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Shear elastic modulus can be used to estimate an index of individual muscle force during a submaximal isometric fatiguing contraction

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Bouillard K, Hug F, Guével A, Nordez A. Shear elastic modulus can be used to estimate an index of individual muscle force during a submaximal isometric fatiguing contraction. *J Appl Physiol* 113: 1353–1361, 2012. First published September 13, 2012; doi:10.1152/japplphysiol.00858.2012.—The present study was designed to determine whether fatigue alters the ability to estimate an index of individual muscle force from shear elastic modulus measurements (*experiment I*), and to test the ability of this technique to highlight changes in load sharing within a redundant muscle group during an isometric fatiguing task (*experiment II*). Twelve subjects participated in *experiment I*, which consisted of smooth linear torque ramps from 0 to 80% of maximal voluntary contraction (MVC) performed before and after an isometric fatigue protocol, beginning at 40% of MVC and stopped when the force production dropped below 30% of MVC. Although the relationships between modulus and torque were very similar for pre- and postfatigue [root mean square deviation ($RMS_{deviation}$) = 3.7 ± 2.6% of MVC], the relationships between electromyography activity level and torque were greatly altered by fatigue ($RMS_{deviation}$ = 10.3 ± 2.6% of MVC). During the fatiguing contraction, shear elastic modulus provided a significantly lower $RMS_{deviation}$ between measured torque and estimated torque than electromyography activity level (5.7 ± 0.9 vs. 15.3 ± 3.8% of MVC). *Experiment II* performed with eight participants consisted of an isometric knee extension at 25% of MVC sustained until exhaustion. Opposite changes in shear elastic modulus were observed between synergists (vastus medialis, vastus lateralis, and rectus femoris) of some participants, reflecting changes in load sharing. In conclusion, despite the fact that we did not directly estimate muscle force (in Newtons), this is the first demonstration of an experimental technique to accurately quantify relative changes in force in an individual human muscle during a fatiguing contraction.

supersonic shear imaging; electromyography; elastography

NEUROMUSCULAR FATIGUE IS DEFINED as an exercise-induced reduction in the muscle's capability to generate force (2, 16). This is a progressive phenomenon that includes important physiological changes that occur before and during the mechanical failure (18). The force produced by individual motor units decreases (14), leading to an increase in the central drive required for maintaining a constant force (15, 27). The relationship between muscle activity level assessed by surface electromyography (EMG) and force is thus altered in a way that depends on the task (12). Consequently, as initially suggested by experimental work (6, 11, 27) and recently confirmed by a modeling approach (10), EMG cannot be used to estimate individual muscle force during a sustained submaximal fatiguing exercise. Since information about muscle force is crucial to study muscle coordination strategies during tasks where neuromuscular fatigue occurs, alternative approaches are necessary.

Considering that direct measurements of individual muscle force cannot be performed noninvasively, numerous modeling approaches have been proposed for nonfatiguing tasks (5, 13) since the first works published in the 1970s (33). However, due to muscle redundancy (i.e., in most tasks, there are more muscles involved than mechanical degrees of freedom), these models remain invalidated (33). A recent study demonstrated that shear elastic modulus measurements performed using an ultrasound elastography technique named supersonic shear imaging (SSI) (1) can provide an accurate index of individual muscle force during nonfatiguing isometric contractions (4). In this latter study, tasks involving only one muscle (i.e., index abduction and little finger abduction) were studied. It was shown that the shear elastic modulus was linearly related to torque or muscle force during isometric linear ramp contraction [from 0 to 60% of maximal voluntary contraction (MVC)], leading to a root mean square (RMS) error of 1.4% of MVC between measured and estimated torque. In contrast to surface EMG measurements that are influenced by several electrophysiological parameters, the shear elastic modulus is a mechanical property. Thus one would expect that the alteration of motor unit discharge characteristics associated with fatigue does not influence the shear elastic modulus. However, putative changes in mechanical properties of muscle occurring with fatigue (31, 34, 35) could preclude the ability to provide an index of muscle force during a fatiguing contraction.

The aim of the present study was twofold. The first aim (*experiment I*) was to determine whether fatigue alters the ability to provide an index of individual muscle force from shear elastic modulus measurements during an isometric fatiguing contraction. For this purpose, it was necessary to investigate a task involving a muscle without synergist, i.e., a task in which the measured torque is produced by only one muscle. Thus we studied little finger abduction, because this movement implies mostly the abductor digiti minimi (26), preventing putative compensations between synergistic muscles. The second aim (*experiment II*) was to test this technique in a muscle group to document change in load sharing during an isometric fatiguing task. To this end, we studied knee extension for which between-muscle compensations have been indirectly observed (i.e., using surface EMG) during low-force fatiguing contractions (24).

MATERIALS AND METHODS

Participants

Twelve healthy volunteers (4 women and 8 men; 24.8 ± 2.2 yr, 173.3 ± 4.6 cm, 58.3 ± 7.2 kg for women; and 23.1 ± 2.2 yr, 179.4 ± 5.8 cm, 74.0 ± 4.9 kg for men) participated in *experiment I*. Eight healthy men (27.3 ± 5.4 yr, 178.9 ± 6.8 cm, 74.6 ± 11.6 kg)

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participated in *experiment II*. All of the participants were informed of the possible risk and discomfort associated with the experimental procedures before giving their written consent to participate. The local ethics committee of Nantes Ouest IV approved the study (CPP-MIP-001), and all of the procedures conformed to the Declaration of Helsinki (last modified in 2004).

Measurements

Ergometer. For *experiment I*, a home-made ergometer was used to measure the torque produced by the little finger abduction, as previously described (4). As this torque is only produced by one muscle, it can be considered to represent an individual muscle force that we intend to predict using shear elastic modulus measurements. Briefly, the participants were seated with their right elbows flexed to 120° (180° corresponds to the full extension of the elbow), and the pronated forearm was supported by a platform; all fingers were extended with the palm facing down. The hand and *fingers 2–4* were immobilized with Velcro straps to prevent any movement during the contractions (Fig. 1). The lateral side of the little finger was in contact with a rigid interface, with the proximal interphalangeal joint aligned with the force sensor (SML-50, Interface).

For *experiment II*, an isokinetic dynamometer (Biodex 3 medical, Shirley, NY) was used to measure knee angle and torque of the right leg. Briefly, participants sat on the dynamometer with the trunk and their right leg flexed at 90°. The torso and waist were strapped to the dynamometer chair to ensure that the participant's body position did not change throughout the experiment. The axis of the dynamometer was aligned with the presumed axis of rotation of the knee.

