Reconstruction accuracy dependence with induced-shear-wave magnitude in Magnetic Resonance Elastography

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Since 1996, Magnetic Resonance Elastography (MRE) holds the promise for absolute quantitation of the mechanical parameters of living tissues [1]. The reproducibility of the technique was challenged [2] while the measurement precision was determined by the uncertainty of the recorded MR-signal phase onto which the inferred motion is encoded [3]. We assumed that the ratio of the resulting total wave amplitude to its related uncertainty, $A_T/\Delta A_T$, should be considered to validate the acquired set of MRE data. Nevertheless, as long as this ratio is greater than unity, the validity of the extracted mechanical parameters might not be questioned. Here, we extract the complex shear modulus, $G = G' + jG''$, by inversion of the three-dimensional equation of motion [4] for a wide range of inferred wave amplitude, starting from zero, in a breast phantom. The shear dynamic, G', and loss, G'', moduli were found to increase with the wave amplitude before reaching a plateau at ratios $A_T/\Delta A_T$ much greater than one. Experiments were carried with standard motion-sensitized refocused field echo (RFE) [1] and motion fractional-encoding fast field echo (FFE) [5], for which sensitivities largely differ, so the relevance of a MRE-validity threshold based on the ratio $A_T/\Delta A_T$ could be exhibited.

I. INTRODUCTION

MRE is generally considered as a non-invasive, robust, and accurate imaging tool for quantifying *in vivo* the mechanical properties of tissues, which are significantly altered by most of diseases. Low frequency shear waves are usually induced from the surface of the body into the targeted tissue and the resulting displacement field therein is recorded by phase contrast MRI [1]. To access the lung and the brain, pressure waves were efficiently guided along the natural pathways through the buccal cavity in humans up to 235 Hz and in small animals up to 630 Hz. Directly applied at the surface of a breast phantom, such guided pressure waves can induce almost arbitrarily-large displacements throughout the phantom [3]. The setup could then serve as a bench test to evaluate in this paper the accuracy of MRE according to the induced-shear-wave magnitude. As we extracted the shear viscoelastic moduli, G' and G'', by inverting the 3D wave equation [4], we assumed that the ratio between the total magnitude of the shear wave generated in the tissue, A_T, and its associated uncertainty, ΔA_T, were the relevant parameter to account for the measurement signal to noise ratio (SNR) and the reconstruction-added noise so to set a validity threshold for MRE.

II. MATERIALS AND METHODS

Pressure wave generation

Monochromatic sinusoidal waves at 85 Hz were generated by a function generator (AFG 3021B, Tektronix, USA) with V_{RMS} up to 2.8284 V. Then they were amplified by a power amplifier (P2500S, Yamaha, Japan) before being transduced into pressure waves by a loudspeaker (12NW100, B&C Speakers, Italy) and guided through a 3.5 m long, 20 mm diameter, altuglas® tube, to the surface of the phantom. The output pressure was recorded by an optical pressure sensor (OPP-M, OpSens, Canada) connected at the end of the waveguide, close to the imaged phantom. The linearity of the excitation system was checked over the range of the applied voltage ($0 \leq V_{RMS} \leq 2.8284$ V).

Experimental setup

Experiments were performed in a 1.5 T scanner (Achieva, Philips Healthcare, The Netherlands) on a breast phantom (Model 073, CIRCS Inc., USA) with two flexible SENSE coils (SENSE Flex-M, Philips Healthcare, The Netherlands). The applied RFE MRE and FFE MRE sequences shared the same geometrical parameters with FOV $= (128 \times 128 \times 76)$ mm^3, isotropic spatial resolution of $(2$ mm$)^3$, and matrix $= (64 \times 64 \times 38)$. TE/TR $= 41/2235$ ms for RFE and $TE/TR = 9.2/450$ ms FFE. Motion-encoding bipolar gradients of 21 mT/m were synchronized and sequentially offset four times with respect to the pressure wave excitation. They were applied during 11.8 ms in the RFE MRE sequence and during 8.2 ms in the FFE MRE sequence. The three spatial components of the displacement field were acquired over acquisition times of 24 min for RFE and 6 min for FFE. The acquisitions were repeated for $0 \leq V_{RMS} \leq 1.2374$ V every 43.8 mV for RFE and for $0 \leq V_{RMS} \leq 2.8284$ V every 176.8 mV for FFE.

Total wave magnitude and uncertainty ratio

For each acquisition, the magnitude of the three components of the recorded displacement field were deduced in every voxel from the sinusoidal time evolution of the propagating wave. The resulting mean magnitudes, (A_X, A_Y, A_Z), yielded the mean total magnitude A_T over the whole phantom volume. SNR maps were calculated to produce the corresponding phase-error maps along the three spatial directions $(\Delta\Phi_{X,Y,Z} = \mathrm{atan}(1/SNR_{X,Y,Z}))$ and to process the average measurement uncertainty of the

component magnitudes and of the total magnitude of the induced shear wave, ΔA_T [1].

III. RESULTS

The sound pressure level at the surface of the breast phantom reached 163.18 dB for RFE and 163.82 dB for FFE (Figure 1). The excitation system remains linear until 162.5 dB as shows the departure of the linear fits from the measured dots for both RFE and FFE experiments.

