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Interfacial State Assessment using Vibration-based Shear Wave Elastography

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Abstract: Fascia hydrorelease is an emerging technique aimed at improving gliding around fascial layers. This study investigates how interfacial conditions around fascia affect shear wave propagation. We conducted Finite-Difference Time-Domain simulations and ex-vivo measurements to assess shear wave phase morphology. Both approaches showed that reduced interfacial adhesion leads to measurable phase shifts. These results suggest that shear wave phase analysis may offer a means to assess fascial adhesion.

Keywords: Shear wave elastography, Phase velocity, Fascia, Hydrorelease, Acoustic morphology

Introduction

Fascia is a connective tissue that links muscles to surrounding anatomical structures, forming a continuous, intricate web throughout the entire body [1]. Inflammation and fibrosis cause abnormal fascia to develop, resulting in increased tissue stiffness and restricted range of motion. These conditions can lead to musculoskeletal disorders like frozen shoulder, significantly reducing patients' quality of life [2]. Hydrorelease (HR) has recently gained attention as a therapeutic technique for addressing disorders related to the fascia. HR is performed under ultrasound guidance to deliver saline into the fascial layers, aiming to normalize fascial mobility [3]. A previous biomechanical study has reported reduced gliding resistance following HR [4], suggesting that the procedure may alter interfacial adhesion conditions.

To evaluate such mechanical properties non-invasively, shear wave elastography (SWE) has emerged as a promising modality. Several studies have applied SWE to the myofascial tissue and estimated its elasticity based on the group velocity of shear waves [5, 6]. However, group velocity-based methods assume that the medium is linear, elastic, and locally homogeneous. These assumptions may not hold for myofascial tissue, which typically measures less than 4 mm in thickness [7]. In contrast, phase velocity-based approaches can be better suited for those thin-layered, potentially heterogeneous structures, and are thus considered more appropriate for assessing the mechanical properties of shear waves in fascial tissue. The influence of interfacial conditions on shear wave propagation, however, remains underexplored and is investigated in this study using a phase velocity-based approach. In this research, we conducted Finite-Difference Time-Domain (FDTD) simulations in which interfacial parameters were introduced to model varying adhesion conditions. In addition, ex-vivo experiments were performed using chicken breast, where HR was applied between the pectoralis major and minor muscles to alter fascial adhesion. The resulting morphological characteristics of shear wave phase were visualized using a Doppler-based phase reconstruction method previously developed by the authors [8, 9].

Shear Wave Reconstruction Method

During shear wave propagation, the ultrasound probe detects a Doppler frequency shift, denoted as $\Delta\phi$. The received signal can be expressed as:

$$y(x,z,t) = a \cdot \exp\left(j\left(2\pi f_0 t + \Delta\phi\left(x,z,t\right)\right)\right), \quad (1)$$

where f_0 is the center frequency of the ultrasound probe, and a is the signal amplitude. Through quadrature detection, the complex Doppler component $\exp\left(j\Delta\phi\right)$ can be extracted. The Doppler phase shift can be related to the axial particle displacement as follows:

$$\exp\left(j\Delta\phi\left(x,z,t\right)\right) = \exp\left(j\frac{4\pi f_0}{c} \cdot u_z\left(x,z,t\right)\right). \quad (2)$$

Here, $u_z\left(x,z,t\right)$ represents the displacement of particles along the z-axis. This displacement can be modeled by the following shear wave equation:

$$u_z\left(x,z,t\right) = u_0 \sin\left(2\pi f_s t + \boldsymbol{k} \cdot \boldsymbol{x}\right), \quad (3)$$

where f_s is the shear wave frequency, u_0 is the displacement amplitude and $\boldsymbol{k}$ is the wavenumber of the shear wave. Accordingly, the spatial phase distribution of the shear wave can be obtained by calculating the argument of the complex Doppler signal.

Simulation

The propagation of shear waves was simulated using the FDTD method. The basic governing equations are derived from the Navier-Stokes equation:

$$\frac{\partial v_x}{\partial t} = \frac{1}{\rho}\frac{\partial \sigma}{\partial z}, \tag{4}$$

$$\frac{\partial v_z}{\partial t} = \frac{1}{\rho}\frac{\partial \sigma}{\partial x}, \tag{5}$$

$$\frac{\partial \sigma}{\partial t} = \mu\left(\frac{\partial v_z}{\partial x} + \frac{\partial v_x}{\partial z}\right) + \eta\left(\frac{\partial \dot{v}_z}{\partial x} + \frac{\partial \dot{v}_x}{\partial z}\right). \tag{6}$$

