Published in final edited form as:

Int J Cardiovasc Imaging. 2016 July ; 32(7): 1103–1111. doi:10.1007/s10554-016-0879-z.

Heart deformation analysis: Measuring regional myocardial velocity with MR imaging

Kai Lin, MD, MS¹, Jeremy D. Collins, MD¹, Varun Chowdhary, MD¹, Michael Markl, PhD¹, and James C. Carr, MD¹

¹Department of Radiology, Northwestern University, 737 N Michigan Avenue, Suite 1600, Chicago, IL 60611

Abstract

The aim of the present study was to test the hypothesis that that heart deformation analysis (HDA) may serve as an alternative for the quantification of regional myocardial velocity. Nineteen healthy volunteers (14 male and 5 female) without documented cardiovascular diseases were recruited following the approval of the institutional review board (IRB). For each participant, cine images (at base, mid and apex levels of the left ventricle [LV]) and tissue phase mapping (TPM, at same short-axis slices of the LV) were acquired within a single magnetic resonance (MR) scan. Regional myocardial velocities in radial and circumferential directions acquired with HDA (V_{rr} and V_{cc}) and TPM (V_r and V_{ϕ}) were measured during the cardiac cycle. HDA required shorter processing time compared to TPM (2.3 ± 1.1 min/case vs. 9.5 ± 3.7 min/case, $p < 0.001$). Moderate to good correlations between velocity components measured with HDA and TPM could be found on multiple myocardial segments ($r = 0.460 - 0.774$) and slices ($r = 0.409 - 0.814$) with statistical significance ($p < 0.05$). However, significant biases of velocity measures at regional myocardial areas between HDA and TPM were also noticed. By providing comparable velocity measures as TPM does, HDA may serve as an alternative for measuring regional myocardial velocity with a faster image processing procedure.

Keywords

MRI; heart deformation analysis; regional myocardial velocity

Introduction

Low ejection fraction (EF) is a traditional marker for the diagnosis of congestive heart failure (CHF), which represents the advanced stage of myocardial damage. However,

Address for correspondence: Kai Lin (jcarr@northwestern.edu) Department of Radiology, Northwestern University, 737 N Michigan Avenue, Suite 1600, Chicago, IL 60611, Tel: (312)695-5577, Fax: (312)926-5991.

Financial disclosure: N/A

Conflict of Interest: N/A

Compliance with Ethical Standards Ethical approval: All procedures performed in studies were in accordance with the ethical standards of the institutional and/or national research committee and with the 1964 Helsinki declaration and its later amendments or comparable ethical standards. The IRB approved this study.

Informed consent: Written informed consent was obtained from all individual participants included in the study.

myocardial dysfunction in its early stage can occur at a regional level and may thus result in only minimal decrease of EF that cannot be easily detected with standard diagnostic tools. Previous clinical studies have proven that the impaired of local left ventricle (LV) wall thickening in systole, which was semi-quantitatively identified by using either echocardiography or magnetic resonance (MR) imaging, is associated with a high incidence of CHF and cardiovascular events [1, 2]. In asymptomatic subjects with prominent cardiovascular risk factors, such as diabetes mellitus (DM) or obesity, impaired regional myocardial motion can be observed without a decrease of EF [3, 4]. Therefore, the accurate description of subtle changes in myocardial motion patterns may be critical for assessing the progression of subclinical cardiovascular diseases and for monitoring individual cardiovascular responses to treatments.

In past decades, many MR imaging techniques have been developed to quantify myocardial mechanics of the heart. More recently, various image processing techniques and principles in the domain of computer vision and machine learning have been applied to measure myocardial displacement, strain and velocity based on standard cine MR images acquired with steady state free precession (SSFP) technique [5–7]. Heart deformation analysis (HDA) is an image processing method which allows for the quantification of myocardial motion on cine images using deformable image registration (DIR) algorithms [8]. Although DIR has been applied to calculate global LV strain, whether it can be used to accurately measure regional LV velocities is still unclear [9]. To address this unmet need, we measured LV velocity components in 19 healthy volunteers using both HDA and tissue phase mapping (TPM), an established phase-contrast MR imaging technique for quantifying myocardial motion by encoding the instantaneous velocity of a pixel into the MR signal phase [10–12]. HDA- and TPM-derived regional velocity components obtained in a single MR scan session were compared and correlated side-by-side. The aim of the present study was to test the hypothesis that HDA may serve as an alternative for the quantification of regional myocardial velocity.

