Selective muscle contraction during plantarflexion is incompatible with maximal voluntary torque assessment

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Abstract

Objective. Large variations in maximal voluntary torque are reported in the literature during isometric plantarflexion contractions. We propose that these differences, which could reach 40% across similar studies, could be explained by differences in the instructions provided, and notably by instructions as to favoring or not multi-joint contractions.

Method. Sixteen participants were placed on an isokinetic ergometer in 3 different positions, supine, prone and seated, with the ankle in the neutral position, and instructed to create maximal force on the footplate by conforming with instructions that favored either isolated (ISOL) or multi-joint (ALL) isometric contractions. Torque, foot kinematics and the electromyographic activity of seven muscles of the lower limb have been recorded.

*Result*s. Joint torques were greater in ALL compared to ISOL (p<.05) with gains of 43.5 [25.4 170.6]%, 42.5 [1.4 194.6]% and 15.3 [9.3 71.9]% in the supine, prone and seated position, respectively (values are given as median [range]). The results suggested that forces created by muscles that do not span over the ankle joint significantly influenced the measured joint torque, notably in the seated position. Nevertheless, the observed gains in torque were associated with greater plantarflexor muscles activation, showing that the ISOL condition may have induced a form of inhibition of these muscles.

Conclusions. The results of this study suggest that using isolated contractions, hence constrained testing protocols, cannot provide optimal conditions for MVC testing, notably for plantarflexor muscles, which seem to be extremely sensitive to such constrained conditions.

Keywords. EMG, maximal voluntary contraction; plantarflexion; multi-joint contraction; concurrent activation

list of abbreviations

ALL: multi-joint contractions condition

ANOVA: analysis of variance

CR: center of rotation

EMG: electromyographic signal

EMGmax: maximal electromyographic (EMG) value obtained over all conditions

GM: *gastrocnemius medialis*

Gmax: *gluteus maximus*

ISOL: isolated contractions condition

MVC: Maximal voluntary contraction

RF: *rectus femoris*

SCoRE: Symmetrical Centre of Rotation Estimation method

SCS: segment coordinate system

SD: standard deviation

Sol: *soleus*

ST: *semi tendinosus*

TA: *tibialis anterior*

VL: *vastus lateralis*

Introduction

Maximal voluntary contraction (MVC) torque is an important measure to evaluate mechanical properties of the muscle and their progress with physical training (Klass et al. 2008; Van Cutsem et al. 1998) or in rehabilitation, to assess the evolution of musculoskeletal diseases and to quantify the beneficial effects of different therapeutic strategies (Moraux et al. 2013; McNeil et al. 2007). MVC at the ankle joint is especially critical to consider due to the important role of the plantarflexor and dorsiflexor muscles in maintaining balance and avoiding fall (Horak et al. 1989). Still, MVCs evaluation requires several precautions to be taken, because mechanical and neural factors could greatly influence torque output. Therefore, the present study will focus on isometric plantarflexion MVCs.

Regarding mechanical factors, even though ergometers have proven to be reliable instrument *per se* (Drouin et al. 2004), many biases are known to affect measurements, such as, (i) gravitational effects, (ii) inertial effects, (iii) compliance of the ergometer moment arm or deformation of the footplate and fasteners compliance, or (iv) misalignments between the axis of rotation of the ergometer relative to that of the joint (Arampatzis et al. 2007; Herzog 1988; Deslandes et al. 2008), that, moreover, represents only an approximation of the actual functional axis of rotation of the joints (Ramos and Knapik 1978; Hicks 1953). In the isometric case, gravitational effects are easily eliminated and the inertial effects are supposed to be negligible (Deslandes et al. 2008). Compliance involves movement of the segment relative to the moment arm of the ergometer, and implies that muscles MVCs cannot be evaluated at the exact intended position. Adjustments can nevertheless be easily performed to correct positional changes observed when the muscles go from the passive to the active state (De Ruiter et al. 2008). Misalignment, on the other hand, is particularly critical for the evaluation of plantarflexion as compared to other group of muscles.

With some simplifications, torque at the ankle can be written (Arampatzis et al. 2007) as:

$$\tau_{ankle} = \frac{r_{ankle}}{r_{dynamometer}} \tau_{dynamometer} \qquad \text{(equation \#1)}$$

with $\tau_{\bullet}$ the torque at either the ankle or the dynamometer axis, and $r_{\bullet}$ the moment arm of the reaction force to either the ankle or the dynamometer. This relation could be rewritten as:

$$\tau_{ankle} = \left(1 + \frac{\Delta r}{r_{dynamometer}}\right) \tau_{dynamometer}; \Delta r = r_{ankle} - r_{dynamometer} \qquad \text{(equation \#2)}$$

highlighting that, for a given misalignment Δr (order of magnitude = 1cm), the bias is lower for knee extension testing (with large $r_{dynamometer}$ relative to Δr; $r_{dynamometer} \approx$ 30-40cm, (Arampatzis et al. 2004; Deslandes et al. 2008) than for plantarflexion testing ($r_{dynamometer} \approx$ 17cm, assuming that the forces on the footplate act at the level of the metatarsophalangeal joint of the big toe, Van Cutsem et al. 1998). In addition, with misalignment, a moment arm is created between the ankle joint and the axis of rotation of the ergometer, and the reactions forces at the level of the ankle joint can thus create a torque on the footplate without any torque on the foot. Moreover, these forces can be easily manipulated by the participant using forces created by muscles not crossing the ankle joint (e.g., knee or hip extensors), and these accessory muscles can then have a mechanical influence on the measured joint torque.

