# The Accuracy of Quantitative MR Elastography in an Anatomically Accurate Diastolic Cardiac Phantom

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**Target audience:** Cardiac radiologists**,** cardiologists, physicians, scientists, and engineers interested in cardiac magnetic resonance elastography (MRE) applications.

**Purpose:** Normal cardiac function is dependent on the mechanical properties of the myocardium[1], which may serve as a valuable predictor of disease. Several groups have investigated the use of Magnetic Resonance Elastography (MRE), a phase-contrast technique capable of making non-invasive stiffness measurements *in vivo*, to probe myocardial stiffness [2-7]. However, a thorough evaluation of the accuracy of the reported stiffness measurements and the optimal experimental settings has not been well established. A typical MRE exam is composed of 3 steps: i) actuation, ii) image acquisition and reconstruction, and iii) inversion. Step one involves the generation of shear waves inside the tissue of interest. This step is performed simultaneously with measurement of the displacement field generated by the propagating shear waves using an MRI phase-contrast technique. Lastly, the acquired shear wave displacement field is used to calculate a stiffness map (elastogram), commonly referred to as wave inversion. The most widely used inversion algorithm (direct or Helmholtz inversion) implicitly assumes interrogated regions are farther than a wavelength from a boundary [8], which may not be the case in a thin-walled structure such as the left ventricle. It has been proposed that applying the curl operator to the 3D displacement field, thus removing longitudinal waves, can account for boundary effects [9-11], but this has not been validated experimentally in such a complex geometry. The purpose of this work, was to evaluate the accuracy of 3D MRE direct inversion on curl wave fields in a morphologically accurate left ventricular phantom using dynamic mechanical testing (DMA) as the reference standard.

**Methods:** An anatomically accurate two-ventricle silicon heart phantom was custom manufactured (The Chamberlain Group, MA, USA) from a segmented ECG-gated computed tomography image volume of a volunteer's heart in diastole. From the same silicon batch as the heart phantom 3 cylindrical samples were poured for mechanical testing purposes. MRE was performed on the heart phantom using a modified spin-echo echo planar imaging sequence, vibration frequencies of 60Hz, 80Hz, 100Hz, 140Hz, 180Hz and 220Hz; TR/TE = 1066-1256/42-58ms; FOV = 28.8 cm; 96x96 image matrix; 8 contiguous 3-mm-thick axial slices; 1-3 motion-encoding gradient pairs each matched to the vibration frequency; x, y, and z motion-encoding directions; and 4 phase offsets spaced evenly over one vibration period followed by 1 no-motion reference scan. MRE post-processing was implemented by taking the curl of the 3D displacement field and performing a 3-dimensional direct inversion of the Helmholtz equation on the resulting wave field. Spatial derivatives were taken using a 3x3x3 "jack-shaped" kernel and the wave speed squared was the reported stiffness value for both DMA and MRE stiffness calculations. The left ventricle of the heart phantom was (semi-automatically) segmented and the octahedral shear strain signal-to-noise ratio (OSS-SNR)[8] was calculated on the curl wave fields. The left ventricle mask was eroded by one pixel in all directions to reduce edge effects. In the remaining volume, the mean and median stiffness values were calculated for pixels with OSS-SNR greater than 3. Immediately after scanning, mechanical testing using a commercially available DMA device (RheoSpectris™ C400, QC, Canada) was performed on the three cylindrical tube samples (9-mm inner diameter). The percent error between the MRE and DMA results (100*(DMA-MRE)/DMA) were calculated at each frequency.

**Results:** Figure 1 shows one component of the vector curl wave field and the corresponding elastograms obtained with cardiac MRE, at frequencies of 60-220Hz. As the vibration frequency increases the wavelength decreases and the wave pattern becomes more and more complex. The DMA testing showed that the heart phantom material had median wave speed squared values of 3.83-4.49 kPa in the 60-220Hz frequency range. The percent error between the DMA measurements and the MRE measurements are plotted in Figure 2. The MRE results generally underestimate the DMA values, and as the frequency of the vibrations increase the MRE measurements start to converge with the DMA results. Convergence occurs when the wavelength to wall thickness ratio approximately approaches 1.

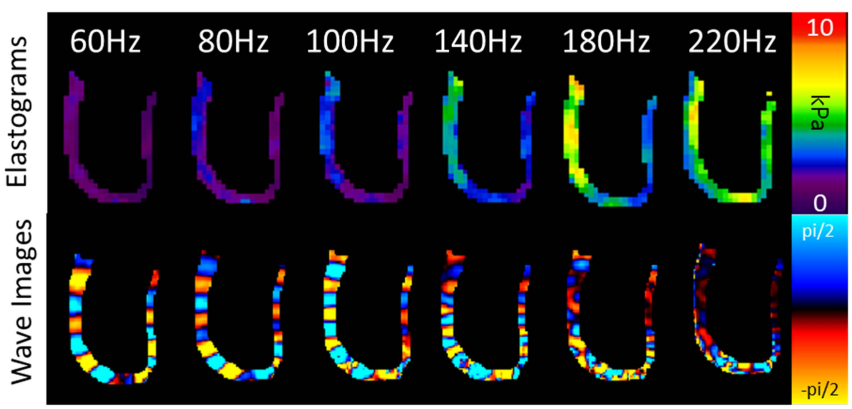


**Figure 1: Sample elastograms (top row) and curled wave images (bottom row) aquired from a cardiac phantom with MRE. Vibrations were applied to the apex of the heart and are propagating upwards.**

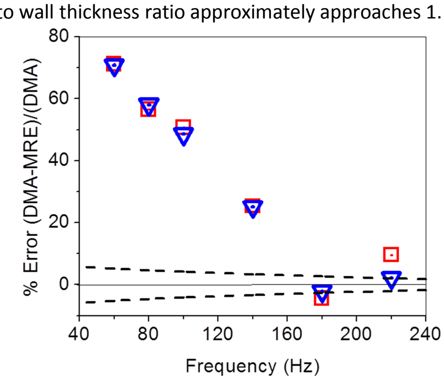


**Figure 2: Percent error comparisons between DMA analysis and the median (red square) and mean (blue triangle) values measured with MRE. The dotted lines are the standard deviation from the 3 sample measurements obtained with the DMA at each frequency.**

**Discussion:** This work demonstrates that MRE direct inversion with curl can produce accurate stiffness measurements within 5% error in an anatomically accurate diastolic cardiac phantom at sufficiently high vibration frequencies. One of the key challenges of in vivo cardiac MRE is getting adequate frequency and amplitude shear waves into the heart in order to obtain accurate stiffness measurements. Due to the high attenuation of shear waves in intermediate tissue between the driver and the heart the recommended 180Hz driving frequency may be challenging to obtain. It is important to recognise however, that lower frequencies may result in a systematic underestimation on the order of 25-70%. Song et al.[9], using ultrasound elastography, reported shear waves speeds in 7 healthy human volunteers to be 1.56±0.36 m/s in end diastole, corresponding to a stiffness of 2.43±0.56 kPa. This suggests that accurate heart stiffness values could be obtained at frequencies of approximately 130 Hz, and above, in diastole (assuming that the ratio of wavelength to wall thickness is the key factor). In conclusion, for an anatomically accurate diastolic heart geometry (3.83-4.49 kPa) phantom accurate stiffness measurements, within 5% error, can be obtained at frequencies greater than 180 Hz using a standard 3D direct inversion algorithm and 3mm isotropic imaging resolution.

**References:**

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