Respiratory motion corrected 4D flow using golden radial phase encoding

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Purpose: To minimize respiratory motion artifacts while achieving predictable scan times with 100% scan efficiency for thoracic 4D flow MRI.

Methods: A 4D flow sequence with golden radial phase encoding (GRPE) was acquired in 9 healthy volunteers covering the heart, aorta, and venae cavae. Scan time was 15 min, and data were acquired without motion gating during acquisition. Data were retrospectively re-binned into respiratory and cardiac phases based on respiratory self-navigation and the electrocardiograph signals, respectively. Nonrigid respiratory motion fields were extracted and corrected for during the \( k-t \) SENSE reconstruction. A respiratory-motion corrected (GRPE-MOCO) and a free-breathing (GRPE-UNCORR) 4D flow dataset was reconstructed using 100% of the acquired data. For comparison, a respiratory gated Cartesian 4D flow acquisition (CART-REF) covering the aorta was acquired. Stroke volumes and peak flows were compared. Additionally, an internal flow validation based on mass conservation was performed on the GRPE-MOCO and GRPE-UNCORR. Statistically significant differences were analyzed using a paired Wilcoxon test.

Results: Stroke volumes and peak flows in the aorta between GRPE-MOCO and the CART-REF showed a mean difference of \(-1.5 \pm 10.3\) mL \((P > 0.05)\) and \(25.2 \pm 55.9\) mL/s \((P > 0.05)\), respectively. Peak flow in the GRPE-UNCORR data was significantly different compared with CART-REF \((P < 0.05)\). GRPE-MOCO showed higher accuracy for internal consistency analysis than GRPE-UNCORR.

Conclusion: The proposed 4D flow sequence allows a straight-forward planning by covering the entire thorax and ensures a predictable scan time independent of cardiac cycle variations and breathing patterns.

KEYWORDS
4D flow, \( k-t \) SENSE, nonrigid motion correction, radial phase encoding
1 | INTRODUCTION

The 4D flow MRI allows for the assessment of time-resolved, 3D and 3-directional velocities.1 Thanks to its increasing availability and robustness, it is to date being applied in several clinical studies such as valvular and cardiovascular diseases.2 Nevertheless, the use of 4D flow MRI in a routine clinical setting or larger clinical studies remains hampered by its long examination times. Although advanced image reconstruction approaches such as parallel imaging or spatiotemporal acceleration techniques3-6 have been able to significantly shorten acquisition times, respiratory motion can still lead to unacceptable long scan time. Respiratory gating is the most commonly used motion suppression technique for Cartesian 4D flow MRI.7-9 For regular breathing patterns, scan efficiencies of 40% to 50% are achieved, but for irregular breathing patterns, this can drop to below 20%,10,11 leading to scan times that are more than 5 times longer than the nominal scan duration.

To overcome the challenge of unpredictably long scan times, several studies have suggested to obtain 4D flow MRI under free-breathing without motion compensation.12-14 Although flow quantification in major vessels did not vary significantly between ungated and gated scans, several studies have demonstrated that respiratory gating is needed for the extraction of wall shear stress and achieves reduced bias, smaller variability, and better agreement to 2D reference scans in phantom and in vivo studies for flow quantification.15,16 Other approaches to reduce motion artifacts in 4D flow MRI include non-Cartesian acquisitions such as radial or spiral schemes.17-21 Although minimizing breathing artifacts, non-Cartesian frequency encoding often lacks robustness to systematic errors such as phase errors due to eddy currents or blurring due to lengthened readout times.22,23

Here, we propose to use motion-corrected image reconstruction for 4D flow MRI to minimize motion artifacts and ensure accurate flow quantification while ensuring 100% scan efficiency and predictable scan times for all subjects. The 4D flow MRI data are acquired with a golden radial phase encoding (GRPE) scheme, which provides a respiratory self-navigator and allows for the estimation of motion information from the data itself.24-28 It allows for a straightforward planning, the acquisition of an isotropic, large field of view (FOV) covering the heart and surrounding vessels, and runs at a predictable scan time of 15 min. Data were re-binned into cardiac phases based on the electrocardiograph (ECG) signal, while breathing motion information was extracted from the central $k$-space lines and corrected for during reconstruction yielding a breathing motion compensated 4D flow dataset (GRPE-MOCO). Flow values were then quantitatively compared with a thin-slab reference Cartesian 4D flow MRI of the aorta running at a similar total scan time (CART-REF). Additionally, GRPE-MOCO was compared with a dataset reconstructed without respiratory motion correction (GRPE-UNCORR). The method was evaluated in 9 healthy volunteers.

