Introduction of sample interval modulation for the simultaneous acquisition of 3D displacement data in MR Elastography

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Introduction: In Magnetic Resonance Elastography (MRE) [1] external vibrations are introduced into the tissue under examination. The tissue vibrations are encoded in the MR signal phase ϕ using standard MRI sequences upgraded with motion encoding gradients (MEG). Hence, tissue mechanical parameters can be calculated from the acquired wave fields. The analysis of MRE data with one motion-encoding direction has revealed the correlation of pathophysiological changes and the mechanical behavior of diverse organs [2]. However, more and more the 3D displacement field is acquired to separate the shear from the compression wave by using the curl-operator [3] and to assess tissue compressibility [4]. **Problem:** In MRE, measurement time is critical. Besides cost factors, long acquisition times potentially decrease the measurement accuracy, since motion may occur and cause misalignment of the images. Further, in conventional 3D MRE, the components of the tissue displacement are acquired in three individual temporally-resolved MRE experiments. Therefore, the components, although attributed to the same point in time, were actually acquired in different physiological states. **Objective:** We developed a concept for the MEG arrangement that is capable of encoding three spatial components of a monofrequency tissue vibration simultaneously. We name our approach *Sampling Interval Modulation* (SLIM)-MRE, since the individual displacement components are observed using different time discretization intervals. In doing so, the components are modulated with different frequencies in the MR signal phase ϕ expressed as a harmonic function of the start time of the MEG. Thus, all displacement components are acquired faster than in conventional MRE and can be derived from the same temporally-resolved MR phase images. We present for the first time, to our knowledge, 3D displacement data that were acquired simultaneously and stored in the same *k*-space. **Theory:** Below, the index j=1, 2, 3 corresponds to the *x*-, *y*- and *z*-direction in the Cartesian system, and to the read-, phase- and slice-direction in the scanner system, respectively. The basic equation of MRE is represented by eq. 1 [1], which describes the encoding of the displacement u_j of an isochromat in the MR signal phase ϕ by applying a magnetic field gradient G_j. Herein, the gyromagnetic ratio of the proton, the duration and the start time of the MEG-component in the j-direction are denoted by γ, T and s_j, respectively. Of special note, in SLIM-MRE, T is kept constant for all MEG-components, while s_j may vary, and we assume sinusoidal functions of the same frequency f for the vibration and for the three MEG-components. Eq. 2 represents the solution of eq. 1 and comprises the initial mechanical phase θ_j at t=0, the amplitude u_j^0 and the encoding efficiency ξ_j of the displacement component u_j. Temporal resolution is expressed by a variation of the MEG-start time. In SLIM-MRE, the sampling interval of the MEG-start time is different for the three components. We set for the sampling interval $\Delta t_j = j/(f \cdot N)$, j = 1, 2, 3, with N being the number of samples and $s_{jn} = n\Delta t_j$, n=0, 1,…, N-1. It is directly perceptible from eq. 3, which represents the discretized expression of eq. 2, that the three displacement components are encoded with different "apparent frequencies", specifically with the 1st, 2nd and 3rd harmonic. The individual components can thus be decomposed by applying a discrete Fourier transform to ϕ_n. An example for the arrangement of the MEGs in SLIM-MRE is shown in fig. 1.

$$\phi = \sum_{j=1}^{3} \phi_j = \gamma \sum_{j=1}^{3} \int_{s_j}^{s_j+T} G_j(t) \cdot u_j(t)\, dt \quad (1) \qquad \phi = \sum_{j=1}^{3} \phi_j(s_j) = \sum_{j=1}^{3} \xi_j u_j^0 \sin\left(2\pi f s_j + \theta_j + \frac{\pi}{2}\right) \quad (2) \qquad \phi_n = \sum_{j=1}^{3} \phi_j(s_{jn}) = \sum_{j=1}^{3} \xi_j u_j^0 \sin\left(2\pi j \frac{n}{N} + \theta_j + \frac{\pi}{2}\right) \quad (3)$$

