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Quantification of Left Ventricular Volumes, Mass and Ejection Fraction using Cine Displacement Encoding with Stimulated Echoes (DENSE) MRI

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Abstract

Purpose—To test the hypothesis that magnitude images from cine Displacement Encoding with Stimulated Echoes (DENSE) MRI can accurately quantify left ventricular (LV) volumes, mass, and ejection fraction.

Materials and Methods—Thirteen mice (C57BL/6J) were imaged using a 7T ClinScan MRI. A short-axis stack of cine T2-weighted black blood (BB) images was acquired for calculation of left ventricular volumes, mass, and ejection fraction (EF) using the gold standard sum-of-slices methodology. DENSE images were acquired during the same imaging session in three short-axis (basal, mid, apical) and two long-axis orientations. A custom surface fitting algorithm was applied to epicardial and endocardial borders from the DENSE magnitude images to calculate volumes, mass, and EF. Agreement between the DENSE-derived measures and BB-derived measures was assessed via coefficient of variation (CoV).

Results—3D surface reconstruction was completed on the order of seconds from segmented images, and required fewer slices to be segmented. Volumes, mass, and EF from DENSE-derived surfaces matched well with BB data (CoVs ~ 11%).

Conclusion—LV mass, volumes, and ejection fraction in mice can be quantified through sparse (5 slices) sampling with DENSE. This consolidation significantly reduces the time required to assess both mass/volume-based measures of cardiac function and advanced cardiac mechanics.

Keywords

magnetic resonance imaging; DENSE; ventricular volume and mass; heart

INTRODUCTION

One of the strengths of MRI is the ability to manipulate the sequence and composition of the excitation pulses to make functional measurements of *in vivo* motion. These functional measures are particularly important in the assessment of heart and vascular function in cardiology as there is growing evidence to suggest that measures of advanced cardiac mechanics (e.g., cardiac strains, torsion, and synchrony) are better predictors of outcomes for patients with cardiovascular disease than ventricular volumes and mass (1,2). Spatial modulation of magnetization (SPAMM)(3,4), strain-encoded (SENC) MRI(5), and displacement encoding (e.g., with stimulated echoes; DENSE) for strain measurement (6–8) are prominent examples of MR acquisition techniques used to make such measurements. An important limitation of these advanced techniques is time: more complex sequences require more time to acquire, whereas the time available in clinical or research imaging protocols is limited and costly. Thus, critical tradeoffs arise between the need to collect routine clinical data (e.g., ventricular mass, volumes, and ejection fraction in the case of cardiac function assessments) and the ability to make use of these more advanced techniques to derive more detailed information.

A novel way to circumvent such tradeoffs is to leverage the data from an advanced sequence like DENSE to obtain both the needed anatomical and functional data, and forego acquisition of the more routine images (such as a cine stack for cardiac function). Similar logic has motivated studies to quantify ventricular volumes from tags (9,10); this ability is also an integral part of full 3-dimensional deformation recovery from tagged imaging (11,12). For DENSE, the additional acquisition time per slice means that fewer slices can be captured over the region of interest, which makes the calculation/rendering of volumes more complicated than for cases where a contiguous slice stack is available. Yet, we hypothesize that these limitations can be overcome by using advanced image post-processing. In this paper, we evaluate this hypothesis in mice by 1) developing an image processing algorithm to reconstruct a full volume of the left ventricular muscle and cavity from sparsely-spaced image data, 2) applying this methodology to DENSE magnitude images for calculation of ventricular volumes, mass, and ejection fraction, and 3) comparing these measures derived from DENSE magnitude images to those obtained from traditional cine black-blood image stacks.

MATERIALS AND METHODS

Mouse models

Thirteen 8-month-old C57BL/6J mice were used for the study. Animals were housed in ventilated cages in a temperature-controlled room with a 14:10 light:dark cycle and provided with enrichment in the form of acrylic huts and nesting material. Four of the mice had diet-induced obesity (48.6 ± 1.1 grams) from a high fat diet (60% of calories from fat; Research Diets #D12492), while the remaining nine mice were normal weight (35.4 ± 2.8 grams; diet with 10% of calories from fat; Research Diets #D12450B). This variable was included to introduce heterogeneity into the sizes and dimensions of the imaged ventricles (i.e., obesity-induced cardiomyopathy vs. normal) and thus provide a more thorough evaluation of the proposed method. All animal procedures conformed to Public Health Service policies for

humane care and use of animals, and all procedures were approved by the institutional animal care and use committee at our institution.

