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Passive *in vivo* elastography from skeletal muscle noise

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Measuring the *in vivo* elastic properties of muscles (e.g., stiffness) provides a means for diagnosing and monitoring muscular activity. The authors demonstrated a passive *in vivo* elastography technique without an active external radiation source. This technique instead uses cross correlations of contracting skeletal muscle noise recorded with skin-mounted sensors. Each passive sensor becomes a virtual *in vivo* shear wave source. The results point to a low-cost, noninvasive technique for monitoring biomechanical *in vivo* muscle properties. The efficacy of the passive elastography technique originates from the high density of cross paths between all sensor pairs, potentially achieving the same sensitivity obtained from active elastography methods. © 2007 American Institute of Physics. [DOI: 10.1063/1.2737358]

Measuring the *in vivo* elastic properties of muscles (e.g., stiffness) provides a means for diagnosing and monitoring muscular activity: muscles typically become “harder” during contraction due to physiological changes.¹ Standard elastography imaging techniques estimate soft tissue (e.g., skeletal muscle and breast) stiffness using propagating shear waves in the human body generated by an active external source (e.g., indentation techniques and ultrasonic radiation force).^{2,3} Instead, we demonstrated a passive (i.e., without active source) *in vivo* elastography technique using cross correlations of contracting skeletal muscle noise recorded with skin-mounted accelerometer sensors (Fig. 1).

During voluntary contractions, low frequency (<100 Hz) and continuous surface mechanical oscillations, also called physiological tremors or muscle noise, occur naturally along the transverse direction of the muscle fibers due to the shortening of the actomyosin filaments along the muscle axis.^{1,4,5} In analogy to electromyography measurements, recordings of these transverse oscillations are called surface mechanomyograms^{5,6} (S-MMGs) (Fig. 1). Given their generation mechanism, S-MMGs are dominated by natural random shear waves propagating with a polarization parallel to the muscle fibers. Furthermore, since most biological materials are nearly incompressible, propagating longitudinal waves travel orders of magnitude faster than shear waves and thus are not typically captured in measurements of low frequency excitations.

This passive elastography technique is based on the cross correlations of S-MMG, interpreted as a random diffuse wave field.^{7,8} An example of a random diffuse wave field in a homogeneous medium is the field produced by the spatial and temporal average of an isotropic distribution of random sources, with spatial-delta-correlated amplitudes. Based on theoretical and experimental results,⁷⁻¹² the relationship in the Fourier domain between the cross-correlation function $C_{12}(\omega)$ of diffuse wave fields recorded by receivers 1 and 2 and the actual Green’s function (or mechanical impulse response), $G_{12}(\omega)$, between these receivers is

$$\langle C_{12}(\omega) \rangle \equiv \frac{\beta(\omega)}{i\omega} (G_{12}(\omega) - G_{21}^*(\omega)), \quad (1)$$

where the symbol “*” stands for complex conjugation and β represents the power spectrum of the diffuse field excitation. Thus, for a uniform noise source distribution or a fully diffuse noise field, the time-domain noise cross-correlation function is a time-symmetric function. Hence, when computed between a skin-mounted sensor pair, the S-MMG cross correlation is expected to yield a coherent wave form equal to the Green’s function between the sensors (up to an amplitude factor), in this instance the local shear wave arrival. This correlation process works by accumulating coherent contributions over time from various natural shear noise sources propagating successively between both sensors. This effectively transforms each passive skin-mounted sensor into a virtual *in vivo* shear wave source.

We conducted a passive elastography study to measure the change of the viscoelastic properties of the *vastus lateralis* muscle (quadriceps knee extensor) of a healthy 28 year old male volunteer without history of musculoskeletal disorder.

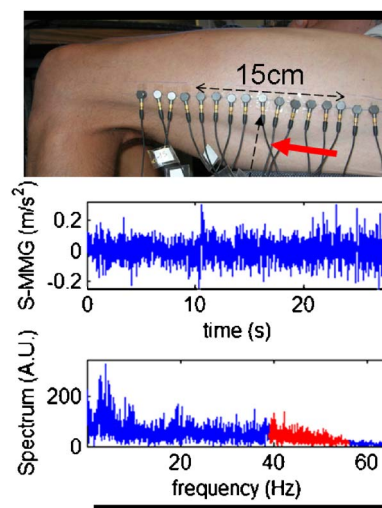


FIG. 1. (Color online) Experimental setup and muscle vibration noise [surface mechanomyogram (S-MMG)] recorded from the *vastus lateralis* during isometric voluntary contractions with a skin-mounted accelerometer. Weight load=10 lb.

