20}|}.
\]

Line profiles in 2D CINE and 3D CINE (MB-PROST and CS-W reconstructions) were drawn at similar anatomical positions. The higher the value of ES, the higher the sharpness. CR was measured as:

\[
CR = \frac{\text{mean (ROI}_{\text{myocardium}})}{\text{mean (ROI}_{\text{blood}})}.
\]

with blood pool regions of interest (ROI) drawn in LV and right ventricle blood pool for 2D CINE and 3D CINE with MB-PROST and CS-W reconstruction methods. Statistical significance for ES and CR were determined with a paired Welch’s t test (significance level of P < .05) and Bonferroni correction under the null hypothesis of equal means for unequal variances.³⁹

Quantitative assessment was conducted for LV function parameters: end-systolic volume (ESV), end-diastolic volume (EDV), and ejection fraction (EF). LV epicardial and endocardial segmentation masks of all (isotropic and slightly anisotropic) 3D CINE LV and 2D CINE acquisitions were automatically determined with the Segment software (version 7917)⁴⁰ and afterward were manually corrected when needed.
3 | RESULTS

3.1 | Comparison of reconstruction methods

A comparison of the proposed MB-PROST reconstruction to PROST (each cardiac phase reconstructed independently) is depicted in Figure 3 and against CS-W in Supporting Information Figure S3 for the isotropic LV acquisition of 4 representative healthy subjects. In Supporting Information Figure S4, MB-PROST is compared against HD-PROST (all nonaligned cardiac phases reconstructed simultaneously), PROST, CS-TV, and CS-W. Best (subject I), average (subjects A and C), and worst (subjects D and F) observed reconstruction results in healthy subjects are depicted. Residual aliasing and blurring are observed in the CS-W images, which can be reduced by the locally low-rank PROST and HD-PROST reconstruction. However, remaining artefacts are observed both in PROST and HD-PROST. PROST does not exploit temporal redundancies in the cardiac dimension, thus limiting the acceleration factor. HD-PROST was not designed to exploit temporal redundancies, thus as expected exploiting redundancies in the temporal dimension of nonaligned cardiac images results in remaining aliasing and blurring. Residual motion-related aliasing is reduced with MB-PROST that considers implicitly motion-aligned patches. Consistent good image quality throughout the cardiac cycle can be appreciated in the temporal profiles with MB-PROST. It shall be noted that CS-W and CS-TV were able to reconstruct comparable results to PROST-type reconstructions in some subjects. Supporting Information Video S1 illustrates the temporal phases of the various reconstruction methods. The overall reconstruction times for a 3D CINE data (isotropic LV coverage) with MB-PROST and CS were ~10 to 15 minutes and ~5 minutes, respectively. The 2D CINE was reconstructed on the scanner with the vendor-provided GRAPPA reconstruction within a few seconds per slice.

3.2 | Proposed 3D CINE reconstruction in comparison to 2D CINE imaging

A midventricular slice in systole and diastole of all healthy subjects is depicted in Figure 4 and Supporting Information Figure S5 for the proposed 3D CINE (isotropic LV coverage scan), with MB-PROST reconstruction in comparison to the conventional 2D CINE scan. 3D CINE (isotropic LV coverage) images show comparable image quality to an anisotropic 2D CINE acquisition, but with a 13-fold shorter acquisition time. The corresponding temporal evolution of the 2D and 3D cardiac CINE for all subjects is shown in Supporting Information Video S2.

The temporal behavior of the proposed 3D CINE approach (isotropic LV coverage scan) for every fourth cardiac phase in 3 representative SA slices is shown in Figure 5 in comparison to the conventional 2D CINE scan at similar slice locations for a healthy subject. All cardiac phases for the same subject are included in Supporting Information Video S3. High spatial and temporal resolution can be achieved with the proposed 3D CINE approach, with higher through-plane resolution (52 slices for isotropic 3D LV coverage scan) in contrast to 2D CINE (16 slices).

