Fast planar steady-state free precession MR elastography on human liver

D. Klatt[1], P. Asbach[1], J. Rump[1], S. Papazoglou[1], U. Hamhaber[2], J. Braun[2], I. Sack[1]

[1]Charite, CCM, Institute of Radiology, Berlin, Germany, [2]Charite, CBF, Institute of Medical Informatics, Berlin, Germany

Introduction: Recent elastographic studies on human liver indicated a relation between degree of liver fibrosis and liver stiffness [1-4]. This opens the perspective for non-invasively characterizing liver fibrosis by elastography in addition to standard MRI examinations.

Problem: Classical MR elastography (MRE) experiments applied to human liver suffer from extended time consumption due to i) the need for slow vibration frequencies, which prolongs the repetition time (*TR*) of the motion sensitive MR sequence and ii) the acquisition of time resolved 3D-vector fields of the harmonic displacement required for data evaluation.

Objective: We propose the application of fast balanced steady-state free precision (SSFP)-MRE for acquiring 2D phase-difference wave images within 3 seconds. The combination of SSFP-MRE with alternating slice subtraction [5] allows the use of low mechanical excitation frequencies below 50 Hz. Moreover, transversal wave excitation allows reducing the dimensionality of the wave inversion problem towards the analysis of a single transverse wave image. Using the new planar SSFP-MRE method 8 healthy volunteers, 1 volunteer with cystic diseased liver and 2 fibrosis patients were investigated within an individual study time of 2 minutes.

Methods: Measurements were performed on a 1.5 T scanner (Siemens-Sonata, Germany). Shear waves were introduced into the liver using a carbon fiber rod connected to a homebuilt acoustic actuator. The transducer was attached beneath the right costal arch and vibrated in head-feet direction by an excitation frequency of 51Hz. The distance between transducer and transverse image plane was about 3-4cm (fig.1a). For image acquisition a balanced SSFP-MRE experiment was used incorporating 150-Hz-oscillating gradients along slice-selection for motion sensitization so that the sequence *TR* was prolonged from approximately 3 to 10 ms. An alternating image reconstruction scheme allowed us to synchronize one vibration cycle with the length of two *TR*'s resulting in one image corresponding to the first half-cycle of the 51-Hz-vibrations and one image with inverse phase-contrast corresponding to the second half-cycle of the vibrations. Both phase contrast images were subtracted from each other for eliminating static phase shifts. Using a matrix size of 128 and 300mm FoV allowed to acquire one phase-difference wave image within 2.6 s. The experiment was repeated 20 times with increasing trigger delay between motion encoding gradient and wave generator for achieving a temporal resolution. Between consecutive experiments a delay of 6 seconds was maintained allowing the examined subject to breath resulting in a total measurement time of 120 s. Reduction of the image resolution to 64 pixel-matrix and temporal resolution to 10 images allowed us to acquire a plane-wave propagation during one single breath hold. Such setup was used to test the 3D-wave form as demonstrated in figs.1b and c and to determine the optimal image slice position for patient studies. Slice thickness was 10 mm in all examinations. The shear modulus was determined from wave data applying an automatic phase gradient technique along profiles running from the edge to the centre of the liver. The wave speeds along all profiles were averaged and converted to one mean shear elasticity using Voigt's model [6] and assuming a constant viscosity of $\eta = 4$Pas, which was previously measured on 2 volunteers.

Results: The multislice experiment (fig.1) shows that the shear waves are spherically emanating from the convex surface of the liver towards its center. The isosurfaces projection of the wave fronts further indicates that the shape of the waves is determined by the shape of the organ and thus, it is possible to analyze a central image slice in order to obtain correct wave velocities. At the slice position demarcated in fig.1c (corresponding to #4 in fig.1b) the curvature of the liver is small so that the effect of a possible overestimation of the wavelengths due to geometrical projection is smaller than the error margins. Wave components along the read-out and phase-encoding directions were negligible small. Fig.2 shows typical wave images acquired at position #4 (fig.1) of the healthy volunteer A and the patient Y with fibrosis (stadium 3). It is visible that in the liver of patient Y the wave lengths are significantly longer than in healthy tissue. However, in patient Z suffering from fibrosis (stadium 1) with severe steatosis no such increase of the propagation speed was found. The elasticity of patient X indicates that liver cysts do not affect stiffness. The resulting shear moduli are summarized in fig.3.