At least two neural factors should be considered in this juncture: motivation and *concurrent activation potentiation* (Ebben et al. 2008a; Ebben et al. 2010), also referred to as *remote voluntary contraction* (Cherry et al. 2010; Ebben et al. 2008b). Motivation is a well-known confound variable influencing performance which can be controlled following several recommendations (see Gandevia (2001) notably for a review). *Concurrent activation potentiation* is much less considered and captures the fact that contraction of accessory muscles (remote contraction) may increase the maximal activation level of primary movers (Ebben et al. 2008b). This phenomenon is commonly attributed to motor irradiation and/or to an increase in spinal excitability (Ebben 2006). Jaw clenching, Valsalva maneuver and hand gripping have been particularly investigated (see Ebben et al. (2008b) for a review), but muscles from adjacent sites also proves to interact with the primary movers (Barry et al. 2008; Devanne et al. 2002; Kouchtir-Devanne et al. 2012). It is therefore likely that muscles that do not span over the ankle joint have also a neural influence on plantarflexor activity and hence on plantarflexion torque.

Since accessory muscles may come into play at both the neural and the mechanical level, the aim of this study was to test the maximal torque produced in plantarflexion using two modalities of instructions aimed at manipulating the degree of involvement of muscles not crossing the ankle joint. Furthermore, the various positioning used in the literature, notably the seated (Moraux et al. 2013; Simoneau et al. 2009), prone (Cresswell et al. 1995; Maganaris 2003) and supine positions (Danneskiold-Samsøe et al. 2009; Simoneau et al. 2007) are likely to favour specific patterns of muscle activity, and thus to influence the results in a different way. Therefore, in this study these three positions have been tested. Offset of the rotation axes, ankle angle deviations and muscle activity have been recorded in order to set apart the neural and mechanical influences of the accessory muscles.

Materials and methods

Participants

16 healthy males participated to the study (mass=76.8 ± 8.5 kg, range =[68-92]; height=1.77 ± 0.07 m, range=[1.62-1.87]; age=26.9 ± 6.4 years, range=[20-41]). All of them were informed of the experimental procedures prior to giving their written consent to participate. The experimental design of the study was approved by the local ethical committee and the experiments were conducted in accordance with the Declaration of Helsinki (last modified in 2004).

General procedure

Ankle torque measurements were performed using an isokinetic dynamometer (Biodex III, Shirley Corporation, NY, USA). The right leg was evaluated in all participants. Participants were equipped of the reflective markers used for kinematic analysis and of the recording electromyographic (EMG) surface electrodes at their arrival at the laboratory (see details in sections *Kinematics* and *Electromyographic acquisition*). We first estimated the position of the real center of rotation of the ankle by moving passively the ankle on the dynamometer footplate in a procedure described in section *Estimation of the ankle rotation axis and center of rotation*. Afterward, participants performed a warm up lasting 5 minutes which consisted of submaximal isometric plantarflexor contractions while seated on the ergometer. Participants were then successively placed in the PRONE, SUPINE or SEATED position in a random order to assess their isometric MVCs. For each of these positions two modalities of instruction were randomly given to the participants. These constitute a total of 3x2=6 randomized conditions and for each of them 3 tries were given to the participant, resulting in a total of 18 MVCs.

Kinematics

A motion analysis system (Vicon Motion System, Lake Forest, CA) equipped with 11 infrared cameras recorded the 3-dimensional position of 11 reflexive markers stuck on the participant and on the dynamometer. Markers were positioned on the right side of the body at the level of the external and internal *maleolli*, *calcaneous* (posterior point of the heel), 1st and 5th Metatarsal Head, fibula's head and tibiale's tuberosity. *4 r*eflexives markers were placed on the dynamometer such that the mid distance between two of the markers corresponds to the position of the dynamometer axis of rotation and that the two others, placed in a more backward position, allowed to recover the direction of this axis. Kinematic data were recorded at a sampling frequency of 200 Hz. Ankle angle represents the angle between the vector going from the *calcaneous* to the midpoint between 1st and 5th Metatarsal Head, and the vector going from the midpoint between fibula's head and tibiale's tuberosity to the midpoint between the two *maleolli* (see Figure 1).

Electromyographic acquisition

Surface EMG was recorded from 7 muscles located on the right side of the body, namely, *tibialis anterior* (TA), *soleus* (Sol), *gastrocnemius medialis* (GM), *vastus lateralis* (VL), *rectus femoris* (RF), *semi tendinosus* (ST) and *gluteus maximus* (Gmax). Prior to electrode application, the skin was shaved and cleaned with alcohol to minimize impedance. Pairs of Ag-AgCl disk electrodes of 8mm diameter with inter electrode-distances of 2cm were placed longitudinally with respect to the underlying muscle fibers arrangement according to the recommendations of Surface EMG for Non-Invasive Assessment of Muscles (SENIAM) (Hermens et al. 2000). The references electrode was placed at the level of the great trochanter. EMG signals were amplified (× 1000), digitized (6-400 Hz bandwidth) at a sampling rate of 1kHz (Biopac System Inc. Goleta, USA), recorded and synchronized using the motion analysis system.

Conditions of MVC testing and recording

The ankle joint torque was acquired with the isokinetic dynamometer and digitally synchronized at a sample rate of 1 kHz using the motion analysis system. During MVCs, participants were positioned on the ergometer and securely stabilized by using two crossover shoulder harnesses and a belt across the abdomen. The right foot was strapped securely to the footplate with the ankle fixed at an angle of 90° i.e., at the neutral position with the sole of the foot perpendicular to the shank, and held in place by a heel block. The axis of the dynamometer was aligned with the anatomical ankle flexion-extension axis, estimated as the line passing through the tips of the *maleolli* (Wu et al. 2002; Lundberg et al. 1989). A clear start and stop signals were given. Each voluntary contraction lasted approximately 3-4 s and 1 minute of rest were given between each contraction (Todd et al. 2004). Participants received no feedback of their performances during the tests.