FIGURE 3  Reconstruction comparison of isotropic 3D CINE (LV coverage scan) in 4 healthy subjects for MB-PROST and PROST. All images were reconstructed from the same dataset acquired with the proposed VD-CASPR sampling. Midventricular slices in short axis for end-systolic and end-diastolic phase are depicted. A line through the largest diameter of the left ventricle in the same short axis slice shows the left and right ventricular function over time. MB-PROST provides most consistent qualitative results. A comparison against compressed sensing with spatiotemporal wavelets (CS-W) is depicted in Supporting Information Figure S3. Temporal behavior of all subjects and reconstruction methods are depicted in Supporting Information Video S1. LV, left ventricle.
The isotropic resolution of the proposed 3D CINE approach allows subsequent reformats in arbitrary views with high spatial resolution. Figure 6 depicts a 2-chamber and 4-chamber reformat from the 3D CINE (isotropic LV coverage) acquisition in comparison to corresponding 2D CINE scans. Consistent high spatial resolution of the 3D CINE in any orientation and over all cardiac phases is shown in Supporting Information Video S4.

3.3 | 3D CINE with isotropic and anisotropic resolution

A comparison of the proposed isotropic 3D resolution to the through-plane slightly anisotropic acquisition (similar to Ref. 15) is demonstrated in Figure 7 for LV and WH coverage, respectively. Due to the same in-plane resolution of the isotropic and slightly anisotropic scan, no significant difference in SA between the 4 acquisitions can be observed. Improved edge delineation through-plane because of the isotropic resolution was observed for the 2-chamber and 4-chamber reformats. The isotropic resolution allows acquisition in arbitrary orientation with subsequent reformatting to the desired angulation. Supporting Information Figure S6 compares the isotropic WH scan natively acquired in SA and in coronal orientations for various oblique reformats. Good visual agreement was observed between SA and coronal orientation acquisitions.

In Figure 8, three representative patient acquisitions are shown for the slightly anisotropic 3D CINE in comparison to the conventional anisotropic 2D CINE. The patients shown reflect the spectrum of challenges encountered in the clinic (arterial fibrillation, strongly varying heart rate) including good exams (constant and low heart rate). In all cases, good agreement of 3D CINE to 2D CINE was achieved.

3.4 | Image analysis and LV functional assessment

A quantitative comparison by line (ES) and region of interest (CR) analysis is given in Figure 9 for the proposed 3D CINE with LV coverage (isotropic and slightly anisotropic) in healthy subjects. ES was significantly improved in 3D CINE with MB-PROST reconstruction compared to 3D CINE with CS-W. ES was highest in conventional 2D cardiac CINE, but no statistically significant difference to the proposed 3D CINE was observed. An increased SD for the anisotropic 3D CINE acquisition in contrast to the isotropic one was observed. ES in systolic phases was on average slightly lower compared with diastolic phases. ES over all cardiac phases and healthy subjects in 2D CINE and 3D CINE (isotropic
with MB-PROST reconstruction) is depicted in Supporting Information Figure S7. ES follows a consistent trend over all cardiac phases between 2D CINE and 3D CINE. Higher temporal resolution in 2D CINE results in less mean deviation for diastolic phases. CR measurements revealed similar trends. CR of 2D CINE was found to be the highest, but not statistically significant different to isotropic 3D CINE with MB-PROST. CR of MB-PROST reconstructions were significantly higher than CS-W in diastolic phases.