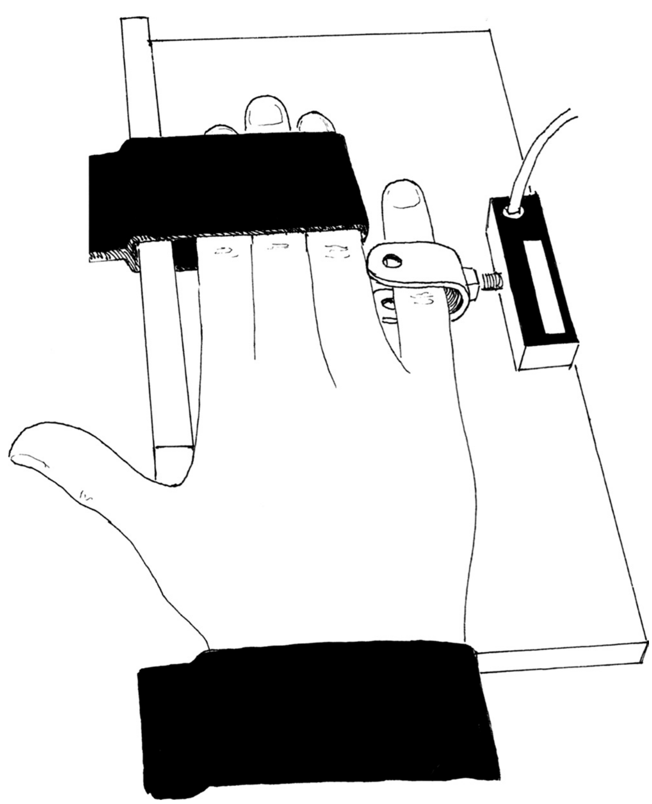

Fig. 1. Experimental setup for *experiment I*. The right pronated forearm was supported on a platform, and all fingers were extended with the palm facing down. The hand and *fingers 2–4* were immobilized with Velcro straps to prevent any movement and compensation during contractions. The little finger was in contact with a rigid interface, with the proximal interphalangeal joint aligned with the force sensor.

EMG. For *experiment I*, bipolar EMG recordings were obtained from abductor digiti minimi, flexor digitorum superficialis, and extensor digitorum. For *experiment II*, bipolar EMG recordings were obtained from vastus medialis, vastus lateralis, rectus femoris, semitendinous, and biceps femoris (long head). For both experiments, we used dry-surface electrodes (1-cm interelectrode distance; Delsys DE 2.1, Delsys, Boston, MA) located over the muscles' belly. The electrodes were placed longitudinally with respect to the underlying muscle fiber's direction, previously determined by ultrasound image for abductor digiti minimi, vastus lateralis, and vastus medialis and with respect to the direction of muscle shortening for the others. A reference electrode was placed at the level of the left wrist (*experiment I*) and the right patella (*experiment II*).

Before electrode placement, the skin was cleaned with a mixture of alcohol and ether to minimize impedance. Signals were amplified (×1,000), band-pass filtered (6–400 Hz), and digitized at a sampling rate of 4,000 Hz (Bagnoli 16, Delsys, Boston, MA). To avoid compensation with the wrist (*experiment I*), a visual feedback of the EMG activity of the flexor digitorum superficialis and the extensor digitorum was continuously displayed on a monitor placed in front of the participants and the experimenter. Participants were instructed to keep these muscles silent, and it was the case for all of the contractions performed.

Shear elastic modulus measurements. An Aixplorer ultrasonic scanner (version 4.2, Supersonic Imagine, Aix en Provence, France) was used in the SSI mode (musculoskeletal preset). As described by Bercoff et al. (1), the system consisted of a transient and remote mechanical vibration generated by radiation force induced by a focused ultrasonic beam (i.e., "pushing beam"). Each pushing beam generated a remote vibration that resulted in the propagation of a transient shear wave. Subsequently, an ultrafast echographic imaging sequence was performed to acquire successive raw radio-frequency data at a very high frame rate (up to 20 kHz). A one-dimensional cross correlation of successive radio-frequency signals was used to determine the shear wave velocity (V_s) along the principle axis of the probe using a time-of-flight estimation. Then, considering a linear (1, 7, 17) and purely elastic (8, 17, 30) behavior, a shear elastic modulus (μ) was calculated using V_s as follows:

$$\mu = \rho V_s^2 \tag{1}$$

where ρ is the density of muscle (1,000 kg/m^3).

The probe was aligned carefully with the direction of muscle fibers for abductor digiti minimi, vastus lateralis, and vastus medialis and with the direction of muscle shortening for rectus femoris. Maps of the shear elastic modulus were obtained at 1 Hz with a spatial resolution of 1 × 1 mm (Fig. 1). The shear elasticity map was chosen to be as large as possible, depending on the muscle depth/thickness (about from 1.0 × 1.5 cm, for the abductor digiti minimi, and 1.5 × 1.5 cm for the quadriceps muscles), to obtain a representative averaged shear elastic modulus value (3, 4).

In *experiment II*, after warm-up and before the MVC, the best location of the probe for each muscle was determined using the echographic image as the region with great-enough muscle thickness, avoiding aponeurosis and tendon, and respecting the muscle fiber direction for vastii. These locations were marked on the skin by waterproof felt-tip pen, assuring that the marks resist to the ultrasound gel. Then, during the experiment, the probe was moved from a mark to another with indispensable check of the ultrasound image (B mode). As contractions were isometric, negligible movements of the muscles (relative to the skin) were expected (and observed).

Protocol

Experiment I. After a familiarization performed on a separate day, the experimental protocol was divided into two sessions separated by 48 h: one was devoted to the SSI measurements, and the other one was devoted to the EMG recordings (randomly assigned). For each experimental session, participants first performed three isometric MVC lasting 3 s, separated by 2 min of recovery. The maximum torque was considered the best performance and was used to normalize subsequent submaximal contractions. To account for putative early changes in musculo-tendinous mechanical properties due to creep phenomenon (28), a conditioning was performed before actual measurements and consisted of a smooth linear torque ramp of 15 s from 0 to 40%. Participants were then asked to perform two smooth linear torque ramps (referred to as "ramp contraction" in this report) of 30 s from 0 to 80% of the previously determined MVC. This range corresponds approximately to the maximal range of torque that can be developed without saturation of the SSI measurement (i.e., 266 kPa). After a 3-min recovery period, participants performed a submaximal isometric fatiguing contraction [referred to as "time-to-exhaustion" (T_{lim}) in this report] that consisted of maintaining their force production at 40% of MVC for as long as possible. Because we observed, during pilot experiments, uncontrollable compensations with wrist adductors below this intensity, the fatigue protocol was stopped when the force production was reduced below 30% of MVC during more than 5 s. Immediately after the end of the T_{lim}, participants performed a new ramp contraction (of 30 s from 0 to 80% of MVC). Finally, this ramp contraction was immediately followed by a MVC to verify that a decrease in MVC occurred and thus to confirm the presence of neuromuscular fatigue. During each contraction, depending on the session, the shear elastic modulus or surface EMG was recorded and synchronized with torque measurements. To control the torque during "ramp contractions" and T_{lim}, a visual feedback was displayed on a monitor placed in front of the participants.

