SSFSE sequence for fast elastography in the presence of susceptibility

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Introduction: Magnetic resonance elastography (MRE) has gained acceptance in recent years as an imaging technique to map tissue stiffness in the liver for the evaluation of fibrosis and cirrhosis [1-3]. One technique studied extensively for such an application uses a modified phase contrast gradient echo (GRE) sequence acquired at multiple phase offsets relative to a continuously applied acoustic wave [1]. The main disadvantage of this method is that each readout must be preceded by motion encoding gradient (MEG) pulses that are about 16.7 msec long for a 60 Hz wave, and both the overall acquisition times and echo times tend to be very long for a GRE sequence. Since subjects may have tissue with high iron concentrations or excessive gas in the stomach, insufficient signal may be collected due to the increased susceptibility. Fast spin echo sequences are much less prone to susceptibility-induced signal losses and can acquire many views with a single set of MEG pulses. However, phase modulations induced by these MEG pulses can accelerate signal decay and cause signal modulations along the echo train. To properly handle such phase modulations, we propose a rapid single shot fast spin echo (SSFSE) sequence for MR elastography that incorporates a technique originally designed for diffusion weighted imaging. The aim of this study is investigate the feasibility of a SSFSE-based MRE sequence for measuring stiffness in the presence of susceptibility.

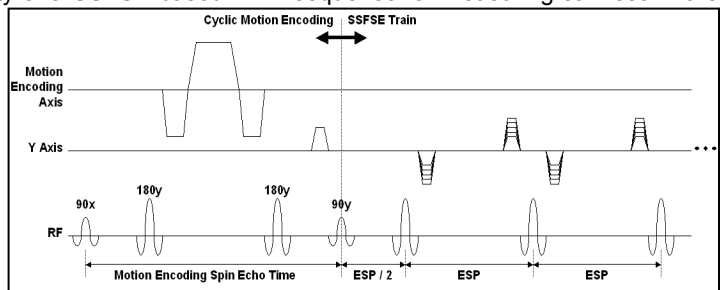

Figure 1. SSFSE based sequence for elastography.

Theory: The SSFSE MRE sequence (figure 1) consists of motion encoding sequence followed by a phase preserving SSFSE sequence as proposed by Alsop [4]. All refocused echoes in the echo train have the same phase information as that at the reference time, defined as the time half an echo spacing period (ESP) before the first refocusing pulse. Fast spin echo CPMG conditions are imposed at the reference time by using a dephasing gradient to spread the signal of each voxel in the transverse plane, followed by a 90 degree pulse centered at the reference time with the same phase as the refocusing pulses to tip all off-axis components into the longitudinal plane. A rephasing gradient before each refocused echo causes spins to realign to the phase they exhibited at the reference time, and a dephasing gradient likewise returns spins to their CPMG-compliant state. Shifting the sequence relative to the applied acoustic waves allows acquisition of different phases of the propagating shear wave. By reversing the polarity of the motion encoding gradients, phase difference techniques can be employed to minimize phase errors caused by motion during other parts of the sequence.

Methods: The SSFSE elastography sequence was applied to yucatan minipigs in a carcinogenesis study under IACUC approved protocol. All imaging was performed on a 1.5T scanner (Signa HDxt, General Electric Healthcare, Waukesha, WI) with an 8 channel torso array coil. Imaging parameters were: TR = 800 ms, TE = 56 ms, bandwidth = +/- 31.25 kHz, matrix = 128x64, FOV = 30.0 x 27.0 cm, slice thickness = 8 mm, slice gap = 4 mm, total acquisition time = 22.4 sec. Phase encoding order was centric. 3 slices were acquired with 4 wave phases. A constant sine wave with a frequency of 60 Hz was applied continuously during the entire sequence with a prototype pneumatic driver. MEG pulses compensated for linear motion were applied in the slice select direction. The GRE based elastography sequence was also applied to the same slices with the same parameters except TR = 150 ms, TE = 23.6, total acquisition time = 1:29 over three breathholds. Shear stiffness was calculated from the wave images using a multi-scale direct inversion algorithm [5]. Both techniques were also applied to a stiffness phantom consisting of variable density gel layers.

Results: Images are shown in figure 2. GRE images showed noticeable signal loss in areas near the stomach and edges of the liver. The few remaining areas with consistent waves generally matched corresponding areas on the SSFSE images. The central portions of the liver suffered from wave interference in most of the subjects when acquired with either technique. SSFSE-based elastograms exhibited broader wave coverage and structures such as the gall bladder were easily visualized. Phantom elastograms were consistent between the two techniques. No susceptibility was observed in the phantom images.

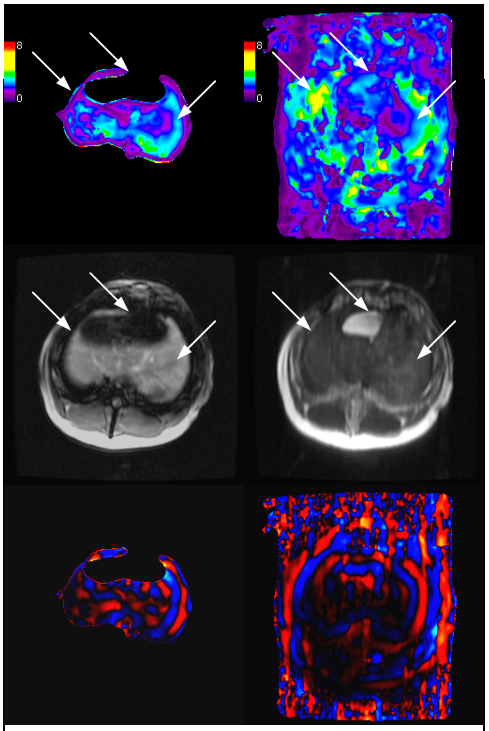

Figure 2. Stiffness maps (top row), magnitude images (middle row) and wave images (bottom row) for the GRE (left) and SSFSE (right) based sequences. Corresponding areas between the images are noted with arrows. Masks were created by a region growing algorithm.

Discussion: We have demonstrated the feasibility and potential benefits of an SSFSE based elastography sequence. This is particularly useful with smaller organs when much of the acquired cross sectional area is lost to susceptibility. While SSFSE images exhibited some residual ghosting and blurring due to motion, their effect on the waves in the phase difference images was relatively small. Better in-vivo data using the GRE sequence is required to validate consistent stiffness measurements between the two techniques.

References: 1. Yin M et al, Clincal Gastroenterol Hepatol 2007; 5:1207-1213. 2. Rouviere et al, Radiology 2006; 240:440-8, 3. Venkatesh SK et al, Am J Roentgenol 2008; 190:1534-40.3. 4. Alsop DC, MRM 1997; 38:527-533. 5. Manduca A et al, Med Image Anal 2001; 5:237-54.