Here, v_x and v_z represent particle velocity in the x and z directions, respectively. ρ is the density of the medium, σ is the stress, μ is the shear mululus, and η is the viscosity coefficient. HR is known to reduce adhesions between myofascial tissue and surrounding tissues. To model this interfacial effect, we introduced adhesion coefficients α_x and α_z, which modify the interaction across interfaces. The modified set of equations incorporating these coefficients is expressed as:

$$\frac{\partial v_x}{\partial t} = \frac{1}{\rho}\cdot\alpha_z\frac{\partial \sigma}{\partial z}, \tag{7}$$

$$\frac{\partial v_z}{\partial t} = \frac{1}{\rho}\cdot\alpha_x\frac{\partial \sigma}{\partial x}, \tag{8}$$

$$\frac{\partial \sigma}{\partial t} = \mu\left(\alpha_x\frac{\partial v_z}{\partial x} + \alpha_z\frac{\partial v_x}{\partial z}\right) + \eta\left(\alpha_x\frac{\partial \dot{v}_z}{\partial x} + \alpha_z\frac{\partial \dot{v}_x}{\partial z}\right). \tag{9}$$

The adhesion coefficients α_x and α_z range from 0 to 1.0, where $\alpha = 1.0$ represents a fully continuous medium. These coefficients modulate the degree of interfacial continuity in the x and z directions, respectively, by scaling the spatial derivatives $\partial/\partial x$ and $\partial/\partial z$. Values less than 1 indicate a reduction in interfacial adhesion, with lower values corresponding to weaker adhesion across the interface.

The resulting phase distributions from the simulation are shown in Fig. 1. The simulation parameters are summarized in Tab. 1. As shown in Fig. 1-a, when the adhesion coefficient was low ($\alpha = 0.1$), a clear phase shift was observed at the interface located at $z = 10$ mm. In contrast, no significant phase shift was observed when the adhesion coefficient was set to $\alpha = 1.0$, representing a fully adhered interface. Fig. 1-b shows the relationship between the adhesion coefficient and the observed phase shift. As α decreased from 0.5 to 0.1, the phase shift at the interface increased from 0.08 rad to 0.47 rad. These results indicate that weaker interfacial adhesion causes greater discontinuities in shear wave propagation, resulting in a greater phase shift.

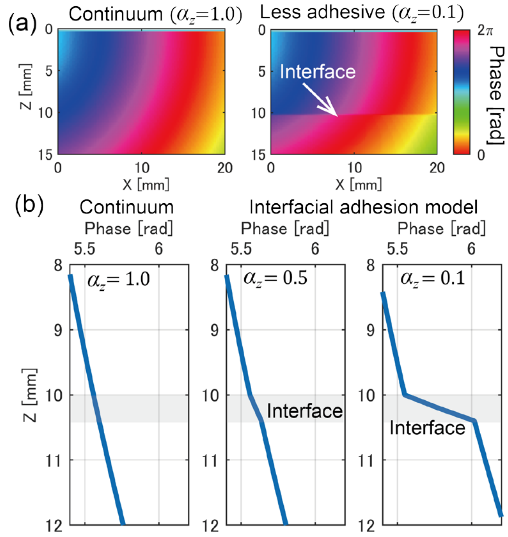

Fig. 1: (a) Phase map from the simulation output at adhesion coefficient $\alpha = 1.0$ (left) and $\alpha = 0.1$ (right). (b) Phase plots at $x = 10$ mm with different adhesion coefficients.

Tab. 1: 2D-FDTD simulation parameters in this study.

Parameter	Value
Model size	50×100 mm
Grid size	0.2 mm
Shear wave frequency	78 Hz (continuous)
Young's modulus	20 kPa
Viscosity η	0.002 Pa·s
Adhesion coefficient α	$0 - 1.0$

Experiments

To experimentally validate the phase shift phenomenon observed in the 2D-FDTD simulation, we conducted an ex-vivo experiment using a commercially available chicken breast tissue. The overall experimental setup is illustrated in Fig. 2-a. Shear wave measurements were performed both before and after the HR procedure to assess changes in interfacial adhesion. The first measurement was taken prior to HR, and the second was acquired 5 minutes after the intervention. The HR procedure was performed using an ultrasound imaging system (Logiq E10, GE Health-