Materials and Methods

In compliance with the Health Insurance Portability and Accountability Act (HIPAA) regulation, 19 asymptomatic volunteers (14 male and 5 female) without documented cardiovascular diseases were recruited following the approval of the institutional review board (IRB) (table 1). Written informed consents were provided by all participants before examinations. All MR scans were performed by 2 certified clinical MR technicians on a 1.5 T scanner (SIEMENS, MAGNETOM, Aera, Germany) with a 18-channel cardiac phased-array coil.

Imaging protocol

A three-plane fast localization sequence was used for anatomic orientation of all images. Four-chamber, two-chamber, and short-axis views were acquired with a black-blood half-Fourier rapid acquisition with relaxation enhancement sequence to identify anatomy for cardiac images.

For each participant, segmented SSFP sequences were used to acquire cardiac cine images in the two-chamber, four-chamber and short-axis of the heart. Imaging parameters were as follows: Repetition time(TR)/Echo time(TE) = 2.8/1.1 ms; flip angle = 65°, voxel size = 2.1×2.1×8.0 mm³, Generalized Autocalibrating Partially Parallel Acquisition (GRAPPA technique) with reduction factor R = 2. Each myocardial slice (with 25 retrospective constructed cardiac phases) was acquired within a breath-hold at end-expiration using retrospective Electrocardiography (ECG) gating. Eight to ten breath-hold scans were acquired to cover the entire heart from base to apex. TPM data were acquired at basal, midventricular and apical short-axis locations (same as selected 3 slices of cine image stack) using a black-blood prepared two-dimensional (2D) cine phase-contrast sequence with tri-directional phase encoding [13, 14]. Imaging parameters were as follows: Slice thickness = 8 mm, in-plane spatial resolution = 2.0 × 2.0 mm², TR/ TE/ Flip angle = 5.2 ms/ 3.4 ms/ 10°, bandwidth = 650 Hz/pixel. TPM images were acquired at high temporal resolution (20.8 ms per cardiac time frame) with prospective ECG gating. Spatio-temporal imaging acceleration (*k-t* parallel imaging PEAK GRAPPA) with a net acceleration factor of R_{net} = 3.6 was employed which allowed data acquisition during breath-holding (breath-hold time = 25 heart beats per slice) [14].

Image processing

HDA—All cine MR images (with two-chamber, four-chamber and short-axis views) were loaded to a dedicated workstation (Dell, STUDIO, SPS 435T) and analyzed with prototype software programmed in Visual C++ (TrufiStrain, Version 1.0, Siemens Corporate Technology, Princeton, NJ) by an experienced observer (___, reader #1, with 9 years of experience in cardiovascular imaging. After anchor points were manually set on landmark anatomic structures (left atrium, aortic root, right atrium, LV base, LV apex, and right atrium), the myocardium borders (epicardial and endocardial) were automatically traced based on a previously described existing algorithm without manually drawing contours [15]. As part of the automatic contour delineation, frame-to-frame elastic image registration was performed to provide motion deformation fields using a DIR algorithm described previously [16]. Each pixel in the deformation field was assigned a 2D displacement vector. To register two images and solve for the 2D displacement vectors, the algorithm minimizes the local cross correlation between the displaced pixels of the moving image and the reference image, while keeping the deformation field smooth. Between any two cardiac time frames for a given LV slice, a dense deformation was computed using gradient descent minimization. The registration is kept inverse consistent by solving for the deformation fields and the inverse deformation fields at every step of the gradient descent minimization. Myocardial velocity components were derived from the deformation fields. Next, the in-plane time-resolved regional myocardial velocities in radial (*V_{rr}*) and circumferential (*V_{cc}*) direction were generated for each myocardial segments (on a standard 16-segment AHA model) in systole (sys) and diastole (dia).

TPM—TPM data were processed at the same workstation using an in-house software package programmed in MATLAB [17]. Reader #1 manually traced the epicardial and endocardial LV contours in basal, midventricular, and apical short-axis slices for all acquired cardiac time frames. The acquired time-resolved velocities in three-directions (*V_x*(*t*), *V_y*(*t*),

$V_z(t)$) were transformed into radial velocities ($V_r(t)$), circumferential velocities ($V_\Phi(t)$), and long-axis velocities ($V_z(t)$). The resulting velocities were thus adapted to the principal motion directions of the heart and represented LV contraction/expansion (V_r), rotation (V_Φ) in the short-axis view of LV. Positive velocities were defined for systolic contraction / shortening / clockwise rotation while negative values indicated diastolic expansion / lengthening / anticlockwise rotation. All velocity data were also mapped on the standard AHA LV model by averaging velocities of all pixels within the myocardial segments.