2 | METHODS

2.1 | Data acquisition

Data were acquired on a 1.5T system (Ingenia, Philips, Best, The Netherlands) using a 28-channel anterior and posterior coil array. Nine volunteers without known cardiovascular disease (4 female; age, 26.9 ± 6.6 years) underwent a 15-min 4D flow GRPE scan and a 4D flow CART-REF scan. The study was approved by the local ethics committee and written informed consent was obtained from all participants before the examination.

2.2 | GRPE

Cartesian frequency encoding was acquired in feet-head (FH) direction (Figure 1A). Phase encoding was carried out along a radial grid in the 2D phase encoding plane $k_y-k_z$.28 Phase encoding was, therefore, not carried out on a Cartesian grid but along non-Cartesian radial lines in the $k_y-k_z$ plane. All phase encoding points along one radial line were obtained before increasing the angle by a golden radial increment of 111.24°. The GRPE scan was planned to cover the entire upper abdomen (i.e., heart, aorta, and pulmonary vasculature) with the following acquisition parameters: repetition time (TR)/echo time: 3.8/2.0 ms, FOV: 250 × 250-270 × 250-270 mm³, acquired voxel size: 2.5 × 2.5 × 2.5 mm³, imaging matrix: 100-108 × 100-108 × 100, flip angle: 7°, readout bandwidth: 84.6 kHz. The scan was not synchronized to the ECG and no respiratory compensation was applied prospectively. The ECG signal, however, was recorded during the scan using a vector cardiogram and used to retrospectively sort the data into 24 cardiac phases. Flow encoding was performed in each direction consecutively using a 4-point symmetric acquisition scheme.29 A velocity encoding value of 150 cm/s was chosen. Acquisition time was fixed to 15:03 min.

2.3 | CART-REF

For comparison, the CART-REF acquisition with parameters according to the 4D flow consensus statement² was acquired. The sagittal slab was planned to exclusively cover the ascending aorta, the aortic arch, and the thoracic descending aorta. Parameters included: TR/echo time: 3.4-3.7/2.0-2.4 ms, FOV: 250 × 125-143 × 50 mm³, acquired voxel size: 2.5 × 2.5 × 2.5 mm³, imaging matrix: 100 × 50-57 × 20, flip angle: 4°, readout bandwidth: 99.5 kHz, velocity encoding value: 150 cm/s. The scan was retrospectively triggered to the ECG signal and a temporal resolution of 32.7-58.5 ms, corresponding to 20-22 heart phases (depending on the heart...
Breathing motion was compensated by placing a pencil-beam navigator on the liver-diaphragm interface in combination with an acceptance window of 6 mm. A parallel imaging (SENSE) factor of 2 was used, resulting in a nominal scan time excluding navigator efficiency of 6:14 ± 0:32 min (range, 5:30-7:03 min).

2.4 | GRPE reconstruction

2.4.1 | Respiratory self-navigator

A respiratory self-navigator was obtained from the central k-space line of each GRPE line. Data acquisition was carried out such that one full GRPE radial spoke was acquired for one flow encoding direction, before repeating the same GRPE radial spoke for the next flow encoding direction. Therefore, a central k-space line was obtained every 100*TR = 380 ms. The overall signal intensities of the central k-space lines vary between different flow encodings; therefore, a respiratory self-navigator was calculated for each flow direction separately.

2.4.2 | Motion estimation

Based on the respiratory self-navigator signal, k-space data of each flow encoding direction was re-binned separately into 8 respiratory phases (RPh1-RPh8) (Figure 1B). The re-binning was carried out independently of the cardiac phase information; therefore, RPh1-RPh8 are averages over the cardiac cycles. Motion states were then reconstructed using autocalibrated iterative SENSE with temporal and spatial total variation regularization. To estimate nonrigid motion fields (M) between respiratory states, a spline-based image registration algorithm was used. A bending energy penalty ensured smooth M allowing a robust motion estimation in the presence of residual undersampling artifacts. A normalized mutual information metric was used to maximize the similarity between images of different respiratory motion states. End-expiration of the flow-compensated data set was used as reference motion state.