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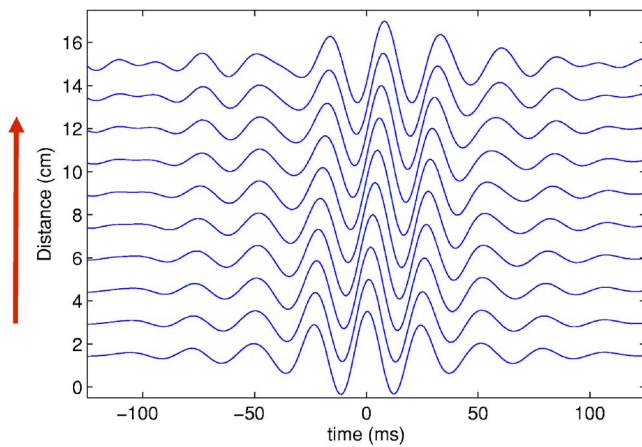


FIG. 2. (Color online) Coherent propagating shear wave profile extracted from stacked cross correlations of muscle noise recorded [in the frequency band (40–55 Hz) only, see Fig. 1, red line] between sensor pairs located on the middle third of the *vastus lateralis* muscle vs increasing sensor separation distance.

ders. Sixteen miniature accelerometers (PCB® A352C65, weight=1.5 g, sensitivity=100 mV/g) were skin mounted along the *vastus lateralis* muscle using a double-sided tape and equally spaced by 1.5 cm ($\pm 20\%$) (Fig. 1). S-MMG were simultaneously recorded on all accelerometers during $T=1$ min long voluntary isometric contraction sequences. Following the protocol described in Ref. 13, increasing weight loads (5, 7.5, 10, and 12.5 lb) were successively applied to the subject's ankle using a pulley system to ensure that the resulting force remained nearly normal to the leg with a constant knee-joint angle of 90° (Fig. 1). For each weight load, the subject rested for a period of 2 min between each of the three contraction sequences, which remained below the point of fatigue.

The *vastus lateralis* is a unipennate muscle: all the muscle fibers are on the same side of the tendon and run mostly parallel in the sagittal plan of the thigh. The pennation angle (i.e., the angle at which the muscle fibers attach with respect to the line of pull of the tendon) is typically smaller than 20° .¹⁴ Given this relatively simple muscle architecture and the low frequencies of the S-MMG records, we assumed that the propagation of shear waves with a polarization parallel to the muscle fibers was one dimensional to first order. Furthermore, to avoid the more anisotropic tendinous extremities and overlapping of other quadriceps muscles, only the 11 accelerometers mounted on the middle third of the *vastus lateralis* were used (Fig. 1).

The recorded S-MMG signals typically display a characteristic spectral shift towards higher frequencies for increasing weight load due to an increase in firing rates of motor units and recruitment of fast twitch motor units.^{5,15} Hence, the S-MMG wave forms were filtered in the high-frequency band (40–55 Hz) (Fig. 1, red line), where the activation of the fast twitch muscle fibers dominates the S-MMG generation mechanism⁵ and more likely results in shear waves propagating along the muscle fibers. At lower frequencies (below 10 Hz), it was unclear whether the muscle surface vibrations originated from locally propagating shear waves or whole limb vibrations (e.g., due to breathing, leg shaking, etc.). Furthermore, using higher frequencies allows for better estimation of the shear wave phase velocity, given the short propagation distances and the increased reliability of the lo-

cal plane wave assumption of uniform shear wave propagation.¹⁶ The continuous S-MMG time series were further truncated to four times the standard deviation. This simple nonlinear process reduces the influence of high-amplitude noise events while preserving the overall phase information of the time series.¹⁷

Assuming uniform shear wave propagation over the 15 cm middle third of the *vastus lateralis* muscle, the computed cross correlations of processed S-MMGs records were spatially averaged over all equidistant sensor pairs. This spatial averaging improved the signal-to-noise ratio of the averaged extracted coherent shear wave form. The stacked averaged cross correlations for increasing sensor separation distance clearly display a propagating shear wave with a polarization parallel to the muscle fibers [Fig. 1(b)], resulting from coherent muscle vibrations. The emergence rate of this coherent wave form is proportional to \sqrt{TB} if the medium remains stationary. Given the time asymmetry of the cross-correlation wave forms,^{10,12,17} this coherent shear wave appears to propagate down the *vastus lateralis* fibers (see Fig. 1, red arrow). Hence, most of the coherent shear noise sources seem to originate from the upper thigh portion where the *vastus lateralis* cross section is the largest and contains the most active muscle fibers. Similar time asymmetry of the cross-correlation wave forms has been observed in acoustic and seismic studies as a result of the nonuniform distribution of noise sources.^{7,9,11}

