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Evidence of changes in load sharing during isometric elbow flexion with ramped torque

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ABSTRACT

This study aimed to: (1) test the repeatability of Supersonic Shear Imaging measures of muscle shear elastic modulus of four elbow flexor muscles during isometric elbow flexion with ramped torque; (2) determine the relationship between muscle shear elastic modulus and elbow torque for the four elbow flexor muscles, and (3) investigate changes in load sharing between synergist elbow flexor muscles with increases in elbow flexor torque. Ten subjects performed ten isometric elbow flexions consisting of linear torque ramps of 30-s from 0 to 40% of maximal voluntary contraction. The shear elastic modulus of each elbow flexor muscle (biceps brachii long head [BB_{LH}], biceps brachii short head [BB_{SH}], brachialis [BA], and brachoradialis [BR]) and of triceps brachii long head [TB] was measured twice with individual muscles recorded in separate trials in random order. A good repeatability of the shape of the changes in shear elastic modulus as a function of torque was found for each elbow flexor muscle (*r*-values: 0.85 to 0.94). Relationships between the shear elastic modulus and torque were best explained by a second order polynomial, except BA where a higher polynomial was required. Statistical analysis showed that BB_{SH} and BB_{LH} had an initial slow change at low torques followed by an increasing rate of increase in modulus with higher torques. In contrast, the BA shear elastic modulus increased rapidly at low forces, but plateaued at higher forces. These results suggest that changes in load sharing between synergist elbow flexors could partly explain the non-linear EMG-torque relationship classically reported for BB during isometric efforts.

1. Introduction

Since the work of Inman et al. (1952), the relationship between muscle activity level assessed using surface electromyography (EMG) and external torque has been established for many muscles. Although a linear relationship has been reported for some small muscles such as the first dorsal interosseous or abductor digiti minimi muscles during isometric contractions (e.g., Del Santo et al., 2007; Milner-Brown and Stein, 1975; Lawrence and De Luca, 1983), a non-linear relationship is frequently reported for larger muscles such as the biceps brachii (BB; e.g., Lawrence and De Luca, 1983; Nordez and Hug, 2010). In addition to problems associated with the effects of antagonist contraction on muscle force and the drawbacks inherent with surface EMG (e.g., crosstalk and/or heterogeneity of muscle activity; Farina et al., 2004), differences in the EMG-torque relationship between muscles are mainly explained by differences in motor unit recruitment strategy (De Luca et al., 1982; Lawrence and De Luca, 1983; Campy et al., 2009). Indeed, muscle force can be graduated by recruitment of additional motor units (population coding) and/or increased discharge rate (rate coding) (Henneman et al., 1974). A non-linear EMG-torque relationship has been argued to arise in muscles that increase force primarily by recruitment of additional motor units (De Luca et al., 1982, Lawrence and De Luca, 1983). This could explain the non-linear relationship between EMG and torque for BB, which relies primarily on population coding (Kukulka and Clamann, 1981), and the linear relationship for the first dorsal interrosseous, which relies primarily on rate coding to increase its force (Milner-Brown and Stein, 1975). Based on this explanation, the non-linear relationship between EMG and torque would be due to myoelectrical phenomena rather than to a non-linear relationship between individual muscle force and joint torque.

Surprisingly, no study has explored the putative influence of changes in load sharing between synergist muscles on non-linear EMG-torque relationships. This is probably explained by the inability to selectively record EMG activity from deep muscles using surface electrodes and the excess selectivity of intramuscular EMG that compromises the representativeness of recordings. Although the first dorsal interosseous is the sole muscle that

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abducts the index finger (Chao, 1989), during elbow flexion the load is shared among four agonist muscles (i.e., BB long head [BB_{LH}], BB short head [BB_{SH}], brachialis [BA] and brachoradialis [BR]) (Murray et al., 2002). Changes in load sharing could partly explain a non-linear EMG-torque relationship for the BB as relative contribution of this muscle and its synergists to torque may change as a function of torque.

Recently, an innovative elastographic technique, known as Supersonic Shear Imaging (SSI), has been shown to reliably estimate the shear elastic modulus (Gennisson et al., 2010; Nordez and Hug, 2010; Shinohara et al., 2010). Since Nordez and Hug (2010) showed that this modulus is linearly related to muscle contraction intensity as recorded with EMG, SSI may provide an alternative non-invasive technique to indirectly estimate muscle activity and would be particularly useful for deep muscles. This technique overcomes some limitations of surface EMG recording, such as crosstalk and signal cancellation, and allows averaging over a wide region of the muscle to be more representative of muscle activity than surface or intramuscular EMG.