2.4.3 | Motion correction

Based on the recorded ECG signal, k-space data were retrospectively re-binned into 24 cardiac phases using a sliding window approach with a window overlap of 20% on each side (Figure 1C). Retrospective arrhythmia rejection was applied to exclude data from heart cycles deviating more than 20% from the average cycle length. For temporal k-t regularization, low spatial and high temporal resolution data (training data) were extracted from the central 25% of the k-space. To minimize truncation artifacts, a 2D...
Gaussian filter was applied and training data were reconstructed using a non-Cartesian iterative SENSE approach. GRPE data were then reconstructed using a self-regularized autocalibrated iterative non-Cartesian $k$-space SENSE approach. At each reconstruction iteration, every cardiac phase was respiratory-motion corrected using the motion fields $M_r$ yielding a respiratory motion-corrected 4D flow reconstruction (GRPE-MOCO). The motion corrected image reconstruction was based on the general matrix description proposed by Batchelor et al. The encoding operator $E$ describing the acquisition process is extended by a sum over all motion states $N$:

$$E = \sum_{r=1}^{N} S_rGFC_cM_r.$$  

$C$ describes the coil sensitivity information for each coil $c$, $F$ is the Cartesian Fourier Operator, $G$ is the interpolation operator from Cartesian to non-Cartesian $k$-space, and $S$ selects the $k$-space points acquired for each respiratory bin $r$. During image reconstruction, an iterative conjugate gradient approach was then used to minimize the difference between the acquired $k$-space data $K$ and the motion-corrected image $I_{MOCO}$:

$$I_{MOCO} = \min_I \|EI - K\|_2^2$$  

$k$-space regularization was added to minimize undersampling artifacts.

Additionally, 4D flow images were reconstructed without respiratory motion correction or respiratory gating by combining all acquired data irrespectively of their corresponding breathing phase (GRPE-UNCORR). This reduces the encoding operator described in Equation 1 to

$$E = SGFC_c.$$  

A schematic overview of the data acquisition and binning into respiratory and cardiac motion states is given in Supporting Information Figure S1, which is available online.

### 2.5 Data analysis

Respiratory motion amplitudes (RMA) were measured as the amplitude of the motion vectors between end-expiration and end-inspiration. RMA was assessed in three regions of interest: right hemi-diaphragm (i.e., where a respiratory navigator would be placed), root of the aorta, and aortic arch. The ratio between RMA of the right hemi-diaphragm and the root of the aorta was also calculated.

Flow quantification was performed by calculating stroke volumes (SV) and peak flows (PF) at three levels along the aorta. Contours were placed in the ascending aorta at the level of the pulmonary artery (AAbb), in the aortic arch (Arch), and in the abdominal descending aorta (DAo) in the CART-REF, GRPE-MOCO, and GRPE-UNCORR data. Bland-Altman analysis comparing results from GRPE-MOCO to CART-REF was performed on SV and PF. Vertical lines in the Bland-Altman plots illustrate the mean bias and the limits of agreement (1.96 to −1.96 standard deviation of the difference).

On the GRPE-MOCO and GRPE-UNCORR data, SV were additionally evaluated based on the conservation of mass: SV in the main pulmonary artery (PA) was compared with the sum of SV in the right PA (RPA) and left PA (LPA). Similarly, the difference in SV in the AAbb (Qs) and PA (Qp) was calculated. Based on the SV, the commonly used metric of Qp/Qs ratios were calculated.

A paired Wilcoxon test was used to evaluate statistical significance of the differences of the compared quantities for CART-REF versus GRPE-MOCO, CART-REF versus GRPE-UNCORR, and internal validation. A $P$-value < 0.05 was considered as statistically significant.

### 3 RESULTS

Data were successfully acquired and reconstructed in all volunteers. An overall acquisition time of 14:46 ± 5:44 min and 15:03 min was achieved in the CART-REF and GRPE sequence, respectively. In the CART-REF sequence, the scan efficiency due to respiration and ECG arrhythmias was 47.7 ± 11.5% and a mean number of 21.7 ± 0.7 heart phases were reconstructed, resulting in a mean temporal resolution of 43.9 ± 7.1 ms. The GRPE sequence was re-constructed to a fixed number of 24 heart phases, resulting in a temporal resolution of 41.7 ± 6.6 ms. The overall undersampling factor for a single cardiac phase combining data from all breathing phases was 3-4 in the GRPE data. Supporting Information Figure S2 visualizes the distribution of $k$-space data in the 2D phase encoding plane $k_y$-$k_z$ for respiratory bins and cardiac phases.

