

A Feasibility Study: MR Elastography as a Method to Compare Stiffness Estimates in Hypertrophic Obstructive Cardiomyopathy and in Normal Volunteers

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Introduction: Hypertrophic cardiomyopathy (HOCM) is a genetic cardiac disease and is most common cause of sudden cardiac death among young people [1]. It is known that HOCM is characterized by left ventricular (LV) hypertrophy, increased ventricular stiffness and impaired diastolic filling [2]. However, *in vivo* assessment of LV stiffness relies upon invasive measurements of LV pressure and simultaneous assessment of LV volume [3,4]. The invasive nature of these measurements has severely limited the understanding of a variety of cardiac disease states. It has been recently demonstrated that magnetic resonance elastography (MRE) [5] can be adapted to measure the shear modulus of myocardium [6-8]. The purpose of this study was to demonstrate the feasibility of using MRE to identify changes in the stiffness of LV in HOCM's when compared to normals.

Methods: *In vivo* cardiac MRE was performed on 18 normal volunteers (13 Female and 5 Male; Mean age: 31 yrs; range: 21-71 yrs) and 2 obstructive HOCM's (2 Female; Ages: 26 and 43yrs). 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 cine MRE sequence [7,8] was used to acquire a 2-short-axis slices 9mm apart at the midventricular level avoiding the papillary muscles in the normal volunteers and for HOCM's at the levels where the muscle was hypertrophic. 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= 36 cm; $\alpha = 30^\circ$; slice thickness= 8 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= 60-100 bpm; views per segment or R-R interval (VPS)= 4,8; 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. 10 cardiac phases were reconstructed and the end-systolic image (magnitude image with smallest chamber) was used for analysis. Each motion-encoding direction was acquired within a heart rate dependent breath hold of ~14 sec in end-expiration. The short-axis image for each volunteer was 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 effective cardiac stiffness using a phase gradient inversion algorithm.

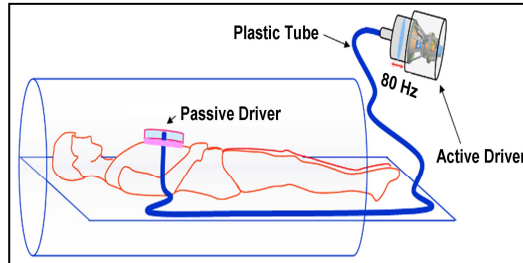


Figure 1. Acoustic Driver. A drum driver is placed on the chest wall. Sound waves are noninvasively transmitted to the driver and into the volunteer. These waves are imaged by MRE and used to calculate stiffness.

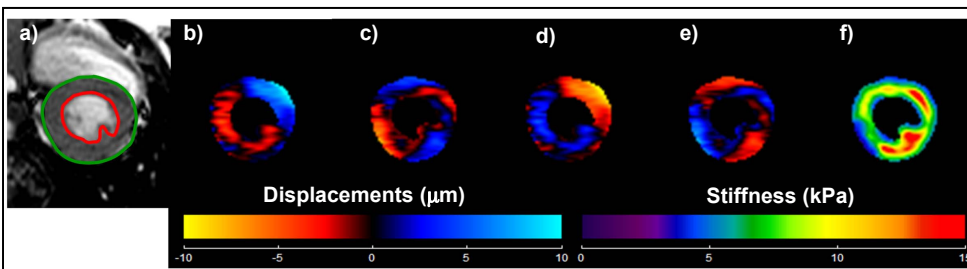


Figure 2. a) Short-axis magnitude image of the myocardium in end-systole with epicardial (green) and endocardial (red) contours in a normal volunteer. **b-e)** The four phases of the through-plane component of the propagating waves and **f)** Weighted stiffness map from 3 encoding directions with a mean stiffness of 7.7 ± 2.7 kPa.

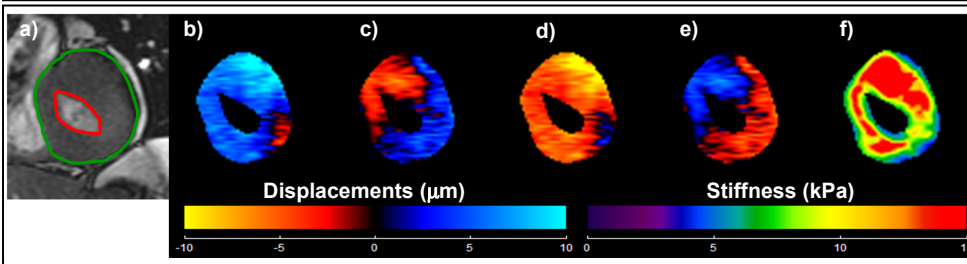


Figure 3. a) Short-axis magnitude image of the myocardium in end-systole in a HOCM. **b-e)** The four phases of the through-plane component of the propagating waves and **f)** Weighted stiffness map from 3 encoding directions with a mean stiffness of 12 ± 6.8 kPa.

Results: Figure 2 (a-e) shows an example of a short-axis magnitude image during end-systole 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 7.7 ± 2.7 kPa. Similarly, Figure 3(a-f) shows the end-systole magnitude and phase images for one of the HOCM's and the weighted stiffness map from 3 encoding directions with a mean stiffness of 12 ± 6.8 kPa. The effective stiffness measurement in end-systole in all volunteers ranged between 3.5 to 7.9 kPa with a mean value of 5.64 ± 1.0 kPa and in HOCM's stiffness values ranged from 12 to 17.2 kPa with a mean value of 14.5 ± 2.2 kPa.

Discussion: These results demonstrate the feasibility of using MRE to identify changes in the stiffness of LV in HOCM's when compared to normals. 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.

References:

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