This study aimed to: (1) compare SSI measures of muscle shear elastic modulus of the 4 elbow flexor muscles between 2 repetitions of isometric elbow flexion with ramped torque; (2) determine the relationship between muscle shear elastic modulus and elbow torque for the four elbow flexor muscles, including the deeply situated BA muscle; and (3) investigate changes in load sharing between synergist elbow flexor muscles with increases in elbow flexor torque. We tested the hypothesis that the non-linear relationship between muscle activity level of biceps brachii and elbow torque during an isometric contraction would be explained by changes in load sharing.

2. Methods

2.1. Participants

Ten healthy volunteers participated in this study (3 women, 7 men; aged 24.9 ± 3.6 years). Participants were informed of the purpose of the study and methods used before providing written consent. The experimental design of the study was approved by the local Ethical Committee and was conducted in accordance with the Declaration of Helsinki (last modified in 2004).

2.2. Measurements

Ergometer. A Biodex system 3 research (Biodex medical, Shirley, NY) isokinetic dynamometer was used to measure elbow angle and torque. The position of the subjects was similar to the position previously described (Gennisson et al., 2005; Nordez and Hug, 2010). Briefly, subjects sat on the dynamometer with their right upper arm and forearm flexed to 90°, and the wrist supinated.

Elastography. An Aixplorer ultrasonic scanner (Supersonic Imagine, Aix-en-Provence, France), coupled with a linear transducer array (4–15 MHz, SuperLinear 15-4, Vermon, Tours, France) was used in SSI mode as has previously been described in detail (Bercoff et al., 2004; Tanter et al., 2008). Briefly, the system consists of a transient and remote mechanical vibration generated by radiation force induced by a focused ultrasonic beam (i.e., pushing beam). Each ''pushing beam'' generates a remote vibration that results in the propagation of a transient shear wave. An ultrafast echographic imaging sequence is then performed to acquire successive raw radio-frequency data at a very high frame rate (up to 20 kHz). One-dimensional cross-correlation of successive radio-frequency signals is used to determine the shear wave velocity (V_s) along the principal axis of the probe using a time-of-flight estimation. Measurements were made from the BB_{SH}, BB_{LH}, BA, BR, and triceps brachii long head (TB). For the superficial muscles, the probe was placed over the muscle belly. For the deep muscle (i.e., Brachialis), it was placed as described by Hodges et al. (2003), in the medial and distal part of arm, near the fold of the joint. For all muscles, the probe was carefully aligned with the direction of shortening of the muscle. Considering a linear elastic behavior, a shear elastic modulus (μ) was calculated using V_s as follows:

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

where ρ is the muscle mass density (1,000 kg/m^3), and μ is the shear elastic modulus (in kPa).

The linear (Gennisson et al., 2003; Bercoff et al., 2004; Catheline et al., 2004; Nordez et al., 2008; Deffieux et al., 2009) and purely elastic (Catheline et al., 2004; Deffieux et al., 2009) behaviors have been most often considered in the studies of elastography for biologic tissues.

Maps of the shear elastic modulus were obtained at 1 Hz (i.e., the maximal sampling rate of SSI measurements of the current version of the ultrasonic scanner) with a spatial resolution of 1×1 mm. This measure was performed in less than 20 ms (Bercoff et al., 2004).

2.3. Protocol

Participants performed three 3-s maximal isometric voluntary elbow flexion efforts (1-min rest between contractions) to determine the maximal voluntary contraction (MVC). Then, they were asked to perform ten isometric contractions (2-min rest between tasks). Each consisted of a smooth linear torque ramp from 0 to 40% of MVC over 30 s. To control the ramping of the torque, the participants had to follow a visual feedback displayed on a monitor placed in front of them. The shear elastic modulus of each muscle (BB_{SH}, BB_{LH}, BA, BR and TB) was measured twice during the ten ramps with individual muscles recorded in separate trials in random order. To ensure consistent location between the two trials, the position of the probe was marked using a waterproof felt-tip pen. Before each trial, participants were asked to completely relax for a 5-s period for measurement of the shear elastic modulus at rest.