#### 3.1 Motion estimation

Figure 2 shows coronal slices in end-expiration (RPh1) and end-inspiration (RPh8) of a respiratory resolved GRPE dataset used for motion estimation. The subtracted image depicts motion state differences in the heart and abdominal organs (Figure 2, red and blue arrows, respectively). After application of the nonrigid motion field ($M_8$), differences are clearly reduced. Animations of the respiratory resolved images for all volunteers are shown in Supporting Information Figure S3.

RMA obtained from the motion fields were 10 ± 5.3 (range, 4.4-21) mm at the right hemidiaphragm, 6.2 ± 3.6 (range, 3.1-14.3) mm at the root of the aorta and 3.7 ± 2.7 (range, 1.2-10.1) mm at the aortic arch. The ratio of RMA between right hemi-diaphragm and aortic root was 0.6 ± 0.1 (range, 0.5-0.7).
3.2 | CART-REF versus GRPE-MOCO

Figure 3 shows magnitude images overlaid with color-coded in-plane velocity vectors in peak-systole as well as net flow curves of the CART-REF (Figure 3, blue) and the GRPE-MOCO (Figure 3, red) datasets in one exemplary volunteer. A longer trigger delay at the start of the cardiac cycle due to the respiratory navigator can be appreciated in the CART-REF flow curves. Figure 4A and B show Bland-Altman plots comparing SV and PF from CART-REF and GRPE-MOCO. Mean differences in SV and PF were $-1.5 \pm 10.3$ mL and $25.2 \pm 55.9$ mL/s, respectively. Differences in SV and PF between CART-REF and GRPE-MOCO are statistically nonsignificant.

3.3 | CART-REF versus GRPE-UNCORR

Mean differences in SV and PF between CART-REF and GRPE-UNCORR were $-0.3 \pm 9.9$ mL and $52.7 \pm 56.1$ mL/s, respectively. Differences in PF between CART-REF and GRPE-MOCO are statistically significant. Supporting Information Table S1 summarizes all comparisons for CART-REF versus GRPE-MOCO and CART-REF versus GRPE-UNCORR and corresponding $P$-values.

3.4 | GRPE-MOCO versus GRPE-UNCORR

Figure 4C-F show Bland-Altman plots for the internal validation using SV differences between the PA and RPA + LPA and between PA (Qp) and AAo (Qs). Pulmonary volumes in the GRPE-UNCORR and GRPE-MOCO show a mean difference of $10.2 \pm 16.6$ mL and $1.5 \pm 16.6$ mL, respectively. Similarly, Qp and Qs in the GRPE-UNCORR data show a higher mean difference of $6.1 \pm 10.3$ mL ($Qp/Qs = 1.08 \pm 0.12$; range, 0.94-1.37) as compared to the GRPE-MOCO data, with a mean difference of $-0.6 \pm 10.0$ mL ($Qp/Qs = 1.00 \pm 0.12$; range, 0.82-1.15). No significant differences were found in the internal validation. All compared values are summarized in Supporting Information Table S2.

Figure 5 shows coronal magnitude images and streamlines in the pulmonary arteries for both reconstructions in 2 exemplary volunteers, one with stronger (left) and one with less respiratory-induced motion (right). Unlike volunteer 2, the magnitude images of volunteer 1 show higher edge sharpness and better visibility of small vessels in the GRPE-MOCO as compared to the GRPE-UNCORR data. In this particular example, streamlines in the pulmonary arteries also indicate that flow in smaller vessels show improved visualization in the GRPE-MOCO data of volunteer 1.

4 | DISCUSSION

The present work proposes a novel 4D flow acquisition technique which is straightforward to plan, runs at a predictable scan time, covers the entire upper abdomen, corrects for breathing motion and uses 100% of the acquired data while running...
at a similar scan time as a standard thin-slab aortic 4D flow sequence.

