Mulit-echo averaging in MR elastography for improved SNR

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Introduction MR elastography (MRE) allows the stiffness of internal human organ to be non-invasively assessed, and has been clinically used in the diagnosis of fibrosis and cirrhosis (1). High spatial resolution and SNR in phase images are key factors for high quality elastogram and conflict with the scan time. Scan time is a critical issue in MRE as breath hold is normally performed during to avoid subject motion in liver scan. The use of a fast acquisition strategy such as EPI partly solves this issue, as the scan may be completed within a single breath hold. However the EPI acquisition provides limited SNR and spatial resolution and suffers from well-known image distortion. In the method, we set to reduce the scan time by using the minimum TR and uses multi-echo acquisition to compensate for the potential loss of phase SNR due to TR reduction.

Method In MRE, the signal phase may be written as [1]. The time dependent term $\theta(t)$ due to B0 field inhomogeneity and local susceptibility is usually suppressed via phase subtraction leaving only the second term $\theta(d,\emptyset)$. The noise power in phase is time dependent and may be written as [2] (2), where σ is the additive noise. Note the factor of 2 is a result of the phase subtraction. In [2] the steady state magnetization M_0 is user controllable by adjusting the flip angle and TR used in scans. In practice a long TR is intentionally used in to achieve a sufficient SNR in MRE scans.

$\theta = \theta(t) + \theta(d,\emptyset)$ [1] $\qquad$ $\sigma_\theta^2(t) = 2\left(\frac{\sigma}{M_0 e^{\frac{-t}{t_2^*}}}\right)^2 \quad M_0 \propto FA, TR$ [2] $\qquad$ $SNR_{ME} = \frac{N\theta(d,\emptyset)}{\sqrt{\sum_{n=1}^{N}\sigma_\theta^2(t_n)}}$ [3]

In practical implementation of MRE, the TR needs to be multiple of the period of the external wave in order to synchronize the motion wave with MEG in consecutive TRs. As a result in a FGRE based MRE sequence, a TR of 50ms that is 3 cycles of the wave (assuming a motion of 60Hz) is used in order to achieve sufficient SNR, while the minimum TR even shorter than 2 wave cycles. We hence propose to utilize the idle time within the TR to place additional echo read-outs as shown in Fig.1 while reducing the TR to 33ms (2 motion cycles). After cancelation of the time dependent phase term, the signal phase attributed to the external motion from the multi-echoes are averaged to compensate for the SNR loss from using a shorter TR. The theoretical phase SNR after *N* echo averaging is given in [3], where $t_n$represents the *n*th echo time. In this way a shorter TR may be used while maintaining the phase SNR with no additional cost. In theory, the same may be achieved by reducing the readout bandwidth (RBW), however a reduced RBW increases the sensitivity to subject motion and B0 inhomogeneity.

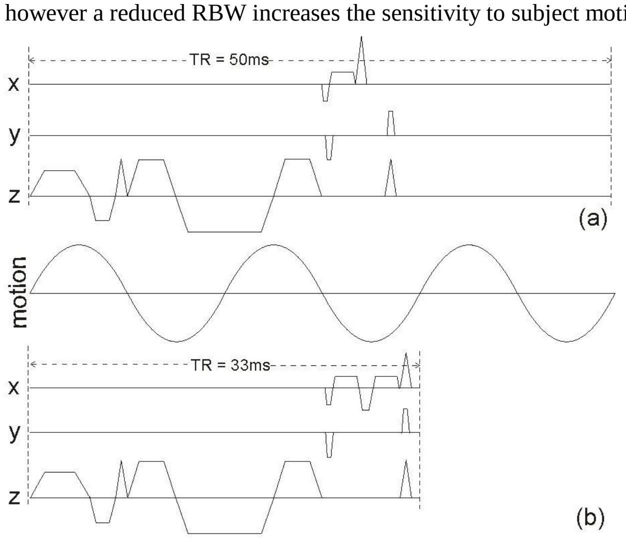

Figure 1: MRE pulse sequence with (a) single echo and a TR of 3 motion cycles (b) dual echo and a TR of 2 motion cycles.

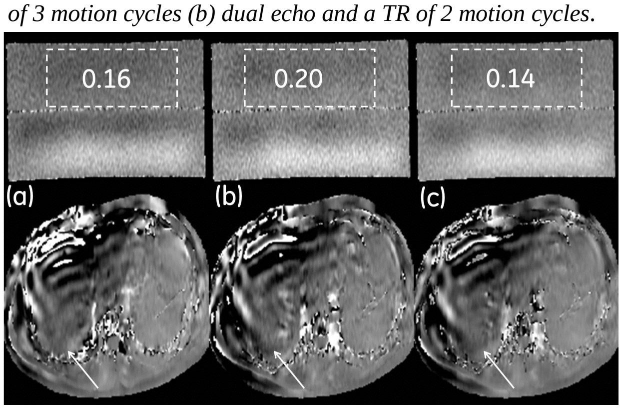

Figure 2: Motion waveform images obtained using (a) single echo acquisition with a TR of (a) 50ms (b) 33ms, (c) dual echo acquisitions with a TR of 33ms. ROI region noise measurements in phantom results are labeled, the arrow indicates the visible SNR improvement in volunteer images.

Experiment The dual-echo acquisition in MRE as illustrated in Fig.1 is implemented in a GE 1.5T body scanner. A phantom (filled with agar gel) and a volunteer were scanned with either single echo (TR = 50ms) or dual echo acquisitions (TR = 33ms). An external driver was used to induce a periodic acoustic wave at 60Hz which was applied to the imaging subject. The following parameters were used: 256x96, FA = 20deg, FOV = 30cm, RBW = 31.25 kHz. With an ASSET factor of 2, the scan times were 19s and 13s respectively with four temporal phases. Consent form was obtained prior to volunteer scan. In phantom scan, each scan was repeated twice to measure the noise level of a selected ROI in a RMS manner [4], where $\theta_1$and θ_2 are the processed phase images from the two scans.

$$noise_\theta = \frac{\sqrt{\sum_{ROI}\left(\frac{\theta_1+\theta_2}{2}\right)^2}}{sd(\theta_1-\theta_2)} \qquad [4]$$

Results The phantom and in vivo phase images from scans using TR = 50ms, TR = 33ms with single echo and TR = 33ms with dual-echo phase averaging are shown in Fig.2 (a),(b),(c) respectively. In the phantom results it is seen that the noise level in the phase increases with a shorter TR as expected (compare Fig.2(a) and (b)). The dual echo phase averaging effectively compensated for the SNR loss (Fig.2.(c)) , as indicated by a lower noise level measurement. Consistent observations are made in the volunteer results: a shorter TR leads to higher level of noisy fluctuations whereas dual echo phase averaging effectively reduces the noise level (arrow).

Discussion & conclusion In MRE, the phase of signal of interest is not TE dependent and hence signal phases from multiple echoes may be directly averaged. The multi-echo averaging effectively increases the SNR in the combined signal phase, and allows us to use a shorter TR to reduce the overall scan time to achieve earlier breath holds. Alternatively, with the current protocol, a matrix size of 256x128 may be easily achieved with a single breath hold (17s), achieving a better spatial resolution.

Reference (1) Mariappan, et al., Clinical Anatomy 2010, 23(5):97-511
(2) Wu, et al., Neuroimage 2012, 59(1):297-305