An elastography-driven biomechanical model for individual muscle force estimation

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***Abstract*— The estimation of muscle forces is a long-standing problem in the field of muscle biomechanics. This is often referred as a classic indeterminate problem, as the unknowns (individual muscles forces) are greater than the measurable properties of the system. Although there exist approaches based mainly on electromyography (EMG), this approach is currently discussed due to the electrical nature of the measures and the susceptibility of the EMG signal to physiological and non-physiological factors. In this context, in recent years, elastography has become a reference method to characterize the longitudinal shear elastic modulus of skeletal muscle. As shown in previous works, this variable is directly related to muscle strength. In this sense, the mechanical and non-electrical nature of the estimates obtained through this methodology makes it a good alternative for calculating muscle forces. Thus, this work proposes a model that, based on muscle elasticity values, allows the calculation of individual muscle forces under a specific loading condition. Particularly, we analyzed the isometric flexion of the elbow joint, where the biceps brachii, brachioradialis, and brachialis muscles act synergically. This way, depending on the muscle and torque level, the model provided reliable force values, which stabilized the external torque on the elbow joint so that the net torque was 0. Furthermore, the model results showed that the ~10% MVC appears to be a breakpoint in the load sharing between these muscles, thus retrieving the phenomenology observed in previous studies regarding their synergistic action during the isometric elbow flexion.**

Keywords—Elastography, static optimization, elbow flexors, individual muscle forces, load-sharing.

I. INTRODUCTION

Understanding the mechanisms by which the central nervous system coordinates the force distribution among muscles requires precise measurement of the force exerted by each muscle [1]. However, when evaluating muscle forces, the overall joint moment is assessed using isokinetic dynamometry. In this way, the combined force exerted by all the muscles involved is measured rather than their individual contributions. Thus, accurately determining the specific forces generated by individual muscles has remained an unresolved challenge in muscle biomechanics. This constitutes a current gap in our knowledge of muscle functionality.

Muscle redundancy, which implies the existence of more muscles than freedom movement degrees, poses a challenge in determining individual muscle forces [2]. Even simple single-joint motor tasks result in an indeterminate problem, where the individual forces are greater than the equations derived from the measurable properties of the system, making a unique solution unattainable [1]. Numerous methods, including musculoskeletal models, have been developed to address this issue [3]. Static optimization, a common approach in the literature, seeks muscle forces to minimize a cost function. These functions are mathematical expressions that intend to optimize certain physiological criteria during specific activities [4, 5], as the central nervous system does. However, it often neglects crucial aspects of muscle physiology related to force production [5]. Alternatively, modified static algorithms have been developed by considering muscle activation and contraction dynamics, but lack validation due to the absence of suitable experimental techniques [3, 1].

Numerous studies have suggested utilizing surface electromyography (EMGs) to estimate individual muscle forces. The earliest studies in this regard have addressed this issue by grouping functionally related muscles and assumed a direct link between the EMG activity of a skeletal muscle and the force it produces [6, 7]. Recently, novel approaches such as employing neural networks and EMG-driven models have also emerged to calculate both joint moments and individual muscle forces [8-14]. However, the validity of this approach is currently under debate due to the inherent electrical nature of EMG measurements. While EMG can evaluate the neural activation of muscles, signals are susceptible to various physiological and non-physiological factors. For instance, signal amplitude can be influenced by electrode placement and tissue conductivity [15]. Additionally, the skeletal muscle force-length relationship implies that different forces can be produced for the same electromyographic activation levels if the muscle operates at varying lengths. Moreover, EMG signals do not capture passive force production [1]. The interference from adjacent muscle EMG signals and the presence of neuromuscular fatigue can also affect the EMG amplitude vs. force relationship [16, 17]. Thus, these factors collectively limit the utility of the EMG for reliably measuring muscle force.

Thus, in recent years, there has been an increasing use of elastography for studying muscle biomechanics. Specifically, shear wave elastography (SWE) has provided a novel way to measure skeletal muscle longitudinal shear elasticity (μ_L) in vivo [18, 19]. The primary advantage of SWE techniques lies in the combination of high-frequency ultrasonic waves with low-frequency waves (100 ~ 1000 Hz), thereby presenting high spatial resolution (< 1 mm) and good contrast in characterizing μ_L [20]. The mechanical and non-electrical

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nature of the measurements obtained by elastography makes this method a promising alternative for estimating muscle forces. In this regard, previous work has demonstrated the potential accuracy in estimating joint torque from μ_L measurements in the abductor muscles of the little finger, as well as revealing a linear correlation between these variables over the entire range of contraction intensity [21, 22]. In addition, elastography has been applied to assess whether fatigue alters the ability to provide an individual muscle strength index and to evaluate load distribution among elbow flexor muscles during isometric contraction [23, 24]. More recently, by assuming a quadratic relationship between shear wave velocity and muscle force, magnetic resonance elastography (MRE) has also been used to characterize forearm muscle forces during isometric tasks [25, 26].

