Application of SLIM-MRE on an In-Vivo Murine Model

Allen Q. Ye[1], Temel K. Yasar[2], Altaf Khan[2], Ziying Yin[1], Dieter Klatt[1], Thomas J. Royston[1], and Richard L. Magin[1]

[1]*Department of Bioengineering, University of Illinois at Chicago, Chicago, IL, United States,* [2]*Department of Mechanical Engineering, University of Illinois at Chicago, Chicago, IL, United States*

Target audience- Researchers working on accelerating magnetic resonance elastography (MRE).

Purpose - In MRE, external vibrations are introduced into tissues and then encoded into the MR phase signal using MRI sequences that include motion encoding gradients (MEG)[1]. From these phase images, tissue stiffness parameters can be extracted that are known to be correlated with pathophysiological changes in soft tissues and organs[2]. However, one of the primary concerns for using MRE during *in vivo* imaging is the length of the scan time. Longer scans lead to artifacts due to motion, higher costs, and in the case of animal models, increased morbidity and mortality due to increased time under anesthesia. *SampLing Interval Modulation* MRE (SLIM-MRE), addresses this problem by simultaneously measuring the motion in three directions, reducing scan time by a factor of three. In this *in vivo* study we validated SLIM-MRE using a 9.4 T animal MR imager to record standard and SLIM MRE data from tissues in a laboratory mouse.

Theory – Total phase accumulation ϕ is a summation of all three phase encoded displacements and each displacement encoded is denoted by an integral term in eq.1, where u_i is the projection of the displacement in ith direction, G_i is the MEG function in the corresponding direction, γ is the gyromagnetic ratio and s_i and τ are the start time and the duration of MEG in ith direction. For harmonic mechanical vibration and same frequency harmonic MEGs, the MR signal phase is represented by a harmonic function of the initial phases θ_i of the displacement projections, initial phases ϑ_i of the MEG projections and scaling factor ϕ_{0i}, eq. 2. In an MRE acquisition in order to obtain ϕ_{0i} and θ_i, multiple discrete MR signal phase acquisitions are performed, by applying the MEG projections with different initial phases with respect to mechanical wave motion. (eq. 3). In SLIM-MRE the sampling interval $\Delta\vartheta_i$ varies in each direction while it is constant for conventional MRE. This enables simultaneous encoding of all displacement projections in one acquisition block in SLIM-MRE [3,4].

$$\phi = \sum_{i=1}^{3} \phi_i = \gamma \sum_{i=1}^{3} \int_0^{\tau_i} G_i u_i \, \mathrm{d}t \qquad (1) \qquad \phi = \sum_{i=1}^{3} \phi_{0\,i} \cos(\vartheta_i - \theta_i) \qquad (2) \qquad \phi_j = \sum_{i=1}^{3} \phi_{0\,i} \cos(\vartheta_{ij} - \theta_i) = \sum_{i=1}^{3} \phi_{0\,i} \cos(j \cdot \Delta\vartheta_i - \theta_i) \qquad (3)$$

Methods – MRI measurements were performed using a 9.4 T Bruker BioSpec 33 cm horizontal bore scanner using a home-made custom 29 mm volume coil on 4 week old Charles River athymic mice. In a typical experiment, the mouse is placed in the MRI with plastic probe tips situated on the flank and with enough compression to ensure vibration transmission into the muscle (**Figure 1**).