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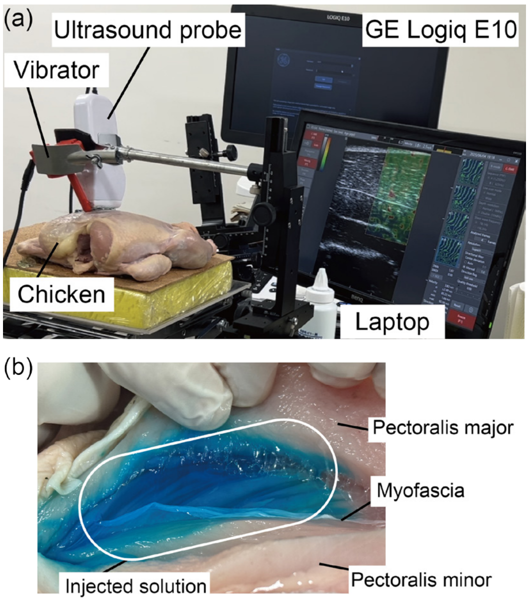

Fig. 2: a) Ex-vivo experimental setup. b) Dissection of chicken breast after HR, validating the successful injection of blue-dyed saline onto the myofascial tissue.

Care, USA) equipped with a linear probe (L2-9VN-D, center frequency: 9 MHz). To alter the degree of adhesion, 2.5 ml of blue-dyed saline solution was injected between the fascia and the overlying pectoralis major muscle. As shown in Fig. 2-b, anatomical dissection confirmed that the solution was correctly delivered into the fascial layer between the pectoralis major and minor muscles. Shear wave measurements were conducted using a separate linear array ultrasound probe (Finggal Link, Japan; 128 active elements, center frequency: 10 MHz). Shear waves were generated using an external mechanical vibrator operating at 78 Hz. The Finggal Link probe was mounted on a 5-axis stage, enabling precise repositioning to the same anatomical location before and after the HR procedure. The shear wave field was reconstructed on a laptop using the Doppler-based phase reconstruction method previously developed by the authors [10].

Results

) The spatial phase distribution of the shear wave is presented as heatmaps in Fig. 3-a. Following HR, a noticeable phase shift was observed, which resembled the simulation results assuming a non-adherent interface. Specifically, a phase shit of 1.2 rad was measured at $x = 4$ mm. As shown in Fig. 3-b, this phase shift extended across 5 mm, indicating a broader spatial spread compared to the sharp transition observed in the FDTD simulation. It is noted that this spread is larger than the fascial thickness observed on the B-mode image, which was approximately 0.5 mm.

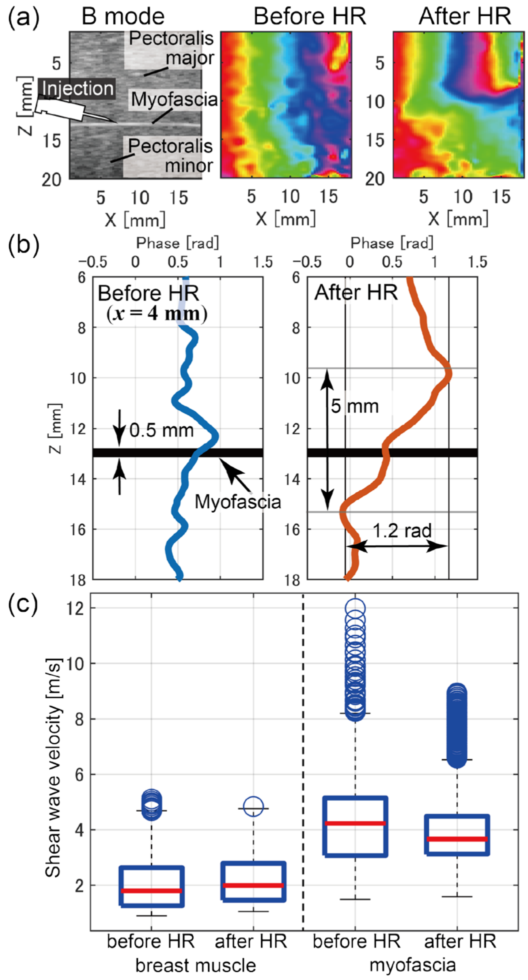

Fig. 3: (a) Resultant B-mode image and shear wave phase map before and after HR. (b) Shear wave phase plot at $x = 4$ mm, before and after HR. (c) Box plots of shear wave velocity within breast muscle and myofascial tissue.

The measured shear wave velocities are summarized as box plots in Fig. 3-c. Within the pectoral muscle, the velocity was 2.02 ± 0.85 m/s, while in the myofascial tissue it increased significantly to 4.35 ± 1.70 m/s. This velocity increase suggests that the fascia possesses higher stiffness than the surrounding muscle tissue, which is consistent with previous elastography studies on fascia [11]. After HR, the shear wave velocity in the myofascial tissue slightly decreased to 4.11 ± 1.60 m/s. This reduction may indicate that the injected saline temporarily decreased fascial elasticity due to the presence of interfacial fluid.