Flow quantification comparisons with standard aortic 4D flow acquisitions revealed differences in-line with previously described alterations attributable to physiological changes. Figure 5 shows that respiratory motion can cause blurring of the individual flow encoded images, which then impairs the quantitative 4D flow estimation leading to flow underestimation. The proposed nonrigid motion correction approach clearly improves image quality as well as flow quantification results. Increased accuracy with respect to CART-REF and Qp/Qs ratio was observed for the GRPE-MOCO data. GRPE-UNCORR data show a higher range of values for Qp/Qs in healthy volunteers of up to 1.37 compared with a maximum value of 1.15 using GRPE-MOCO. A Qp/Qs ratio > 1.5 can be used for classification of significant shunts; therefore, GRPE-MOCO could reduce the risk of false classification. The analysis of PF shows significant underestimation in the GRPE-UNCORR data and nonsignificant differences in the GRPE-CORR data. The effect of noncompensated motion on PF is in-line with a previously study using computational fluid mechanics.

In contrast to a typically respiratory-gated Cartesian acquisition, the proposed self-gating approach does not require the acquisition of a respiratory navigator, which assures the homogenous coverage of the entire cardiac cycle while the signal remains in the steady-state. The temporal resolution of the respiratory self-navigator obtained in this study was 380 ms, which was sufficient for all subjects. Nevertheless, for patients with very rapid breathing, a higher temporal resolution might be required. Although ECG information is recorded and used for re-binning, heart cycle variations do not influence data acquisition and arrhythmic heart cycles can be treated retrospectively.
The amplitudes of respiratory motion vary strongly between different volunteers. The ratio between motion of the right hemi-diaphragm and aortic root was on average 0.6 as reported in previous studies. Nevertheless, the range of the ratio was between 0.5 and 0.7, which means that a respiratory navigator placed on the right hemi-diaphragm with a window of 6.0 mm would have reduced the motion in the aortic root for 1 subject to 3.0 mm but for another only to 4.2 mm. The respiratory motion estimation presented here does not suffer from such subject specific variations and ensures accurate local motion minimization. Although residual motion in each respiratory bin may still be present, it remains small compared with the spatial resolution of the acquisition. For example, for the aortic root, the average residual motion amplitude is below 1 mm. Additional intra-bin motion correction could, however, be applied to further reduce this remaining motion.

In combination with the proposed motion correction approach, all acquired GRPE data were used for the final 4D

**FIGURE 4** Bland-Altman plots for validation. SV (A) and (B) PF values are compared for CART-REF versus GRPE-MOCO. Three aortic contours were analyzed corresponding to the 3 different markers in (A) and (B). C-F, conservation of mass validation in the GRPE data. SV in PA versus LPA+RPA (C,E) and AAO versus PA (D,F) are shown. Lower mean differences and standard deviations can be seen in GRPE-MOCO as compared to GRPE-UNCORR.

**FIGURE 5** Qualitative comparison of GRPE-UNCORR and GRPE-MOCO. A,B,E,F, Coronal magnitude images of two exemplary volunteers. The arrow depicts a better delineation of the aortic arch in GRPE-MOCO in volunteer 1. C,D,G,H, Streamlines from the PA at peak systole. In volunteer 1, improved visualization of flow in small vessels in the GRPE-MOCO data is demonstrated.
flow reconstruction. Therefore, the predictability of scan time is independent of irregularities or changes in the breathing pattern during the scan. As unpredictable scan times and scan aborts due to low acceptance efficiencies and breathing position drifts currently hamper the wider use of 4D flow, the hereby presented approach is expected to alleviate this hurdle and make 4D flow easier integrable in clinical routine protocols. Even in the current study in which only healthy subjects were scanned, scan times for the respiratory-gated Cartesian acquisition showed variations from 9 to 28 min, which is in agreement to previous studies that have shown navigator efficiencies below 20%.10,11

For the iterative non-Cartesian k-t SENSE reconstruction, the number of iterations and the weight of the k-t regularization were optimized based on one dataset. Supporting Information Figure S4 shows flow curves for different reconstruction parameters in one subject. Based on this analysis, the number of iterations and the weights were set to 12 and 3, respectively. As scan parameters were left unchanged between volunteers, these parameters were left constant throughout all reconstructions. The retrospective separation of data leads to a random distribution of k-space points in each cardiac phase. The distribution of data in each cardiac phase is different between different flow encodings and volunteers but leads to similar undersampling artifacts. Due to the undersampling properties of GRPE, this iterative reconstruction technique could minimize these incoherent undersampling artifacts and ensure high quality 4D flow data. More advanced image reconstruction approaches such as compressed sensing40 or k-t PCA41 could be used to further improve image quality and allow for a reduction in scan time.