The Bland-Altman plots of LV function parameters in Figure 10 reveal high agreement between 3D CINE (healthy subjects: isotropic and slightly anisotropic LV coverage scan; patient: slightly anisotropic LV coverage scan) with MB-PROST and 2D CINE. On average, a bias for ESV, EDV, and EF of $-0.1\, \text{mL}$, $-0.9\, \text{mL}$, and $-0.1\%$, respectively, was observed for the proposed isotropic 3D CINE, with all observations lying inside the confidence interval for ESV, EDV and EF of $\pm 1.7\, \text{mL}$, $\pm 1.2\, \text{mL}$, and $\pm 1.2\%$ (10 healthy subjects), respectively. The slightly anisotropic 3D CINE showed larger deviations for all LV function parameters with a bias of $-0.1\, \text{mL}$, $-0.8\, \text{mL}$, and $-0.3\%$ (10 healthy subjects + 15 patients), and most measurements lying inside the $\pm 4.7\, \text{mL}$, $\pm 3.2\, \text{mL}$, and $\pm 4.2\%$ (10 healthy subjects + 15 patients) confidence interval for ESV, EDV, and EF, respectively. In the
10 healthy subjects, confidence intervals for ESV, EDV, and EF were ±6.1 mL, ±3.8 mL, ±5% with a bias of −0.3 mL, −0.4 mL, 0.3%, respectively. In the 15 patients, confidence intervals for ESV, EDV, and EF were ±3.7 mL, ±2.6 mL, and ±3.6%, with a bias of −0.1 mL, −0.8 mL, and −0.3%, respectively.
In this work, we present an isotropic 3D CINE acquisition within a single breath-hold of ~20 seconds for LV and ~26 seconds for WH coverage together with a novel multi-bin 3D patch-based low-rank reconstruction (MB-PROST). Shorter breath-hold durations of ~13 to 15 seconds with slightly anisotropic resolutions for LV coverage were studied in a patient cohort. MB-PROST exploits redundancies in the cardiac dimension by considering implicit motion alignment between patches of distinct cardiac phases. The isotropic nature of the acquisition allows reformats in arbitrary views (SA, 2-chamber, 4-chamber) without loss of resolution and does not demand a double-oblique imaging orientation simplifying workflow and reducing planning time, which in turn increases patient comfort. The proposed 3D CINE provides thus a potential alternative to 2D CINE in clinical routine.

The proposed VD-CASPR 3D CINE sampling scheme guarantees incoherent undersampling within and between cardiac phases, which is beneficial for the reconstruction. Based on the interleaved acquisition order fulfilling the alternating angle increments, a more careful sampling mask preparation is required instead of rotating a prepared mask of one cardiac phase, as for example with spiral phyllotaxis sampling patterns. A comparison for rotation of (1) a fixed sampling pattern, (2) fixed tiny golden angle increment, and (3) the proposed sampling is shown in Supporting Information.

![Figure 10](image_url) Extracted ventricular function parameters, (A) ESV, (B) EDV, and (C) EF for isotropic (in 10 healthy subjects) and slightly anisotropic (in 10 healthy subjects and 15 patients) 3D CINE in comparison to conventional 2D CINE. Stated values for slightly anisotropic 3D CINE depict the pooled bias and confidence intervals of healthy subjects and patients. EDV, end-diastolic volume; EF, ejection fraction; ESV, end-systolic volume.

## 4 | DISCUSSION

In this work, we present an isotropic 3D CINE acquisition within a single breath-hold of ~20 seconds for LV and ~26 seconds for WH coverage together with a novel multi-bin 3D patch-based low-rank reconstruction (MB-PROST). Shorter breath-hold durations of ~13 to 15 seconds with slightly anisotropic resolutions for LV coverage were studied in a patient cohort. MB-PROST exploits redundancies in the cardiac dimension by considering implicit motion alignment between patches of distinct cardiac phases. The isotropic nature of the acquisition allows reformats in arbitrary views (SA, 2-chamber, 4-chamber) without loss of resolution and does not demand a double-oblique imaging orientation simplifying workflow and reducing planning time, which in turn increases patient comfort. The proposed 3D CINE provides thus a potential alternative to 2D CINE in clinical routine.