2.4. Data analysis

The shear elasticity modulus value was averaged over a circular region located in the middle of the map (from 0.5 to 1.5 cm in diameter, depending on muscle thickness) (Fig. 1). This region was slightly moved to take into account for the slight muscle displacements observed during the isometric contraction. Due to technical limitations of the ultrasonic scanner, measurements saturated at 100 kPa, limiting the range of analysis for some muscles, for most participants. If one value in the circular region reached 100 kPa, this measurement of the trial and all the following measurements were discarded from further analysis. The five shear elastic measurements performed at rest before each contraction were averaged to obtain a resting value for each muscle.

The shear elastic modulus/torque relationship was plotted for each ramp of each subject, and the shear elastic modulus was linearly interpolated to obtain a value at every 1% elbow flexion MVC. For each muscle, a mean pattern was obtained by averaging the shear elastic modulus values across subjects. The best trial of each subject (i.e., the ramp for which there were few or no single shear elastic modulus measurements rejected due to saturation) was retained for the averaging.

2.5. Statistical analysis

Data distributions consistently passed the Shapiro-Wilk normality test (Statistica®V6, Statsoft, Maison-Alfort, France). Values are reported as mean ± SD. The level of significance was set as $P < 0.05$.

To satisfy the first aim, the repeatability of the shape of the shear elastic modulus/torque relationship between the two trials was assessed for each subject and each muscle by calculating the Pearson's correlation coefficient (r). For the second aim the shape of the modulus-torque relationship was quantified in 2 ways. First, linear and polynomial regressions were fitted to the data. The order of polynomial required to produce an R^2 value greater than 0.99 was identified and the equation recorded. Second, the threshold torque at which the elastic modulus changed from the value at rest was identified. This threshold was identified using a repeated-measures ANOVA for each muscle (random factor - participant, between subject factor - torque) (Statistix®, Tallahassee, FL, USA). If a main effect was identified for ''torque'' (i.e. modulus was significantly changed as a function of torque) Duncan's post-hoc test was used to identify the first torque increment at which the modulus was different from the modulus value recorded during the rest periods between ramp contractions. Duncan's test is less conservative than other tests such as Tukey's, but this was deemed appropriate considering the exploratory nature of the present study and the required sensitivity to detect changes in shear elastic modulus from rest values. For aim three, load sharing was considered qualitatively by comparison of the shape and threshold torque values.

Additional analysis was conducted for depiction of the rate of change in elastic modulus between muscles during different stages of the contractions. The 0-16% of MVC was the maximal range of torque common to all muscles and all subjects. Thus, the relative change in shear elastic modulus was quantified for each muscle on three stages of 1% MVC increments in torque: (i) between 4 and 5% MVC, (ii) between 9 and 10% MVC, (iii) between 14 and 15% of MVC. If the shear elastic modulus of all muscles changed in an identical manner with increased torque (i.e., no change in load sharing), the relative change on each of the three 1% MVC increments would be the same for all muscles.