Cardiac MRI

Cardiac MRI was performed on a 7-Tesla Bruker ClinScan (Bruker, Ettlingen, Germany) equipped with a 4-element phased array cardiac coil and a gradient system with a maximum strength of 450 mT/m and a maximum slew rate of 4500 mT/m/s. Animals were anesthetized with isoflurane using a precision vaporizer delivering 1.5–2.5% isoflurane in oxygen at a rate of 1.0 L/min. Once anesthetized, three legs were shaved for placement of cutaneous ECG electrodes required for cardiac gating. A diaphragm to sense breathing was placed under the abdomen for respiratory gating in order to minimize motion artifact. A rectal thermometer was used to monitor core temperature. During scanning, all vital signs including heart rate, respiratory rate, and core temperature were continuously monitored with a fiber optic system (SA Instruments, Inc, Stony Brook, NY). Body temperature was maintained between 36 and 37°C with a heated water blanket.

The protocol for cardiac black blood imaging in mice using a double inversion recovery gradient echo pulse sequence (13). Immediately following electrocardiogram R-wave trigger detection, a combination of nonselective/selective 180° pulses was applied with the gradient echo readout occurring on the subsequent R-wave. In this way, 9 ventricular short-axis slices (spanning LV apex to base with no slice gaps) and two long-axis slices (2- and 4-chamber views) were acquired per mouse (see Figure 1A). This sequence is the standard for ventricular function quantification in animals at high field strengths, as opposed to balanced steady-state free precession (SSFP) techniques frequently used in humans, because of the susceptibility of SSFP to off-resonance artifacts at high field strength (13). Image details are provided in Table 1. Each slice took 2–3 minutes to acquire depending on the heart and respiratory rates.

The acquisition protocol for cine DENSE in mice has also been described in detail previously (6,14). Immediately after an electrocardiogram R-wave trigger detection, a displacement encoding module consisting of radiofrequency and gradient pulses was applied, which stores position-encoded longitudinal magnetization. This initial encoding was followed by successive applications of a readout module, consisting of a radiofrequency excitation pulse, a displacement un-encoding gradient, and an interleaved spiral k-space trajectory. This sequence creates 3 images: a magnitude image and two phase images (encoded for 'x' and 'y' displacements, respectively). This study exclusively utilized the magnitude image. We acquired 3 short-axis images and 2 long-axis images (2- and 4-chamber views) for each mouse (see Figure 1A). The short-axis images were planned perpendicular to the 4-chamber long-axis image such that the mid-ventricle slice was positioned at 50% of the end-systolic ventricular length, with the apical and basal slices placed 20% of the end-systolic ventricular length above and below the mid-ventricle. Image details are provided in Table 1. Each two-dimensional image acquisition took 6–9 minutes depending on the heart and respiratory rates.

Image Processing

Figure 1 presents a visual summary of the image segmentation steps from both the DENSE and black blood images. For the end diastolic cardiac phase, the endocardial and epicardial borders were traced for both the long-axis (Figure 1A, 1B) and short-axis (Figure 1C) slices, while only the endocardial borders were segmented at end systole in all slices (Figure 1D for short-axis example). Papillary muscles were excluded from the myocardial mass in all cases. Segmentation was performed using a custom software written in MATLAB (Mathworks, Inc., Natick, MA), that displayed both short- and long-axis images in a 3-dimensional user interface based on the spatial coordinates provided by the MR scanner in order to facilitate accurate segmentation, especially at the apical and basal slices. Additionally, the 3D surface model required user specification of valve points from the long-axis slices to constrain the base of the model at the valve plane. Figure 1E shows an example of the endocardial ('x') and epicardial ('o') valve points marked on a 4-chamber image. As implemented, the entire segmentation protocol required approximately 2–3 minutes per slice.

