SLIM-MRE without prolonged echo time for the simultaneous acquisition of the 3D displacement vector applied to in vivo mouse brain

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Target Audience: Researchers working on acquisition protocols for the 3D displacement field in Magnetic Resonance Elastography (MRE).

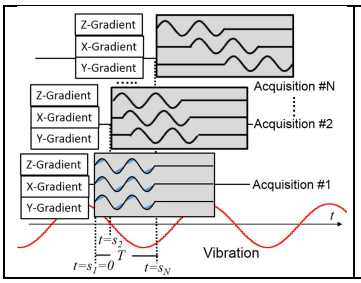
Introduction: MRE is a noninvasive imaging technique capable of determining tissue mechanical properties, which correlate with pathology [1, 2]. While there are MRE approaches acquiring and analyzing only one component of the displacement, it has become prevalent to consider all three components for data analysis. Conventionally, all three components of the displacement are acquired individually in consecutive scans. *SampLe Interval Modulation* (SLIM)-MRE [3, 4] represents a novel technique for the simultaneous acquisition of all components of the 3D displacement vector in only one temporally-resolved experiment. Therefore SLIM-MRE may facilitate clinical applications by accelerating the exam and reducing the potential for errors due to bulk motion of the patient in between the consecutive scans. Of note, the SLIM concept of motion encoding is not bound to a specific sequence type, but can be implemented into most sequence types used in MRE [3].

Problem: Current implementations of the SLIM encoding concept involve a direction-dependent mutual shifting of motion encoding gradients (MEG) relative to the applied vibration [3, 4]. Even when using symmetry and periodicity relations of MEGs and harmonic vibration the necessary increase of the echo time TE has to be 25% of the vibration period at a minimum. Therefore, the acceleration of scan time is achieved at a cost of SNR reduction due to increased T2 relaxation [4].

Objective: We present a novel implementation of the SLIM-MRE concept without prolonged TE. Modulation of the sampling interval is achieved by circular shifting of the phase of the MEG components at their onset, rather than by mutual shifting of their start times. The presented concept is embedded into a spin-echo multiple slice sequence with the Agilent operating software VnmrJ and applied to in vivo mouse brain.

Theory: The basic equation of MRE is described by the integral eq. 1, where ϕ is the accumulated MR signal phase, u_i and G_i represent the projections of the displacement and of the MEG, respectively, γ is the gyromagnetic ratio, and τ corresponds to the duration of each MEG projection. Assuming a harmonic vibration and bipolar MEGs, eq. 1 can be solved as a harmonic function of the initial phases θ_i of the displacement projections, of the initial phases ϑ_i of the MEG projections and of a scaling factor ϕ_{0i} (see eq. 2). Of note, in the presented formulation the origin of the time axis coincides with the onset of all MEG projections. The amplitude of the vibration projection can be calculated from the scaling factor ϕ_{0i}, which is a function of the cycle number and of the period of the MEG. In an MRE acquisition the quantities ϕ_{0i} and θ_i are determined, which are needed for the reconstruction of mechanical parameters. For this purpose, multiple discrete MR signal phase acquisitions ϕ_j are performed. In the presented implementation of SLIM-MRE, we apply the MEG projections with different initial phases ϑ_{ij} at their onsets (eq. 3). The shift of the initial phase $\Delta\vartheta_i$ varies in each direction. This allows for simultaneous encoding of all displacement projections in one MRE acquisition block by using one frequency bin per encoding direction (see eq. 3).