For 5 randomly chosen healthy subjects, reviewer #1 re-analyzed all cine and TPM images with a 1-month interval to test the intra-observer variability. A second reviewer (___, with 2 years of experience of clinical radiology) independently analyzed those cases with the same methods to evaluate inter-observer variability.

Statistical analysis

All continuous variables were reported as mean $\pm$ one standard deviation (SD). Total processing times for each case were recorded and compared with paired t-tests. Average measures of velocities at each time point were calculated for three LV slices (from 6 segments at the Base and Mid levels and 4 segments at the Apex level). On a per-segment basis and a per-slice basis, peak systolic and peak diastolic velocities acquired with HDA and TPM were related with the Pearson's correlation coefficient (r). In order to directly compare HDA- and TPM data that were acquired using different temporal resolutions, HDA derived myocardial velocity-time curves were interpolated to the TPM temporal resolution (20.8 ms). The time curves showing overall changes of velocities during the cardiac cycle (before and after interpolation) were separately drawn to demonstrate the correlations of HDA- and TPM-derived velocities. Associations between HDA and TPM-derived velocity data at adjusted time points were investigated using linear regression models. Bland-Altman plots were applied to evaluate intra-, inter-observer variability of velocities acquired with different methods. Statistical analysis was performed using SPSS software (Version 13.0, Chicago, IL). A p value < 0.05 was considered to be statistically significant.

Results

SSFP cine and TPM images were successfully acquired in all 19 volunteers. All images were eligible for quantitative analysis and resulted in 57 LV slices (304 myocardial segments) for the comparison of two sets of velocity data from MR imaging. Compared to TPM (with semi-automatic analysis), HDA was able to generate LV contours automatically and required significantly shorter processing times (2.3 ± 1.1 minutes/case vs. 9.5 ± 3.7 minutes/case, $p < 0.001$). Figure 1 shows a typical LV plane processed with HDA and TPM for measuring myocardial velocity.

On a per-segment basis, there were significant ($p < 0.05$) correlations (moderate to good) between velocity measures acquired with HDA and TPM on multiple myocardial segments ($r = 0.460 - 0.774$). Figure 2 shows side-by-side comparisons and correlations of segmental velocity reads for both methods.

On a per-slice basis, significant correlations ($r = 0.409 - 0.814$) of peak velocity components measured with HDA and TPM in systole and diastole were found. However, significant differences in peak myocardial velocities acquired with HDA and TPM were observed for most myocardial short axis locations. See table 2 for the summary of correlations and differences of peak velocities in systole and diastole acquired with HDA and TPM. For each subject, the original myocardial motion patterns at the “base”, “mid” and “apex” levels of the LV within a cardiac cycle were presented in the supplemental data (a).

After the normalization of TPM and cine MR images (time window was set at 900 ms) to enroll velocity data from all participants with different R-R intervals, HDA- and TPM-derived velocity-time curves in all subjects, represented by 42 velocity data points with the same time interval (20.8 ms), were similar (figure 3). The linear regression models showed the adjusted velocity measures separately described by HDA and TPM were highly correlated. However, different slopes of fit lines were found for individual LV slices (figure 4).

There was good intra- and inter-observer agreements of major myocardial measures for HDA (peak V_{rr-sys} , V_{rr-dia} , V_{cc-sys} and V_{cc-dia}) and TPM (peak V_r-sys , V_r-dia , $V_{\Phi-sys}$ and $V_{\Phi-dia}$) in 5 volunteers (80 myocardial segments). See the supplemental data (b) for the Bland-Altman plots showing intra- and inter-observer variations for observing myocardial velocities using two different methods.

Discussion

In the present method study, we applied HDA to acquire myocardial velocity measures based on standard cine SSFP images, with the benefit of not requiring additional scans. Our data shows that regional peak systolic and diastolic myocardial velocities in radial and circumferential directions derived from HDA and TPM demonstrated good correlation between techniques in a cohort of healthy volunteers. In addition, HDA required significantly less image processing time than TPM.

