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Short communication

# Improvements to Hoang et al.'s method for measuring passive length–tension properties of human gastrocnemius muscle *in vivo*

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ABSTRACT

While the passive mechanical properties of a musculo-articular complex can be determined using the relationship between the articular angle and the passive torque developed in resistance to motion, the properties of different structures of the musculo-articular complex cannot be easily assessed. Recently, an elegant method has been proposed to estimate the passive length–tension properties of gastrocnemius muscle–tendon unit (Hoang et al., 2005). In the present paper, two improvements of this method are proposed to decrease the number of parameters required to assess the passive length–tension relationship from 9 to 2. Furthermore, these two parameters have physical meaning as they represent a passive muscle–tendon stiffness index ($\alpha$) and the muscle–tendon slack length ($l_0$). $\alpha$ and $l_0$ are relevant clinical parameters to study the chronic effects of aging, training protocols or neuromuscular pathologies on the passive mechanical properties of the muscle–tendon unit. Eight healthy subjects performed passive loading/unloading cycles at $5^\circ$/s with knee angle at 6 knee angles to assess the torque–angle relationships and to apply the modified method. Numerical optimization was used to minimize the squared error between the experimental and the modeled relationships. The experiment was performed twice to assess the reliability of $\alpha$ and $l_0$ across days. The results showed that the reliability of the two parameters was good ($\alpha$: ICC=0.82, SEM=6.1 m$^{-1}$, CV=6.3% and $l_0$: ICC=0.83, SEM=0.29 cm, CV=0.9%). Using a sensitivity analysis, it was shown that the numerical solution was unique. Overall, the findings may provide increased interest in the method proposed by Hoang et al. (2005).

## 1. Introduction

In humans, the passive mechanical properties of a musculo-articular complex, including structures spanning the joint (Riemann et al., 2001), can be determined using the relationship between the articular angle and the passive torque developed in resistance to motion (Gajdosik, 2001; Magnusson, 1998; McNair et al., 2001, 2002; Nordez et al., 2008, 2009). However, these properties represent global passive joint mechanical properties, and the contributions of different structures of the musculo-articular complex cannot be easily determined. In that framework, Hoang et al. (2005) have recently proposed an elegant method to assess the passive length–tension properties of human gastrocnemius muscle–tendon unit *in vivo*. Therefore, this methodology has important applications for understanding the behavior of the muscle–tendon complex *in vivo* as demonstrated by recent works by the same group (Hoang et al., 2007a, b, 2009).

The method proposed by Hoang et al. (2005) is based on the reasonable assumption that the gastrocnemius muscle–tendon unit is the major structure that crosses both ankle and knee joints and that it produces a significant passive torque during the ankle motion. Experiments are then performed at different knee angles to obtain its influence upon the passive torque–angle relationships of the ankle. Thereafter, numerical optimization is used to minimize the squared error between the experimental and the modeled relationships. In the method of Hoang et al. (2005), 3 parameters are required to model each of the 3 contributions to the external torque: (i) mono-articular structures in plantar aspect of the ankle, (ii) mono-articular structures in dorsal aspect of the ankle and (iii) bi-articular structures crossing ankle and the knee joints (i.e., the gastrocnemius muscle–tendon unit). The passive force–length relationship of the *gastrocnemii* is then reconstructed using the level arm estimated from the Grieve's Model (Grieve et al., 1978) and the parameters determined for (iii). However, using such optimization with 9 parameters, it is quite difficult to verify the uniqueness of the numerical solution, and the parameters estimated for (i) and (ii) may influence the parameters of (iii). The presence of several numerical solutions may explain

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the lack of reliability for some parameters in the study of Hoang et al. (2005).

In the present paper, two techniques are utilized to improve the methodology proposed by Hoang et al. (2005). They involve decreasing the number of parameters required to reconstruct the passive force–length relationship. These parameters could be relevant for clinical studies that would aim to assess changes in the passive muscle–tendon mechanical properties.

