

A Single-Element Focused Transducer Method for Harmonic Motion Imaging

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Abstract— The feasibility of the harmonic motion imaging (HMI) technique for simultaneous monitoring and generation of focused ultrasound therapy using two separate focused ultrasound transducer elements was previously shown [1]. In this study, a new HMI technique is described that images tissue displacement induced by harmonic radiation force excitation using a single focused ultrasound element. First, wave propagation simulation models were used to compare the use of one Amplitude-Modulated (AM) focused beam versus two overlapping focused beams as previously implemented for HMI [2]. Simulation results indicated that, unlike the two-beam configuration, the AM beam produced a consistent, stable focus for the applied harmonic radiation force. The AM beam thus offered the unique advantage of sustaining the application of the spatially-invariant radiation force. Experiments were then performed on gelatin gel phantoms and in-vitro tissues. The radiation force was generated by a 4.68 MHz focused transducer using a low-frequency Amplitude-Modulated (AM) RF-signal. A 7.5 MHz single-element, imaging transducer was placed through the center of the focused transducer so that the diagnostic and focused beams were aligned. Consecutive RF signals were acquired with a PRF of 5 kHz and the displacements were estimated using 1D cross-correlation. Finally, taking advantage of the real-time capability of our method, the change in the elastic properties was monitored during focused ultrasound (FUS) ablation of in-vitro tissues. Based on the harmonic displacements, their temperature-dependence, and the calculated acoustic radiation force, the change in the regional elastic modulus was monitored during heating. In conclusion, the feasibility of using an AM radiation force for HMI for simultaneous monitoring and treatment during ultrasound therapy was demonstrated in phantoms and tissues in-vitro. Further study of this method will include stiffness and temperature estimation, ex-vivo and in-vivo.

Keywords: *Acoustic radiation force, Harmonic Motion Imaging, Displacement, Oscillatory, Noninvasive measurement*

I. INTRODUCTION

Palpation is commonly used by physicians to distinguish cancerous from normal tissue in areas such as the breast and the prostate. However, this technique can be incapable of detecting lesions deep in tissues such as the liver. Several studies have shown that the acoustic radiation force can be generated, and thus capable of perturbing, at variable depths within the tissue [1-7]. The acoustic radiation force produces localized displacements of the soft tissue in the region of the focal zone. The displacements of the tissue can be monitored in real-time using consecutively acquired RF signals [2] [3] [4].

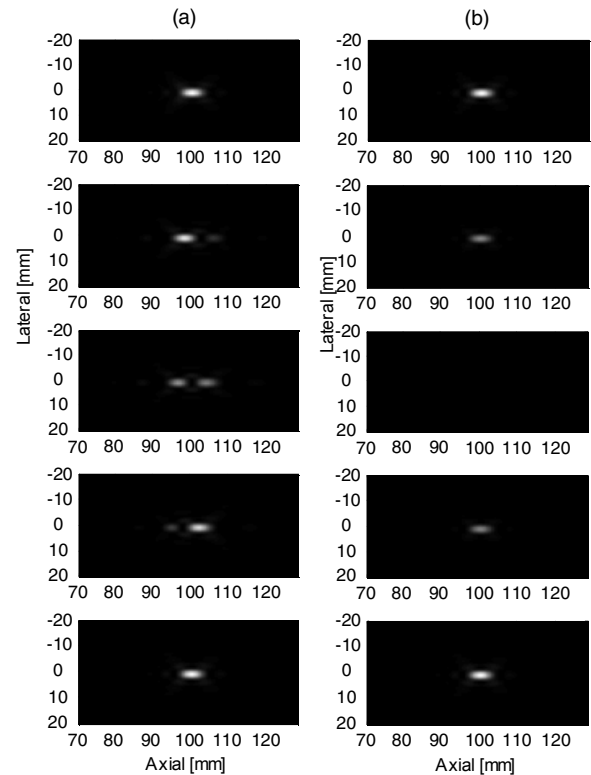


Figure 1. The acoustic radiation force field at 4 ms intervals (from top to bottom) produced by (a) two overlapping focused ultrasound beams at two different frequencies $\Delta f = 50$ Hz and (b) one-beam modulated at 50 Hz. Note the spatial invariance of the radiation force field using the one-beam configuration versus the alternative.