Recently, myocardial velocity has been identified as an important indicator for identifying cardiovascular disease in its early stage. From the technical viewpoint, several tissue tracking techniques have been applied to detect potentially impaired myocardial function as a quantitative imaging biomarker for the evaluation of the development of cardiovascular diseases [18–20]. Since cine SSFP MR images are routinely acquired for the calculation of EF and myocardial mass in clinical practice and most cardiovascular research studies, the application of such techniques provides a valuable opportunity for acquiring “historical” information of myocardial motion on existing cardiovascular MR imaging data. For example, feature tracking (FT) is an existing method for the description of myocardial motion or deformation [7]. Features of a target, such as corner-, edge-, or speckle-like features, can be extracted and tracked in series image frames (acquired chronologically) based on the brightness and/or color using various FT algorithms, including particle filter, mean shift, and Kanade-Lucas-Tomasi (KLT) tracker [21]. Mechanical indices, including strain and velocity, can then be calculated based on the moving trajectory of the target in a three-dimensional (3D) space.

Developed for the same purpose, HDA allows to effectively tracking of all structures within an image series using the DIR approach, which is constrained since the heart moves as a continuum. Hence, this principle of DIR is distinct from regional FT of discrete material points [22]. The DIR-based myocardial strain measures have been validated by using MR tagging as a reference [9]. However, the accuracy of DIR-based regional myocardial velocity has not been assessed in human subjects. In the present study, we validated the HDA method in assessment of myocardial motion by relating DIR-based velocity measures to the results of TPM. TPM represents an established MR imaging principle, phase-contrast imaging, for the quantification of regional myocardial velocity [23]. Using bipolar motion encoding gradients, TPM can directly measure the underlying tissue motion velocity of each image pixel with high temporal resolution. Researchers used TPM to detect differences of myocardial motion patterns among healthy subjects, patients with dilated cardiomyopathy (DCM) and patients with left bundle branch block (LBBB) [24]. In addition, Rider et al. demonstrated lower peak myocardial velocity in asymptomatic subjects with obesity, a prominent risk conditions for cardiovascular diseases, as compared to volunteers with normal body weight using TPM [4].

However, despite close ties between HDA- and TPM-derived measures, we also noticed that inter-method correlations of velocity measures were not significant in some myocardial segments or slices. In addition, there are discrepancies of regional myocardial peak velocities measured with two methods on the same imaging plan. Therefore, HDA and TPM are not completely interchangeable. We believed that several factors may jointly contribute to those phenomena. First, the inherent technical differences of the two methods may possibly contribute to the variances between the two sets of velocity measures. Since we were using retrospective gating for cine and perspective gating for TPM, differences of temporal resolution may contribute to the discrepancies of measures derived from those two techniques. Second, the automatic border segmentation algorithm used in the HDA analysis was based on signal contrast between the blood-pool and myocardium. Therefore, the HDA-derived velocity measures may be affected by “external structures” especially when the deformation field covers “too much” or “too less” of the LV area. Compared to high myocardium-blood pool contrast in cine SSFP magnitude images, TPM images usually suffer a low signal to noise ratio (SNR) in the myocardium. As a result, inaccurate contours due to poor signal contrast between the myocardium and blood-pool may possibly affect the velocity measures. Third, the amplitude of the HDA and TPM curves in figure 2 looks more comparable for radial than for circumferential velocities. This may be explained by the fact that there is more image contrast information in the radial direction (i.e. during radial thickening) than in the circumferential direction (i.e. during twisting and circumferential shortening). This could introduce an possible underestimation of circumferential velocities for the HDA approach. Therefore, further studies are needed to investigate how to effectively eliminate the caliber systemic bias between HDA and TPM for the comprehensive interpretation of myocardial velocity data from different sources in cardiovascular risk estimation.

Our study has limitations. First, we did not use a “third party” reference to evaluate the accuracy of either HDA or TPM for quantifying myocardial velocities. However, there is not a well-accepted “normal range” defined by such a “gold standard” for noninvasively

measuring regional myocardial velocities in clinical practice [25]. Second, our study only included radial and circumferential velocity components for comparisons. Limited by the fixed MR imaging protocol, we were unable to acquire cine images with multiple long-axis views, which are necessary to match all myocardial segments present on short-axis TPM images. Third, we only tested one HDA software package (developed on one typical DIR algorithm) in healthy volunteers, further studies are needed to test and compare the clinical values of HDA methods to other similar technical platforms, such as FT, in cardiovascular risk estimation for special patient groups. Nevertheless, the motion patterns on septum and free wall of the LV are different. Such a diversity help to test the capability of HDA in detecting regional myocardial motion indices in different scales.