Positions

Three positions were tested, PRONE, SUPINE and SEATED. For PRONE and SUPINE positions the participants were lying on the dynamometer chair with the hip and the knee fixed at an angle of 0° (=full extension for both). In these positions the thigh was stabilized using a belt. For the SEATED position, the chair was lifted up at an angle of 90° from the horizontal and the knee and hip joints were both placed at an angle of 90°.

Instructions

For each position, MVCs were performed with two different modalities of instructions named ISOL and ALL. In the isolation condition (ISOL), participants were required to produce a force by rotating the footplate as hard as possible and to handle the shoulder harnesses. In this condition, they were invited to use only their calf muscles. In a second condition (ALL), the participants were invited to grip the ergometer handle and to use all the possible means to create forces against the footplate.

Estimation of the ankle joint rotation center

The ankle joint rotation center was estimated using the Symmetrical Centre of Rotation Estimation (SCoRE) method (Ehrig et al. 2006). Briefly, the position of the center of rotation (CR) between two segments is determined by assuming a constant contact point between each and use the relation

$$CR = \mathbf{o}_1 + \mathbf{R}_1\mathbf{u} = \mathbf{o}_2 + \mathbf{R}_2\mathbf{v} \qquad \text{(equation \#3)}$$

where $\mathbf{o}_1$ and $\mathbf{o}_2$ are arbitrary points on segments #1 (the foot) and #2 (the leg), $\mathbf{R}_1$ and $\mathbf{R}_2$ are the rotation matrix transforming the segment coordinate system (SCS) to the global coordinate system and $\mathbf{u}$ and $\mathbf{v}$ are the vector linking respectively $\mathbf{o}_1$ and $\mathbf{o}_2$ to CR in the foot and leg coordinate system respectively. The SCSs were defined according to Wu et al. (2002). For the estimation of the CR position, participants were seated on the ergometer chair with solely their right foot strapped on the footplate connected to the moment arm of the dynamometer and the ankle joint was moved passively at full but comfortable range of motion for about 10 flexion-extension cycles in order to localize an accurate joint center. The values of $\mathbf{u}$ and $\mathbf{v}$ were then used to estimate the position of the CR relative to the SCSs (foot and leg) in all experimental conditions.

Data analysis

EMG signals were filtered with a bandpass filter (4th order Butterworth) between 20 and 400 Hz. Linear envelopes for each muscle were obtained by low-pass filtering the fully rectified raw EMG signals with a 9 Hz low-pass filter (2nd order Butterworth, zero lag, (Shiavi et al. 1998). For each condition, the averaged value between -150 ms and 150 ms around the peak torque event was extracted (Figure 2) and then normalized by the maximal value obtained over all conditions (=EMG_{max}). These calculations were performed for each muscle and each participant independently.

Joint torque and kinematic data were filtered by a 15 Hz low pass filter (2nd order Butterworth filter (Winter 1990)). Joint torque was corrected for gravity by subtracting the baseline, and for each condition the maximal value reached over the three tries given to the participant was extracted for analysis (Figure 2).

Statistics

Normality of the data has been checked using Shapiro-Wilk's tests. For normally distributed data two-way repeated measure ANOVAs (instruction=ALL and ISOL × position=SUPINE, PRONE and SEATED) were performed after checking for violations of sphericity using Mauchly’s test. Post-hoc analyses were then performed using Bonferroni method (Maxwell 1980). For non-normal distribution non parametric Friedman ANOVAs (one-way repeated measures ANOVA on ranks) was chosen. Wilcoxon rank sum tests associated with Bonferroni-Dunn corrections were used when the null hypothesis was rejected.

The different biases mentioned in the introduction were rallied in kinematic deviations. They include i) the ankle angle changes (in degrees) during the test due to the compliance of the ergometer moment arm, deformation of the footplate and fasteners compliance; ii) the alignment errors (in mm) between the axis of the dynamometer and the functional ankle joint centre of rotation in horizontal and vertical axis during the rest and the MVC. Kinematic deviations were compared to the reference using one-sample Student’s t-tests (reference value=0). A description of the axes is given in Figure 1.

We assessed the relationships between torque and other variables (i.e., kinematic deviations and EMG activity) using Pearson's correlation coefficient (r). For these analyses, values of each variable and for each participant were converted to z-scores, calculated by subtracting the average (over all conditions) and dividing the result by the SD. Because correlation analysis is very sensitive to the presence of outliers in the data (Chatterjee and Hadi 1986), normality of each variable was checked and values of |z-score|>2.58 (corresponding to the 99^{th} percentile of the distribution) were discarded from the analysis (Burke 2001). All available data were used (3 tries x 3 positions x 2 instructions x 16 subjects).

All statistical analyses were performed with the Statistica® software (Statistica®V6, Statsoft, Maison-Alfort, France). Values are reported as mean ± SD for normally distributed data and as median [range] instead. A *p*-value below .05 was considered statistically significant.

