

Motion correction in MR-elastography

Marion Tardieu¹, Marie Poirier-Quinot¹, Emeline Lamain¹, Ralph Sinkus², Luc Darrasse¹, and Xavier Maître¹

¹IR4M (UMR8081), CNRS, Univ Paris-Sud, Orsay, France, ²Centre de Recherches Biomédicales Bichat-Beaujon (UMR773), CRB3, Inserm, Paris, France

Introduction

MR-elastography (MRE) aims at characterizing the properties of tissues by motion sensitized phase measurements. The displacement field induced by the propagation of a shear wave within the targeted tissue is recorded over the mechanical cycle along the three spatial directions. Hence, the shear viscoelastic moduli may be inferred from the inversion of the wave equations [1]. Long acquisition times may lead to a motion-induced mismatch of the displacement field components as a result of physiological, cardiac, respiratory or physical, patient motions. Spatial normalization over the whole data set provides primary motion correction for improved localization and reduced variance [2]. However, such a normalization process involves 3D rotations, which furthermore mix up the displacement field components encoded onto the MR signal phase. In this work, secondary motion correction was successfully added to account for it. It was implemented and tested onto an easily-motion-controlled whole-brain MR-elastography data set where motion could be added. This approach improved processing of MR-elastography data for rotation motion of already less than a degree.

Materials and methods

In vivo brain MRE acquisition: Acquisitions were performed in a 1.5 T scanner (Achieva, Philips Medical Systems, The Netherlands) at CIERM, Bicêtre Hospital, France. Pressure waves were generated by a loudspeaker at 43 Hz and transmitted to the oral cavity along a waveguide. Motion encoding gradients of 25.2 mT·m⁻¹ were synchronized with the pressure waves. Eight snapshots of the wave propagation were acquired during the oscillatory cycle, along the three spatial directions, leading to a whole data set free of motion artefact.

Simulated motion artefact brain phantom: Step 1 - To actually simulate patient motions during the acquisition within the scanner, the acquired displacement fields, namely the signal phases, were correspondingly rotated before applying the spatial 3D rotations on the complex data matrix. Motions with different amplitudes were then simulated by degrading the whole data set by 3D rotations of varying angles, within the realistic range from 0.5 to 5°, between the three spatial encoded directions. This represents the initial motion-artefacted data set used from now on.

Spatial and phase correction: Step 2 - Spatial normalization was performed on the motion-artefacted data set using SPM8 (UCL Institute of Neurology, London, United Kingdom) so the angles of the 3D rotations simulated in Step 1 could be inferred.

Step 3 - The transformation matrix was constructed from the inferred angles and was applied to the spatially normalized components of the displacement fields.

Results

On Figure 1.a, the mean standard deviation of the difference between the original motion-free and the corrected motion-simulated signal phases is represented at the three correction steps for the different motion-simulated angles from 0.5° to 5°. Plots of Figure 1.b and c similarly represent the mean standard deviation of the differences for the dynamic, G_d , and loss, G_l , shear moduli at the same reconstruction steps. These plots give a fair trend of the mean errors left on the displacement field, G_d , and G_l when no motion correction (Step 1), primary (Step 2) or secondary (Step 3) correction is applied.

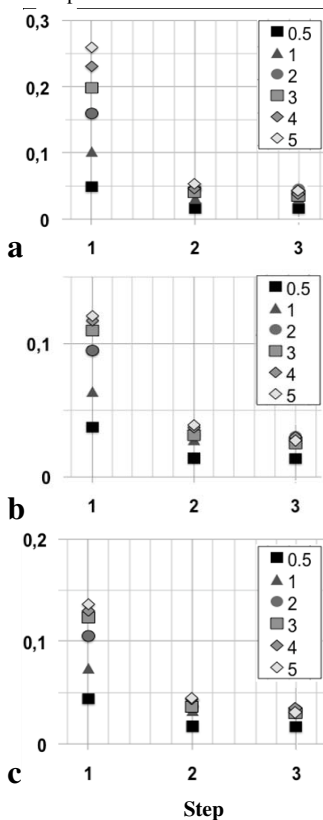


Figure 1 Evolution of the mean standard deviation of the difference between the original motion-free and the corrected motion-simulated signal phases (a), G_d (b), and G_l (c) at correction steps 1, 2, and 3 for the different motion-simulated angles from 0.5° to 5°.

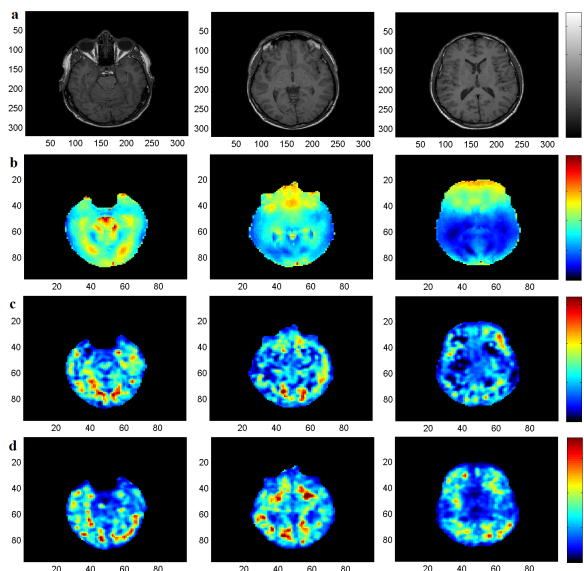


Figure 2 Anatomy (a) Displacement field total magnitude map (b) Dynamic shear modulus map G_d (c) Loss shear modulus map G_l (d) after full motion correction for a 3° simulated motion distortion in three out of forty-three slices acquired in the brain of a healthy volunteer.

Figure 2 shows the resulting maps of the total amplitude of the displacement field, G_d , and G_l with respect to the magnitude image of three out of the forty-three slices acquired in the brain after full motion correction.

Discussion

Improvements on the *mean phase error* increase with the angle of the induced motion. They range between 66% for 0.5° and 83% for 5°. The major gain is obtained at Step 2 through spatial normalization. Nevertheless, at Step 3, phase correction still carries almost 12% to 20% *mean error* reduction for 3° to 5°. The overall proposed correction is robust and overcomes the reconstruction added noise.

The maps of the displacement field total amplitude as well as the maps of G_d and G_l , shown in Figure 2, reveal high left-right symmetry of the brain structures after full motion correction.

More generally speaking, MR-elastography of moving organs like liver, lung, or heart would highly benefit from processing motion correction as developed here.

References

- [1] Sinkus *et al.* Magn Reson Imaging **23**:159-165 (2005)
- [2] Ashburner *et al.* Clinical MRI/Developments in MR **7**:26–28 (1997)