The proposed VD-CASPR 3D CINE sampling scheme guarantees incoherent undersampling within and between cardiac phases, which is beneficial for the reconstruction. Based on the interleaved acquisition order fulfilling the alternating angle increments, a more careful sampling mask preparation is required instead of rotating a prepared mask of one cardiac phase, as for example with spiral phyllotaxis sampling patterns. A comparison for rotation of (1) a fixed sampling pattern, (2) fixed tiny golden angle increment, and (3) the proposed sampling is shown in Supporting Information.
Figure S1B. A more evenly distributed and less redundant sampling is achieved with the proposed trajectory. Moreover, lower side-band energy and larger maximal incoherence are obtained, as seen in Supporting Information Figure S1C. The out-inward sampling together with a tiny golden increment has been shown before to minimize the effect of eddy currents,\(^\text{23,25}\) which was also empirically observed in phantom measurements (Supporting Information Figure S8) in our study. Cartesian acquisitions can be intrinsically more sensitive to motion than non-Cartesian approaches but provide a uniform resolution throughout the entire FOV while also allowing for arbitrary, noncubic FOV dimensions and enable faster reconstruction algorithms. Recently, faster non-Cartesian reconstructions were proposed\(^\text{41}\), which, however, are still slower than their Cartesian counterparts.

The implicit motion alignment of patches over all cardiac phases with MB-PROST, together with soft-weighting among neighboring cardiac phases, exploits spatial and temporal redundant information. Our results show that in the majority of cases, MB-PROST clearly outperforms PROST, CS-W, and CS-TV reconstructions. For some cases in which less fat-related aliasing occurred, similar performance was observed between MB-PROST and CS-W/CS-TV. The implicit motion alignment in MB-PROST provides a strong reduction in motion-related aliasing of the highly accelerated acquisitions. CS-TV reconstruction with total variation reduction in motion-related aliasing of the highly accelerated acquisitions, similar to an XD-GRASP reconstruction,\(^\text{19}\) effectively smoothes out aliasing artifacts. This can come at the cost of a blurred temporal evolution, as shown in Supporting Information Figure S4. The comparison to CS-W and CS-TV only focused on the reconstruction side and did not include the proposed acquisitions in Ref. 15,19. Our experiments therefore only suggest that the proposed highly undersampled Cartesian trajectory used in our work is less suited for CS-W and CS-TV reconstructions.

The inclusion of an explicit motion alignment (based on image registration) in a locally low-rank reconstruction has been recently shown for 2D cardiac CINE.\(^\text{34}\) However, as shown in Supporting Information Figure S2 and in Ref. 28, the combination of compression on local and nonlocal scales (as in PROST/HD-PROST) outperforms local low rank methods by exploiting nonlocal redundant image content. Thus, the combination of HD-PROST with explicit motion alignment could be of interest and will be investigated in future work. The proposed reconstruction shares some similarities with the PRICE algorithm\(^\text{35}\) and with the method in Ref. 42, which uses a weighted patch-based regularization framework for the reconstruction of undersampled 2D cardiac CINE MR images. Although their algorithm relies on the selection of a weighted \(l_p\) distance to promote sparsity, as in CS, here we exploit the intrinsic spatiotemporal low-rank property of self-similarity groups (built from similar 3D patches) to perform an efficient and robust 3D + time denoising.

The implicit motion alignment in MB-PROST provides the best and most consistent results throughout the entire cardiac cycle. In 1 subject, we compared MB-PROST to HD-PROST reconstruction (i.e., jointly reconstructing all cardiac phases) without any explicit motion alignment in the temporal direction. A larger sharing dimension can potentially enable a better recovery, but the lack of motion alignment and/or correction leads to a higher rank as in PROST or MB-PROST and hence is not feasible for this CINE application. MB-PROST is not limited to reconstruction of cardiac motion-resolved CINE images but could also be utilized for reconstructing any other dynamic datasets, such as cardiac MR fingerprinting\(^\text{43}\) or cardiac 4D flow. Future work will also investigate its potential extension to multi-contrast dynamic imaging.