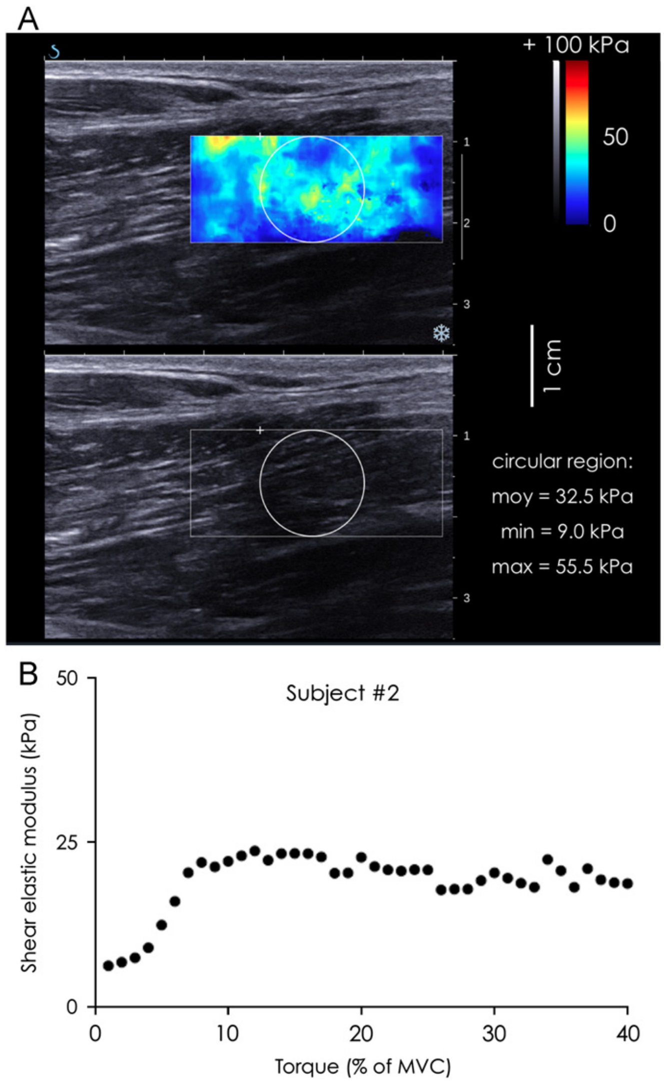

Fig. 1. A. Typical example of shear elastic modulus measurement of the brachialis muscle. The region of interest (colored region) was chosen using an echographic image. For the sake of clarity, the echographic image without the color map is depicted at the bottom of the figure. The shear elastic modulus was measured over the circular region. B. Individual example of changes of the shear elastic modulus of the *Brachialis* with external torque. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

3. Results

The maximal torque recorded during the MVC efforts was 57 ± 18 N.m, ranging from 28 to 87 N.m. The 40% MVC (target peak torque) was 22.8 ± 7.2 N.m. Of the 100 ramps (10 ramps × 10 subjects), the 100 kPa maximal value of the shear elastic modulus was reached 34 times before the 40% of MVC was reached. Consequently, the full ramp could not be analysed and instead data were analysed for each muscle up to the highest percentage torque that was available across the subjects. Thus, the isometric contractions were analyzed up to 34.5 ± 4.7% (BB_{SH}); 34.9 ± 7.9% (BB_{LH}); 24.1 ± 7.1% (BR); 37.4 ± 6.6% (BA); and 38.9 ± 3.8% of MVC (TB).

Fig. 1 depicts an individual example of changes in shear elastic modulus of BA. Table 1 reports the mean correlation coefficients between the two ramps for each muscle. Excluding the data for TB, which was minimally active in the elbow flexion task, individual *r*-values ranged from 0.85 to 0.94 which demonstrates good repeatability of the shape of the changes in shear elastic modulus as a function of elbow flexion torque.

Table 1
Mean correlation coefficients (r) between the shear elastic modulus/torque relationships obtained during the two ramps.

Muscle	r
BB_{SH}	0.93 ± 0.08
BB_{LH}	0.94 ± 0.04
BA	0.93 ± 0.05
BR	0.85 ± 0.12
TB	0.19 ± 0.52[a]

BB_{SH}: Biceps brachii short head, BB_{LH}: Biceps brachii long head, BR: Brachioradialis, BA; Brachialis, TB: Triceps brachii long head.

[a] The low *r*-value obtained in TB is explained by no increase in shear elastic modulus for this muscle.

Fig. 2 depicts, for each muscle, the mean shear elastic modulus/torque relationship averaged across subjects. A significant main effect of torque on the shear elastic modulus was identified for all muscles, except TB, indicating a significant increase in muscle shear elastic modulus with torque (Table 2). However, the modulus increased in a non-linear manner. Relationships were best explained by a second order polynomial. For all muscles except TB and BA an R^2 of greater >0.99 ($P<0.05$) could be achieved with a second order polynomial (Fig. 2, Table 2). Increasing the order of the polynomial to a 4th order equation increased the R^2 for BA to 0.978 ($P<0.05$), suggesting a more complex relationship between torque and elastic modulus for BA compared to the other muscles. Inspection of Fig. 2 and the results of the ANOVA show that the shear elastic modulus of BB_{SH} and BB_{LH} had an initial slow change at low torques and did not increase above resting values until 10% ($p=0.030$) and 11% ($p=0.012$) of MVC for BB_{LH} and BB_{SH}, respectively, followed by an increasing rate of increase in modulus with higher torques; at higher torque levels (BB_{LH} contractions > 20% MVC; BB_{LH} contractions > 21% MVC) a shear elastic modulus was significantly increased with force increments of 3–4% MVC. In contrast to the two heads of BB, the BA shear elastic modulus increased rapidly at low forces and was significantly higher than the rest value with contractions at 5% of MVC ($P=0.008$, Table 2). However, at higher force levels there was no further increase in shear elastic modulus as the relationship plateaued, i.e. at torque increments above 8% the shear elastic modulus did not undergo further change and was not different to the final value at 35% MVC (Fig. 2). After a short period of no change at low torque levels, the BR shear elastic modulus increased significantly when torque reached 6% of MVC ($P=0.011$). The TB shear elastic modulus was not significantly changed during the isometric ramps, suggesting no influence of antagonist co-activation on the relationships between torque and estimated of activity.