In conclusion, by providing comparable velocity measures as TPM does, HDA may serve as an alternative for measuring regional myocardial velocity with a faster image processing procedure.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

This study was supported by two grants from the National Institute of Health (R01HL117888 and K01HL121162)

List of abbreviations

HDA	Heart deformation analysis
TPM	Tissue phase mapping
MRI	Magnetic resonance imaging
V_{rr}	HDA-derived radial velocity
V_{cc}	HDA-derived circumferential velocity
V_r	TPM-derived radial velocity
V_Φ	TPM-derived circumferential velocity

References

1. Yan RT, Bluemke D, Gomes A, Burke G, Shea S, Liu K, et al. Regional left ventricular myocardial dysfunction as a predictor of incident cardiovascular events MESA (multi-ethnic study of atherosclerosis). *Journal of the American College of Cardiology*. 2011; 57(17):1735–44. doi: 10.1016/j.jacc.2010.10.060. [PubMed: 21511109]
2. Cicala S, de Simone G, Roman MJ, Best LG, Lee ET, Wang W, et al. Prevalence and prognostic significance of wall-motion abnormalities in adults without clinically recognized cardiovascular disease: the Strong Heart Study. *Circulation*. 2007; 116(2):143–50. doi:10.1161/CIRCULATIONAHA.106.652149. [PubMed: 17576870]
3. Ernande L, Thibault H, Bergerot C, Moulin P, Wen H, Derumeaux G, et al. Systolic myocardial dysfunction in patients with type 2 diabetes mellitus: identification at MR imaging with cine