One potential limitation is the acquisition of a very large FOV. Although acquisition time is similar to a standard 4D flow acquisition of a sagittal slab covering the aorta, the added value of hemodynamic information in the entire region may be questioned. In this study, the acquisition of a square FOV with isotropic spatial resolution was prescribed due to implementation ease; however, a freely adaptable FOV could be implemented and beneficial for the assessment of specific regions.42 In the current feasibility study, the straightforward planning and improved signal-to-noise ratio when using the large FOV was, however, preferred.

As compared to so-called pseudo-radial or pseudo-spiral Cartesian k-space sampling,43-46 the used GRPE leads to a variable density of acquired phase encoding points located on non-Cartesian coordinates. This requires gridding operation and, therewith, longer reconstruction times.

The use of motion correction was shown to increase accuracy of flow quantities. However, the co-registration of images from different respiratory phases assumes invariant hemodynamics for different respiratory states. This assumption may not be valid in specific vessels such as the vena cavae or in specific anatomies such as Fontan patients who show a respiratory dependent hemodynamic.47 In these cases, the current technique may be used to extract respiratory-dependent 4D flow datasets (5D flow).

5 | CONCLUSIONS

The proposed GRPE 4D flow technique allows for motion-corrected 3D and 3-directional velocity quantification by exploiting 100% of the acquired data. A predictable scan time, straightforward planning, and large geometrical coverage is achieved and may help integration into clinical applications.

CONFLICTS OF INTEREST

Dr. Kilian Weiss is an employee of Philips Healthcare. Dr. Daniel Giese is an employee of Siemens Healthcare since December 2018. All of Dr. Giese’s input related to the submitted manuscript, however, was performed prior or outside of his duties at Siemens. All other authors have no conflict of interest to declare.

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REFERENCES

1. Markl M, Kilner PJ, Ebbers T. Comprehensive 4D velocity mapping of the heart and great vessels by cardiovascular magnetic resonance. J Cardiovasc Magn Reson. 2011;13:7.
2. Dyverfeldt P, Bissell M, Barker AJ, et al. 4D flow cardiovascular magnetic resonance consensus statement. J Cardiovasc Magn Reson. 2015;17:72.
3. Carlsson M, Töger J, Kanski M, et al. Quantification and visualization of cardiovascular 4D velocity mapping accelerated with parallel imaging or k-t BLAST: head to head comparison and validation at 1.5 T and 3 T. J Cardiovasc Magn Reson. 2011;13:55.
4. Schnell S, Markl M, Entezari P, et al. k-t GRAPPA accelerated four-dimensional flow MRI in the aorta: effect on scan time, image quality, and quantification of flow and wall shear stress. Magn Reson Med. 2014;72:522–533.
5. Giese D, Wong J, Greif GF, Bucher M, Schaeffter T, Kozerke S. Towards highly accelerated Cartesian time-resolved 3D flow cardiovascular magnetic resonance in the clinical setting. J Cardiovasc Magn Reson. 2014;16:42.
6. Knobloch V, Boesiger P, Kozerke S. Sparsity transform k-t principal component analysis for accelerating cine three-dimensional flow measurements. Magn Reson Med. 2013;70:53–63.
7. Dyverfeldt P, Bissell M, Barker AJ, et al. 4D flow cardiovascular magnetic resonance consensus statement. J Cardiovasc Magn Reson. 2015;17:72.
10. Nguyen TD, Spincemaille P, Cham MD, Weinsaft JW, Prince MR, Markl M, Harloff A, Bley TA, et al. Time-resolved 3D MR velocity mapping at 3T: improved navigator-gated assessment of vascular anatomy and blood flow. J Magn Reson Imaging. 2007;25:824–831.

11. Kolbitsch C, Prieto C, Smink J, Schaeffer T. Highly efficient whole-heart imaging using radial phase encoding-phase ordering with automatic window selection. Magn Reson Med. 2011;66:1008–1018.

12. Nordmeyer S, Riesenkampff E, Crelier G, et al. Flow-sensitive four-dimensional cine magnetic resonance imaging for offline blood flow quantification in multiple vessels: a validation study. J Magn Reson Imaging. 2010;32:677–683.