Results

Torque

The results showed that MVCs were significantly affected by the positions ($F(2,30)=13.2$, $p<.001$, $\eta_p^2=.60$) and the instructions provided ($F(1,15)=54.7$, $p<.001$, $\eta_p^2=.80$; Figure 3). Post-hoc analyses showed that MVCs were significantly greater in the SUPINE and PRONE positions compared to the SEATED position (pooled data: SUPINE=146.0 ± 40.5 N.m and PRONE=145.7 ± 38.9 N.m *vs.* SEATED=118.5 ± 31.2 N.m, $p<.001$). Torque was greater in the ALL condition compared to the ISOL condition for each position ($p<.001$), corresponding to gains of 43.5 [25.4 170.6]%, 42.5 [1.4 194.6]% and 15.3 [9.3 71.9]% for the SUPINE, PRONE and SEATED position respectively. Gains were significantly lower in SEATED compared to SUPINE ($Z=3.15$, $p<.001$) and PRONE ($Z=2.43$, $p=.015$) but were similar between SUPINE and PRONE ($Z=1.55$, $p=.121$).

Muscle activation

EMG variables were not normally distributed. The activity level of TA was greater in ALL compared to ISOL (pooled data: ALL=89.6 [6.5 100]% *vs.* ISOL=65.3 [2.7 100]%, $Z=3.21$, $p=.001$) but this effect was present in the SUPINE and PRONE positions only ($Z=3.31$, $p<.001$, and $Z=2.53$, $p=.011$ respectively). Analysis revealed no main effect of the position on the activity of TA ($\chi^2= 4.75$, $p=.093$; Figure 4).

Overall positions, Sol activity was greater in ALL compared to ISOL (ALL=85.7 [6.0 100]%) *vs.* ISOL=66.1 [0.7 100]%, $Z=2.84$, $p=.004$) but post-hoc analysis revealed significant differences in the SUPINE position only ($Z=2.43$, $p=.015$). There was no main effect of the position on the activity of SOL ($\chi^2= 2.25$, $p=.325$).

Activity of GM was greater in ALL compared to ISOL (i.e., pooled data: ALL=88.1 [15.8 100]% *vs.* ISOL=61.5 [1.9 98.2]%; Z=4.24, $p<.001$). These differences held for the SUPINE and PRONE positions (Z=2.84, $p=.004$ and Z=2.74, $p=.006$ respectively) but no differences were found in the SEATED position (Z=1.76, $p=.08$). Analysis revealed a main effect of the position ($\chi^2=17.69$, $p<.001$) i.e., GM was significantly less activated in the SEATED position (=49.9 [15.3 100]%,) compared to the PRONE (=79.4 [1.9 100]%., Z=3.78, $p<.001$) and SUPINE (=0.83 [0.09 1]%, Z=3.72, p<.001) positions.

ST was maximally activated in the PRONE position in 10 out of 16 participants. The activity of ST was significantly higher in ALL compared to ISOL in SUPINE (Z=3.46, p<.001) and SEATED (Z=3.00, $p=.003$), but no differences were found in the PRONE position (Z=0.67, $p=.502$). A main effect of the position was found ($\chi^2=6.94$ p =.03116) i.e., there was higher ST activity in the PRONE position but differences were significant only when compared to the SUPINE's (i.e., pooled data=74.6 [3.6 100]% *vs.* 8.0 [0.8 100]%, Z=3.22, $p=.001$).

VL, RF and Gmax activities possess the same patterns among the experimental conditions and were maximally activated in the SEATED position in most participants i.e., in 15, 13 and 15 out-of 16 participants respectively (Figure 4). Friedman ANOVA confirmed the effect of position on VL, RF and Gmax (VL: $\chi^2= 31.75$, $p<.001$; RF: $\chi^2= 17.44$, $p<.001$; Gmax: $\chi^2= 38.31$, $p<.001$). These muscles were significantly more activated in the SEATED compared to the SUPINE and PRONE positions (merged values in SEATED position: VL=97.1 [1.4 100]%, RF=62.5 [0.6 100]% and Gmax=90.7 [1.3 100]%) *vs.* SUPINE + PRONE: VL=6.2 [0.3 90.5]%, RF=6.8 [0.3 100]% and Gmax=5.1 [0.4 54.9]%, Wilcoxon Z-values ranged from 3.23 to 4.75, p<.001). Analyses indicated that the activity of VL, RF and Gmax were greater in the ALL compared to the ISOL condition in the 3 positions tested (Wilcoxon Z-values and p-values ranged from Z=2.84, $p=.004$ to Z=3.51, $p<.001$; Figure 4).

Kinematic deviations

Kinematic results and statistics are summarized in Table 1. During MVCs the ankle joint angle varied of -9.73 ± 4.15° in average i.e., from 91.0 ± 5.2° to 81.3 ± 4.9° overall conditions. From rest to MVC and overall conditions, the CR varied on $\mathbf{e}_x$ of ΔX=+14.8 ± 8.0mm (i.e., from Xrest=9.7 ± 22.1mm to XMVC=24.7 ± 24.6mm) and of ΔY=-0.71 ± 9.6mm on $\mathbf{e}_y$ (i.e., from Yrest=3.9 ± 13.1mm to YMVC=3.0 ± 15.0mm).

Correlations

All the results on correlation analyses are summarized in Table 2 and indicated that torque was significantly correlated with the activity of the plantar flexors in each position. VL, RF, ST and kinematic variables (ΔY and the variation in joint angle, Δθ) were found to be significantly related to torque depending on the position and on the instruction (see Table 2).

Discussion

The aim of this study was to point out the differences in torque output during maximal voluntary contraction (MVC) in isometric plantarflexion when activating either isolated or global muscle (conditions named ISOL and ALL respectively). The ALL condition was associated with higher EMG activities in most of the recorded muscles, notably in plantarflexor muscles, and was associated with higher joint torque compared to ISOL.