Surface Fitting Model

The surface fitting algorithm was designed to construct smooth epicardial and endocardial surfaces that approximate the boundary points specified by the segmentation masks, but that are sharply cut at the valve plane. From these surfaces, the ventricular mass, volumes, and ejection fraction (EF) can be quantified. To accomplish this goal, separate consideration was given to: 1) defining the 3-dimensional valve plane; 2) defining the endocardial and epicardial surface models and their intersection with the valve plane; and 3) calculating the resulting volumes.

1. **Valve Model-** Two sets of valve points were specified in the segmentation software corresponding to the endo- and epicardial intersections and they were averaged together to form a singular set of mid-myocardial valve points. A best-fit plane was specified from the averaged valve points and the points were then projected onto that plane to compute the convex hull. Any points located inside the convex hull were removed as they were considered redundant. The Fourier Descriptor coefficients (15) were computed (up to the second order) from the valve points on the convex hull and evaluated at a preset number of points ($K = 100$) to form the valve boundary (Figure 2A). This boundary was shifted in both normal directions to the centers of the original epicardial and endocardial valve points, creating two identical parallel valve boundaries that span the intersection with the myocardium (Figure 2D).
2. **Smooth Surface Model-** A closed form expression of the smoothest interpolator for a set of arbitrarily located points on a sphere does not exist; an approximation of this interpolator is referred to as a pseudo thin plate spline, which is the approach we retained for this model (16). Wahba (16) proposed a class of pseudo thin plate splines on the sphere and provided the corresponding closed form expression, which has the following form:

$$f(\hat{u}) = \alpha_0 + \sum_{n=1}^N \alpha_n \psi(\hat{u} \cdot \hat{u}_n). \quad [1]$$

In Eq. 1 $\hat{u}$ represents a unit vector (i.e., a direction in which the function needs to be evaluated); $\hat{u}_n$ represent a set of N unit vectors; $a_0, \dots, a_N$ are model coefficients; and function $\psi: [-1,1] \mapsto \mathbb{R}$ defines the type of the pseudo thin plate splines. We use ψ for the case of $m = 2$ in (16), that is:

$$\psi(x) = \frac{1}{2\pi} \left[\frac{1}{2} q(x) - \frac{1}{6} \right], \quad [2]$$

where

$$q(x) = \begin{cases} \left[(12W^2 - 4W) \ln \left(1 + \frac{1}{\sqrt{W}} \right) - 12W \sqrt{W} + 6W + 1 \right] / 2 & -1 \leq x < 1 \\ \frac{1}{2} & x = 1 \end{cases} \quad [3]$$

and

$$W = \frac{1-x}{2} \quad [4]$$

While Wahba proposed to use the model given by Eq. 1 as an interpolator (16), here it is used as an approximator. To use the model as an approximator, we uniformly sampled the sphere with $N = 1000$ unit vectors, $\hat{u}_n$. Let $\vec{v}_m = v_m \hat{v}_m$ ($m = 1, \dots, M$), represent the boundary points to be approximated with the surface model, while $\vec{w}_k = w_k \hat{w}_k$ ($k = 1, \dots, K$), represent the valve boundary points. The goal is to determine the coefficients of the model given by Eq. 1 that result in a smooth surface which approximates the surface boundary points as closely as possible while satisfying the constraints of the valve boundary. To ensure more fidelity of the surface near the valve boundary (i.e., to improve the ability of the surface model to approximate the valve boundary), unit vectors $\hat{w}_k$ were added to the set of unit vectors $\hat{u}_n$ at which the surface model is defined. The above requirements result in the following optimization problem: find coefficients $a_0, \dots, a_N$ that minimize

$$S = (1-\gamma) \frac{1}{M} \sum_{m=1}^M [f(\hat{v}_m) - v_m]^2 + \gamma \frac{1}{K} \sum_{k=1}^K [f(\hat{w}_k) - w_k]^2 + \lambda \frac{1}{N} \sum_{n=1}^N a_n^2. \quad [5]$$