Fig. 3 shows the change in relative increase in shear elastic modulus between muscles with increments in torque at different levels. At the increment in force from 4–5% MVC there is a large change in BA and BR modulus compared to the heads of BB. At the increment from 14–15% this relationship has changed and the biggest change is for BB_{SH}. This demonstrates a torque-dependent change in sharing load between muscles.

4. Discussion

The data of the present study show a unique relationship between torque and shear elastic modulus for each muscle, and inspection of these data suggest that changes in the load sharing

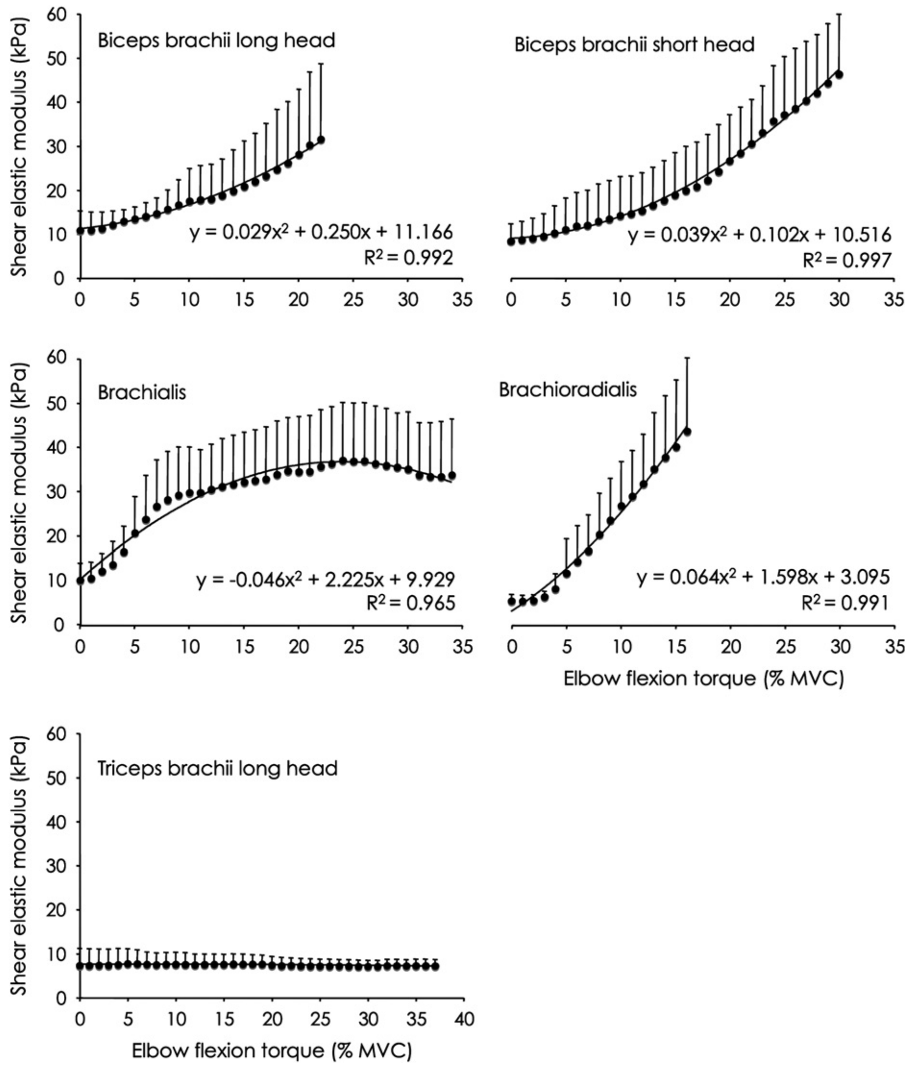

Fig. 2. Mean shear elastic modulus/torque relationship averaged across the 10 subjects. Standard deviation and line of best fit (second order polynomial) are shown along with the associated equation and R^2. BB_{SH}: Biceps Brachii Short Head, BB_{LH}: Biceps Brachii Long Head, BR: Brachioradialis, BA: Brachialis, TB: Triceps Brachii.