Very large differences were observed between ALL and ISOL, with gains on joint torque of about 40% in average (Figure 3). Lower torque in seated position could be attributable to muscle mechanics, i.e., force-length relationships (Maganaris 2003), and to impairments in motor units recruitment, as already reported for this particular joint angle configuration (i.e., knee and ankle joint angles set at 90°of flexion) (Cresswell et al. 1995; Kennedy and Cresswell 2001). In mine with our findings, a previous study showed that plantarflexion torque could be significantly enhanced (~+26%) in multi-joint compared to isolated plantar flexion (Hahn et al. 2011). However, this study remained inconclusive regarding the differences in EMG activity resulting from these two conditions and used different methodologies to assess joint torque in the multi-joint and isolated contractions, i.e., they used inverse dynamic calculations and ergometer measurements, that proved to provide different results (Herzog 1988; Kaufman et al. 1995; Arampatzis et al. 2004). Sasaki et al.(1998) observed an increase in plantarflexion torque linked to jaw clenching, but the conclusions relied on integrated electromyographic activity per unit of time rather than EMG level, and did not check for the influence of mechanical factors, as they focused on jaw clenching only.

Interestingly, the value of 40% found in the present study fits well to differences with that observed in similar studies examining ankle MVC, that is, values ranging from 134 up to 186

N.m, despite similar populations and protocols (Danneskiold-Samsøe et al. 2009; Cresswell et al. 1995; Maganaris 2003). More precisely, considering isometric plantarflexion MVCs in the supine and prone positions, and a population of young male adults, literature reports MVC values ranging from ~134 N.m [e.g. 142 ± 42 , N=10 (Danneskiold-Samsøe et al. 2009) or 134 ± 23 N.m , N=10 (Cresswell et al. 1995)] up to ~186 N.m [e.g. 172 ± 15 , N=8 (Maganaris 2003) or 186 ± 28 N.m (N=9) (Hahn et al. 2011)]. These differences may pertain to differences in the participants' fitness (i.e., more or less trained participants), but the results suggest that an explanation also bears on the nature of the instructions (ALL vs. ISOL).

Mechanical factors

Misalignment has been shown to induce bias of ~10% in the estimation of joint torque (Arampatzis et al. 2007; Deslandes et al. 2008), but this factor is not expected to create large differences among studies, as the positioning of the foot is expected to be carefully executed

Given the equation #1, positive deviations of CR in the x-direction, that decreases $\frac{r_{ankle}}{r_{dynamometer}}$, decreases the effectiveness of the ankle torque. The misalignments observed in this study on the X-axis are positive and then, they are not likely to explain the gains in torque. Nevertheless, misalignments may allow the auxiliary muscles, through joint reaction forces transmission, to influence the ankle torque.

One can first observe that positioning have an effect on the activity of knee extensors, knee flexors and hip extensors muscles (Figure 4). For example, the seated position was associated with higher level of activity for VL, RF and Gmax and the prone position was associated with higher ST activity (Figure 4) but the correlation values between these muscles and the torque produced remained modest (Table 2), suggesting that the mechanical influence of these muscles is small. Furthermore, despite the fact that differences were observed in ST activity between

PRONE and SUPINE, no differences were found in torque between these two positions. Additionally, the large increase in Gmax, VL and RF activity (Figure 4) did not preclude to the force deficit associated with the seated position (Figure 3). As a consequence, no major mechanical influence of these muscles is expected on the produced torque. Notwithstanding, at least two observations forbid ruling out the influence of such forces. Firstly, despite significant and positive correlations of the activity of Sol and GM with torque in the seated position (Table 2), the increase in torque was not associated with significant increases in the activity of these muscles i.e., SOL (p=.255) and GM (p=.079). And in this particular position, VL, RF and ST were also correlated with torque (Table 2). These observations strongly suggest that the forces created by muscles that do not span over the ankle joint significantly influenced the measured joint torque, at least in the seated position. Secondly, provided that torque is mainly related to plantarflexor activity, as the relation between EMG and torque tends to be convex toward tension at high force levels (Perry and Bekey 1981; Lawrence and De Luca 1983), a given increase in torque in this portion of the curve should have been associated with a larger increase in EMG and not with a similar one (Figure 3 and Figure 4). Suggesting that even in the supine and prone positions, plantarflexors are not the sole contributors of the increase in torque.

Neural factors

What can explain the higher muscle activity level in ALL compared to ISOL observed in this study? First, motivation is not likely to explain the differences observed between ALL and ISOL. Although motivation has not been explicitly assessed in this study, differences in motivation are not expected, because the tests were randomized. Additionally, we found high reliability between the 3 trials within all the conditions tested, with an average correlation coefficient of .95 (range=[.90 .98]; model corresponding to case #1 in McGraw et al. (1996)) and an average coefficient of variation of 5.8% (range=[3.2 8.2]). These reliability values are highly consistent with previous reports testing MVCs in plantarflexion (Webber and Porter 2010; Todd et al. 2004; Sleivert and Wenger 1994) and can be taken as evidences that MVC testing conditions carried out here can be truly compared to those imposed in previous studies. As a consequence, the lower values observed in ISOL are not likely to be due to a lack of motivation from the participants.

