Calculation of Myocardial Stiffness Using a Rapid Multiphase MRE in a Heart Simulating Phantom

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Introduction: Numerous studies [1-3] have attempted to quantify the mechanical properties of the heart by calculating myocardial left ventricular (LV) chamber stiffness based on cardiac pressure-volume (P-V) measurements of LV filling [4]. Unfortunately, these P-V methods are technically challenging and invasive. Magnetic resonance elastography (MRE) [5] is a noninvasive technique that can measure the mechanical properties of an object undergoing cyclic deformation [6]. However, this technique involved long scan times, acquiring only one phase of the motion. The aim of this study was to develop and evaluate a rapid multiphase MRE technique, to compare shear stiffness to the previous single-phase technique, and to assess the stability and reliability of the stiffness estimates with changes in MRE imaging parameters and frequency of motion.

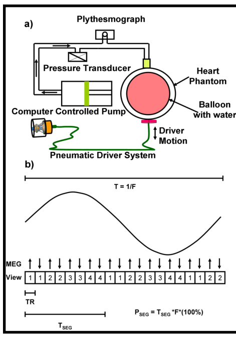

Figure 1: (a) Experimental Setup. (b) Timing diagram of the multiphase MRE (4 VPS).

Methods: *Setup and Acquisition:* All imaging was performed on a GE 1.5T Signa Excite scanner (GE Health Care, Waukesha, WI). A hollow spherical phantom was constructed with silicone rubber (Wirosil, BEGO, Germany) of 10-cm diameter and 1-cm thick to mimic the LV. Figure 1(a) shows the experimental setup and includes the phantom with a flexible inner bladder filled with water connected to a computer-controlled flow pump. The flow pump system produced programmable periodic pressure variations within the phantom cavity. A pressure transducer was used to provide real-time measurements of the line pressure, which were considered to be equal to the pressure within the phantom. An attached pulse plythesmograph produced a waveform which was used to gate the MRE pulse sequences. Synchronized, continuous mechanical stress waves at 200 Hz were generated in the phantom by an acoustic driver placed in contact with the bottom surface of the phantom. All data acquisitions were performed using conventional gradient echo sequences adapted to perform MRE. To reduce motion artifacts from the swirling water in the phantom, chemical presaturation was used to saturate the water signal (the chemical shift of water relative to Wirosil at 1.5 T is 290 Hz). *Single-Phase MRE:* A single-phase MRE sequence [6] was used to sample 8 phases of the phantom undergoing dynamic deformation (pressure range: 55-90 mmHg) at 18 bpm by varying the time delay between the trigger from the pulse plythesmograph and the start of the imaging sequence. Imaging parameters included: TE/TR = 10.9/3333 ms; since only one view, i.e. one line of k-space is acquired per period; α= 30°; slice thickness = 10 mm; acquisition matrix = 256x64; FOV = 14 cm; receiver bandwidth = ±16 kHz; mechanical excitation frequency = 200 Hz; 4 MRE time offsets; and motion encoding gradients (MEG) applied separately in the x and y directions to measure the in-plane motion. *Multiphase MRE:* Multiphase MRE experiments were performed while cyclically varying the pressure within the phantom (range: 56-93 mmHg) at several different frequencies (18, 27, 39, 51, 60, and 72 bpm). Imaging parameters were the same as in the single-phase acquisitions except for TE = 9.8 ms and TR = 35 ms. The multiphase MRE acquisition strategy is shown in Figure 1(b). The time taken to collect a segment of data is T_{SEG} = 2*TR*views per segment (VPS), where the factor of 2 accounts for the alternating MEG polarity used for the phase-contrast reconstruction. The fraction of the deformation period required to acquire a segment can be expressed as the percentage $P_{SEG} = T_{SEG}*(1/T)*(100\%)$, where T is the pulsation time (i.e. (1/T) = F, where F is the pulsation frequency). View sharing was performed to reconstruct 10-20 phases throughout the pressure cycle. The VPS were varied for each pulsation frequency. To determine the deviation of stiffness estimates with varying VPS, the mean square error (MSE) was calculated between stiffness estimates at the smallest P_{SEG} (4 VPS) to those of higher P_{SEG}'s at each frequency. *Calculation of shear stiffness:* Shear stiffness estimates for each acquisition were obtained using a thin spherical shell inversion [6].