Harmonic motion imaging (HMI) technique estimates unidirectional tissue displacements remotely induced by the acoustic radiation force [1]. In this paper, a new HMI method is discussed, in which the harmonic radiation force is produced by a single focused ultrasound element locally within a medium such as, gel phantoms and bovine liver. The displacements are measured at the same location of force application using a separate imaging transducer. One major advantage of the HMI technique is that the displacements are measured during application of the acoustic radiation force so that this method can be used for monitoring of the mechanical properties of tissues during focused ultrasound (FUS) therapy.

In a previous study, we proposed a similar technique for monitoring of focused ultrasound therapy using two separate focused ultrasound transducer elements working at different

frequencies (f and $f + \Delta f$) [1][6]. The two overlapping focused beams produced an acoustic radiation force field moving at the difference frequency (Δf) (Fig. 1(a)). In this paper we propose to use one amplitude-modulated focused ultrasound beam to generate the harmonic acoustic radiation force (Fig. 1(b)). The AM beam thus offered the advantage of sustaining the application of the radiation force at the same tissue region *and* a simpler transducer design.

The feasibility of the technique is shown in tissue-mimicking phantoms with different stiffnesses. An experiment is then performed in a tissue mimicking phantom containing a stiffer inclusion. This shows that HMI is able to map the mechanical properties of the material with high resolution. Finally, FUS therapy combined with HMI is performed in in-vitro tissue samples. The displacements of the tissue are monitored during ablation and the results show that HMI is able to detect the formation of lesions in real-time.

II. METHODS

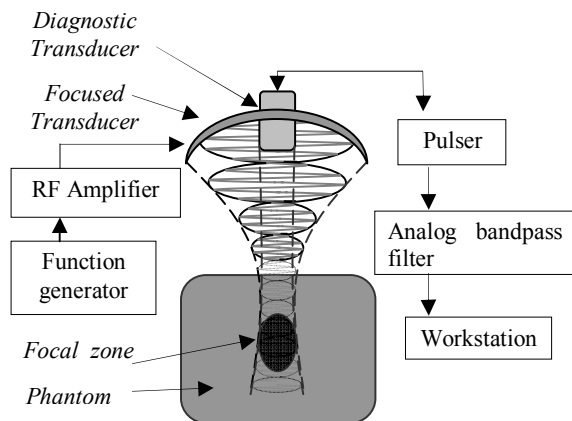


Figure 2. Experimental setup to generate harmonic acoustic radiation force and measure displacement.

Our experiments were performed on two models of tissue mimicking phantoms and in-vitro tissues (bovine liver). The experimental setup is shown in Fig. 2. We chose gelatin gel material for our tissue mimicking phantoms. Phantom preparation was completed according to Hall et al. [8]. Five homogeneous phantoms with different stiffness (20 kPa, 30 kPa, 40 kPa, 50 kPa, and 60 kPa) and a 20 kPa tissue mimicking phantom with 40 kPa inclusion were made.

Acoustic radiation force was generated by a 4.68 MHz focused transducer using a low-frequency Amplitude-modulated (AM) Radio Frequency (RF) signal. A function generator (Agilent (HP) 33120A) was used to produce the RF signal at 4.68 MHz (Fig. 3(a)). The amplitude of the RF signal was then modulated using a second function generator that generates a low frequency modulation (Fig. 3(b)). Therefore, the focused transducer generated a pressure field shown in Fig. 3(c) and a modulated acoustic intensity plotted in Fig. 3(d).