## 2. Material and method

Eight healthy males (24.3 ± 3.0 years, height: 177.0 ± 7.8 cm and weight: 72.2 ± 7.7 kg) volunteered to participate in the present study. All procedures were approved by the University Ethics Committee and all subjects signed a document of informed consent.

After a familiarization session, the maximal range of motion in dorsiflexion was determined. The subject was placed into the dynamometer and the foot was slowly manually dorsiflexed. The subject stopped the motion when the maximal tolerable stretch was reached (Nordez et al., 2006, 2008). This test was repeated three times, and the maximal angle of these three trials was considered as the maximal range of motion in dorsiflexion. The maximal range of motion was determined with the knee set at 0° and 80°. Then, the subjects performed 5 loading/unloading cycles on a Biodex dynamometer (Biodex medical, Shirley, NY, USA) set in a passive mode at 5°/s at each of the following knee angles (Fig. 1A): 0, 15, 30, 45, 60 and 80° in a randomized order (0° = knee fully extended), with 5 min rest between each set of the 5 cycles. For all these loading/unloading cycles, the maximal angle in plantarflexion was 20° of the plantarflexion. The maximal angle in dorsiflexion for the cycles with the knee angle at 0° and 80° was set at 80% of the maximal range of motion previously determined at the corresponding knee angles. At 15°, 30°, 45° and 60° of the knee angle, the maximal angle in dorsiflexion was increased with a constant step between 80% of the maximal range of motion at 0° until 80% of the maximal range of motion at 80°. Since no significant changes in passive musculo-articular mechanical properties occurred between the fourth and the fifth cycle (McNair et al., 2002; Nordez et al., 2009), the fifth cycle was considered for the analysis for each knee angle test. Subjects were tested twice at the same hour with two days between sessions.

Bipolar electrodes (Delsys DE 02.3, Delsys Inc, Boston, USA) were placed on *gastrocnemius medialis*, *gastrocnemius lateralis*, *soleus* and *tibialis anterior* with a 10 mm interelectrode distance according to the recommendations of the surface electromyography for the non-invasive assessment of muscle project (SENIAM) (Hermens et al., 2000). Using electromyographic feedback, the subjects and the experimenter were able to visualize any muscle activity, and subjects were asked to be as relaxed as possible. Electromyographic activity was normalized to activity recorded from a maximal isometric contraction performed at the end of the testing session. A requirement was that EMG activity was less than 1% of that recorded during the maximal contraction (McNair et al., 2001, 2002). This was achieved in all subjects.

Mechanical (torque, angle and angular velocity provided by the Biodex dynamometer) and electromyographic signals were collected synchronously at 1000 Hz with analog/digital converter (Bagnoli 16, Delsys Inc, Boston, USA). All the collected data were stored in a computer hard drive for further analyses.

The improvements of the original methods were

1. An optimization procedure performed on the differences between the torque–angle relationships obtained at 0, 15, 30, 45, 60° and 80° of the knee angle (Fig. 1B). Thus, contributions of mono-articular structures (i) and (ii) were removed and the contributions of the gastrocnemius could be determined directly, according to the following equation:

$$T_k - T_{80} = m_G(F_{Gk} - F_{G80}) \tag{1}$$

where $T_k$ and $F_{Gk}$ are the ankle torque and the gastrocnemius force determined at different knee angles (0°, 15°, 30°, 45° and 60°); $T_{80}$ and $F_{G80}$ are the ankle torque and the gastrocnemius force determined at 80° of knee angle, respectively, $m_G$ is the gastrocnemius level arm assessed using Grieve et al. model (1978). Using this new optimization scheme, the number of parameters required to determine $F_G$ dropped from 9 to 3.