Thus, beyond the previously referred advances, it is still necessary to develop an approach that provides estimates of individual muscle forces expressed in N, comprehensively considering the relationship between the muscle contraction dynamics, the change of μ_L, and force generation. In this regard, in our previous work, we integrated both the ligand-binding framework, the SRS principle, and the theory of acousto-elasticity [27], to account for the dynamics of longitudinal deformation of muscle during the isometric contraction. As the above depends on cross-bridge formation, which underlies the muscle force-generating mechanism, this approach may be useful for calculating individual muscle forces. Thus, this work aims to integrate such developments within an elastography-driven static optimization model, which allows calculating the individual forces produced by the biceps brachii (BB), brachioradialis (BR), and brachialis (BA) muscles during the isometric elbow flexion.

II. Materials and Methods

A. Elastography-driven static optimization model

The elbow joint during isometric flexion was modeled considering a one-degree-of-freedom joint in the sagittal plane. The shoulder and elbow flexion angles were set at θ = 90°, and the forearm in a supine position. Under these conditions, the BB, BR, and BA muscles contract, acting synergistically through the sum of their internal torques (τ_{int}):

$$\sum \tau_{int} = \sum_{i=1}^{3} \vec{F_i} \times r_i \qquad (1)$$

Here, the notation $i = (1, 2, 3)$ denotes the individual muscle forces ($\vec{F_i}$) and their corresponding lever arms (r_i), for the BB, BR, and BA muscles, respectively. We included the following inequality constraints regarding the maximal and minimal values of $\vec{F_i}$ as a function of the joint torque (τ):

$$0 \leq \vec{F_i}(\tau) \leq \vec{F}_{i_{max}} \qquad (2)$$

where $\vec{F}_{i_{max}}$ is the maximal theoretical force developed by the corresponding muscle. Given the arm position considered in the model, the force acting on the forearm ($\vec{F}$) is the only force contributing to the external torque on the system. Thus, based on all the above, the following balance equation must also be satisfied:

$$\overrightarrow{F_{BB}}.r_{BB} + \overrightarrow{F_{BR}}.r_{BA} + \overrightarrow{F_{BA}}.r_{BA} = \vec{F}.L = \tau \qquad (3)$$

where L is lever arm corresponding to $\vec{F}$.

In this way, Eq. (1) was solved numerically as a function of the constraints of Eqs. (2) and (3), by using the function fmincon of the optimization toolbox of Octave/Matlab (GNU Octave Team/MathWorks Inc.). This function requires an initial approximate value of the individual forces for each torque level ($\vec{f_i}(\tau)$) to find the minimal optimal solution. Thus, such forces were calculated using the $C_i(\tau)$ coefficients described in [27]:

$$\vec{f_i}(\tau) = C_i(\tau).\overrightarrow{Fm}_{i_{max}} = \left(1 - \frac{\mu_L(0)}{\mu_L(\tau)}\right).\overrightarrow{Fm}_{i_{max}} \qquad (4)$$

Here, $\mu_L(0)$ and $\mu_L(\tau)$ are the muscle longitudinal shear elastic modulus for the corresponding *i*-muscle at rest (τ = 0) and contracted according to τ level. Meanwhile, $\overrightarrow{Fm}_{i_{max}}$ is the maximal theoretical contractile force developed by the muscle belly, which was calculated as a function of the maximal theoretical stress of the skeletal muscle ($\sigma_{i_{max}}$) and the corresponding physiological cross-sectional area of the *i*-muscle belly ($PCSA_i$):

$$\overrightarrow{Fm}_{i_{max}} = PCSA_i.\sigma_{i_{max}} \qquad (5)$$

Concerning the value of $\sigma_{i_{max}}$, several studies has assigned values ranging from 220 to 360 kPa for mammalian muscles [28-31]. Here, we assumed a $\sigma_{i_{max}}$ value of 330 kPa for the muscles of the upper limb [31].