Adipose tissue and epicardial fat can introduce strong aliasing in highly accelerated 3D cardiac CINE. Fat suppressed acquisitions, such as water-selective excitation or fast interrupted steady-state\(^\text{44-47}\) can help to reduce the impact of fat, but they demand longer acquisition times for similar parametrization and acceleration. Fat-suppression approaches for the proposed 3D CINE acquisition were explored in 4 subjects\(^\text{48}\) but the extended breath-hold duration or increased acceleration factors required to maintain the similar acquisition time did not justify the reduced fat-related aliasing for single breath-hold CINE. In some subjects and patients, residual fat-related aliasing can be observed with its severity depending on the angulation of the FOV. Acquisition in nonoblique views may provide a solution.

In this study, we do not report a contrast-to-noise ratio analysis because the proposed denoising-based reconstruction can bias the results. A fairer comparison to CS and 2D CINE is hence achieved by computing CRs. The CR in 3D CINE was on average slightly lower than in 2D CINE. Whereas MB-PROST CR values were not significantly different to 2D CINE, CS-W CR values were significantly worse to 2D CINE and MB-PROST. These differences in contrasts are expected because the maximum achievable flip angle within specific absorption rate limitations of 3D CINE is lower than in 2D CINE. Moreover, the saturation of the ventricle blood pool and inflow of saturated blood into the atrial pool by the 3D slab-selective excitation affect the contrast of 3D cardiac CINE imaging. In contrast, the slice-selective 2D CINE can benefit from inflow of unsaturated blood. However, the obtained contrast is still comparable and sufficient for reliable extraction of ventricular functions. Similar observations hold for ES measurements.

Isotropic resolution scans allow acquisition in nonoblique orientations and reformatting into arbitrary orientations after acquisition, reducing planning time. For this study, the proposed 3D CINE was acquired in SA to allow an easier and fairer comparison to the conventional 2D CINE acquired in SA. Moreover, less blurring in reformats along
the slice direction was observed with isotropic 3D CINE in comparison to the corresponding anisotropic acquisitions (Figures 6 and 7). Improved ES and CR were achieved in isotropic acquisitions in comparison to the corresponding anisotropic acquisitions for both systolic and diastolic phases. However, SDs for CRs in anisotropic resolutions were smaller and comparable to 2D CINE, benefitting from the thicker slices. The retrieved ventricular functions showed less variation with tighter confidence intervals for the isotropic resolution scans. Nevertheless, the anisotropic 3D CINE still provided reliable results within confidence intervals.

For the patient study, spatial resolution was traded for temporal resolution and faster acquisition. A slightly anisotropic resolution of $1.9 \times 1.9 \times 2.5$ mm ($\sim$13-15 seconds) is feasible and provided satisfactory results. Due to some patients’ inability to cope with longer breath-holds, faster acquisitions should be favored.

WH coverage is possible with the proposed approach but demands an extended breath-hold duration of $\sim$26 seconds, which is not clinically feasible. Incomplete or unstable breath-holds can result in less consistent ventricular function parameters compared to LV. As indicated by the patient study, shorter breath-hold durations of $\sim$13 to 15 seconds are feasible for LV coverage. Future studies will thus investigate further reduction of scan time for WH coverage or LV isotropic resolution in $\sim$15 seconds. This may be accomplished by increasing the acceleration factor, which may demand a different reconstruction, such as a deep learning-based reconstruction, due to the increased aliasing. If subjects are unable to hold their breath for 15 seconds, an extension to free-breathing acquisitions with a periodic sampling of the k-space center line for respiratory self-gating is conceivable. The prospective electrocardiogram triggering could then be changed to a retrospective triggering/gating. The proposed sampling scheme used in this study already intrinsically supports this functionality. However, free-running acquisitions demand proper handling of fat suppression to mitigate fat-related motion aliasing.

5 | CONCLUSION

In this work, we present an isotropic 3D Cartesian CINE acquisition with an efficient multi-bin patch-based implicitly motion-aligned reconstruction. The proposed 3D cardiac CINE imaging can be acquired within a single breath-hold of $\sim$20 seconds for LV coverage and in $\sim$26 seconds for WH coverage, showing good agreement with clinical 2D cardiac CINE scans in terms of left ventricular functional assessment. Moreover, the proposed 3D CINE approach enables data acquisitions in a nonoblique orientation with reformatting to arbitrary orientation, which greatly reduces planning times.

