Desynchronized motion encoding in rapid steady-state free precession MR elastography

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Introduction: The mechanical resistance of soft tissue to shear deformations can be non-invasively measured by MR elastography (MRE) [1]. Regardless the success of MRE in volunteer examinations and in clinical pilot studies, it has remained a relatively slow technique compared to rapid acquisition schemes of other flow- or motion-quantification MRI methods [2]. Here, we propose the use of low-frequency shear vibrations matched to the repetition time of the MRI sequence so that the motion encoding frequency is desynchronized with the vibration frequency. Therefore, a lower motion-encoding efficiency than in conventional MRE is achieved. However, as experiments in this study show, this trade-off is compensated by i) the gain of deflection amplitude due to decreased vibration frequencies and ii) the increase of signal amplitude due to shorter echo times. The new technique is designed for rapid elastographic studies on tissues which exhibit short T_2-relaxation times making SSFP-MRE [3] useful for in-vivo examinations of skeletal muscle, myocardium or liver.

Theory: The vibration is described by a sinusoidal oscillation of an initial displacement field $\mathbf{u}_0$ with frequency f_v. The interaction of a harmonic deflection with magnetic field gradients results in a spin phase φ given by:

$$\varphi = \varphi_0 + \mathbf{u}_0 \underbrace{\gamma \int_\tau (\mathbf{G}_{SS} + \mathbf{G}_{ME} + \mathbf{G}_{PE} + \mathbf{G}_{RO}) \sin(2\pi f_v [t + \Delta t]) dt}_{\boldsymbol{\Phi}} \quad (1)$$

with γ the gyromagnetic ratio, the off-resonance phase φ_0 caused by field inhomogeneities, the gradients for slice selection ($\mathbf{G}_{SS}$), phase encoding ($\mathbf{G}_{PE}$), read out ($\mathbf{G}_{RO}$) and motion encoding ($\mathbf{G}_{ME}$) and Δt as the position of the trigger to the wave generator relative to the gradient timing. The term $\boldsymbol{\Phi}$ in eq.1 denotes a phase gradient that can be used for characterizing the sensitivity of MRE sequences for harmonic motion encoding:

$$HENC = \frac{\pi}{|\boldsymbol{\Phi}|} \quad (2)$$

equivalently to the *VENC*-quantity in flow MRI [2], i.e. *HENC* is defined by the harmonic wave amplitude that is required to produce $\pm\pi$ phase shift.

Methods: For theoretically analyzing the motion-sensitivity of SSFP-MRE sequences (fig. 1) *HENC* was calculated using gradient waveforms corresponding to our experiments (fig. 2). Experiments were performed on a 1.5 T scanner (Siemens Magnetom Sonata, Erlangen, Germany). For MRE-image acquisition, a standard balanced SSFP sequence was sensitized to motion using a trapezoidal, bipolar gradient ($\mathbf{G}_{ME}$) between phase-encoding gradient and read-out gradient with variable direction, frequency (f_{ME}) and period number (n_{ME}). For in vivo experiments on the human liver a costume-built actuator based on a loudspeaker was used. 20 phase-difference wave images were acquired by a trigger increment of *TR*/20. Further acquisition parameters for in vivo liver examinations were: α = 50°, *TR*/*TE* = 20/18.3, $\mathbf{G}_{ME}$ parallel to $\mathbf{G}_{SS}$, f_{ME} = 119 Hz, n_{ME} = 2, f_v = 50 Hz. Total scan time was approximately 2 minutes.

Results and Discussion: The sensitivity of MRE sequences to harmonic motion is well quantified using the new *HENC* variable. Figure 2 shows that SSFP-MRE provides motion sensitivity comparable to the classical way of MRE motion encoding, i.e. f_{ME} is matched to f_v. The desynchronized motion encoding in SSFP-MRE allows decreasing f_v so that larger wave amplitudes are gained by the same length of $\mathbf{G}_{ME}$. It is additionally visible in figure .2 that SSFP-MRE is sensitive to higher harmonic frequencies: A single cycle of the motion encoding gradient provides the highest spectral sensitivity near 1/TR (fig. 2a) whereas two gradient cycles relatively elevate the sensitivity of SSFP-MRE to higher harmonic vibrations (fig. 2b). The in vivo experiment on human liver was adjusted to increase the sensitivity to the 2nd harmonic vibration. Therefore, it was possible to measure the wave propagation speed of both fundamental and higher harmonic oscillations simultaneously with 1.35 ± 0.16 m/s and 1.48 ± 0.16 m/s, respectively. From the dispersion of both wave speeds a shear modulus of 1.9 ± 0.4 kPa and a viscosity of 2.0 ± 1.0 Pa s were determined. It is a promising result that SSFP-MRE is applicable for in vivo liver studies even with using the very long *TR*'s of 20ms. If T_2*-dephasing is stronger than in our experiments we propose to apply n_{ME} = 1 yielding a *TR* = 11.7 ms for the current experimental setup. Then, the vibration could be adjusted corresponding to the synchronization schemes of fig.1a, 1b or 1c with f_v = 85.4, 42.7 or 21.7 Hz, respectively.