B. Data sources

Given the exploratory purpose of the present work, we tested the present model using the $C_{BB}(\tau)$, $C_{BR}(\tau)$, $C_{BA}(\tau)$, and τ data corresponding to three of the subjects evaluated in our previous work [27]. Such coefficients were calculated from the respective μ_L values measured during isometric elbow flexion ramps performed between 0-30% of the maximal voluntary contraction (MVC) over 15 s. The $PCSA_i$ values were estimated from the corresponding echographic images obtained with SSI during the measurements performed at rest. In this regard, we considered the muscle a cylinder whose diameter corresponds to the maximum muscle height obtained from such images. Regarding r_i, they were estimated using the equations provided by [32]. Finally, we derived the Fimax forces by linear regression from data on stress and cross-sectional area of the BB tendon provided in the literature [33, 34]. In this respect, we assumed the same mechanical properties for tendons inserted at the elbow joint. This yielded the plausible values of 2060, 848, and 1507 N for the BB, BR, and BA muscles, respectively.

III. Results

Fig. 1A shows the results of the $\vec{f_i}$ forces as a function of joint torque for the three subjects studied. As can be seen, the internal torque determined by such forces was not enough to satisfy the equilibrium condition of Eq. (3) over the entire range of contraction intensity. Thus, between 0 - 30% MVC, $\vec{f_i}(\tau)$ ranged from 0 – 184.05 ± 61.92 N (BB), 0 – 53.36 ± 29.09 N (BR), and 0 – 144.47 ± 80.50 N (BA).