Acknowledgment

This work was supported by the following grants: EPSRC EP/P032311/1, EP/P001009/1, and EP/P007619/1; Wellcome EPSRC Centre for Medical Engineering (NS/A000049/1); and the Department of Health via the National Institute for Health Research (NIHR) Cardiovascular Health Technology Cooperative (HTC) and comprehensive Biomedical Research Centre, awarded to Guy’s & St Thomas’ NHS Foundation Trust in partnership with King’s College London and King’s College Hospital NHS Foundation Trust. The views expressed are those of the author(s) and not necessarily those of the NIHR or the Department of Health and Social Care.

Conflicts of Interest

We declare that the author Radhouene Neji is an employee of Siemens Healthcare.

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overspread functions (PSF) for the respective sampling patterns. C, Central line through point sampling points of all cardiac phases, i.e. \#sampled = 16 (red)

angle increment (right). Masks show the summed up sampled temporal redundancies. In Proceedings of the European Society for Magnetic Resonance in Medicine (ESMRMB); 2019; Rotterdam, The Netherlands.

**SUPPORTING INFORMATION**

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

**FIGURE S1** A, Variable-density CArtesian acquisition with Spiral PRofile ordering (VD-CASPR) trajectory for a sampling space of size \( N_y = 48, N_z = 24 \). SP = 2 exemplary spiral-like arms are depicted with \( R = 16 \) sampling points along each spiral arm. The fully sampled elliptical central region \( c = 15\% \) is colored in blue. B, Impact of interleaved CINE sampling for \( N_y = 160, N_z = 52, L = 10, R = 20, SP = 528 \). \( N_C = \) 16 cardiac bins for (i) a fixed sampling pattern with \( SP = 33 \) is rotated \( N_C = 16 \) times by alternating tiny golden \((23.62°)/\text{golden}(111.24°)\) increment (left); (ii) proposed sampling but with a fixed tiny golden angle \((23.62°)/\text{golden angle increment (middle); and (iii) proposed sampling with acquisition-specific tiny golden angle \((27.2°)/\text{golden angle increment (right). Masks show the summed up sampling points of all cardiac phases, i.e. \#sampled = 16 (red) indicates maximal redundancy. C, Central line through point spread functions (PSF) for the respective sampling patterns. The shaded area indicates the side-band energy \( E \) (calculated over \( k_x \) and \( k_z \): the lower the better) and the achievable coherence \( \mu \) (maximum of side-band; the lower the better).

**FIGURE S2** Compression comparison between several low-rank techniques. A, A reference patch (red box) is used to find 20 similar patches and to build a low-rank self-similarity matrix. In locally low-rank (LLR) this matrix is created by collecting all patches located at the same spatial position than the reference patch in all the cardiac phases. In PROST, the most similar patches are collected spatially in one single cardiac phase. MB-PROST selects those patches by looking in both spatial and cardiac phase dimensions. For the three techniques, the reference patch is vectorized and placed in the first column of each self-similarity matrix (red lines). Global low-rank (GLR) which does not take into account redundancies on a patch scale is also shown. B, The normalized singular values are shown for the four techniques and show a faster decay with an effective rank of 1 for MB-PROST.

**FIGURE S3** Reconstruction comparison of isotropic 3D CINE (LV coverage scan) in four healthy subjects for MB-PROST, PROST and compressed sensing with spatio-temporal wavelet regularization (CS-W). All images were reconstructed from the same dataset acquired with the proposed VD-CASPR sampling. MB-PROST provides most consistent qualitative results. Mid-ventricular slices in short axis for end-systolic and end-diastolic phase are depicted. A line through the largest diameter of the left ventricle in the same short axis slice shows the left and right ventricular function over time. Temporal behaviour is depicted for all subjects in Supporting Information Video S1.