Methods: The experimental setup for a similar MRE experiment in the same 11.7 T Bruker vertical MRI system has been described before [5]. We prepared an inhomogeneous phantom consisting of agarose beads (0.7% by water) embedded in an agarose matrix (1.1 % by water). The sample bin (⌀=9 mm) was driven by a piezostack using a monofrequency actuation of 5 kHz. One transversal slice was acquired employing a gradient echo-sequence upgraded with 5 cycles of 5 kHz-MEGs in *x*- *y* and *z*- direction. For motion encoding, we used the SLIM concept introduced in the Theory section and illustrated in fig. 1; however, the available sequence programming tools forced us to apply the MEG-components successively. Still, SLIM principles were obeyed, specifically choosing different sampling intervals for the start time of the MEG-components, multidirectional encoding within one TR and saving of multidirectional images in one temporally resolved *k*-space. Additional sequence parameters used were: TR/TE = 200/6.94 ms; slice thickness = 1 mm; FOV=10x10mm^2; matrix size=128^2; flip angle = 30°. At each time step, two acquisitions were conducted with inverse MEGs to clear biases due to constant field inhomogeneity. Duration of SLIM-MRE was ~6.8 min. For comparison, conventional MRE experiments were performed in which the 3D displacement was acquired in three individual experiments for each direction resulting in a total measurement time of ~20.5 min. In conventional and SLIM-MRE, 2D local frequency estimation (LFE) was applied to the images and the wave length images were averaged over the ROI, which corresponded to the largest bead within the image slice. **Results:** SLIM-MRE was successfully applied to the phantom (fig. 2). The wave amplitude varies for the different encoding directions. Amplitudes are the strongest for motion encoding in *z*-direction, as the actuator used provided motion perpendicular to the image slice. Still, within the bead *x* and *y*-displacements are clearly visible, which were caused by transmission and reflection of wave energy at the spherical bead boundaries. Wave images corresponding to the same encoding direction are very similar. Thus, independent of standard MRE and SLIM-MRE, the average LFE-derived wave length λ over the ROI was (0.6±0.1) mm, (0.6±0.1) mm and (0.5±0.1) mm for encoding in the *x*-, *y*- and *z*-direction, respectively. **Discussion and Conclusion:** We introduce the concept of SLIM to MRE for acquiring all three motion components of a monofrequency vibration simultaneously and storing them in the same *k*-space. We have shown that the use of eight samples for temporal resolution is sufficient for being able to decompose the three displacement components, which were encoded with different apparent frequencies in the MR phase. Therefore, SLIM-MRE is 1.5-3 times faster than conventional MRE, where four to eight samples are acquired for each component. As a drawback of SLIM-MRE, the application of multiple MEGs with different start time increases the minimum echo time compared to conventional MRE. This increase, however, can be minimized to $1/f$ by taking advantage of the periodicity of the harmonic functions. As an example, the 8th sample of the MEG *z*-component corresponds to 525 µs, which is equivalent to 125 µs due to the periodic nature of the 5 kHz-vibration. Finally, as a concept for the MEG-arrangement, SLIM-MRE is applicable to all standard sequences commonly used in MRE.

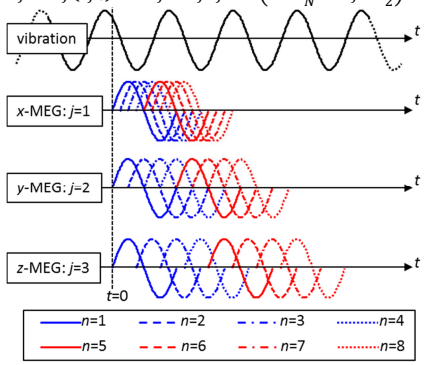

Fig. 1: MEG arrangement relative to the mechanical actuation signal in SLIM-MRE. For better visualization, only one MEG-cycle is illustrated per component and time step n. The start of all MEGs in the first time step (n=1) coincides with t=0. Using f=5kHz, the sampling interval with respect to the start time of the *x*-, *y*- and *z*-MEG corresponds to 25 µs, 50 µs and 75 µs, respectively.

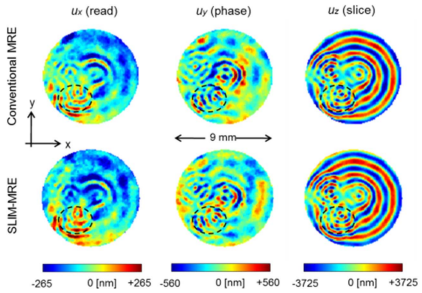

Fig. 2: Complex wave images in the axial plane at 5 kHz acquired using conventional MRE (top) and using SLIM-MRE (bottom). One displacement component is shown per column as indicated in the heading. The ROI is demarcated with dashed lines.

References: [1] Muthupillai et al., Science 269, 1854-1857 (1995); [2] Glaser et al., JMRI 36, 757-774 (2012); [3] Sinkus et al., Magn Reson Imaging 23, 159-165 (2005); [4] Hirsch et al., MRM, doi: 10.1002/mrm.24499 (online); [5] Yasar et al., MRM, doi: 10.1002/mrm.24495 (online). [6] Knutsson et al., Proc IEEE ICIP-94, 36-40 (1994). **Acknowledgement**: NIH support EB007537, EB012142.