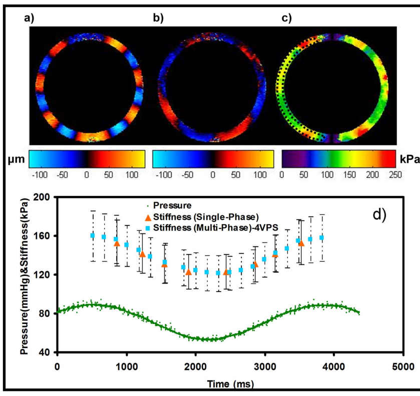

Figure 2: (a,b) Radial and circumferential displacements (c) stiffness map at a pressure of 67.5 mmHg at 60 bpm (P_{SEG} = 28%). The mean stiffness in the ROI shown with the dotted line is 142 ± 24 kPa. (d) Pressure and stiffness vs time for single-phase and multiphase MRE sequences at 4 VPS.

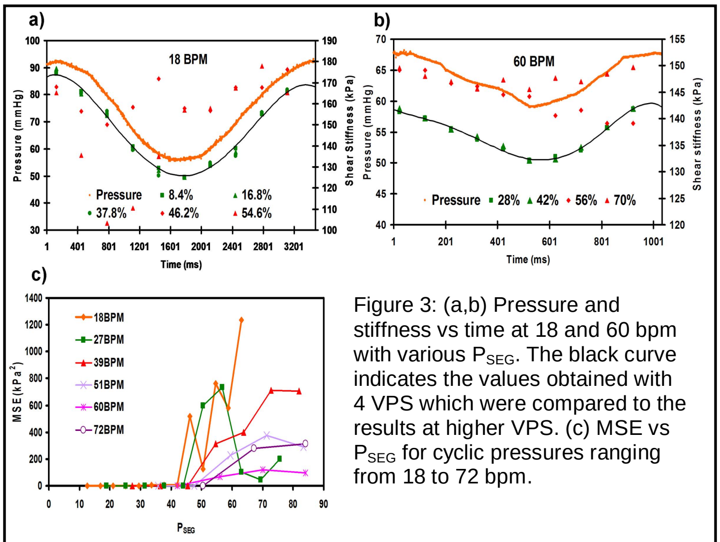

Figure 3: (a,b) Pressure and stiffness vs time at 18 and 60 bpm with various P_{SEG}. The black curve indicates the values obtained with 4 VPS which were compared to the results at higher VPS. (c) MSE vs P_{SEG} for cyclic pressures ranging from 18 to 72 bpm.

Results: Figures 2(a-c) show the radial and circumferential components of displacement for a single slice through the center of the phantom obtained at an inflation pressure of 67.5 mmHg at a frequency of 1 Hz (60 bpm) and 4 VPS (P_{SEG} = 28%) and its corresponding stiffness map. An average stiffness value of 142±24 kPa was obtained from the region of interest shown. Figure 2(d) shows the MRE-derived shear stiffness values obtained under cyclic pressure conditions (18 bpm) acquired with both the single-phase (one pressure point at a time) and multiphase (20 pressure points, 4 VPS (P_{SEG} = 8.4%)) GRE MRE sequences. For reference, the pressure waveform measured using the inline pressure monitor is also shown. The error bars represent ±1 SD of the shear stiffness measurements. Figures 3(a, b) show plots of shear stiffness and pressure versus time for cyclic pressures of 18 bpm and 60 bpm, respectively. The plots show the mean shear stiffness measurements obtained at different P_{SEG} for each time point. For clarity, figure 3(a) only includes stiffness estimates up to P_{SEG} = 54.6%. The plots show that the shear stiffness measurements at 18 and 60 bpm become less accurate when P_{SEG} is greater than about 42%. Figure 3(c) shows the MSE of the stiffness estimates at various P_{SEG} (i.e., different VPS) for frequencies ranging from 18 bpm to 72 bpm. The MSE starts to significantly increase when P_{SEG} is greater than about 42% for all frequencies tested.

Discussion: The above results indicate that stiffness estimates are reproducible using both single-phase and multiphase MRE sequences as long as P_{SEG} <42%. Acquisition times for this multiphase MRE acquisition are ~16 sec. per offset and sensitization direction at 60 bpm and 4 VPS, which can be performed in a breath-hold. Therefore, multiphase MRE may be practical for future in vivo cardiac tissue studies.

References: **1.** Zile, M.R,et al, Circulation, 2002.105(11):1387-93.
2. Badenhorst,D,et al,Cardiovas.Research,2003.57(3):632-41.
3. Pislaru, C.J,et al,Circulation, 2004.109(23):2905-10.
4. Mirsky,I,et al,Prog.inCardiovas.Diseases,Vol.32 No4,1990;291-318.
5. Bensamoun S.F,et al,JMRI, 2007;26(3):708-713.
6. Kolipaka,A,et al,In: Proc.16th Annual Meeting of ISMRM, 2008 (p.1031).