In ALL, participants were allowed to grasp the ergometer, which is not generally allowed in studies measuring ankle MVCs (cf. a representative setup in Figure #1 of Simoneau et al, (2009), so that *concurrent activation potentiation* could be induced (Ebben et al. 2008a; Ebben et al. 2010; Cherry et al. 2010; Ebben et al. 2008b). Jaw clenching or Valsalva maneuver have been reported to improve the level of maximal activation of the contracting muscles (Ebben et al. 2008b; Sasaki et al. 1998). However, most of these studies focused on the knee extensors muscles (Ebben et al. 2008b; Ebben et al. 2010), and not on plantarflexors. Furthermore, improvements in peak torque due to jaw clenching and Valsalva maneuver have been reported to be of ~15% for the quadriceps muscles (Ebben et al. 2008b; Ebben et al. 2010), that is, far less than the differences observed in the present study (i.e., ~40%). This may suggest a particular sensitivity of the plantarflexors to the phenomenon. In fact, contrary to other muscles such as elbow flexors (Herbert et al. 1998) or ankle dorsiflexors (Kent-Braun et al. 2002), activation of plantarflexor muscles is rarely complete (Todd et al. 2004; Cresswell et al. 1995). Without training or adequate testing conditions, plantarflexor muscles are not maximally activated by volition, and only reach about 80-90% of voluntary activation (Cresswell et al. 1995; Maffiuletti et al. 2002). This is in line with the finding that the neural drive to these muscles can be significantly improved by a strength training (Shield and Zhou 2004) or imagined strength training (Zijdewind et al. 2003; Sidaway and Trzaska 2005), whereas such is not the case for the elbow flexors, for example which possess a high initial level of voluntary activation (Herbert et al. 1998). The work of Devanne and collaborators (Devanne et al. 2002; Kouchtir-Devanne et al. 2012) is particularly interesting in this respect. They observed that the excitability of the cortical neurons associated with the *first dorsal interosseus* was lower during isolated (contraction of *first dorsal interosseus* only) vs. global muscle contractions (precision grip implying the thumb and the finger). This indicates that the cortical excitability of a given muscle depends on its functional interconnections at the cortical level. These findings support the idea that isolated contractions, which explicitly or implicitly (through instructions) require a selection of the contracting muscles, may induce inhibition, incompatible with the objectives of MVC testing. Allowing global muscle activation or not is then a critical aspect of the instructions

Conclusions

This study reports that activation of plantarflexor muscles are superior during global muscle activation compared to isolated joint contraction, entailing very large differences in motor-output. It emphasizes the pertinence of using isolated vs. unconstrained MVC testing protocols, notably for muscles that are not maximally activated by volition.

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Tables

	SUPINE		PRONE		SEATED	
	ALL	ISOL	ALL	ISOL	ALL	ISOL
X rest	**22,5 ± 13,5**	**20,9 ± 14,6**	**-14,8 ± 11,3**	**-14,7 ± 16,0**	**20,0 ± 15,3**	**22,1 ± 16,4**
Y rest	3,9 ± 9,7	4,4 ± 10,8	-5,2 ± 12,8	-6,0 ± 11,3	**12,4 ± 10,4**	**15,4 ± 9,7**
X MVC	**39,3 ± 14,1**	**36,1 ± 13,0**	-2,6 ± 13,6	-4,4 ± 17,0	**41,0 ± 18,0**	**36,2 ± 15,4**
Y MVC	-0,7 ± 10,3	**11,6 ± 11,8**	**-12,5 ± 12,5**	1,5 ± 10,1	1,9 ± 14,1	**18,1 ± 12,1**
ΔX	**16,9 ± 5,8**	**15,2 ± 7,3**	**12,2 ± 12,7**	**9,8 ± 4,9**	**21,0 ± 5,3**	**14,1 ± 4,9**
ΔY	-4,6 ± 4,7	**7,3 ± 3,5**	**-7,1 ± 8,1**	**7,5 ± 7,5**	**-10,6 ± 9,3**	2,7 ± 5,4
Δθ ankle	**-6,71 ± 3,81**	**-7,70 ± 2,67**	**-9,51 ± 3,32**	**-9,63 ± 4,80**	**-14,29 ± 3,14**	**-10,70 ± 2,51**

Table 1. ***Kinematic variables***. X and Y are the position of the ankle joint (estimated by the SCoRe method) relative to the axis of rotation of the dynamometer in the x-direction and y-direction at rest (**X rest** and **Y rest**) and at the peak torque event (**X MVC** and. **Y MVC**), given in mm. ΔX = X MVC- X rest, and ΔY =Y MVC- Y rest. Δθ is the difference in joint angle in degree between MVC and rest, in degree. Bolded values indicate a significant difference from 0 (t-test for single mean; *p*<.05).Values are given as mean ± SD.

	SUPINE		PRONE		SEATED	
torque vs.	**ISOL**	**ALL**	**ISOL**	**ALL**	**ISOL**	**ALL**
TA	0.10	**0.55****	0.29	**0.67****	0.29	0.10
Sol	-0.11	**0.42****	**0.41***	**0.64****	**0.59****	0.18
GM	**0.40***	0.16	0.30	**0.57****	**0.42***	**0.55****
VL	-0.28	-0.11	0.22	0.25	**0.38***	**0.35***
RF	-0.24	-0.10	0.19	**0.34***	0.24	0.16
ST	**0.31***	0.16	0.17	0.00	**0.54****	0.27
Gmax	-0.29	-0.19	0.22	0.19	0.16	0.04
ΔX	0.17	0.01	0.23	0.03	0.27	0.24
ΔY	**0.30***	-0.17	-0.08	-0.24	-0.06	-0.29
Δθ	-0.07	0.21	-0.28	-0.21	-0.26	**-0.37***
N	46	47	36	37	42	37

Table 2. ***Correlation coefficients.*** N refers to the number of values used to compute the Pearson's r. Bolded values indicate significant correlations (*:p<.05;**:p<0.001).