- displacement encoding with stimulated echoes. *Radiology*. 2012; 265(2):402–9. doi:10.1148/radiol.12112571. [PubMed: 22929334]
4. Rider OJ, Ajufo E, Ali MK, Petersen SE, Nethononda R, Francis JM, et al. Myocardial tissue phase mapping reveals impaired myocardial tissue velocities in obesity. *The international journal of cardiovascular imaging*. 2015; 31(2):339–47. doi:10.1007/s10554-014-0548-z. [PubMed: 25301275]
 5. Taylor RJ, Umar F, Moody WE, Meyyappan C, Stegemann B, Townend JN, et al. Feature-tracking cardiovascular magnetic resonance as a novel technique for the assessment of mechanical dyssynchrony. *International journal of cardiology*. 2014; 175(1):120–5. doi:10.1016/j.ijcard.2014.04.268. [PubMed: 24852836]
 6. Ceelen F, Hunter RJ, Boubertakh R, Sommer WH, Armbruster M, Schilling RJ, et al. Effect of atrial fibrillation ablation on myocardial function: insights from cardiac magnetic resonance feature tracking analysis. *The international journal of cardiovascular imaging*. 2013; 29(8):1807–17. doi:10.1007/s10554-013-0287-6. [PubMed: 24002783]
 7. Hor KN, Gottliebson WM, Carson C, Wash E, Cnota J, Fleck R, et al. Comparison of magnetic resonance feature tracking for strain calculation with harmonic phase imaging analysis. *JACC Cardiovascular imaging*. 2010; 3(2):144–51. doi:10.1016/j.jcmg.2009.11.006. [PubMed: 20159640]
 8. Lin K, Collins JD, Lloyd-Jones DM, Jolly MP, Li D, Markl M, et al. Automated Assessment of Left Ventricular Function and Mass Using Heart Deformation Analysis: Initial Experience in 160 Older Adults. *Acad Radiol*. 2016; 23(3):321–5. doi:10.1016/j.acra.2015.10.020. [PubMed: 26749328]
 9. Phatak NS, Maas SA, Veress AI, Pack NA, Di Bella EV, Weiss JA. Strain measurement in the left ventricle during systole with deformable image registration. *Medical image analysis*. 2009; 13(2):354–61. doi:10.1016/j.media.2008.07.004. [PubMed: 18948056]
 10. Delfino JG, Johnson KR, Eisner RL, Eder S, Leon AR, Oshinski JN. Three-directional myocardial phase-contrast tissue velocity MR imaging with navigator-echo gating: in vivo and in vitro study. *Radiology*. 2008; 246(3):917–25. doi:10.1148/radiol.2463062155. [PubMed: 18223122]
 11. Pelc LR, Sayre J, Yun K, Castro LJ, Herfkens RJ, Miller DC, et al. Evaluation of myocardial motion tracking with cine-phase contrast magnetic resonance imaging. *Investigative radiology*. 1994; 29(12):1038–42. [PubMed: 7721545]
 12. Arai AE, Gaither CC 3rd, Epstein FH, Balaban RS, Wolff SD. Myocardial velocity gradient imaging by phase contrast MRI with application to regional function in myocardial ischemia. *Magnetic resonance in medicine : official journal of the Society of Magnetic Resonance in Medicine / Society of Magnetic Resonance in Medicine*. 1999; 42(1):98–109.
 13. Jung B, Ullmann P, Honal M, Bauer S, Hennig J, Markl M. Parallel MRI with extended and averaged GRAPPA kernels (PEAK-GRAPPA): optimized spatiotemporal dynamic imaging. *Journal of magnetic resonance imaging : JMRI*. 2008; 28(5):1226–32. doi:10.1002/jmri.21561. [PubMed: 18972331]
 14. Jung B, Honal M, Ullmann P, Hennig J, Markl M. Highly k-t-space-accelerated phase-contrast MRI. *Magnetic resonance in medicine : official journal of the Society of Magnetic Resonance in Medicine / Society of Magnetic Resonance in Medicine*. 2008; 60(5):1169–77. doi:10.1002/mrm.21764.
 15. Jolly M-P, Guetter C, Lu X, Xue H, Guehring J. Automatic Segmentation of the Myocardium in Cine MR Images Using Deformable Registration. *Statistical Atlases and Computational Models of the Heart Imaging and Modelling Challenges*. 2012; 7085:98–108. doi:10.1007/978-3-642-28326-0_10.
 16. Guetter, C.; Xue, H.; Ched'hotel, C.; Guehring, J. Efficient symmetric and inverse-consistent deformable registration through interleaved optimization. *Biomedical Imaging: From Nano to Macro, 2011 IEEE International Symposium on* 2011; p. 4
 17. Collins J, Sommerville C, Magrath P, Spottiswoode B, Freed BH, Benzuly KH, et al. Extracellular volume fraction is more closely associated with altered regional left ventricular velocities than left ventricular ejection fraction in nonischemic cardiomyopathy. *Circulation Cardiovascular imaging*. 2015; 8(1) doi:10.1161/CIRCIMAGING.114.001998.
 18. Schneeweis C, Lapinskas T, Schnackenburg B, Berger A, Hucko T, Kelle S, et al. Comparison of myocardial tagging and feature tracking in patients with severe aortic stenosis. *The Journal of heart valve disease*. 2014; 23(4):432–40. [PubMed: 25803969]

19. Buss SJ, Krautz B, Hofmann N, Sander Y, Rust L, Giusca S, et al. Prediction of functional recovery by cardiac magnetic resonance feature tracking imaging in first time ST-elevation myocardial infarction. Comparison to infarct size and transmural by late gadolinium enhancement. *International journal of cardiology*. 2015; 183:162–70. doi:10.1016/j.ijcard.2015.01.022. [PubMed: 25675901]
20. Onishi T, Saha SK, Delgado-Montero A, Ludwig DR, Onishi T, Schelbert EB, et al. Global Longitudinal Strain and Global Circumferential Strain by Speckle-Tracking Echocardiography and Feature-Tracking Cardiac Magnetic Resonance Imaging: Comparison with Left Ventricular Ejection Fraction. *Journal of the American Society of Echocardiography : official publication of the American Society of Echocardiography*. 2015 doi:10.1016/j.echo.2014.11.018.
21. Shi, jT; Carlo. Good Features to Track. *IEEE Conference on Computer Vision and Pattern Recognition (CVPR'94)*; 1994. p. 8
22. Boldea V, Sharp GC, Jiang SB, Sarrut D. 4D-CT lung motion estimation with deformable registration: quantification of motion nonlinearity and hysteresis. *Medical physics*. 2008; 35(3): 1008–18. [PubMed: 18404936]
23. Petersen SE, Jung BA, Wiesmann F, Selvanayagam JB, Francis JM, Hennig J, et al. Myocardial tissue phase mapping with cine phase-contrast mr imaging: regional wall motion analysis in healthy volunteers. *Radiology*. 2006; 238(3):816–26. doi:10.1148/radiol.2383041992. [PubMed: 16424246]
24. Foell D, Jung BA, Germann E, Staehle F, Bode C, Hennig J, et al. Segmental myocardial velocities in dilated cardiomyopathy with and without left bundle branch block. *Journal of Magnetic Resonance Imaging*. 2013; 37(1):119–26. doi:10.1002/jmri.23803. [PubMed: 22987362]
25. Marwick TH, Leano RL, Brown J, Sun JP, Hoffmann R, Lysyansky P, et al. Myocardial strain measurement with 2-dimensional speckle-tracking echocardiography: definition of normal range. *JACC Cardiovascular imaging*. 2009; 2(1):80–4. doi:10.1016/j.jcmg.2007.12.007. [PubMed: 19356538]