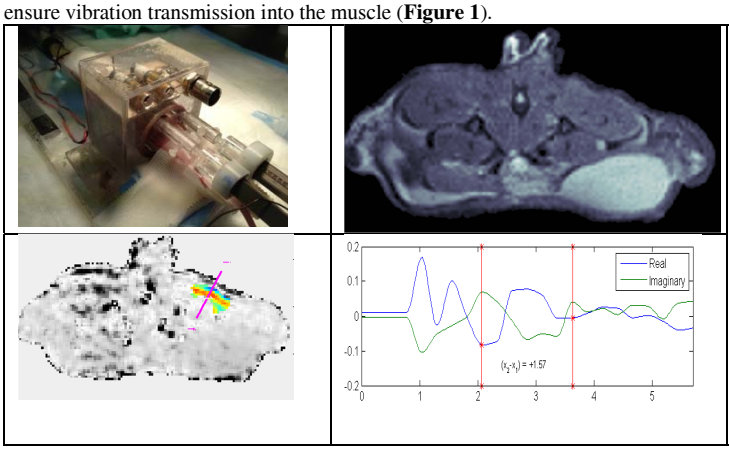

Figure 1 (top left): Experimental setup. There were two piezo ceramic stacks driving both flanks of mouse at the same time. A homemade mouse body RF coil has been used to fit both actuation tips and mouse without excessive pressure on the mouse. **Figure 2 (top right):** T2 weighted magnitude image of the selected slice. Depressions of the actuation tips can be seen on the right flank. **Figure 3 (bottom left):** Displacement map1500 Hz harmonic wave in the slice direction. Region of interest (ROI) is highlighted with color. Linear profile is taken across the purple line. **Figure 4 (bottom right):** Real and imaginary part of linear profile taken from analytical wave images shown in Figure 3. Red vertical lines indicate borders of ROI.

Actuation was achieved using a piezo ceramic stack at 1500 Hz. Low-resolution fast spin-echo (SE) T2 weighted images were acquired first using a rapid acquisition with refocused echoes (RARE) sequence for localization purposes. Next, standard 1-direction MRE was performed in the phase, read, and slice directions individually and SLIM-MRE was performed to encode motion in all three directions at once. Both regular and SLIM MRE sequences used the same imaging parameters of TR=2000 msec, TE = 23.454 msec, Rare factor = 4, resolution: 128x128, FOV: 3x3 cm, 7 slices of 1 mm thickness, power of MEG in each direction was 40% of maximum gradient amplifier power, and number of MEG cycles was 5 for all MEGs. Custom-written MATLAB programs were used to calculate shear stiffness (μ) values using a 2-D direct inversion method. For the sake of demonstration of SLIM-MRE we picked a region of interest on a muscle in the first slice, where there was visible wave motion in all three directions. **Table 1** presents the median value of real part of the estimated shear stiffness.

Results – SLIM-MRE was successfully applied to the mouse and as can be seen from **Table 1**, median shear stiffness estimation of both methods are similar, within the range of error of the MRE process with units provided in kPa.

Direction / Method	Units in kPa		
	Read	Phase	Slice
SLIM-MRE	4.35	4.05	3.47
Conventional MRE	4.06	4.22	3.64
Difference	0.29	0.17	0.17

Table 1: A comparison of median shear stiffness between SLIM-MRE and conventional MRE in all three directions.

Discussion and Conclusion - We demonstrated for the first time the application of SLIM MRE for an *in-vivo* murine MRE experiment. We have shown that the stiffness estimation obtained by SLIM-MRE and conventional MRE have the same value within a 10% error. Since SLIM-MRE is 3 times faster than conventional MRE, it is expected to have less image registration artifacts; however, this may contribute to different results between the two methods. There are also other factors that can cause discrepancies, such as concomitant gradient terms[5] (whenever gradients are on in 2 or more directions at the same time, they create additional magnetic fields that can introduce additional phase accumulation into MR images). Additional data needs to be acquired to analyze these artifacts. Finally, since all MEGs in SLIM require different time shifts, this may increase the echo time by 25% of one period for the 8 time steps MRE scan. However, this increase would not pose a problem for small animal models where one period is usually less than 1 msec.

Acknowledgments - This study was supported by the NIH Grant NIBIB-EB007537 (Magin) and TL1TR000049 (Ye).

References – [1] Muthupillai et al., Science 269, 1854-1857 (1995); [2] Glaser et al., JMRI 36, 757-774 (2012); [3] Klatt et al., ISMRM 21, 2444 (2013); [4] Klatt et al., Phys Med Biol, In Press (2013); [5] Bernstein MA, Zhou XJ, Polzin JA, King KF, Ganin A, Pelc NJ, Glover GH. Magn Reson Med. 1998;39(2):300-8