Experiment II. First, participants performed two MVC lasting 3 s for both knee extension and knee flexion, separated by 2 min of recovery. The maximum torque was considered the best performance and was used to normalize subsequent submaximal contractions. Second, to test the repeatability of muscle shear elastic measurement without fatigue, participants were asked to perform six isometric knee extensions of 15 s at 25% of the previously determined MVC, separated by 1 min of recovery. During each contraction, the shear elastic modulus of one of the knee extensors (vastus medialis, vastus lateralis, or rectus femoris) was measured in random order, such that each muscle was studied twice. Third, after a 5-min recovery period, participants performed a submaximal isometric fatiguing contraction that consisted of maintaining 25% of MVC for as long as possible. In contrast to *experiment I*, where we aimed at observing a decrease in torque, it was important to maintain the torque constant during this submaximal exercise to better interpret putative changes in load sharing. So, this fatigue protocol was stopped when the force decreased by >5% from the required target during 3 s. As the participants were instructed to maintain the torque, a decrease by >5% can be considered as a task failure due to fatigue (24). During this contraction, the ultrasonic probe was alternatively placed (same order as the one used for testing repeatability) over the three quadriceps muscles for 5 s (i.e., five shear elastic modulus measurements) until the end of the contraction. Surface EMG signals of vastus medialis, vastus lateralis, or rectus femoris were recorded and synchronized with torque and shear elastic modulus measurements.

Data Analysis

Data processing was performed using MATLAB scripts (Mathworks, Natick, MA). For both ramp contractions (*experiment I*) and submaximal fatiguing contractions (*experiments I* and *II*), the RMS of the EMG signal (EMG-RMS) was calculated using a 1-s time-averaging window. EMG-RMS values were normalized to the maximal value achieved over 150-ms during MVC contractions to limit signal cancellation (23).

Maps of the shear elastic modulus were exported from the software (version 4.2, Supersonic Imagine, Aix en Provence, France) in "mp4" format, sequenced in "jpeg". A region of interest (ROI) was defined in each elasticity map (i.e., obtained each second) as the largest muscular region available, avoiding aponeurosis, tendon, and bone ($\approx$1.5 cm^2 for abductor digiti minimi and $\approx$1.8 cm^2 for quadriceps muscles; Fig. 2).

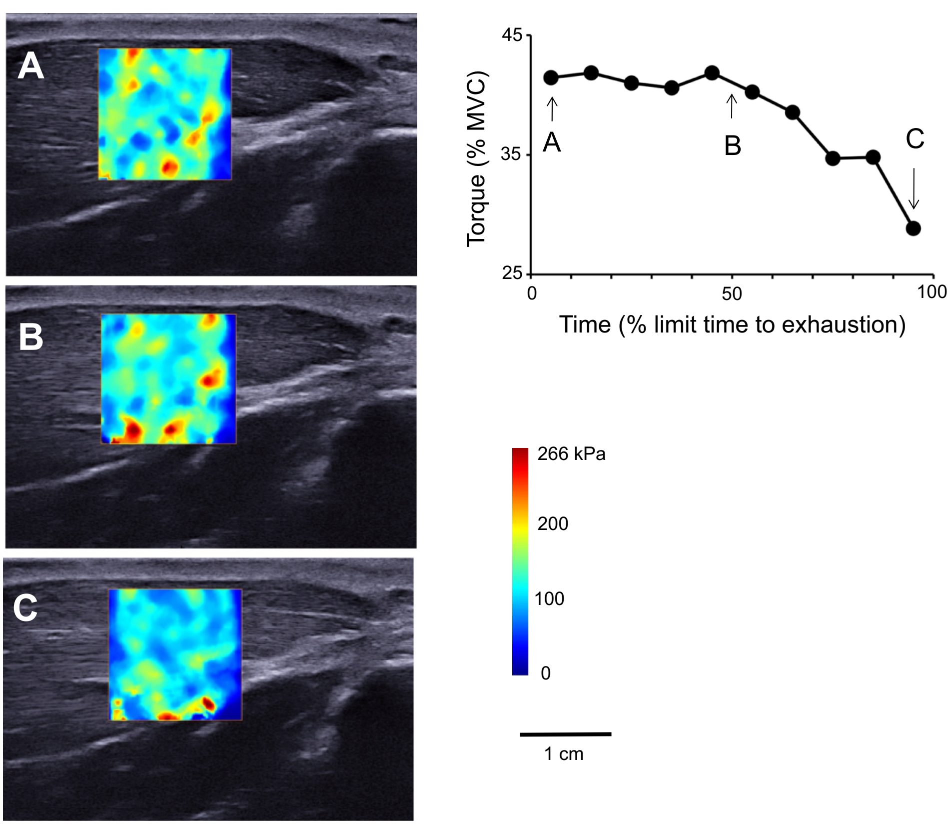

Fig. 2. Typical example of shear elastic modulus measurements of the abductor digiti minimi during the fatiguing contraction (*experiment I*). The start (*A*), middle (at ~50% of the total duration; *B*), and end of the time to exhaustion (T_{lim}) contraction (*C*) are shown. The colored region represents the shear elasticity map, with the scale to the *right* of the figure. To obtain a representative value, the shear elastic modulus (in kPa) was averaged over the greatest muscular area, avoiding aponeurosis. MVC, maximal voluntary contraction.

A mean shear elastic modulus value was thus calculated over this ROI. Due to limitations of the current version of the ultrasonic scanner, shear elastic modulus measurements saturated at 266 kPa, limiting the range of analysis for most of the participants. This value was never reached for submaximal fatiguing contractions. For "ramp contractions", once any value (corresponding to a 1 mm × 1 mm subregion) within the ROI reached 266 kPa, there was no further analysis at that force and any higher force.