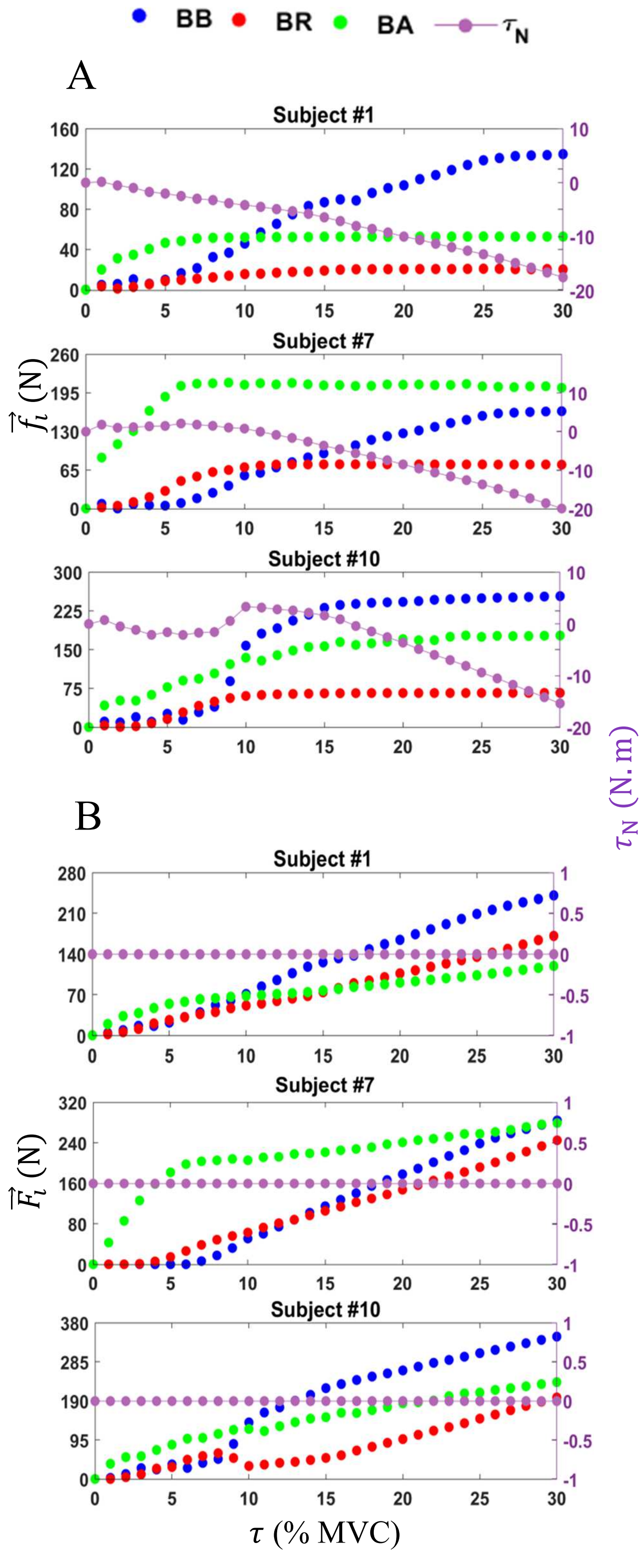

Fig. 1. A: Initial approximates of force values calculated from eq. (4) in subjects 1, 7, and 10 of [27] to optimize the estimation of individual muscle forces of the BB, BR, and BA muscles through the model. The net torque (τ_N) is different from zero in all cases for each contraction intensity level. B: Individual muscle force values calculated by the model from the initial approximated forces of A. The net torque determined by $\vec{F}_{BB}$, $\vec{F}_{BR}$, $\vec{F}_{BA}$, and $\vec{F}$ is zero in all cases for each contraction intensity level.

On the other hand, Fig. 1B shows the individual forces for the BB, BR, and BA muscles as a function of joint torque, resulting from the application of the model to the same subjects. As can be seen, using the $\vec{f_i}(\tau)$ values as initial approximations of the individual muscle forces determined optimal solutions that nulled the net torque for each contraction intensity level. In this sense, the resulting $\vec{F_i}(\tau)$ forces obtained by the model comprised values ranging from 0 – 290.65 ± 52.99 N, 0 – 204.66 ± 37.42 N, and 0 – 211.40 ± 82.45 N, for BB, BR, and BA, respectively.

IV. Discussion

The goal of the present study was to develop a biomechanical model able to calculate individual muscle forces based on elastography measurements. The model is based on the formalism developed in our previous work, thus combining the ligand-binding framework, the acousto-elasticity theory, and the short-range stiffness principle [27]. The resulting $C_i(\tau)$ coefficients of this approach describe the dynamics of the longitudinal shortening of each muscle, which is intrinsically associated with force generation during an isometric contraction. Thus, such coefficients are the key element of the model, since they are considered for calculating the $\vec{f_i}$ forces (Eq. (4)), which are the inputs of the present elastography-driven static optimization algorithm.

In this context, the model provided plausible muscle force values for the synergistic action of the elbow flexors muscles during their isometric contraction. In this regard, the results agreed, both qualitatively and quantitatively, with the biomechanical and functional aspects of the BB, BR, and BA muscles discussed in previous elastographic studies [24, 20, 27, 35-37]. Thus, the results for the BA denoted its capacity to produce the first torque effects at low contraction intensity levels, thus contributing to developing precise movements and stabilizing the elbow joint. Since the BA is a uniarticular muscle, having approximately half of the lever than the BB at 90º of elbow flexion, the previous results are consistent with its anatomical and functional features [27]. Concerning the BB and BR, the results reflect their capacity to additionally stabilize the shoulder and wrist joints by producing torques determined by comparable force levels to BA helped by their long-moment arm [24, 27].

As observed in Figs. 1A and 1B, at ~10% MVC, both the $\vec{f_i}(\tau)$ and $\vec{F_i}(\tau)$ curves intersect, especially those of the BB and BR muscles. In this way, beyond this point, and up to the end of the contractions, $\vec{f}_{BR}(\tau)$ remains stable while $\vec{f}_{BB}(\tau)$ increases. Likewise, $\vec{F}_{BB}(\tau)$ and $\vec{F}_{BR}(\tau)$ continue increasing from their intersection point, but the BB reaches higher force values. Therefore, based on the above, the ~10% MVC seems to behave as a breakpoint concerning the load-sharing of the elbow flexors synergistic muscles, especially for the BB and BR. In this regard, the model retrieves the phenomenology described in previous works, about the significance of such a level of contraction intensity in the early load-sharing between the elbow flexors muscles during their isometric contraction [24, 27]. This could be related to the inflection point of the $C_i(\tau)$ curves of those muscles whose longitudinal deformation during isometric contraction describes cooperative positive dynamics, such as the BB and BR muscles between 0-30% MVC [27].

Based on all the above, the proposed model makes a significant contribution to the previous efforts for characterizing individual muscle forces by shear wave elastography [1, 22-24, 27]. Future developments of this model may have important derivations, both for basic research in muscle biomechanics, as well as for clinical applications in related fields.

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