

In Vivo Cardiac MR Elastography in a Single Breath Hold

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Introduction: Left ventricular (LV) myocardial stiffness is an important marker of cardiac disease [1-3]. However, *in vivo* assessment of LV stiffness relies upon invasive measurements of LV pressure and simultaneous assessment of LV volume [1-3]. The invasive nature of these measurements has severely limited the understanding of a variety of diseases, the most notable being heart failure with preserved ejection fraction (HFPEF) [1]. It has been recently demonstrated that magnetic resonance elastography (MRE) [4] can be adapted to measure the shear modulus of myocardium but with the limitations of acquisition speed resulting in multiple breath holds to collect the data required for processing [5,6]. The purpose of this study was to demonstrate the feasibility of *in vivo* cardiac MRE to measure cardiac stiffness at the end-diastolic and end-systolic phases of the cardiac cycle in a single breath hold.

Methods: *In vivo* cardiac MRE was performed on 5 normal volunteers. All imaging was performed in a 1.5 Tesla MRI scanner (Signa Excite, GE Health Care, Milwaukee, WI). The volunteers were positioned in the supine position and placed feet first into the scanner as shown in Figure 1. A gradient echo MRE sequence [7,8] was used to acquire a short-axis slice at the midventricular level avoiding the papillary muscles. Mechanical waves were introduced into the heart by a pneumatic driver system as shown in Figure 1. Imaging parameters included TE/TR= 9.3/12.5 ms; FOV= 35 cm; $\alpha = 30^\circ$; slice thickness= 10 mm; acquisition matrix= 220x64; phase FOV= 1-0.7; receiver bandwidth= ± 62.5 kHz; SENSE acceleration factor of 2; mechanical motion frequency= 80 Hz; heart rate= 60-65 bpm; views per segment or R-R interval (VPS)= 8; 4 MRE time offsets; and bipolar 6.25-ms duration (160-Hz) 2.3 G/cm motion-encoding gradients (MEG) applied separately in the x, y, and z directions to measure the in-plane and through-plane tissue motion. Positive and negative MEG amplitudes were used on alternate views and a phase contrast reconstruction was performed to obtain images of tissue displacement. Specific trigger delays were used to obtain data in the end-systolic and end-diastolic phases of the cardiac cycle. Each motion-encoding direction was acquired within a heart rate dependent breath hold of ~14 sec. The short-axis images for each volunteer were masked with epicardial and endocardial contours to obtain only the left ventricular (LV) myocardium as shown in Figure 2. The x, y and z components of motion were analyzed to determine the phase-difference signal-to-noise ratio (PDSNR) and the effective cardiac stiffness using a direct inversion algorithm.

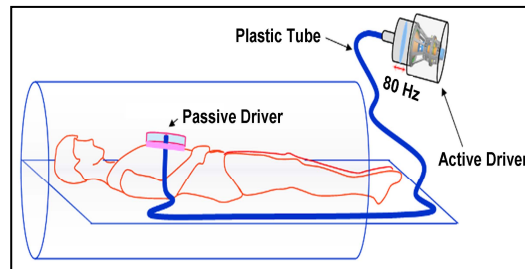


Figure 1. Acoustic Driver. A drum driver is placed on the chest wall. Sound waves are noninvasively transmitted to the driver and into the volunteer. These waves are imaged by MRE and used to calculate stiffness.

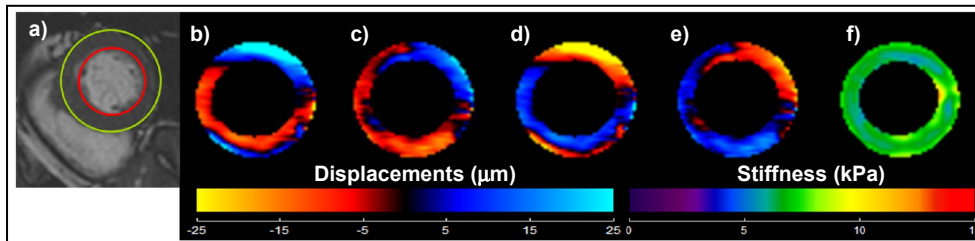


Figure 2. a) Short-axis magnitude image of the myocardium in end-diastole with epicardial (green) and endocardial (red) contours. **b-e)** The four phases of the through-plane component of the propagating waves and **f)** Weighted stiffness map from 3 encoding directions with a mean stiffness of 6.5 ± 0.6 kPa.

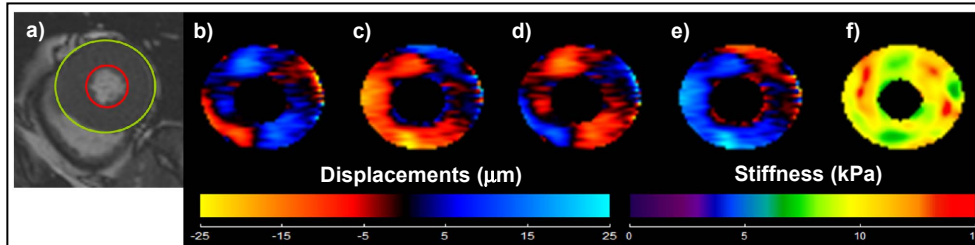


Figure 3. a) Short-axis magnitude image of the myocardium in end-systole. **b-e)** The four phases of the through-plane component of the propagating waves and **f)** Weighted stiffness map from 3 encoding directions with a mean stiffness of 9.8 ± 1.5 kPa.

Results: The PDSNR was calculated in each volunteer during end-diastole and end-systole from the motion in all encoding directions. The PDSNR in end-diastole ranged from 5.0 to 21 and in end-systole from 5.1 to 14. Figure 2 (a-e) shows an example of a short-axis magnitude image during end-diastole with the contours used for delineating the LV myocardium and the phase images of the through-plane component of the propagating waves for one of the volunteers. Figure 2(f) shows the weighted stiffness map from 3 encoding directions with a mean stiffness of 6.5 ± 0.6 kPa. Similarly, Figure 3(a-f) shows the end-systole magnitude and phase images for the same volunteer and the weighted stiffness map from 3 encoding directions with a mean stiffness of 9.8 ± 1.5 kPa. The effective stiffness measurement in end-systole in all volunteers ranged between 8.3 to 11.3 kPa and in end-diastole was 6.5 to 8.7 kPa. The effective stiffness measurements were comparable between volunteers for each cardiac phase. When no external motion was applied, no discernible waves were present in the myocardium, indicating that cardiac MRE is insensitive to the physiologic motion of the heart.

Discussion: These results demonstrate the feasibility of performing *in vivo* cardiac MRE in a single breath hold. The current implementation is also capable of obtaining multiple phases of the cardiac cycle in a cine mode within a breath hold (not shown). Because the current inversion algorithm does not take into account the heart geometry, anisotropy, and 3D wave propagation effects, an effective rather than absolute estimate of shear modulus (i.e. stiffness) is obtained. Future studies will be used to determine the impact of factors such as geometry on these stiffness estimates to determine if corrections are required that can account for the radius and thickness of the LV myocardium. Furthermore, this technology will be evaluated as a means for diagnosing HFPEF and hypertrophic cardiomyopathy.

References:

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