Figures

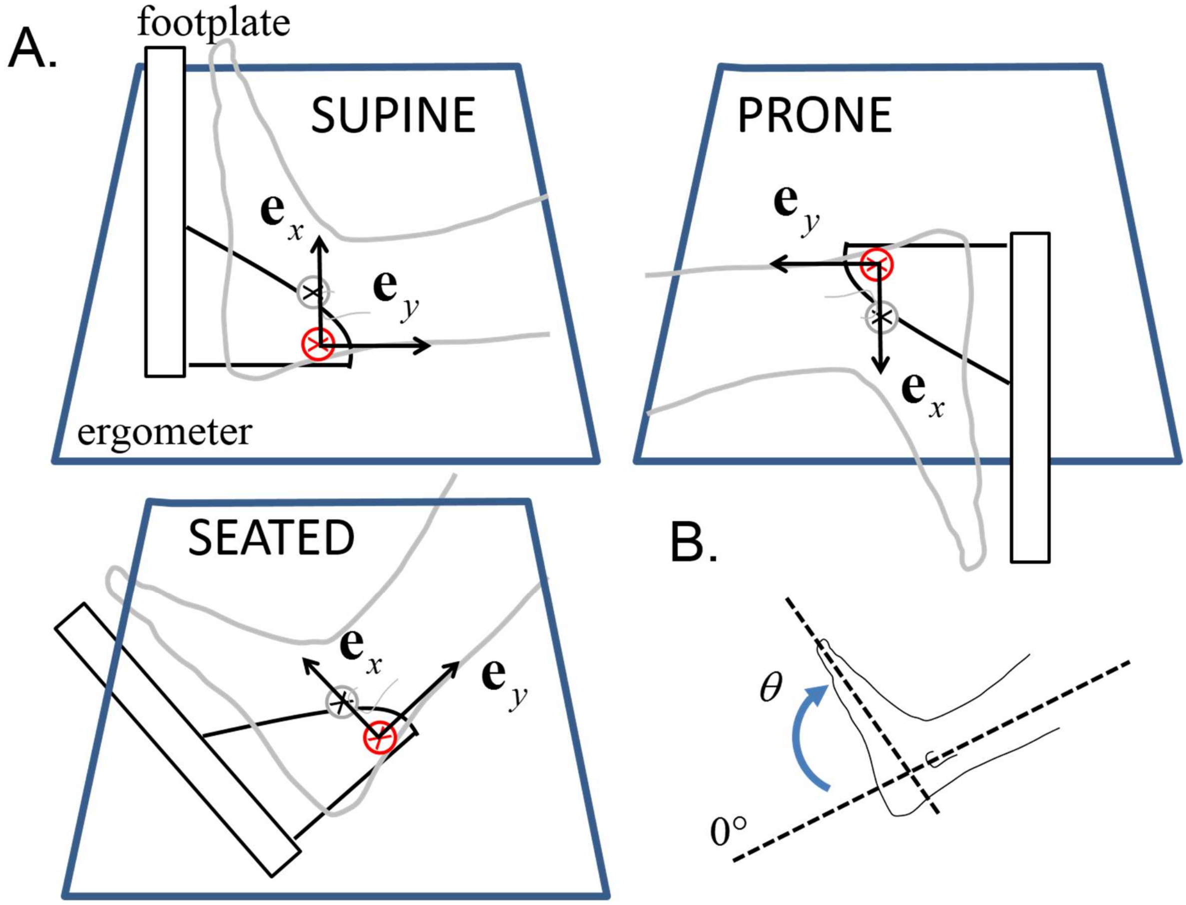

Figure 1. Position of the foot relative to the ergometer. **A.** The circled red cross designs the ergometer axis of rotation and the circled black cross the ankle axis of rotation (=CR). In all positions, the X-axis associated with the vector $\mathbf{e}_x$ is the axis parallel to the footplate and pointing toward the participant toes and the Y-axis asociated with the vector $\mathbf{e}_y$ is the axis orthogonal to it and pointing toward the participant leg. The origin is centered at the level of the ergometer axis of rotation. **B.** Definition of the ankle angle (θ).