For *experiment I*, the EMG-RMS/torque and shear elastic modulus/torque relationships obtained for "ramp contractions" were fitted to a linear model (4, 34) to calculate slope (*a*) and *y*-intercept (*b*). To assess the fit for the first ramp, the coefficient of determination (R^2) and the error of estimation [i.e., RMS deviation (RMS_{devR1})] were calculated for both EMG and SSI measurements (4). The change of the EMG-RMS/torque and shear elastic modulus/torque relationships with fatigue was assessed by calculating of a RMS deviation between the first and the third ramp ($RMS_{devR1\text{-}R3}$), respectively, achieved before and after T_{lim}.

Previously determined "*a*" and "*b*" coefficients during the first ramp contraction were used to estimate torque during the T_{lim} from both EMG-RMS and shear elastic modulus values. T_{lim} was splitted into 10 equal time-windows; the averaged values of both measured and estimated torques were calculated for each. RMS deviation ($RMS_{devTlim}$) was then calculated to quantify the error of estimation throughout the contraction. Finally, median frequency (MF) of the EMG signal was computed during T_{lim} as an indirect evaluation of neuromuscular fatigue (29).

For *experiment II*, to test the repeatability, an average value of shear elastic modulus over the entire contraction was calculated for each 15-s contraction. For the submaximal fatiguing contraction, each series of five successive shear elastic modulus measurements was averaged to obtain a representative value. Thus we obtained 4–11 averaged values per muscle, depending on the duration of the contraction, and thus on the participant.

Statistical Analysis

For *experiment I*, we first determined the intrasession repeatability of the EMG-RMS/torque and shear elastic modulus/torque relationships of the two "ramp contractions" (R1 vs. R2) performed before the T_{lim} (i.e., without fatigue). For this purpose, the equations of the linear regressions were used to estimate EMG-RMS or shear elastic modulus values at 15, 30, and 50% of MVC (50% of MVC being the maximal value of torque common to all subject for the shear elastic modulus/torque relationship, due to shear elastic modulus measurements saturation). For each of the three torque levels, the repeatability was assessed by the intraclass correlation coefficient (ICC), standard error of measurement (SEM) and coefficient of variation (CV), as recommended by Hopkins (19).

For all data, normality testing consistently passed the Kolmogorov-Smirnov normality test (Statistica V10, Statsoft, Maison-Alfort, France), and thus the values are reported as a mean ± SD throughout the text and Figs. 4 and 5. To verify that T_{lim} induced neuromuscular fatigue, MVC obtained before and after the T_{lim} were compared using a paired *t*-test. Also, a two-way repeated-measures ANOVA [random factor: participant; between-subject factor: method (torque measured during SSI session and torque measured during EMG session) and time (10 equal time-windows)] was used to verify that the torque decreased during T_{lim} contraction. Finally, for EMG session, an additional one-way repeated-measures ANOVA [random factor: participant; between-subject factor: time (10 equal time-windows)] was used to test the main effect of time on the MF during the T_{lim} contraction.

The effect of fatigue on the ability to estimate torque was tested for both ramp and T_{lim} contractions. First, a paired *t*-test was used to compare the $RMS_{devR1\text{-}R3}$ between EMG and SSI. Subsequently, a two-way repeated-measures ANOVA [random factor: participant; between-subject factor: methods (EMG and SSI) and time (10 equal time-windows)] was used to test the effects of the method and the time on the $RMS_{devTlim}$. Post hoc analyses were performed using Tukey's method. The level of significance was set as $P < 0.05$.

For *experiment II*, the intrasession repeatability of the shear elastic modulus of each leg extensor muscle at 25% of MVC (measured during the six 15-s contractions) was assessed by the ICC, the SEM, and the CV. No statistical analysis was performed due to high interindividual variability of change in shear elastic modulus during the fatiguing submaximal exercise. Indeed, a lack of significant difference would not have indicated an absence of change, as we observed opposite changes between subjects in some cases.

RESULTS

Experiment I

Accuracy and repeatability of torque estimation during ramp contractions. Among the 36 "ramp contractions" (3 ramps × 12 participants), the saturation level of the shear elastic modulus at 266 kPa was reached 29 times before the end of the ramp. Consequently, "ramp contractions" were analyzed up to 58.3 ± 14.2% of MVC.

Figure 3 depicts an example of a typical relationship obtained between EMG-RMS and torque and between shear elastic modulus and torque. Mean R^2 of the linear regression associated to the first ramp performed without fatigue was 0.947 ± 0.020 (range: 0.912–0.981) for EMG-RMS/torque relationships and 0.982 ± 0.009 (range: 0.968–0.992) for shear elastic modulus/torque relationships. The RMS_{devR1} values associated with this fitting of the first ramp was 3.9 ± 1.3% of MVC (range: 1.6–5.6% of MVC) for EMG-RMS/torque relationship and 1.9 ± 0.9% of MVC (range: 0.7–3.1% of MVC) for shear elastic modulus/torque relationships.

Results concerning the assessment of repeatability between the two ramp contractions performed without fatigue are shown in Table 1. For both techniques (EMG and SSI), ICC values were high (from 0.957 to 0.985) and SEM values low (from 1.2 to 4.0% of the maximal EMG-RMS value for EMG and from 4.5 to 8.2 kPa for SSI). In most cases, the CV was lower than 10% (except for the EMG at 15% of the MVC),

Table 1. *Repeatability of both EMG-RMS and shear elastic modulus estimation during the two first ramps (without fatigue) at three different contraction levels*

	15% of MVC			30% of MVC			50% of MVC		
	ICC	SE	CV, %	ICC	SE	CV, %	ICC	SE	CV, %
EMG-RMS	0.979	1.3	20.6	0.980	2.3	9.1	0.975	4.0	9.4
Shear elastic modulus	0.957	4.5	9.8	0.985	4.7	4.9	0.981	8.3	4.7

Standard error of measurement (SE) is expressed in % for electromyography (EMG)-root mean square (RMS) and in kPa for shear elastic modulus. MVC, maximal voluntary contraction; ICC, intraclass coefficient correlation; CV, coefficient of variation.