The first term in Eq. 5 corresponds to matching the surface boundary points, the second term corresponds to matching the valve points and the third term controls the smoothness of the surface. Parameter γ takes a value from $[0,1]$ and it controls the relative importance of the first two terms: when $\gamma = 0$ the second term disappears, when $\gamma = 1$ the first term disappears, when $\gamma = 0.5$ the two terms have the same weight. For the present study, $\gamma = 0.9$, which gave more weight to the valve term (since experiments demonstrated that the valve boundary points needed to be approximated more closely than the boundary points). λ controls the importance of the smoothness term relative to the point matching (first two) terms, and was set to $\lambda = 0.0001$ for the present study. Note that all three terms are normalized (divided by the number of terms in the respective summations), which means that λ did not need to be adjusted for different numbers of boundary points.

In order to minimize S we took derivatives with respect to the model coefficients and set them equal to zero, i.e.,

$$\frac{\partial S}{\partial \alpha_0} = 0, \quad [6]$$

and

$$\frac{\partial S}{\partial \alpha_i} = 0, \quad i=1, \dots, N. \quad [7]$$

Equations 6 and 7 lead to the following system of $N + 1$ linear equations and $N + 1$ unknowns ($\alpha_0, \dots, \alpha_N$).

$$\alpha_0 + \sum_{n=1}^N \alpha_n \left[\frac{1-\gamma}{M} \sum_{m=1}^M \psi(\hat{v}_m \cdot \hat{u}_n) + \frac{\gamma}{K} \sum_{k=1}^K \psi(\hat{w}_k \cdot \hat{u}_n) \right] = \frac{1-\gamma}{M} \sum_{m=1}^M v_m + \frac{\gamma}{K} \sum_{k=1}^K w_k, \quad [8]$$

and for $i = 1, \dots, N$:

$$\alpha_0 \left[\frac{1-\gamma}{M} \sum_{m=1}^M \psi(\hat{v}_m \cdot \hat{u}_i) + \frac{\gamma}{K} \sum_{k=1}^K \psi(\hat{w}_k \cdot \hat{u}_i) \right] + \sum_{n=1}^N \alpha_n \left[\frac{1-\gamma}{M} \sum_{m=1}^M \psi(\hat{v}_m \cdot \hat{u}_n) \psi(\hat{v}_m \cdot \hat{u}_i) + \frac{\gamma}{K} \sum_{k=1}^K \psi(\hat{w}_k \cdot \hat{u}_n) \psi(\hat{w}_k \cdot \hat{u}_i) \right] + \frac{\gamma}{N} \alpha_i = \frac{1-\gamma}{M} \sum_{m=1}^M v_m \psi(\hat{v}_m \cdot \hat{u}_i) + \frac{\gamma}{K} \sum_{k=1}^K w_k \psi(\hat{w}_k \cdot \hat{u}_i) \quad [9]$$

Once the above system (Equations 8 and 9) is solved for $\alpha_0, \dots, \alpha_N$, the surface point in direction $\hat{u}$ is $f(\hat{u})\hat{u}$. This resulting surface is triangulated everywhere except inside the valve boundary, which is capped flat. The above procedure was done for both endocardial (Figure 2B) and epicardial (Figure 2C) surfaces.

3. Volume Calculation- To calculate the volume enclosed by the fitted surfaces (i.e., the internal volume of the endocardium or the ventricular mass enclosed between the epi- and endocardial surfaces), the Divergence theorem was used. This theorem states that for a vector field, $\vec{F}$, and a closed surface, S , the following holds:

$$\int_V \text{div } \vec{F} dV = \oint_S \vec{F} \cdot \hat{n} dS, \quad [10]$$

where $\hat{n}$ is the surface normal and V the volume enclosed by the surface. By selecting $\vec{F} = x\hat{i}$, it follows that $\text{div } \vec{F} = 1$ and consequently

$$V = \int_V dV = \oint_S x n_x dS. \quad [11]$$

Since the surfaces are triangulated, Equation 11 can be modified to

$$V = \sum_{t=1}^T n_x^t \frac{x_1^t + x_2^t + x_3^t}{3} A^t, \quad [12]$$

where T is the number of triangles; index t denotes the triangle number; n_x^t is the x component of its normal vector; x_1^t , x_2^t , and x_3^t are the x coordinates of its three vertices; and A^t is its area.