**FIGURE S4** Reconstruction comparison of isotropic 3D CINE in two healthy subjects for MB-PROST, HD-PROST, PROST, compressed sensing with spatial wavelet and temporal total variation regularization (CS-TV) and compressed sensing with spatio-temporal wavelet regularization (CS-W). All images were reconstructed from the same dataset acquired with the proposed VD-CASPR sampling. The best and worst observed healthy subject is depicted. MB_PROST performs consistently well in all subjects. Mid-ventricular slice in short axis for end-systolic and end-diastolic phase are depicted. A line through the largest diameter of the left ventricle in the same short axis slice shows the left and right ventricular function over time.

**FIGURE S5** End-systolic and end-diastolic images of five healthy subjects of the proposed single breath-hold isotropic 3D CINE (LV coverage scan) with MB-PROST reconstruction in comparison to a conventional multiple breath-held 2D cardiac CINE scan. Good agreement of 3D CINE to 2D CINE was achieved. Temporal behaviour is depicted in Supporting Information Video S2.

**FIGURE S6** Qualitative comparison of reformats in two different isotropic 3D CINE acquisitions of short axis (SA) orientation and coronal orientation (bottom) with whole-heart (WH) coverage in healthy subject C. Isotropic resolution enables acquisition in arbitrary orientation with reformatting to multiple views (e.g., 2 chamber, 4 chamber, SA and coronal). For SA acquisition, the coronal reformat shows high agreement with the coronal image of the coronal acquisition (bottom) and vice versa for the SA scans.

**FIGURE S7** Quantitative edge sharpness (ES) analysis for line profiles in conventional 2D cardiac CINE and isotropic 3D CINE with MB-PROST over all cardiac phases (2D CINE: 20 phases, 3D CINE: 16 phases) and healthy subjects. ES is measured in lines through right ventricle and myocardium orthogonal to epicardium.

**FIGURE S8** Phantom measurements to examine impact of trajectory (outward only or out-inward) and angle increment (golden/golden angle or tiny golden/golden angle). The
proposed sampling pattern (right column) is not affected by eddy currents as indicated by the arrows. The out-inward trajectory provides strongest artefact reduction.

**TABLE S1** Proposed multi-bin PROST (MB-PROST) reconstruction algorithm

**VIDEO S1** Cardiac motion-resolved images of isotropic 3D CINE in four healthy subjects for MB-PROST, PROST and compressed sensing with spatio-temporal wavelet regularization (CS-W). All images were reconstructed from the same dataset acquired with the proposed VD-CASPR sampling.

**VIDEO S2** Cardiac motion-resolved images of all ten healthy subjects of the proposed isotropic 3D CINE with MB-PROST reconstruction in comparison to a conventional 2D CINE.

**VIDEO S3** Cardiac motion-resolved images of isotropic 3D CINE and conventional 2D CINE in healthy subject F. Similar anatomical slice locations to 2D CINE have been selected for 3D CINE, i.e. not all 3D CINE slices are shown.

**VIDEO S4** Qualitative comparison of cardiac motion-resolved reformats (2 chamber and 4 chamber) in isotropic 3D CINE to conventional 2D CINE for two healthy subjects. CINE was acquired with left ventricle coverage in short axis orientation. Similar anatomical positions were selected for comparison.

**VIDEO S5** Cardiac motion-resolved short-axis images from 3D CINE (1.9 × 1.9 × 2.5 mm) and 2D CINE (1.9 × 1.9 × 8 mm) in three patients. Patient 1 had arterial fibrillation with a heart rate of ~45 bpm, patient 2 had a constant heart rate of 70 bpm and patient 3 had a strongly varying heart rate of 60-90 bpm.

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**How to cite this article:** Küstner T, Bustin A, Jaubert O, et al. Isotropic 3D Cartesian single breath-hold CINE MRI with multi-bin patch-based low-rank reconstruction. *Magn Reson Med*. 2020;84:2018–2033. [https://doi.org/10.1002/mrm.28267](https://doi.org/10.1002/mrm.28267)