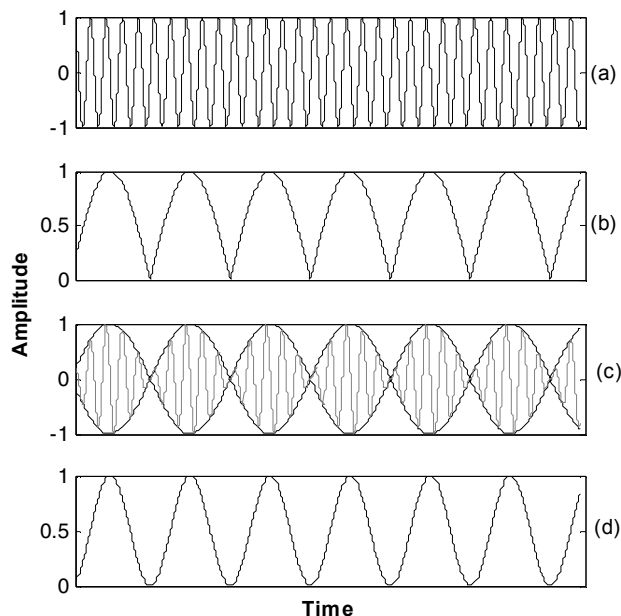


Figure 3. (a) High frequency input, (b) low frequency modulation, (c) AM signal output of function generator, and (d) normalized acoustic intensity generated at the focus

Amplitude-modulated (AM) frequencies were varied from 10 Hz to 100 Hz. The output of the function generator could be adjusted from 100 to 600 mVpp and then amplified by a 50dB RF-amplifier (EIN 3100L). The sonication time was adjusted to induce 100 oscillations at the frequency of the modulation. A 7.5 MHz single-element, diagnostic transducer was placed through the center of the focused transducer so that the diagnostic and focused beams were properly aligned. A bandpass analog filter (Reactel, Inc.) was used to filter out the spectrum of the focused beam. Consecutive RF signals were acquired with a Pulse Repetition Frequency (PRF) of 5 kHz (Panametrics 5051PR). An acquisition board (Gage Applied Technologies) was used to capture RF data with sampling frequency of 80 MHz. A 1D cross-correlation technique was used to calculate axial (along the ultrasound beam axis) displacements between two successive RF images [1].

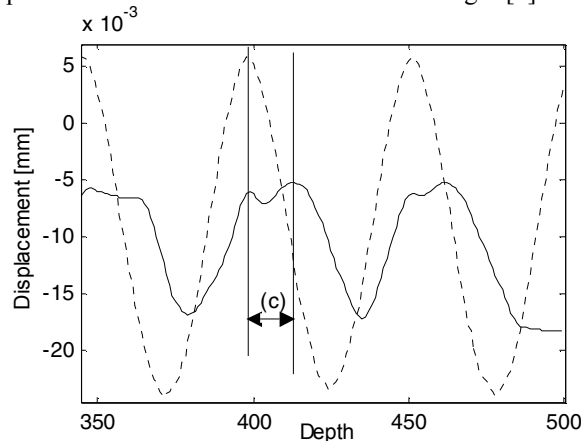


Figure 4. (a) Output displacement in gelatin gel, (b) normalized input radiation force intensity, and (c) phase shift

This method is simple to implement, computationally efficient, and provides an accurate estimation of small displacements (on the order of 10 μm). The amplitude of the displacement variation was measured, and a phase shift was calculated between the amplitude-modulated signal and the displacement (Fig. 4).

III. RESULTS

A. Tissue-mimicking phantom experiments

Fig. 5(a) and (b) show results from the five gelatin gels of different stiffness. In order to create an effective acoustic radiation force, the intensity of the focused beam used was equal to 658.5 W/cm^2 at an AM frequency of 50 Hz. The intensity of the focused beams is calculated according to

$$I = \frac{P^2}{2\rho c} \quad (1)$$

where, $\rho = 1.1 \text{ g}/\text{cm}^3$, $c = 1.5 \cdot 10^5 \text{ cm}/\text{s}$ and P = acoustic pressure at the focus for $V = 500 \text{ mVpp}$. Our results show that the displacement amplitude decreases from 10.3 microns to 4.15 microns with gel stiffness increases from 20 kPa to 60 kPa (Fig. 5(a)). The HMI displacements clearly indicate the stiffness variation. Phase shifts decrease from -66.4° to -30.4° consistent with increasing gel stiffness (Fig. 5(b)).