Endpoints and Statistics

The study objective was to evaluate the accuracy with which the novel surface fitting algorithm described could be used in conjunction with sparsely sampled (with respect to coverage of the LV volume) DENSE magnitude images to quantify LV end-diastolic and end-systolic volumes (EDV, ESV), EF, and LV mass.. These standard volume measures were calculated directly from the black blood short axis stack as the product of the segmented area and the slice thickness.

To first evaluate the accuracy of the surface fitting algorithm, 3D surfaces were created from the full segmented black blood image stack and compared to the sum-of-slices results. Subsequently, the 3D models from the DENSE images were compared to the black blood segmentation data for the primary evaluation of interest. Segmentations were done with the operator blinded to the volume results to avoid bias. A paired t-test was used to compare volume and mass results between methods and the Bland-Altman limits of agreement were assessed (17). Additionally, a modified mean coefficient of variation (CoV) was used to compare variability of measuring a given variable, X , between methods ('3D' surface vs. sum-of-slices ('SS')) over the Z mice to the absolute magnitude of the measurement means, as follows:

$$CoV [\%] = \frac{\sum_{i=1}^Z [St.Dev.(X_{3D} - X_{SS})_i] / Z}{|\sum_{i=1}^Z [(X_{SS})_i] / Z|}. \quad [13]$$

Finally, inter- and intra- user variability of the DENSE-derived volume and mass measures was quantified using CoV and intra-class correlation. Statistical tests were performed using SPSS version 21 (IBM, Armonk, NY).

RESULTS

The proposed surface reconstruction method was implemented and successfully able to converge to a solution and calculate the resulting 3D volume from the segmented set of slices on the order of seconds.

Black Blood 3D Surface vs. Black Blood Slice Summation

Table 2 summarizes the volume, EF, and mass results from the cine black blood images using either the 3D surface fitting or area summation methods. The table also provides the mean pair-wise difference, the 95% limits of agreement, and the CoV for each end point. The corresponding Bland-Altman plots are shown in Figure 3. There was excellent agreement between the two methods, with all CoVs $\leq 8\%$ and all mean differences within 5 units. There was a slight bias for the 3D surface to under-estimate ventricular volumes, as evidenced by the fact that the upper limit of agreement was ≤ 0 for both EDV and ESV.

Black Blood Slice Summation vs. DENSE 3D Surface

Table 3 summarizes the volume, EF, and mass results and comparisons (mean differences, limits of agreement, CoVs, and t-test) between the black blood (sum-of-slices) and DENSE data using the 3D surface fitting algorithm. The corresponding Bland-Altman plots are shown in Figure 4. The CoVs and limits of agreement ranges were slightly higher than for the black blood 3D surface comparison. However, the overall agreement was still very good, with all CoVs $\leq 11\%$ and very little change in the mean differences (as compared to Table 2).

Reproducibility

Table 4 reports the results of the inter- and intra-user reproducibility analyses. With respect to CoV, all results were $\leq 8\%$, while the intra-class coefficients were all ≥ 0.81 .

DISCUSSION

To take advantage of the capabilities of advanced but lengthy MRI acquisition protocols, such as DENSE, it is beneficial to maximize the data derived from such acquisitions and thus replace routine anatomic sequences. This study demonstrated a proof of concept for this paradigm by using cine DENSE magnitude images along with a novel 3D surface fitting algorithm in place of a stack of cine black blood images to quantify left ventricular volumes, mass, and EF in mice. Despite reduced numbers of slices (9 SA for black blood vs. 3 SA, 2 LA for DENSE), and generally reduced SNR in the DENSE magnitude images, we report sufficient accuracy and reproducibility of this approach as evidenced by low CoVs ($\leq 11\%$) among volume, EF, and mass measures from DENSE compared to black blood image segmentation.