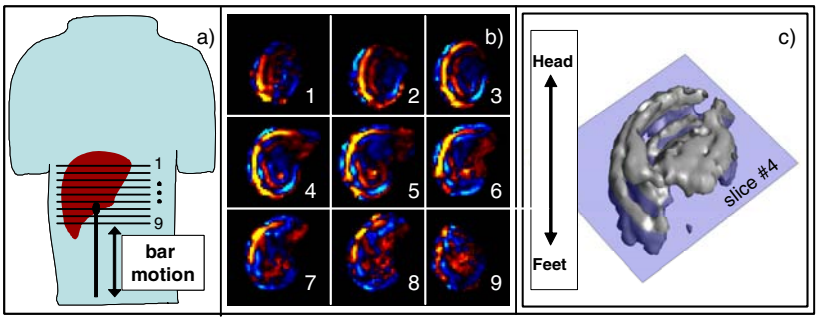

Fig. 1: Multislice SSFP-MRE measurements on *in vivo* human liver. **a:** Transverse image slice and transducer position relative to the examined liver. **b:** Nine transversal slices acquired at positions shown in fig.1a. **c:** 3D Surface plot of constant wave phase in the liver. The demarcated plane was used as image slice in all subsequent volunteer and patient studies.

Discussion and Conclusion: Planar SSFP-MRE considerably accelerates elastography studies on liver by a factor of 10 from 20 min [2,3] to 2 min. A further reduction of the temporal resolution to 8 wave images might be possible without loosing information. Then planar SSFP-MRE on the liver could be accomplished within a single breath hold. A mean shear stiffness of $\bar{\mu} = 2{,}5 +/- 0{,}8$ kPa was found among 8 healthy male volunteers. Our findings on two patients suggest that the shear stiffness can provide an important means to characterize liver diseases in addition to biopsy and standard MRE methods.

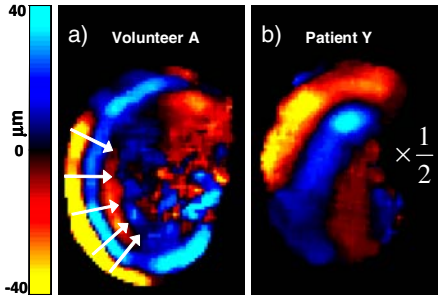

Fig.2: Fast planar SSFP-MRE measurements on *in vivo* human liver. **a:** Shear waves in the liver of volunteer A. The arrows symbolize profiles for automatically determining wave speeds. **b:** Shear waves in the diseased liver of patient Y. Note that the color bar values have to be divided by 2 for real deflections.

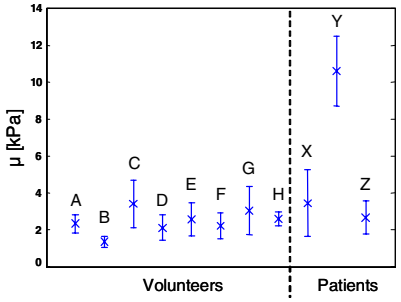

Fig. 3: Shear modulus of 8 healthy volunteers (A-H), 1 volunteer with cystic diseased liver (X) and 2 fibrosis patients (Y,Z).

References [1] Sandrin et. al., Ultrasound in Med. & Bio. 29(12), 1705-1713 (2003)**;** [2] Sinkus et al., ISMRM 2005, p. 624**,** [3] Peeters et al., ISMRM 2005, p. 339; [4] Rouviere et al., ISMRM 2005, p. 340; [5] Rump et al., ISMRM 2005, p. 2384, [6] Catheline et al., J. Acoust. Soc. Am. 105 (5), 2941-2950 (1999).