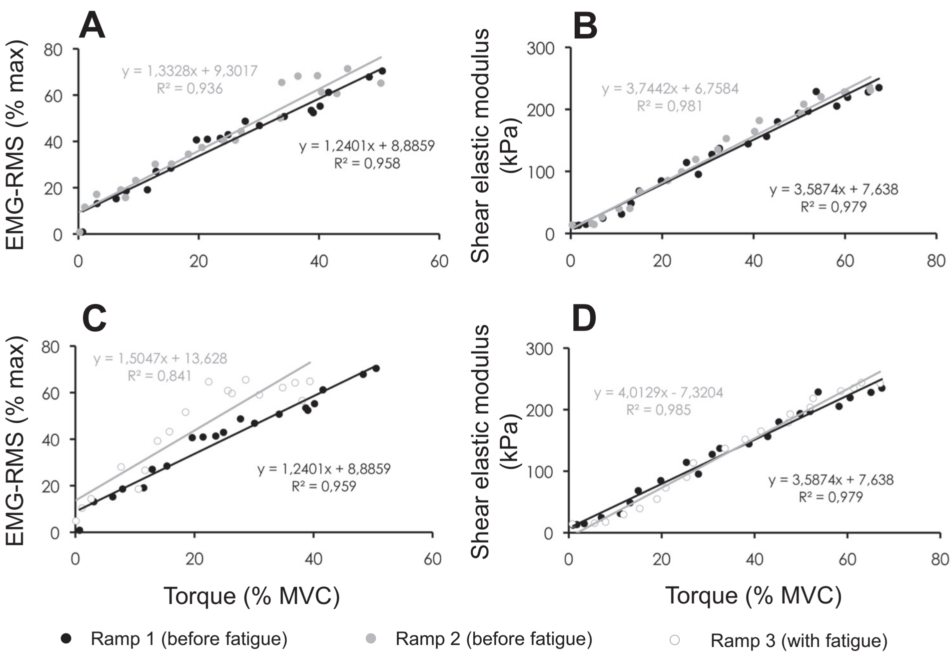

Fig. 3. Typical electromyography (EMG)/torque and shear elastic modulus/torque linear relationships (*participant 1*). *A* and *B*: the relationships obtained for EMG-root mean square (RMS) and shear elastic modulus without fatigue (first two ramps), respectively. They were used to assess the repeatability of the measurements. *C* and *D*: linear regressions obtained without (first ramp) and with (third ramp) fatigue for EMG-RMS and shear elastic modulus, respectively.

indicating good repeatability of ramp contractions for both EMG and SSI.

Occurrence of fatigue during T_{lim} contraction. The average duration of the T_{lim} contraction was 227.5 ± 124.2 s during the EMG session and 227.6 ± 83.7 s during the SSI session. MVC significantly decreased by −30.0 ± 17.8% ($P < 0.0001$) after the T_{lim} contraction. ANOVA revealed a significant main effect of time ($P < 0.0001$) on the torque during the T_{lim} contraction, indicating a decrease in torque after 90% of the T_{lim}. Finally, MF of the abductor digiti minimi EMG signal significantly decreased during T_{lim} contraction from 110 ± 28 to 76 ± 22 Hz ($P < 0.0001$).

Effect of fatigue on the ramp contractions. A significantly higher $RMS_{devR1\text{-}R3}$ value (i.e., between *ramps 1* and *3*; $P = 0.047$) was found for EMG-RMS/torque relationships (10.3 ± 12.6% of MVC) than for shear elastic modulus/torque relationships (3.7 ± 2.6% of MVC). This means that the EMG-RMS/torque relationship was more significantly affected by fatigue.

Effect of fatigue on the accuracy of individual muscle torque estimation. Figure 4, *A* and *B*, depicts an individual example of changes in the measured and estimated torques during T_{lim} contraction for both EMG and SSI sessions. It clearly shows that torque estimated using SSI followed changes in the recorded torque, while it was not the case for the torque estimated using EMG. This result is confirmed by Fig. 4, *C* and *D*, that depicts averaged changes across all the participants. ANOVA revealed a main effect of "method" ($P < 0.0001$) and "time" ($P < 0.0001$) and an interaction effect "method × time" ($P = 0.0001$) on the $RMS_{devTlim}$. More precisely, we found a significantly greater $RMS_{devTlim}$ for the EMG session (15.3 ± 3.8% of MVC) than for the SSI session (5.7 ± 0.9% of MVC), indicating an overall more accurate estimation of muscle torque using SSI compared with EMG. Moreover, post hoc revealed that $RMS_{devTlim}$ significantly increased during EMG session (from 9.5 ± 3.6% of MVC at the beginning to 20.0 ± 12.6% of MVC at the end of the T_{lim}), whereas it remained constant during SSI session (Fig. 5).

Experiment II

Repeatability of shear elastic modulus during submaximal knee extension. Table 2 depicts the results of repeatability of the EMG-RMS and the shear elastic modulus measured during the 15-s nonfatiguing contractions performed at 25% of MVC. ICC values were high (from 0.898 to 0.982) and SEM values were relatively low (<1.5% of $EMG\text{-}RMS_{max}$ and <5.9 kPa for EMG-RMS and shear elastic modulus, respectively). In most cases, the CV was lower than 10% (except for the shear elastic modulus of the rectus femoris, CV = 13.2%), indicating an overall good repeatability of measurements.

Changes in load sharing. The average T_{lim} of the constant-load isometric knee extension was 243.5 ± 102.4 s (range: 145–418 s).

EMG data from antagonist muscles (i.e., long head of biceps femoris and semitendinosus) could not be systematically analyzed. For three participants, the EMG signal of these muscles was lost, certainly because of the electrode location between the thigh and the seat. For the five participants exhibiting usable EMG signals for biceps femoris, EMG-RMS increased by 1.5 ± 3.3% of the maximal value between the beginning and the end of the submaximal fatiguing contraction. Thus the change in antagonist activity was negligible.

Figure 6 shows, for each recorded muscle, the change in EMG-RMS and shear elastic modulus across the fatiguing contraction for the four participants who exhibited the most significant changes in load sharing, which is defined here as opposite changes in shear elastic modulus between synergistic muscles (data from the four others participants are shown in Fig. 7). Depending on the participant, different changes in load sharing were observed, e.g., *1*) between vastus lateralis/rectus femoris and vastus medialis after 75% of T_{lim} in *participant 1*;