Figure 1.

A 42 years old female volunteer. Myocardial velocity was acquired with HDA and TPM on the base level of the LV. The myocardium-blood pool contrast of cine SSFP is superior than that of TPM images.

a A short-axis SSFP image without contours.

b A short-axis SSFP images with contours generated by HDA.

c A TPM image (at the same image plane) without contours.

d A TPM image without contours (manually drawn).

Figure 2.
Comparisons of myocardial velocity measures between HDA and TPM. Radial and circumferential peak LV velocities were mapped onto the AHA 16-segment model.
* Indicates correlations between segmental velocity reads with Pearson correlation coefficient (r) > 0.4 and p < 0.05.

Figure 3.

Time-resolve velocity-time curves describing the changes of velocity (HDA vs. TPM) during the cardiac cycle. The individual graphs represent radial and circumferential time-resolved velocities averaged over all 19 subjects. The temporal resolutions of cine images (retrospective gating, 25 frames) and TPM images (perspective gating with a fixed temporal resolution) were interpolated to a common temporal resolution (20.8 ms) for direct comparisons. The error bars represent inter-individual standard deviations. A 900 ms cut-off point was applied for all subjects with different R-R intervals. Note: The peak velocities at three LV levels shown on this figure could be different from the averages of original peak velocity for all subjects.

Figure 4.

Linear regression models show the correlations of average time-resolved velocity measures (at three LV levels) in radial and circumferential directions (after adjustment for different temporal resolutions and R-R intervals) acquired with HDA and TPM.

Table 1

The description of subjects in the present study

	Volunteers (N = 19)
Male (%)	15 (79)
Age (years old)	50.1 ± 15.5
Weight (kg)	89.6 ± 19.5
Height (cm)	173.1 ± 10.1
Heart rate (beats/minute)	69.7 ± 10.2
Systolic blood pressure (mmHg)	98.0 ± 17.1
Diastolic blood pressure (mmHg)	76.6 ± 12.9

Table 2

Comparisons and correlations of myocardial velocities (at a per-slice basis) acquired with HDA (V_{rr} and V_{cc}) and TPM (V_r and V_Φ) in systole and diastole. All numbers represent mean velocities over all 19 subjects.

	Peak radial velocities			Peak circumferential velocities		
	V _{rr} (cm/s)	V _r (cm/s)	<i>r</i>	V _{cc} (cm/s)	V _Φ (cm/s)	<i>r</i>
Base (sys)	2.96±0.48	2.77±0.52 *	0.686▲	-0.69±0.67	-1.80±1.06 *	0.644▲
Mid (sys)	3.30±0.54	2.77±0.46 *	0.561▲	-1.54±0.60	-2.80±0.99 *	0.715▲
Apex (sys)	3.12±0.98	2.51±0.54 *	0.087	-1.17±0.52	-3.06±1.00 *	0.267
Base (dia)	-2.63±0.62	-3.04±1.19	0.490▲	0.74±0.29	1.06±0.67	0.814▲
Mid (dia)	-2.72±0.62	-2.51±1.27	0.420▲	0.98±0.36	1.61±0.81 *	0.546▲
Apex (dia)	-2.21±0.78	-2.75±1.18 *	0.009	0.90±0.50	2.59±0.97 *	0.409▲

* indicates statistical significance of differences between velocity measures acquired with HDA and TPM (paired t-test, $p < 0.05$).

▲ indicates statistical significance of correlations between velocity measures acquired with HDA and TPM (Pearson correlation coefficient [*r*], $p < 0.05$)