For a given weight load and viscoelastic model of the muscle fibers,^{2,18} the absolute shear modulus μ (ratio of transverse stress to strain) and dynamic shear viscosity ν (ratio of shear stress to velocity gradient) can be estimated from the measured shear wave phase velocity V once a viscoelastic model of the muscle fibers has been selected. Here, a simple Voigt model is used since it was shown to be well suited to estimate the viscoelastic response of soft tissues for low frequencies excitations (0–500 Hz).^{19,20} The relationship between the phase velocity $V(f)$ of a plane shear wave and the mechanical properties of a homogeneous isotropic viscoelastic Voigt medium is given by

$$V(f) = \sqrt{\frac{2(\mu^2 + 4\pi^2 f^2 \nu^2)}{\rho(\mu + \sqrt{\mu^2 + 4\pi^2 f^2 \nu^2})}}, \quad (2)$$

where $\rho=1100$ kg/m³ is the mass density (assumed to be constant). Both parameters μ (kPa) and ν (Pa s) were estimated from the best fit between the Voigt model [Eq. (2)] and the measured frequency dispersion $V(f)$, in a least-mean square sense in the frequency band (40–55 Hz) (Ref. 19) (see Fig. 3). The shear wave phase velocity dispersion $V(f)$ (change of velocity versus frequency) is estimated by (1) estimating the time-frequency distribution of each S-MMG cross-correlation wave forms (Fig. 2), using a continuous Morlet wavelet transform,²¹ and then (2) computing the slope of the shear wave arrival time for increasing sensor separation distances.

The passive estimates of μ and ν appear highly correlated with weight load and hence muscle contraction level (Fig. 4), in agreement with previously reported active isometric experiments.^{2,3,18} The relatively high values of μ and ν may result from the strong contraction effort, the specific muscle fibers' composition, and the thin adipose layer of the subject.^{5,18} The variations in experimental errors between different weight loads (see Figs. 3 and 4) may be due (1) to

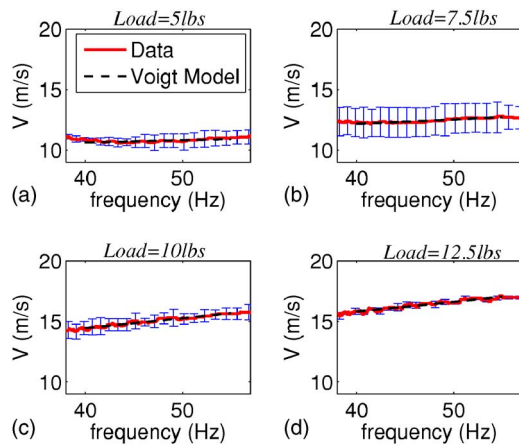


FIG. 3. (Color online) Measured mean shear wave phase velocity $V(f)$ during isometric contractions using increasing weight load (from 5 to 12.5 lb) in the frequency band (40–55 Hz) (red line). Dashed black lines represent least-mean-square fit to the data according to the Voigt model [Eq. (S2)]. The corresponding best-fit values of the shear modulus and shear viscosity parameters are shown in Fig. 4. Error bars represent one standard deviation for three contraction sequences (blue line).

uncontrolled fluctuations in the tension of the pulley system and also (2) to a physiological origin: the possible transition at intermediate loads (e.g., here 7.5 lb) in the proportion of various types of activated fast-twitch muscle fibers^{5,15} (e.g., fast-oxidative type IIA versus fast-glycolytic type IIB) which could influence the effective global viscoelastic properties of the whole muscle. Furthermore, the measured shear wave velocities may also be overestimated due to the diffraction effects caused by the skin interfaces and the impossibility of dissociating the displacements due to the longitudinal waves,

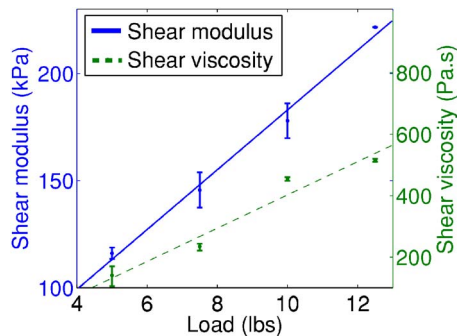


FIG. 4. (Color online) Variations of the mean shear modulus (blue, left axis) and shear viscosity (green, right axis) of the muscle fibers for increasing weight load during isometric contractions. Error bars indicate one standard deviation over three contraction sequences. Straight lines display a linear fit to the data.

shear waves, and coupling terms in the very near field of the noise sources regions.²² Estimating the size of the very near-field region would require a precise knowledge of the muscle contraction generation mechanism, i.e., the location and directionality of the S-MMG vibrations sources, which is a topic still under investigation.²³

The passive elastography technique presented in this work, which solely uses muscle noise recordings, clearly points toward a low-cost, noninvasive technique for monitoring biomechanical *in vivo* muscle properties. Each skin-mounted passive sensor becomes a virtual *in vivo* shear wave source. Higher spatial resolution, as is currently obtained from active elastography methods,^{2,3} could be achieved with this passive technique by using currently available dense surface sensor arrays (e.g., microsensors or laser interferometry).

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