To some extent, the variability in these comparisons is a result of differences in the method of volume quantification. As shown in Table 2, even when the underlying data source was identical, the surface volume generally under-estimated the stacked areas. The inclusion of the valve ‘plane’ (in actuality, an amalgamation of the mitral and aortic valve planes) as a constraining factor on the 3D surface was likely a contributor to this bias. Since the valves are typically not aligned with nor easily visualized by ventricular short axis slices, inclusion of part of the aortic root or atrial volume into basal ventricular segmentations is not uncommon. These partial volumes would be included in the area summation, but cropped from the 3D surface by registration of the valve points. Similarly, the finite slice thicknesses of the short-axis stack will tend to over-estimate volumes (particularly at the apex) by assuming uniform area through the slice. Thus, the 3D surface method applied to the black blood image stacks may, in fact, be more accurate than the traditional area summation method, at least in cases of regular and concentric LV shape. The systematic under-sizing bias was no longer present when comparing the 3D surfaces derived from DENSE to the area summation method from black blood images, indicating that differences in image quality and slice numbers between data acquisition are also an important component of the overall variation, as expected.

The ability to use DENSE to quantify ventricular function can result in significant cost savings with respect to total scan time. From the present study, acquiring 5 DENSE slices (with only 2-dimensional displacement encoding) required a minimum of 30 minutes (based on the low end estimate of time per slice). Extending the acquisition to also incorporate through-plane motion (18,19) would obviously extend that time requirement even further. To acquire a cine short axis stack of black blood images with 2- and 4-chamber long-axis views (11 total images in the present protocol) adds an additional 22 minutes (based on the low-end time estimates). By instead using the DENSE images for quantification of both strains and volumes/mass, those 22+ minutes could be utilized to make additional MRI measures, or cut out to save both cost and post-processing time. Alternatively, this approach could also be used in conjunction with traditional cine imaging (black blood or SSFP) to reduce the number of slices required to quantify ventricular volumes and similarly reduce scan time.

Although a direct comparison of normal function and obesity-mediated ventricular dysfunction was not a focus of this study, the inclusion of the obese mice in the study design was notable and important for several reasons. First, only using normal mice could artificially reduce CoVs by providing a homogenous data set. Furthermore, assessing 'disease' function in mice is typically of greater importance and interest than 'normal' function. In that sense, obese mice offered a relevant disease model for the present work since they are known to have cardiac remodeling and hypertrophy (20), and are theoretically more difficult to image due to problems with fat artifacts and generally poorer health/tolerance of anesthesia.

There are several limitations to note. First, we have evaluated the accuracy of DENSE-based left ventricular volume and mass calculations using CMR data from mice, which cannot be directly extrapolated to human data. However, mouse models of ventricular function are extensively used for basic science and translational studies into cardiac disease, so demonstrating proof of concept with this model is an important contribution to small-animal research and will also likely directly translate to humans. Second, the DENSE sequence had lower temporal resolution than the black blood sequence, which may have contributed to the uncertainty (i.e., over-estimation) in end systolic volume quantification; however, this potential effect was not enough to create a significant difference in the means between the two data sets. A significant difference ($p < 0.05$) between the mean ejection fractions was observed (Table 3); however, the low CoV measure for EF suggests that the magnitude of this difference was small compared to the expected measurement values. Finally, no formal optimization of the surface fitting parameters was conducted, which may have led to additional improvements in model accuracy. Even without optimization, the model was good enough to yield results with low variability compared to a gold standard measurement; yet collection of more data sets in future work would provide an optimization training set to potentially achieve additional improvements in accuracy.

In conclusion, we utilized the magnitude images from cine DENSE MRI in conjunction with a novel 3D surface fitting algorithm to accurately and reproducibly quantify left ventricular volumes, ejection fraction, and mass in mice. With this approach, a single DENSE acquisition using as few as 5 slices can be used to evaluate both traditional (volumes, mass

and ejection fraction) and advanced (strains, torsion, synchrony) measures of cardiac function and mechanics without need for a dedicated cine anatomical (e.g., black blood) MR acquisition. This advancement greatly reduces the amount of time required to assess advanced cardiac function using MRI, and could ultimately improve the efficiency of the technique during routine practice.

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Figure 1.