13. Kanski M, Töger J, Steding-Ehrenborg K, et al. Whole-heart four-dimensional flow can be acquired with preserved quality without respiratory gating, facilitating clinical use: a head-to-head comparison. BMC Med Imaging. 2015;15:20.

14. Valverde I, Nordmeyer S, Uribe S, et al. Systemic-to-pulmonary collateral flow in patients with palliated univentricular heart physiology: measurement using cardiovascular magnetic resonance 4D velocity acquisition. J Cardiovasc Magn Reson. 2012;14:25.

15. Uribe S, Beerbaum P, Sorenson TS, Rasmusson A, Razavi R, Schaeffer T. Four-dimensional (4D) flow of the whole heart and great vessels using real-time respiratory self-gating. Magn Reson Med. 2009;62:984–992.

16. Dyverfeldt P, Ebbers T. Comparison of respiratory motion suppression techniques for 4D flow MRI. Magn Reson Med. 2017;78:1877–1882.

17. Bastkowski R, Weiss K, Maintz D, Giese D. Self-gated golden-angle spiral 4D flow MRI. Magn Reson Med. 2018;80:904–913.

18. Petersson S, Sigfridsson A, Carlhall CJ, Ebbers T. Retrospectively gated intra-cardiac 4D flow CMR using spiral k-space trajectories. J Cardiovasc Magn Reson. 2013;15(Suppl 1):O64.

19. Sigfridsson A, Petersson S, Carlhall CJ, Ebbers T. Spiral readouts for 4D flow MRI. J Cardiovasc Magn Reson. 2012;14(Suppl 1):W31.

20. Santelli C, Schaeffer T, Kozerke S. Radial k-t SPIRiT: autocalibrated parallel imaging for generalized phase-contrast MRI. Magn Reson Med. 2014;72:1233–1245.

21. Johnson KM, Lum DP, Turski PA, Block WF, Mistretta CA, Wieben O. Improved 3D phase contrast MRI with off-resonance corrected dual echo VIPR. Magn Reson Med. 2008;60:1329–1336.

22. Bornert P, Schomberg H, Aldefeld B, Groen J. Improvements in spiral MR imaging. MAGMA. 1999;9:29–41.

23. Lechner SM, Sipila PT, Wiesinger F, Kerr AB, Vogel MW. Spiral imaging artifact reduction: a comparison of two k-trajectory measurement methods. J Magn Reson Imaging. 2009;29:1485–1492.

24. Prieto C, Uribe S, Razavi R, Atkinson D, Schaeffer T. 3D undersampled golden-rod phase encoding for DCE-MRA using inherently regularized iterative SENSE. Magn Reson Med. 2010;64:514–526.

25. Winkelmann S, Schaeffer T, Koehler T, Eggers H, Doessel O. An optimal radial profile order based on the Golden Ratio for time-resolved MRI. IEEE Trans Med Imaging. 2007;26:68–76.

26. Kolbitsch C, Ahlman MA, Davies-Venn C, et al. Cardiac and respiratory motion correction for simultaneous cardiac PET/MR. J Nucl Med. 2017;58:846–852.

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

28. Buerger C, Clough RE, King AP, Schaeffer T, Prieto C. Nonrigid motion modeling of the liver from 3-D undersampled self-gated golden-radial phase encoded MRI. IEEE Trans Med Imaging. 2012;31:805–815.

29. Pelc NJ, Bernstein MA, Shimakawa A, Glover GH. Encoding strategies for three-direction phase-contrast MR imaging of flow. J Magn Reson Imaging. 1991;1:405–413.

30. Kolbitsch C, Neji R, Fenchel M, Mallia A, Marsden P, Schaeffer T. Respiratory-resolved MR-based attenuation correction for motion-compensated cardiac PET-MR. Phys Med Biol. 2018;63:135008.

31. Block KT, Uecker M, Frahm J. Suppression of MRI truncation artifacts using total variation constrained data extrapolation. Int J Biomed Imaging. 2008;2008:184123.

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

33. Tsao J, Boesiger P, Pruessmann KP. k-t BLAST and k-t SENSE: dynamic MRI with high frame rate exploiting spatiotemporal correlations. Magn Reson Med. 2003;50:1031–1042.