Table 2
Relationship between torque and shear elastic modulus.

Muscle	Threshold torque (Duncan test)	R^2 of linear fit	Equation of second order polynomial	R^2
BB_{LH}	10% MVC	0.956	$Y=0.029x^2+0.250x+11.166$	0.992
BB_{SH}	11% MVC	0.940	$Y=0.039x^2+0.102x+10.516$	0.997
BR	6% MVC	0.980	$Y=0.064x^2+1.598x+3.095$	0.991
BA	5% MVC	0.681	$Y=-0.046x^2+2.225x+9.929$	0.965
TB	–	–	–	–

The threshold torque corresponds to the first torque increment at which the modulus was different from the modulus value recorded during the rest periods. BB_{SH}: Biceps brachii short head, BB_{LH}: Biceps brachii long head, BR: Brachioradialis, BA; Brachialis, TB: Triceps brachii long head. MVC, Maximal Voluntary Contraction.

could partly explain the non-linear EMG/torque relationship classically reported for BB during isometric efforts.

The good repeatability of the shear elastic modulus measured by SSI reported here corroborates earlier data for BB (Nordez and Hug, 2010). As the shear elastic modulus is measured along the probe direction, it represents the muscle behaviour along the shortening direction and not along the muscle fiber direction. Gennisson et al. (2010) showed that the shear elastic modulus measurement is sensitive to the angle of rotation between the ultrasonic probe and the muscle fibers and, thus, to the pennation angle. Those data suggested the shear elastic modulus decreases as pennation angle increases. As BB_{SH} and BB_{LH} are fusiform muscles (Murray et al., 2002), no change of pennation angle is expected during the isometric contraction in their central region. However, both BR (Lieber et al., 1992) and BA (Herbert and Gandevia, 1995; Hodges et al., 2003) have pinnate structure and the pennation angle may influence the shape of the relationship between muscle activity level and the shear elastic modulus. As the pennation angle increases with torque (Herbert and Gandevia, 1995; Hodges et al., 2003), this means that the increase in shear elastic modulus would have been slightly underestimated. The pennation angle of BR is low (about 2°) and is minimally affected by the contraction (Lieber et al., 1992). Hodges et al. (2003) showed a small increase in the pennation angle of the BA of $\sim$7.7° as the torque increased from rest to 50% of MVC. In addition, these authors reported the increase in pennation angle was non-linear and occurred mostly at contraction intensities below 10% MVC with very minor changes for higher contraction intensities. Thus, our method may underestimate the increase in BA shear elastic modulus at the beginning of the isometric ramp contractions. Because our results showed that the main increase of the BA shear elastic modulus occurred within this range, from 0 to about 10% of the MVC, the

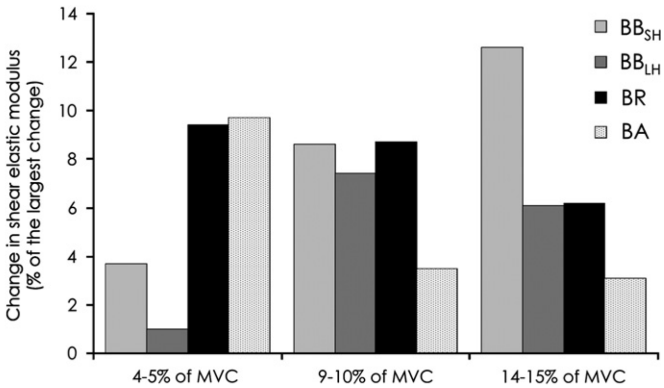

Fig. 3. Load sharing between elbow flexor synergist muscles. The change in shear elastic modulus as a percentage of the largest change between rest and 16% MVC are shown for 3 different 1% MVC increments in force (5% MVC – increment from 4–5% MVC; 10% MVC – increment from 9-10% MVC; 15% MVC – increment from 14–15% MVC). The distribution of change in shear elastic modulus changes as a function of torque with greater changes in brachialis (BA) at low torque and greater changes in biceps brachii short head (BB_{SH}) and biceps brachii long head (BB_{LH}) at high torques. BR: Brachioradialis.

putative effects of the changes in the muscle pennation angles (i.e., underestimation of the increase of the muscle activity level for BA) would not affect our main result showing that the BA activity plateaued for higher torque level.