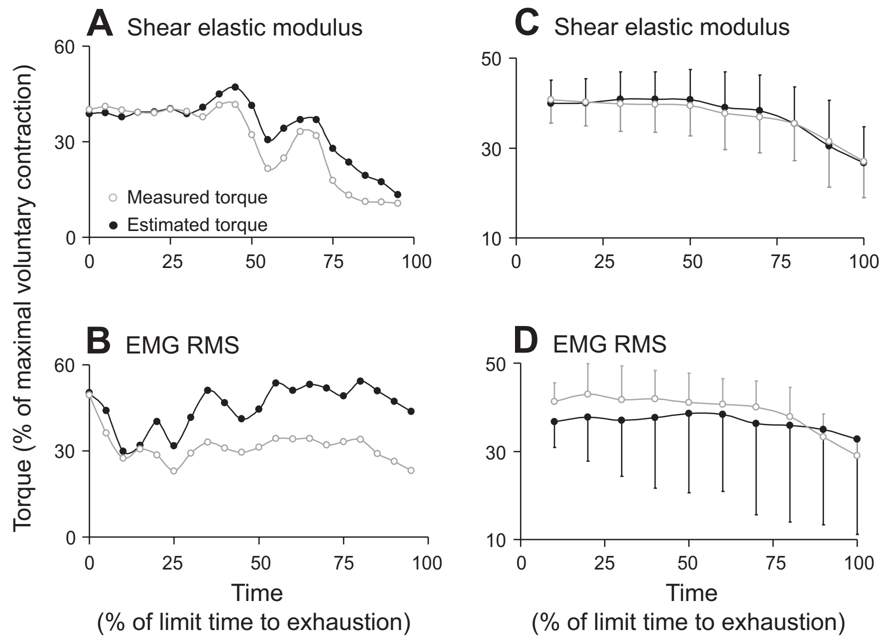

Fig. 4. Individual (*A* and *B*) and averaged (*C* and *D*) changes of both the measured and estimated torque throughout the fatiguing contraction (*experiment I*). Note that, for the typical individual example (*participant 4*), values were averaged over 20 equal time-windows for both the supersonic shear imaging (SSI) session (*A*) and EMG session (*B*) to provide more precision. *C* and *D*: averaged results across all the participants. Values are means ± SE.

2) mainly between rectus femoris and vastus lateralis in *participant 6*. In addition, for *participant 2*, an increase for rectus femoris with no extreme changes for other synergists (vastii) was observed.

An increase in EMG-RMS for a given muscle was observed without any increase in its shear elastic modulus, i.e., in its force (e.g., vastus lateralis and vastus medialis for *participant 2*). However, an increase in EMG-RMS was also associated with an increase in shear elastic modulus of the muscle in other participants (e.g., rectus femoris in *participants 2* and *6*).

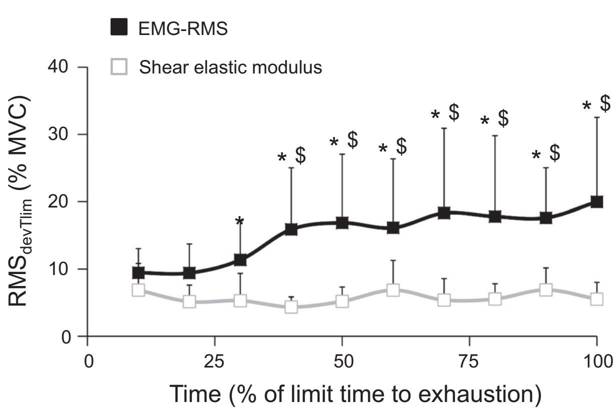

Fig. 5. Mean error of torque estimation throughout the fatiguing contraction (*experiment I*). For each subject, RMS deviation of T_{lim} ($RMS_{devTlim}$) values (in %MVC) were averaged over 10 equal time-windows for both EMG and SSI sessions. Mean $RMS_{devTlim}$ across subjects was significantly greater for EMG session (mean value across the contraction: 15.3 ± 3.8% of MVC) than for SSI session (mean value across the contraction: 5.7 ± 0.9% of MVC), indicating an overall more accurate estimation of torque with shear elastic modulus than EMG-RMS. Moreover, mean $RMS_{devTlim}$ significantly increased during EMG session, whereas it remained constant during SSI session. Values are means ± SE. *Significant difference ($P < 0.05$) between the two methods (EMG-RMS and shear elastic modulus). $Significant increase compared with the first value at 10% of T_{lim}.

DISCUSSION

One of the main challenges in biomechanics is experimentally estimating muscle force [for a review, see Erdemir et al. (13)]. The present study validates our hypothesis that the shear elastic modulus, averaged across a representative muscle region, also follows changes in torque during a fatiguing isometric contraction, and thus it is an accurate index of individual muscle force under such experimental conditions. We further demonstrated that this method could highlight changes in the load sharing between synergistic muscles during fatiguing constant-load isometric contraction. The present study is a step forward in showing that individual muscle force can be experimentally estimated during a wide range of contractions and conditions.

Assessment of Fatigue (Experiment I)

MVC measured immediately after the third ramp contractions of *experiment I* (i.e., after the T_{lim} contraction) was significantly decreased by ~30% compared with the first MVC performed without fatigue. During the T_{lim} contraction, both torque and MF significantly decreased. According to the definition of the neuromuscular fatigue (2, 12) and the decrease in MF (32), these results confirm that the T_{lim} contractions induced neuromuscular fatigue and thus allow us to conclude that we, in fact, estimated an index of individual muscle force during a fatiguing task.

Estimation of Muscle Force From EMG (Experiment I)

Our results obtained during the EMG session are in accordance with the well-known incapability to estimate muscle force from surface EMG during fatiguing contractions (6, 10, 27). Indeed, $RMS_{devTlim}$ significantly increased during the T_{lim}, reaching 20.0 ± 12.6% of MVC at the end of the exercise (Fig. 5). This impairment in the precision of the force estimation may be explained by the alteration of the EMG activity level-force relationship when fatigue occurs (11). Indeed, fa-

Table 2. *Repeatability of both EMG-RMS and shear elastic modulus values obtained for vastus medialis, vastus lateralis, and rectus femoris during isometric knee extension at 25% of MVC*

	Vastus Medialis			Vastus Lateralis			Rectus Femoris		
	ICC	SE	CV, %	ICC	SE	CV, %	ICC	SE	CV, %
EMG-RMS	0.950	1.4	8.0	0.969	1.5	7.60	0.982	0.6	5.7
Shear elastic modulus	0.928	5.9	8.2	0.898	5.7	6.7	0.973	3.4	13.2

SE is expressed in % for EMG-RMS and in kPa for shear elastic modulus.

tigue induces changes in the shape of the motor unit action potentials due to changes in the shape of the intracellular action potential and/or in the propagation velocity of these potentials. Because these changes do not relate fully to the change in the motor unit twitch forces (9, 11), EMG amplitude (i.e., sum of motor unit action potentials) and force level (i.e., sum of motor unit twitches) change differently with fatigue.