Overview of image processing steps: cine data were acquired from mice using either black blood (top row) or DENSE (bottom row) algorithms in both short- and long-axes. A) shows the number and orientation of these slices with respect to a 4-chamber view. For end diastolic phases, the endo- and epicardial surfaces were segmented, as shown in B) and C) for 2-chamber and short axis images, respectively. However, it is noted that the long-axis black blood images were not used for slice area summation. At end systole, only the endocardium was segmented (D). Finally, the endo- and epicardial valve points at their intersection with the myocardium were marked on the long-axis images (E) for definition of the valve plane in the 3D surface model.

Figure 2.

Components of the 3D surface reconstruction: A) The valve boundary (in pink) calculated from the specified points (in blue), as shown in Figure 1E; B) Posterior lateral view of the endocardial surface (shown in yellow) with valve boundary intersection used for calculation of ventricular volumes; C) Posterior lateral view of the epicardial surface (dark green) encompassing the endocardium, used for calculation of ventricular mass; D) The valve boundary extruded through the epicardial and endocardial surfaces.

Figure 3.

Bland-Altman plots comparing the left ventricular (LV) A) end diastolic volume, B) end systolic volume, C) ejection fraction, and D) mass results between the 3D surface model and segmented area summation method both derived from the black blood image stack. The blue markers denote measurements taken from the mice on a low fat diet, while the red markers correspond to the high fat diet mice.

Figure 4.

Bland-Altman plots comparing the left ventricular (LV) A) end diastolic volume, B) end systolic volume, C) ejection fraction, and D) mass results between the 3D surface model from DENSE magnitude images and the area summation method using the black blood image stack. The blue markers denote measurements taken from the mice on a low fat diet, while the red markers correspond to the high fat diet mice.

Table 1

Imaging parameters for the two scanning protocols

	TR/TE (ms)	Matrix size	FOV (cm)	Signal average	Frames/cardiac cycle	Slice thickness (mm)	No. of Images (SA/LA)
Black Blood	4.0/1.9	128 × 128	2.5 × 2.5	2	25–40	1	9/2
DENSE	7.1/0.67	128 × 128	3.2 × 3.2	3	13–18	1	3/2

TR- repetition time; TE- echo time; FOV- field of view; SA- short axis

Table 2

Left ventricular function assessment from cine black blood images comparing the sum of slices ('Slice sum') to '3D surface' reconstruction as means of quantification

	Slice sum	3D Surface	Mean Difference	B-A limits	CoV (%)
EDV (μL)	64 ± 9	59 ± 9	-5	-7, -2	5
ESV (μL)	24 ± 7	21 ± 6	-3	0, -5	8
EF (%)	63 ± 7	64 ± 6	2	-2, 5	2
LV Mass (mg)	108 ± 9	107 ± 8	-1	-5, 3	1

* Data presented as mean ± standard deviation; B-A limits: Bland-Altman 95% limits of agreement; CoV: mean coefficient of variation

Table 3

Comparison of left ventricular function as derived from the sum of slices from cine black blood images ('Slice sum') and '3D surface' reconstruction from DENSE images

	Slice sum	3D DENSE	Mean Difference	B-A limits	CoV (%)	P-value, T-test
EDV (μL)	64 ± 9	60 ± 5	-3	-23, 16	8	0.21
ESV (μL)	24 ± 7	25 ± 5	1	-8, 11	11	0.27
EF (%)	63 ± 7	58 ± 9	-5	-22, 12	8	0.04
LV Mass (mg)	108 ± 9	104 ± 10	-4	-19, 12	5	0.08

* Data presented as mean ± standard deviation; B-A limits: Bland-Altman 95% limits of agreement; CoV: mean coefficient of variation

Table 4

Inter- and Intra-user reproducibility of DENSE segmentations

	Intra-user		Inter-user	
	CoV (%)	ICC	CoV (%)	ICC
EDV (μL)	4	0.90	5	0.83
ESV (μL)	6	0.87	8	0.81
LV Mass (mg)	4	0.88	4	0.86

* CoV: mean coefficient of variation; ICC: intra-class coefficient