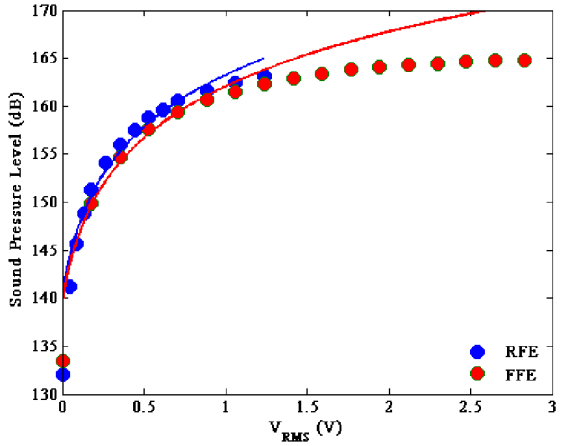

Figure 1: Sound pressure level (dB) at the surface of the breast phantom at the output of the waveguide versus the applied root mean square voltage V_{RMS} for RFE acquisitions (blue dots) and FFE acquisitions (red dots). The lines are the corresponding linear fits performed for $0 \leq V_{RMS} \leq 1$ V.

Figure 2 shows the extracted mean dynamic and loss moduli obtained in the breast phantom as a function of the shear wave total magnitude A_T for RFE and FFE. ΔA_T was (1.06 ± 0.35) µm μm for RFE and (3.52 ± 0.20) µm for FFE. The viscoelastic moduli were normalized for the plots by their maximal value. Both curves exhibit an exponential growth that reaches a plateau within the normalized relative uncertainties $\delta G'$ and $\delta G''$ at different values of A_T: around 9.2 μm for both G' and G'' for RFE and between 46 μm and 95 μm for G' and G'' for FFE. For RFE, after reaching a maximal value, the viscoelastic moduli decrease down by 10% while A_T is increased.

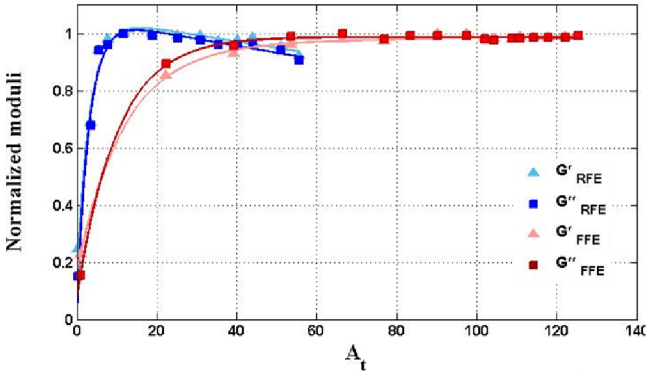

Figure 2: Normalized values of mean shear viscoelastic moduli G' (pale) and G'' (dark) as a function of to the total shear wave magnitude A_T (μm) for RFE (blue dots) and FFE (red dots). Lines are guiding bi-exponential fits.

The normalized mean viscoelastic moduli become similar when they are plotted as a function of the ratio $A_T/\Delta A_T$ (Figure 3). They all reach a plateau within the normalized relative uncertainties $\delta G'$ and $\delta G''$ for $A_T/\Delta A_T$ between 16 and 17 for RFE and between 20 and 38 for FFE.

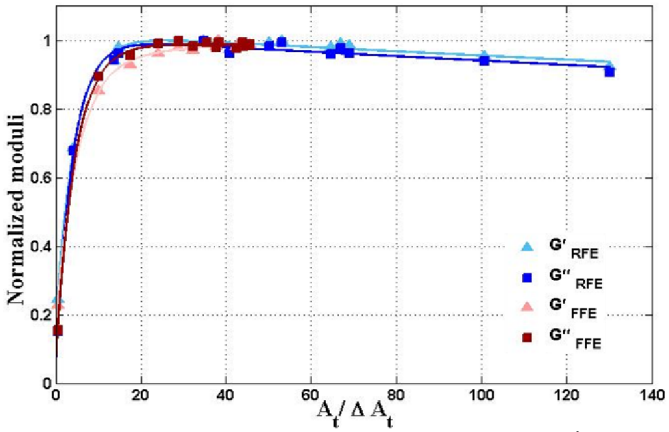

Figure 3: Normalized values of mean shear viscoelastic moduli G' (pale) and G'' (dark) as a function of $A_T/\Delta A_T$ for RFE (blue dots) and FFE (red dots). Lines are guiding bi-exponential fits.

IV. DISCUSSION – CONCLUSION

RFE acquisitions overall lasted eight hours over which the breast phantom had become *softer* under the continuous applied pressure wave. Thereafter, the decreasing trend followed by the viscoelastic moduli for large A_T corresponds to an effective decay of s G' and G'' over time. Figure 3 show a rather general behavior of the mechanical properties with respect to the ratio $A_T/\Delta A_T$. It effectively sets a threshold above which MRE data can be considered as valid. Hence, total shear wave magnitude 16 to 38 times greater than the measurement uncertainty ΔA_T is required to secure absolute quantification of the mechanical properties *in vivo*. Any MRE study below this threshold is prone to inaccurate mechanical characterization and to intra- and inter-subjects misinterpretation.

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