Estimation of Muscle Force From SSI (Experiment I)

The first main result of the present study is that shear elastic modulus measured by SSI provides an accurate index of muscle force when neuromuscular fatigue occurs. During the fatiguing task, the changes in the torque estimated from shear elastic modulus measurements followed the changes in the torque measured, leading to a constant $RMS_{devTlim}$ (i.e., from 6.9 ± 4.0% of MVC at the beginning to 5.5 ± 2.5% of MVC at the end of the T_{lim} contraction, Fig. 3). This error is similar to the error previously reported by Bouillard et al. (4) during a nonfatiguing isometric contraction with random changes in torque (4.5 ± 2.5% of MVC), suggesting that the ability to provide an accurate index of individual muscle force using SSI is unaffected by fatigue. Moreover, the linear relationships between shear elastic modulus and torque were few affected by fatigue (i.e., $RMS_{devR1\text{-}R3}$ = 3.7 ± 2.6% of MVC). It is well known that the muscle-tendon creep phenomenon that occurs during sustained isometric contractions induces changes in muscle-tendon elastic properties, but

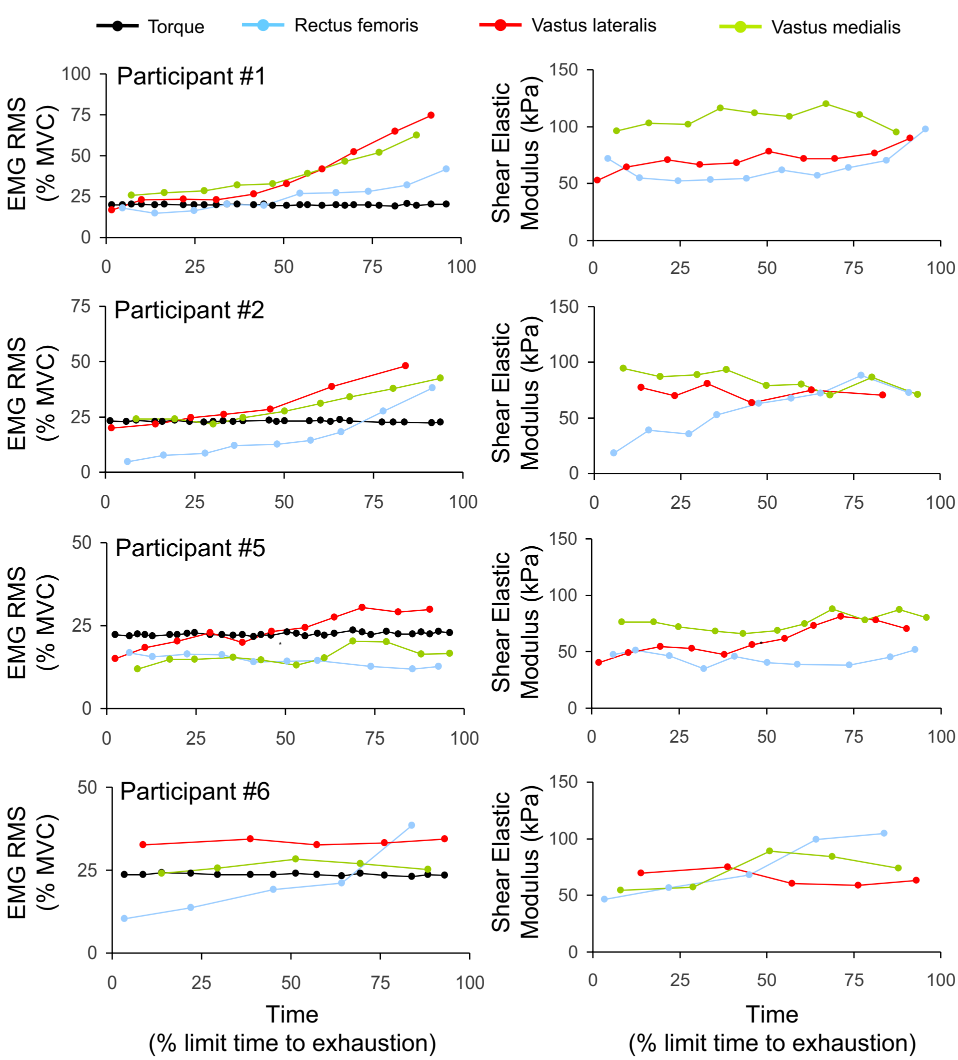

Fig. 6. Evidence of change in load sharing during the constant-load submaximal contraction (*experiment II*). Five successive shear elastic modulus values were alternatively recorded from vastus medialis, vastus lateralis, and rectus femoris during the submaximal fatiguing contraction. These five values were averaged to obtain a more representative value. Only data from the four participants who exhibited the most significant changes in load sharing (defined here as opposite changes in shear elastic modulus between synergistic muscles) are shown in this figure. Note that three points are missing (i.e., two for vastus lateralis for *participant 2*, and one for rectus femoris for *participant 5*) because of the incapability to precisely reposition the probe on the correct location in <15 s (i.e., taking into account the marks on the skin and in a way to obtain a good ultrasound image). The results for the other participants are shown in Fig. 7. Error bars are not displayed because the very low standard deviations make them unreadable (i.e., too small compared with the graph scale).