Conclusion: The proposed SSFP-MRE technique considerably accelerates in vivo MR elastography in the low-frequency audio regime. Furthermore, SSFP-MRE is sensitive to anharmonic vibrations. The superior SNR gained by balanced SSFP combined with large deflection amplitudes of low-frequency shear vibrations as used in SSFP-MRE potentially extends the field of in vivo MR elastography applications.

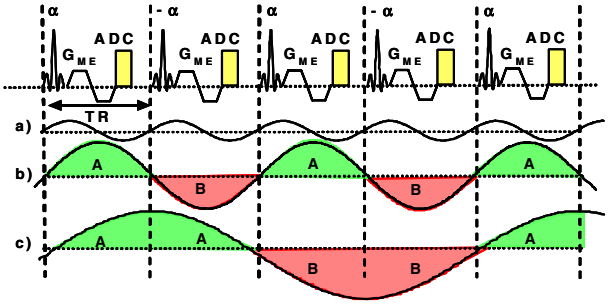

Fig. 1 Sketch of the balanced SSFP-MRE sequence used for in vivo MR elastography. **a)** Vibration synchronized to 1/*TR*. For measuring phase-difference images $\mathbf{G}_{ME}$ is toggled yielding two images A and B by two consecutive experiments. **b)** In contrast to a) two images with reverse motion-phase contrast are simultaneously measured by synchronizing f_v to 1/(2*TR*) similarly to BASEL [4]. **c)** Further extension of the vibration period with maintaining the same short *TR* (f_v = 1/(4*TR*)).

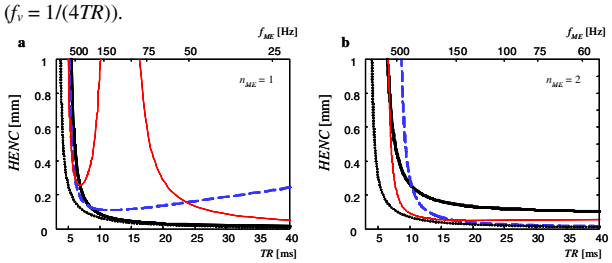

Fig. 2: Harmonic motion encoding efficiency *HENC* for SSFP-MRE assuming a minimum SSFP-*TR* (without $\mathbf{G}_{ME}$) of 4 ms. For simplicity, solely $\mathbf{G}_{ME}$ was considered to be motion sensitive. For comparison the dotted graph shows the motion sensitivity of a conventional MRE experiment (f_v = f_{ME}). The black, blue and red graphs display HENC for vibrations synchronized to 1/*TR*, 2/*TR* and 3/*TR*, respectively. **a:** one motion encoding gradient cycle **b:** two cycles of $\mathbf{G}_{ME}$.

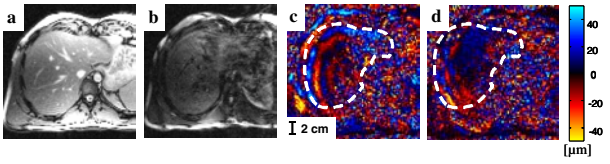

Fig. 3: Anatomic SSFP image without a motion encoding gradient **a)** and inserted $\mathbf{G}_{ME}$ with f_{ME} = 119 Hz, n_{ME} = 2 **b).** Despite the loss of half of the initial signal-to-noise-ratio of a), the corresponding phase images were analyzable for the 1st **(c)** and 2nd **(d)** harmonic vibration.

References:

[1] Muthupillai, R. et al., Science 1995; 269: 1854; [2] Bernstein, M. et al., MRI Pulse Sequences 2004, Elsevier 2004 [3] Rump, J. et al., ISMRM 2005: 2384 [4] Bieri, O. et al., ISMRM 2005: 97