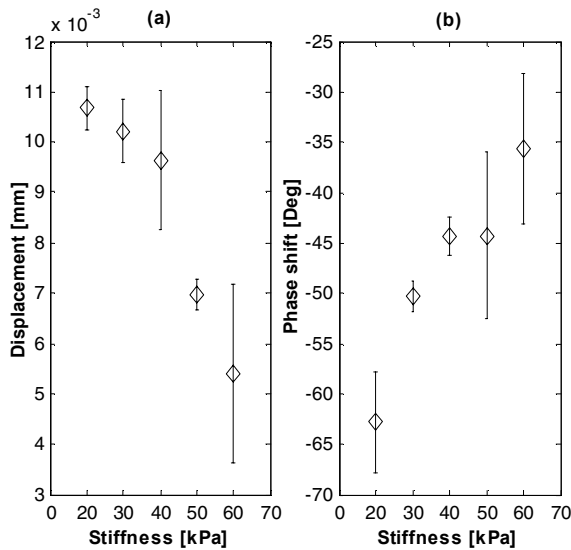


Figure 5. a) Displacement measurement and (b) phase shift in five gelatin gels.

Secondly, we performed an experiment in a gelatin gel containing a cylindrical inclusion twice stiffer than the background (20 kPa) (Fig. 6) The transducer was moved along a 2D grid using a computer-controlled positioner (Velmex, Inc.). A $20 \times 20 \text{ mm}^2$ area was raster-scanned with a step size of 1mm. The 2D maps of displacement amplitude and phase shift are shown in Fig. 7(a) and (b), respectively. The average displacement in the inclusion is 3.3 microns and the phase shift is -34° (Fig. 7(a)). The average displacement in the

surrounding gel is 6.1 microns and the phase shift is -65.9° (Fig. 7(b)). These results consistent with the inverse relationship between displacement and elastic modulus [6]. The phase shift relationship between amplitude-modulated input signal inducing the acoustic radiation force, and measured displacements can be used to estimate the viscoelastic properties in tissue-mimicking phantoms as well as in-vitro tissues.