2. The exponential model used in the study of Hoang et al. (2005) to compute $F_G$ was replaced with the Sten–Knudsen Model (Sten-Knudsen, 1953; Nordez et al., 2006):

$$F_G = \frac{E_0}{\alpha}(e^{\alpha(l-l_0)} - 1) \quad \text{if } l > l_0 \tag{2}$$

$$F_G = 0 \quad \text{if } l < l_0$$

where $l$ is the gastrocnemius length (assessed using Grieve's model), $E_0$ and $\alpha$ are parameters determined using the optimization that concerns the muscle

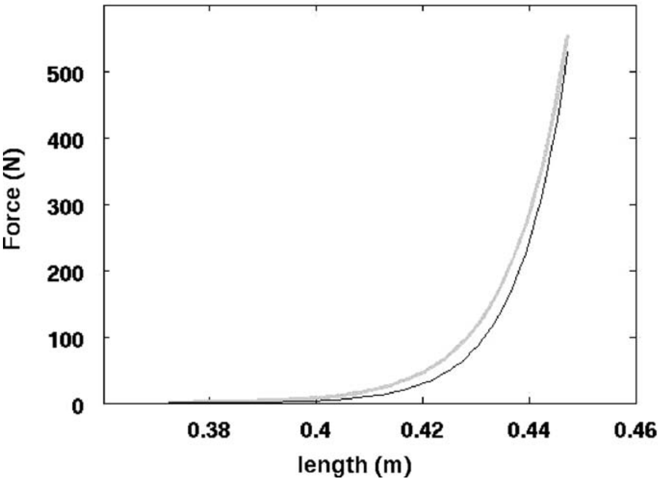

**Fig. 2.** Typical example of gastrocnemii's force–length relationships obtained using the original method of Hoang et al. (2005) (gray line) and the proposed method (black line).

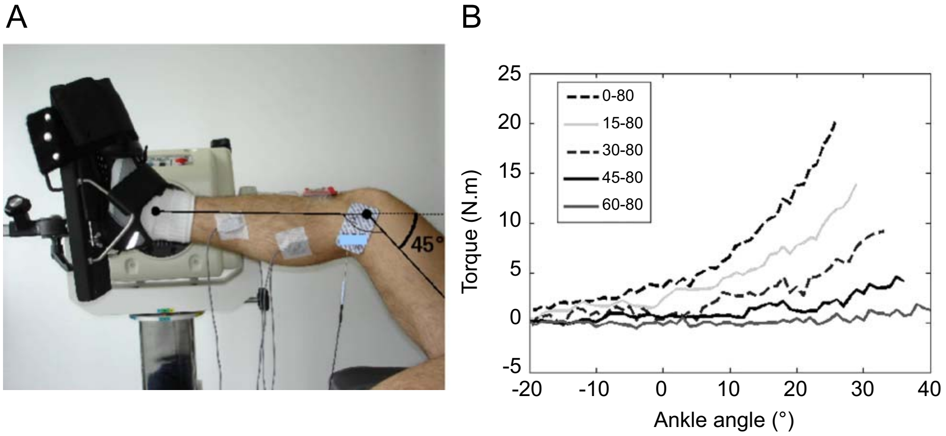

**Fig. 1.** (A) Position of the subject on the dynamometer and (B) typical example of plantarflexors muscles' torque–angle relationships: differences between torque–angle relationships obtained at the different knee angles and at 80° were calculated to perform the optimization (see text).

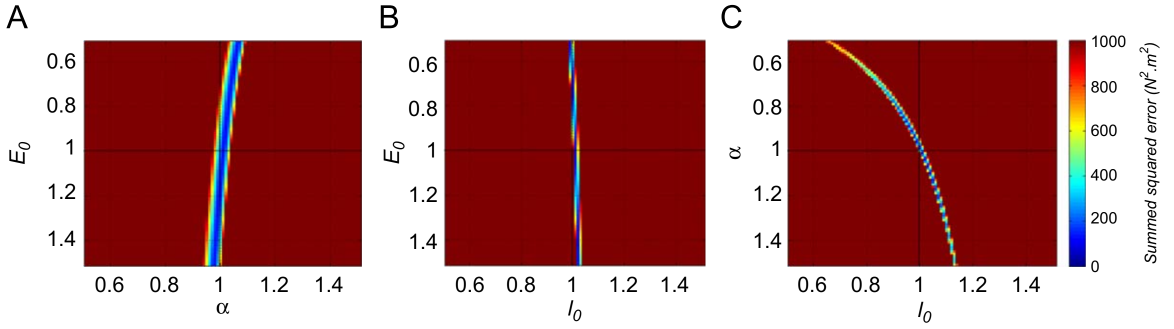

**Fig. 3.** Typical results of the sensitivity analysis. Color maps of the summed squared difference between the experimental and the modeled results as functions of $E_0$ and $\alpha$ (A), $E_0$ and $l_0$ (B) and $\alpha$ and $l_0$ (C) (see material and method section for parameter signification). Parameter values are expressed as the ratio of values found with the optimization (i.e., 1.0 represents the actual value found with the optimization).