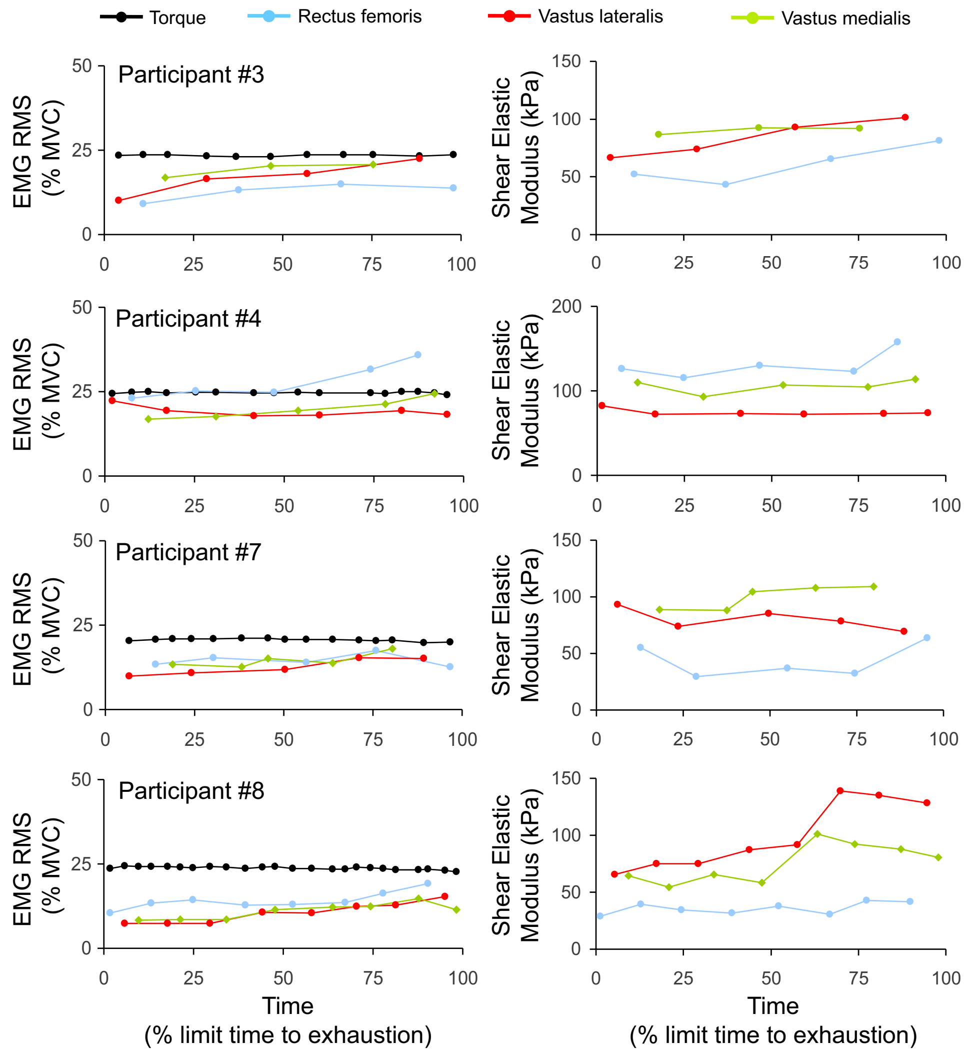

Fig. 7. Evidence of change in load sharing during the constant-load submaximal contraction. Data are from the four participants who exhibited the least significance changes (*experiment II*). Five successive shear elastic modulus values were alternatively recorded from vastus medialis, vastus lateralis, and rectus femoris during the submaximal fatiguing contraction. These five values were averaged to obtain a more representative value. The fact that the stiffness of all the three muscles either stays the same or increases in some participants (e.g., *participant 4*) could be explained by sharing load strategy, implying the unmeasured vastus intermedius, which exhibits a similar volume as the vastus lateralis. Error bars are not displayed because the very low standard deviations make them unreadable (i.e., too small compared with the graph scale).

only during the first few seconds of exercise (e.g., 25, 28). Therefore, the conditioning performed in the present study was sufficient to eliminate the creep effects on the shear elastic modulus measurements.

Evidence of Changes in Load Sharing (Experiment II)

The second main result of the present study is that shear elastic modulus measurements performed by SSI can be used to document changes in load sharing that occur between synergist muscles during submaximal isometric fatiguing contraction. Our analysis did not account for the relative cross-sectional area of the muscle or their moment arms, and thus we cannot estimate the relative contribution of each muscle to knee torque production. But, the opposite changes in shear elastic modulus between knee extensors observed in several participants (Fig. 6) can be interpreted as changes in load sharing (3). Indeed, accounting for the relatively good repeatability of the shear elastic modulus measurements of the knee extensors at 25% of MVC, changes observed (e.g., +90 kPa for rectus femoris in *participant 2*, Fig. 6) can be attributed to changes in muscle coordination rather than to variations inherent to the technique. Notably, we did not record data from vastus intermedius, the fourth main knee extensor, which also could have been involved in changes in load sharing.

As clearly shown in *experiment I*, EMG amplitude cannot be used to accurately study muscle coordination during a fatiguing exercise. Individual results from *experiment II* confirm that fatigue dependence of the surface EMG amplitude is not predictable. Indeed, for some participant's muscles, EMG-RMS increased during a constant-load exercise to compensate for the decrease in the force that occurs in fatigued muscle fibers (e.g., vastus lateralis in *participant 2*; Fig. 6). However, in other participant's muscles (e.g., rectus femoris in *participant 2*; Fig. 6), EMG-RMS and force (indirectly assessed by shear elastic modulus herein) changed in similar ways. In other words, an increase in EMG activity level during a submaximal constant-load fatiguing exercise cannot be interpreted only as a sign of neuromuscular fatigue (20, 32), because it can also indicate an increase in muscle force to compensate for fatigue in other muscles, i.e., change in muscle coordination (22). Other studies are necessary to further understand the high interindividual variability reported for change in shear elastic modulus, i.e., variability in muscle coordination during fatigue.

Limitations

Despite the great interest of the technique in the field of biomechanics, several limitations have to be point out. First, due to the saturation at 266 kPa and the low temporal resolution (1 Hz), tasks that can be studied are limited to midlevel (below ≈40–80% of MVC, depending on the muscle) isometric contractions. However, both the level of saturation and the acquisition frequency are mainly due to "software" limitations that could be improved in the near future. Second, shear elastic modulus is measured within a relatively small region (between 1 and 2 cm^2) with regard to the volume of some muscles. Future investigations should be performed to determine whether the putative heterogeneity of muscle recruitment could really affect the measurements. Finally, we estimated an index of individual muscle force that can be used to accurately quantify relative changes in force in an individual human. It should be kept in mind that the actual muscular force (in Newtons) could not be directly estimated from SSI.

Conclusion and Perspectives

The present study is the first demonstration of an experimental technique that can accurately quantify relative changes in force in an individual human muscle during isometric fatiguing contraction. This result opens considerable perspectives into various fields, such as physiology, motor control, and biomechanics. For instance, it is difficult to dissociate the effects of neuromuscular fatigue and the putative compensations between muscles because of the fatigue dependence of the surface EMG signal (21, 22). The ability to precisely provide an index of force by SSI would lead to new studies concerning the adaptations of muscle coordination during fatiguing tasks, to better understand the mechanisms of intermuscular compensation.

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DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the author(s).

AUTHOR CONTRIBUTIONS

Author contributions: K.B., F.H., A.G., and A.N. conception and design of research; K.B. performed experiments; K.B. analyzed data; K.B., F.H., A.G., and A.N. interpreted results of experiments; K.B., F.H., and A.N. prepared figures; K.B., F.H., and A.N. drafted manuscript; K.B., F.H., A.G., and A.N. edited and revised manuscript; K.B., F.H., A.G., and A.N. approved final version of manuscript.

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