Although a positive linear relationship was previously reported between shear elastic modulus and EMG activity level (Nordez and Hug, 2010; Gennisson et al., 2005), that does not signify that they reflect the same physiological/mechanical phenomena. Compared to EMG activity level, shear elastic modulus is more linked to mechanical factors and would not be sensitive to the motor unit recruitment strategy (i.e., to electrophysiological process). Nordez and Hug (2010) reported a non-linear EMG/torque relationship for BB, and a similar shaped relationship between shear elastic modulus and torque. These results suggest that this non-linear relationship can be explained by factors other than the activation pattern of the motor units as classically reported in the literature (Lawrence and De Luca, 1983; Zhou and Rymer, 2004). The patterns of change in shear elastic modulus reported here highlight putative torque-dependent changes in load sharing between the elbow flexor synergist muscles (Figs. 2 and 3). The characteristic shape of the change in shear elastic modulus of both BB heads with increasing torque demonstrated little change initially, but with increasingly large increments in modulus at higher torques. In contrast, the shear elastic modulus of the BR and BA increased rapidly at low torques and then that for BA plateaued. The relative change in shear elastic modulus at low and higher torques (Fig. 3) supported this observation. One interpretation of these differences in shape of curvature is that torque is primarily produce by preferential activity of BR and BA at low torque levels, and the increase in torque after ~10% of MVC is mainly due to the BB. This provides evidence of changes in the load sharing between the synergist muscles as external torque increased. Because BB has a moment arm that is approximately two times higher than BA at 90° of elbow flexion (Murray et al., 1995; Murray et al., 2002), the changes in the load sharing might be explained by different functions of these muscles. A shorter moment arm (i.e., BA) would provide an advantage for the force graduation (i.e., exertion of a low force level with precision), while a longer moment arm (i.e., BB) could provide an advantage for the production of high torque levels.

Despite the similarity in shape of the shear elastic modulus-torque relationships across subjects, it is true that a variance was observed. This could be explained by some anatomical particularities (e.g., differences in the moment arms, cross-sectional areas) and, more certainly, by variability of muscles recruitment between subjects. Indeed, although general shapes of the relationships were similar (e.g., strong increase followed by plateauing of the brachialis shear elastic modulus), individual variability was observed as it can be classically observed in most studies that concern muscle coordination (e.g., Hug et al., 2010). The perspective to combine shear elastic modulus measurements with moment arm and cross-sectional area measurements would be of high interest in the way to precisely study the compensations between individual muscle torques.

This study identified a key limitation in the SSI method for estimation of shear elastic modulus during muscle contraction. Notably, only low torque contractions could be tested due to the saturation of the measurements at 100 kPa. However, recent software/hardware improvements are likely to partially resolve this problem in the future versions of the echographic device.

5. Conclusion and perspectives

Measurement of the shear elastic modulus of each elbow flexor muscle during an isometric ramp contraction provides evidence of torque-dependent changes in the load sharing between synergist elbow flexor muscles. The failure to identify this in previous work is due to the lack of data for several key synergist muscles involved in the task (i.e., in particular BA).

Due to the non-linearity of the mechanical properties of biological tissues (Fung, 1993), muscle stress is linked to its elastic modulus. Thus, one would expect that measurement of muscle elasticity would provide an indirect measurement of muscle stress. However, the shape of the relationship between the shear elastic modulus and muscle stress is not known. This information is essential to determine the mathematical model (e.g., linear, polynomial, exponential, etc.) required for the estimation of the muscle stress from the shear elastic modulus measurement. Based on previous experiments (Nordez and Hug, 2010) and on experimental data obtained using magnetic resonance elastography (Dresner et al., 2001), we hypothesize that this relationship would be linear. To test this hypothesis we plan on establishing the nature of the relationship between shear elastic modulus and torque for a muscle without synergist. Combined with estimates of both physiological cross-section area and moment arm, measurement of muscle stress could be used to estimate individual muscle force that could provide considerable insight into various scientific fields (e.g., neuromuscular physiology, motor control, robotic and biomechanics) (Erdemir et al., 2007).

Conflict of interest statement

None.

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