stiffness, and $l_0$ the gastrocnemius slack length determined using the optimization.

Using this exponential model the gastrocnemius stiffness ($S_G$, i.e., the derivative of the force in respect to the muscle length) can be written as

$$S_G = \alpha \times F_G + E_0 \quad (3)$$

Therefore, the parameters of the Sten-Knudsen Model have physical meaning since $\alpha$ represents a stiffness index, $E_0$ represents the stiffness for a null force (Nordez et al., 2006) and $l_0$ the gastrocnemius slack length. It was then hypothesized that $E_0$ has a much lower influence than $\alpha$ and $l_0$. Thus, the number of important parameters required to determine the passive force–length relationship of the gastrocnemius muscle could be decreased to 2.

The 3 parameters of the Eq. (2) were determined by minimizing the squared difference between the experimental and the modeled (Eqs. (1) and (2)) responses using Matlab (The Mathworks, Natick, USA) and the optimization toolbox (Levenberg–Marquard algorithm). Initial values of the three parameters were chosen as 0.03 N m$^{-1}$, 70 m$^{-1}$ and 0.3 m for $E_0$, $\alpha$, and $l_0$, respectively. A pilot study had shown that the results were unaffected by these initial values. A sensitivity analysis was performed on the three remaining parameters to determine whether the solution was unique. The 9 parameters of the initial method were also calculated as described in Hoang et al. (2005) to compare the obtained force–length relationships.

## 3. Results

For each subject, the reconstructed force–length relationship assessed using the novel method slightly underestimated the force in comparison to the force obtained using the original method (Fig. 2).

Reliability was good for $\alpha$ (ICC=0.82, SEM=6.1 m$^{-1}$, CV=6.3%) and $l_0$ (ICC=0.83, SEM=0.29 cm, CV=0.9%) while the reliability was low for $E_0$ (ICC=0.49, SEM=0.26 N m$^{-1}$, CV=128%).

Fig. 3 is a typical example for one subject but similar results were found for all the subjects. It can be seen in Fig. 3A and B that $E_0$ has no notable influence on the force–length relationship, validating our initial hypothesis and showing that only two important parameters ($\alpha$ and $l_0$) were required to assess the passive mechanical properties of the gastrocnemius muscle–tendon unit. In addition, Fig. 3C shows that the numerical solution is unique.

## 4. Conclusion

It was demonstrated that the passive mechanical properties of the gastrocnemius can be determined by a unique numerical solution using only 2 reliable parameters ($\alpha$ and $l_0$) in comparison to the 9 parameters that are required in the method proposed by Hoang et al. (2005). Thus, our findings may provide increased interest in the method proposed by Hoang et al. (2005).

It can be noticed that the Hoang's method includes the extraction of the mono-articular structure contributions. These contributions cannot be directly assessed with the method proposed in the present paper. For that purpose, once the bi-articular contribution is assessed using the improved method, the mono-articular contribution can be easily estimated using a second optimization on the data removed by the differences of the Eq. (1).

Changes in the passive muscle force can be due to both muscle resting length and/or tissue stiffness changes. Therefore, $l_0$ and $\alpha$ could be used to dissociate changes related in muscle–tendon length and the changes in muscle–tendon stiffness, respectively. Thus, these parameters seem relevant for clinical studies that might be interested in the effects of aging, training protocols or neuromuscular pathologies on the passive mechanical properties of the muscle–tendon unit.

## Conflict of interest

None of the authors have any financial and personal relationships with other people or organizations that could inappropriately influence this work.

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