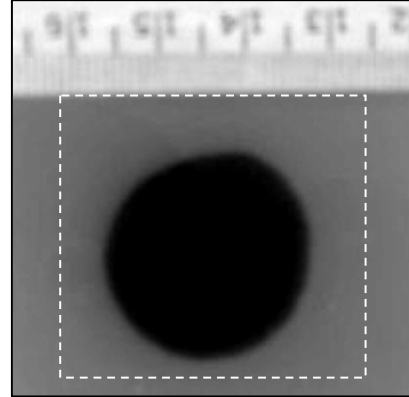


Figure 6. Optical image of a phantom with an inclusion

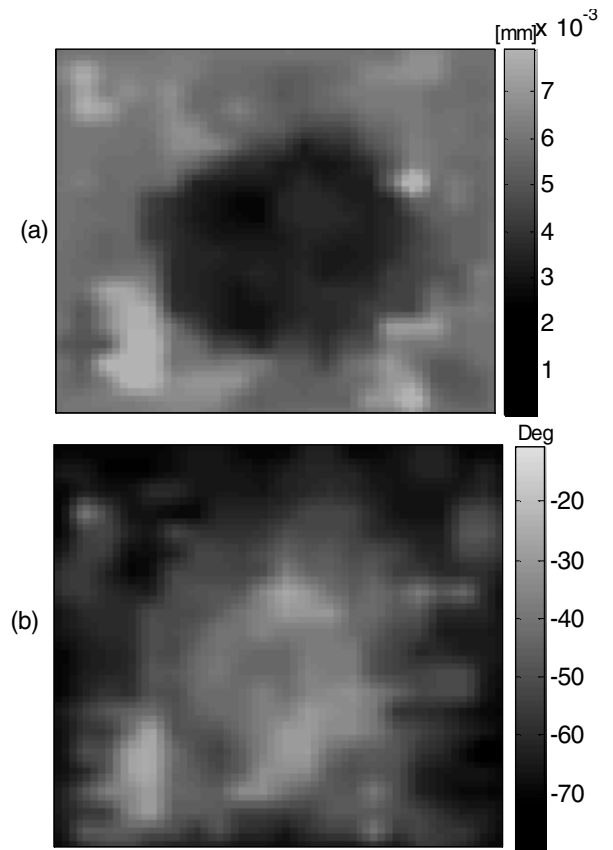


Figure 7. (a) Displacement measurement and (b) phase shift in a gelatin gel phantom with inclusion.

B. Monitoring of FUS ablation

Finally, an ablation tissue experiment was performed on in-vitro tissue samples. A piece of $50 \times 50 \text{ mm}^2$ bovine liver submerged in degassed water. In this experiment, the intensity of the focused ultrasound beam was 948.23 W/cm^2 (calculated according to (1) for $V=600 \text{ mVpp}$) at the focus, in order to simultaneously generate the harmonic radiation force and the tissue ablation. The frequency of the modulation was 50 Hz and the total sonication time was approximately 288 sec. The displacements were monitored in real-time. The variation of the amplitude and the phase shift are shown in Fig. 8(a) and (b). The oscillatory displacement amplitude and displacement-force phase shift start to rapidly decrease beyond 120 s of continuous sonication, possibly indicating tissue coagulation beyond that sonication period.

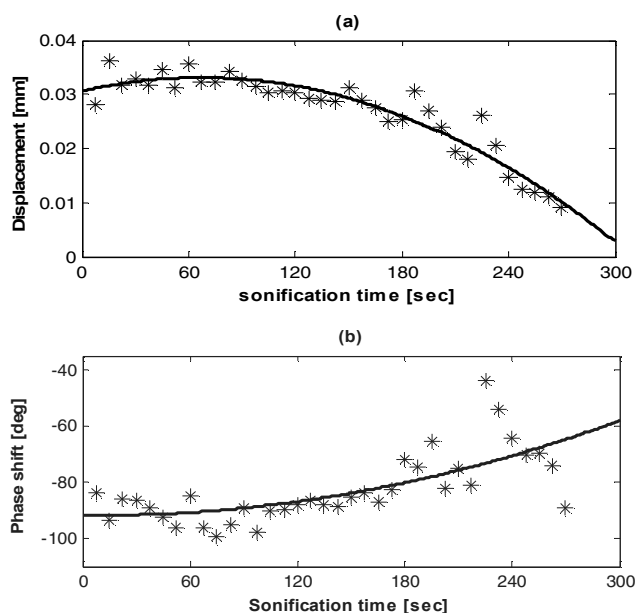


Figure 8. (a) Displacement measurement and (b) phase shift in a bovine liver during sonification.

Furthermore, it should be noted that the effect on the echoes induced by the change of the speed of sound with temperature induces a low frequency shift of the speckle, and was successfully separated from the higher frequency displacements induced by the harmonic radiation force and imaged here. This technique is therefore able to accurately monitor the heating process and possibly detect the time of coagulation.

IV. CONCLUSION

The feasibility of using an amplitude-modulated radiation force for harmonic motion imaging and simultaneous monitoring of tissue elasticity variation during ultrasound therapy was shown. This method was demonstrated in tissue-mimicking phantoms and in-vitro tissues. A variation in phase shifts in Fig. 8(b) shows a decrease in phase shift during sonication that may have possibly occurred following coagulation and lesion formation. Further study of this method will include stiffness, viscoelastic properties measurements, as well as temperature and coagulation mapping, both ex-vivo and in-vivo.

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