34. Hansen MS, Baltes C, Tsao J, Kozerke S, Pruessmann KP, Eggers H. k-t BLAST reconstruction from non-Cartesian k-t space sampling. Magn Reson Med. 2006;55:85–91.

35. Batchelor PG, Atkinson D, Irazarraval P, Hill DL, Hajnal J, Larkman D. Matrix description of general motion correction applied to multishot images. Magn Reson Med. 2005;54:1273–1280.

36. Markl M, Wallis W, Harloff A. Reproducibility of flow and wall shear stress analysis using flow-sensitive four-dimensional MRI. J Magn Reson Imaging. 2011;33:988–994.

37. Hanneman K, Sivagnanam M, Nguyen ET, et al. Magnetic resonance assessment of pulmonary (QP) to systemic (QS) flows using 4D phase-contrast imaging: pilot study comparison with standard through-plane 2D phase-contrast imaging. Acad Radiol. 2014;21:1002–1008.

38. Wang Y, Riederer SJ, Ehman RL. Respiratory motion of the heart: kinematics and the implications for the spatial resolution in coronary imaging. Magn Reson Med. 1995;33:713–719.

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

40. Lustig M, Donoho D, Pauly JM. Sparse MRI: the application of compressed sensing for rapid MR imaging. Magn Reson Med. 2007;58:1182–1195.

41. Pedersen H, Kozerke S, Ringgaard S, Nehrkoe K, Kim WY. k-t PCA: temporally constrained k-t BLAST reconstruction using principal component analysis. Magn Reson Med. 2009;62:706–716.

42. Larson PZ, Gurney PT, Nishimura DG. Anisotropic field-of-views in radial imaging. IEEE Trans Med Imaging. 2008;27:47–57.

43. Prieto C, Doneva M, Usman M, et al. Highly efficient respiratory motion compensated free-breathing coronary MRA using...
golden-step Cartesian acquisition. *J Magn Reson Imaging*. 2015;41:738–746.

44. Liu J, Saloner D. Accelerated MRI with CIRcular Cartesian UnderSampling (CIRCUS): a variable density Cartesian sampling strategy for compressed sensing and parallel imaging. *Quant Imaging Med Surg*. 2014;4:57–67.

45. Cheng JY, Zhang T, Ruangwattanapaisarn N, et al. Free-breathing pediatric MRI with nonrigid motion correction and acceleration. *J Magn Reson Imaging*. 2015;42:407–420.

46. Zhu Y, Guo Y, Lingala SG, Lebel RM, Law M, Nayak KS. GOCART: GOlden-angle CArtesian randomized time-resolved 3D MRI. *Magn Reson Imaging*. 2016;34:940–950.

47. Korperich H, Barth P, Gieseke J, et al. Impact of respiration on stroke volumes in paediatric controls and in patients after Fontan procedure assessed by MR real-time phase-velocity mapping. *Eur Heart J Cardiovasc Imaging*. 2015;16:198–209.

**SUPPORTING INFORMATION**

Additional supporting information may be found online in the Supporting Information section at the end of the article.

**FIGURE S1** Schematic illustration of the re-binning scheme into respiratory and cardiac phases. From left to right the time course of a GRPE sequence is shown. The four different flow encodings are illustrated in gray, blue, red, and yellow. From top to bottom increasingly finer time steps down to subsequent TRs are shown.

**FIGURE S2** Distribution of GRPE *k*-space data in the 2D phase encoding plane *k*$_y$-*k*$_z$ for respiratory and cardiac bins of two volunteers (a and b). For each volunteer six out of eight respiratory and out of 24 cardiac bins are shown.

**FIGURE S3** Respiratory resolved images for all nine volunteers included in this study.

**FIGURE S4** Evaluation of reconstruction parameters. Data for one volunteer were reconstructed using different numbers of iterations (a) and different *k*-*t* regularization strengths (b). Note that the peak flow decreases with higher *k*-*t* regularization. 12 iterations and a *k*-*t* regularization strength of 3 was used for all reconstructions.

**TABLE S1** Mean ± standard deviation of SV and PF over all volunteers. The *P*-values show if differences between CART-REF versus GRPE-MOCO and CART-REF versus GRPE-UNCORR values are statistically significant.

**TABLE S2** Mean ± standard deviation of SV for internal validation of the GRPE-MOCO and the GRPE-UNCORR data. The *P*-values show if differences between the compared values are statistically significant.

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