Discussion

Our results indicate that interfacial adhesion between fascia and surrounding tissue affects shear wave phase behavior, with detectable phase shifts observed in

both FDTD simulations and ex-vivo experiments.
However, the observed phase shift in the ex-vivo experiment extended across 5 mm, which is substantially greater than the fascial thickness of about 0.5 mm observed in B-mode imaging. This discrepancy is likely attributed to the limited spatial resolution associated with the relatively low shear wave frequency (78 Hz) used in this study.
Anatomical differences between the chicken breast model and human fascia should also be considered. For instance, the human fascia lata is approximately 1 mm thick [12], notably thicker than the chicken fascia used here. Such structural differences may influence shear wave propagation and affect the applicability of these findings to human tissue.
Furthermore, the shear wave velocity in the myofascial tissue after HR was 4.11 ± 1.60 m/s, consistent with previously reported values (3.8–5.1 m/s) [11]. However, this may have been influenced by the observed phase shifts. Future work should aim to decouple these effects to improve elasticity assessment.
Finally, while the FDTD simulation modeled interfacial adhesion using simplified coefficients α_x and α_z, biological interfaces are more complex. Future simulations should incorporate additional physical parameters, such as friction and interfacial fluid viscosity, to better represent tissue interactions. This may offer deeper insights into how HR improves fascial mobility in clinical contexts.

Conclusion

This study demonstrated that the interfacial adhesion between the fascia and adjacent tissues significantly affects the propagation characteristics of shear waves. Specifically, both FDTD-based numerical simulations and ex-vivo experiments using chicken breast confirmed that reduced interfacial adhesion leads to a distinct phase shift in shear wave propagation. These findings suggest that shear wave elastography has the potential to evaluate fascial adhesion. Future work may focus on both shear wave velocity and waveform morphology to enable a deeper and more comprehensive assessment of tissue characteristics.

Acknowledgements

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References

[1] D. Marić et al. "FASCIA–THE FORGOTTEN TISSUE". In: *Exercise and quality of life* 6.1 (2014), p. 31.

[2] H. Kanamoto et al. "Effect of Ultrasound-Guided Hydrorelease of the Multifidus Muscle on Acute Low Back Pain". In: *Journal of Ultrasound in Medicine* 40.5 (2021), pp. 981–987.

[3] K. Shiwaku et al. "Fascial Ultrasound-Guided Injection: Where Do We Really Inject?" In: *Cureus* 17.2 (2025).

[4] K. Shiwaku et al. "Biomechanical effects of fascial hydrorelease: a cadaveric study". In: *BMC Musculoskeletal Disorders* 26.1 (2025), p. 306.

[5] L.-R. Liao et al. "Reliability of shear-wave elastography in assessing the stiffness of the nuchal fascia and the thickness of upper cervical muscles". In: *Medical Engineering & Physics* 120 (2023), p. 104039.

[6] T. L.-W. Chen et al. "Ultrasound elastographic assessment of plantar fascia in runners using rearfoot strike and forefoot strike". In: *Journal of Biomechanics* 89 (2019), pp. 65–71.

[7] J. M. Fabrikant and T. S. Park. "Plantar fasciitis (fasciosis) treatment outcome study: plantar fascia thickness measured by ultrasound and correlated with patient self-reported improvement". In: *The Foot* 21.2 (2011), pp. 79–83.

[8] N. Tano et al. "Continuous Shear Wave Elastography for Liver Using Frame-to-Frame Equalization of Complex Amplitude". In: *Ultrasonic Imaging* 46.3 (2024), pp. 197–206.

[9] M. Tabaru et al. "Examination of rapid adjustment system based on screen score obtained using continuous shear wave elastography". In: *Journal of Medical Ultrasonics* 51.3 (2024), pp. 407–418.

[10] R. Koda et al. "Application of continuous shear wave elastography method with multiple frequency selection to liver viscoelasticity measurement". In: *Japanese Journal of Applied Physics* 63.4 (2024), 04SP82.

[11] G. Schillizzi et al. "Evaluation of plantar fasciopathy shear wave elastography: a comparison between patients and healthy subjects". In: *Journal of Ultrasound* 24 (2021), pp. 417–422.

[12] S. Otsuka et al. "Investigation of the association between human fascia lata thickness and its neighboring tissues' morphology and function using B-mode ultrasonography". In: *Journal of Anatomy* 239.5 (2021), pp. 1114–1122.