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



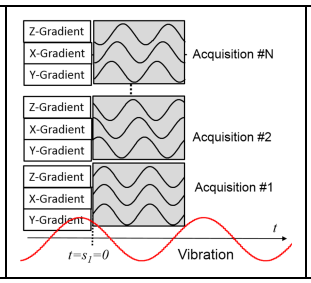
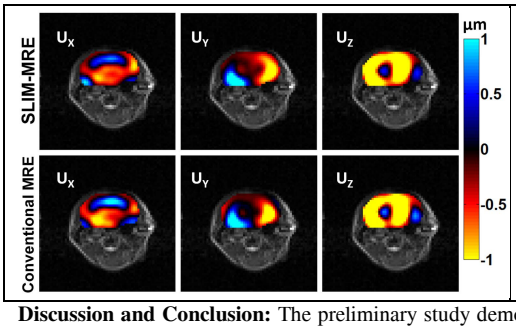


Figure 1 (left): Original implementation method of SLIM-MRE using varied delayed start time and mutual gradient shifting. All MEGs have the same phase at their onset. **Figure 2 (right):** Current implementation using varied MEG initial phase without mutual shifting. No increase in the TE time is needed.

Methods: Cerebral MRE was performed on 2 healthy C57BL-6 8 month old female mice. All animal experiments were conducted under the approval of UIC's Office of Animal Care and Institutional Biosafety. Both the SLIM-MRE and conventional MRE sequences were run using a modified spin echo sequence. Scans were performed on an Agilent 9.4 T magnet using the following imaging parameters; 1000 Hz mechanical frequency; 25 G/cm MEG amplitude; 10 MEG cycles; 1000 ms repetition time TR; 16.1 ms echo time TE; FOV 24 mm x 24 mm; 20 slices (16 slices mouse 1); 0.375 mm^3 isotropic voxel size. The sample number was N=8 and the sample interval of the MEG start phase was direction dependent with $\Delta\vartheta_i = i2\pi/N$, i = 1, 2, 3. Complex wave images U were taken from the Fourier-transformed samples. The complex shear modulus was determined from the 3D low-pass filtered U by applying the curl operator and subsequent algebraic inversion of the Navier Equation [5]. Median values were calculated using a manually created ROI within the central 12 slices (8 slices for mouse 1).

Results:



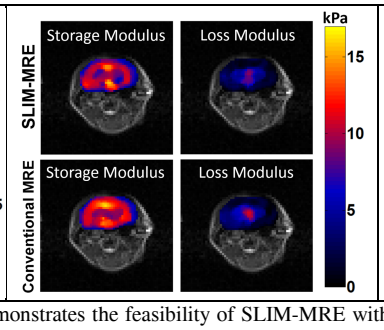


Figure 3 (left): The real part of the complex wave image for all spatial directions comparing SLIM-MRE (top) and conventional MRE (bottom). Image is of a central slice of mouse 2. **Figure 4 (right):** Complex shear modulus maps of the same slice and mouse as in Figure 3. The ROI-averaged G-values determined with SLIM-MRE (conventional MRE) were for mice 1 and 2, 10.2+i2.86 kPa (10.1+i2.72 kPa) and 11.0+i3.58 kPa (10.3+i3.70 kPa), respectively.

Discussion and Conclusion: The preliminary study demonstrates the feasibility of SLIM-MRE without increased TE by using MEGs with varying initial phases and without mutual shift. Only subtle differences in amplitude and phase of the wave images acquired with both methods are visible. This is likely due to varying gradient moment nulling characteristics in the presented implementation of SLIM-MRE (see Figure 2). Still, inversion of the wave field data yields similar images of the complex modulus for SLIM-MRE and conventional MRE and the median values vary by less than 8%. In conclusion SLIM-MRE with varying start phase MEG wave forms enables the rapid acquisition of the 3D displacement vector in one third of the time of conventional motion encoding concepts without the need for prolonged echo times and therefore the presented implementation of SLIM can be considered as an alternative strategy for simultaneous multidirectional motion encoding in situations when T2 is critically short. **References:** [1] Muthupillai et al., Science 269, 1854-1857 (1995); [2] Glaser et al., JMRI 36, 757-774 (2012); [3] Klatt et al., Phys Med Biol 58, 8663-8675 (2013); [4] Klatt et al., JMRI (in print); [5] Manduca et al., Med Imag Anal 5, 237-254 (2001).