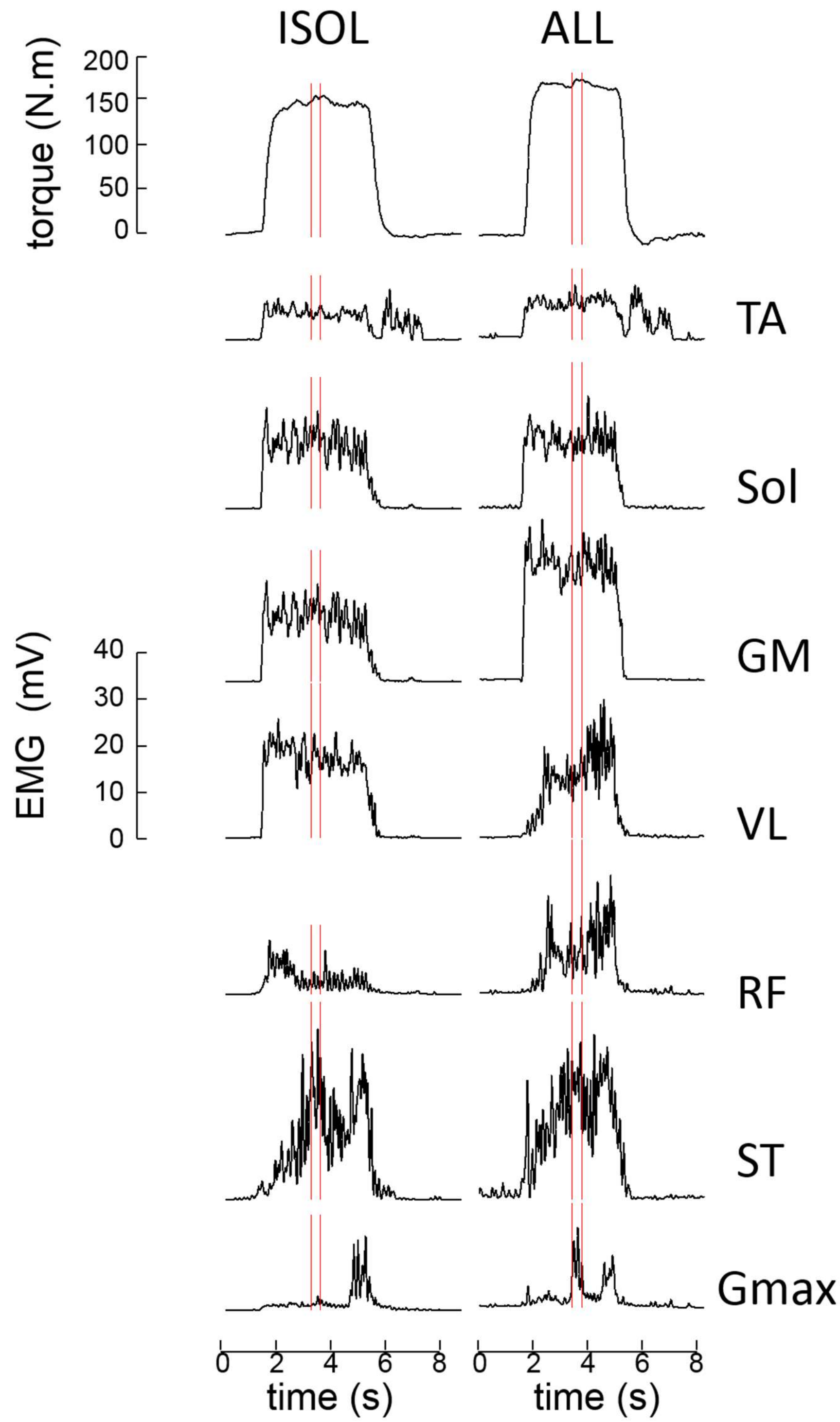

Figure 2. Example of torque and EMG data for a typical participant. Condition = supine position. Smoothed torque and EMG envelope are processed as indicated in section *Data analysis*. Vertical lines indicate the region around peak torque used for analysis.: *tibialis anterior*. **SOL**: *soleus*. **GM**: *gastrocnemius medialis* **VL**: *vastus lateralis*. **RF**: *rectus femoris*. **ST**: *semi tendinosus*. **Gmax**: *gluteus maximus*.

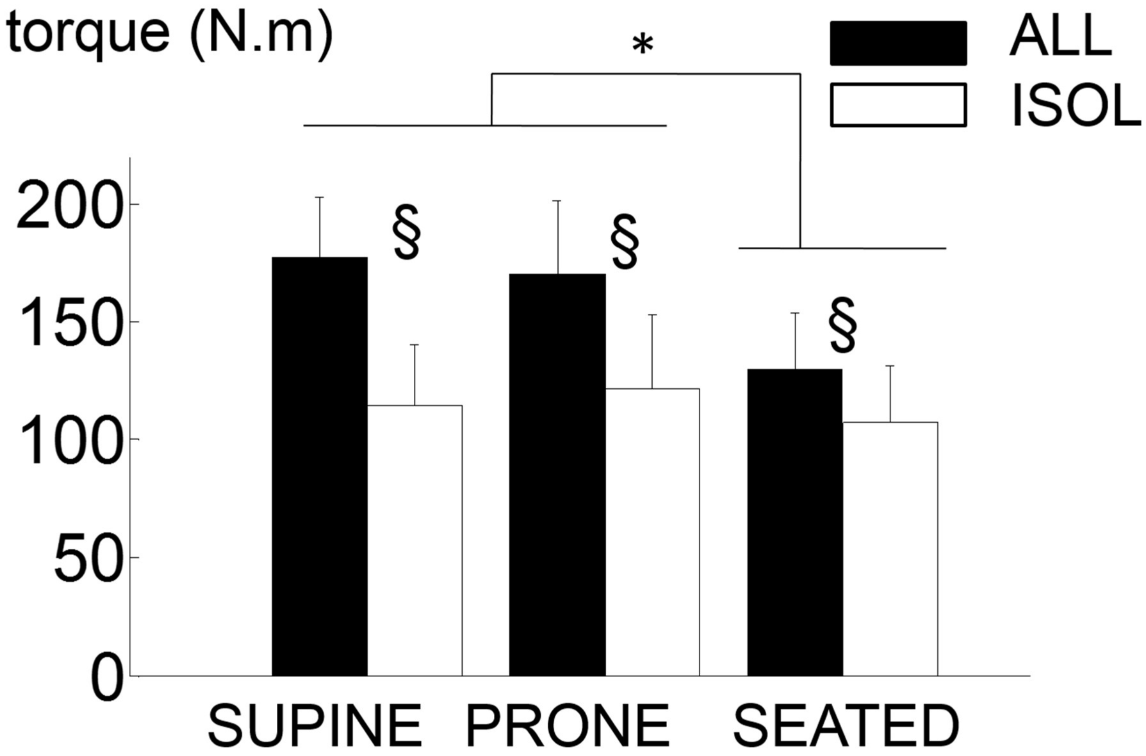

Figure 3. **Torque data**. Bars represent the mean and error-bars one SD. § indicates a significant difference (p<.05) between ALL and ISOL. * indicates a significant difference between [SUPINE and PRONE] *vs*. SEATED (*p*<.05). See section RESULTS for details.

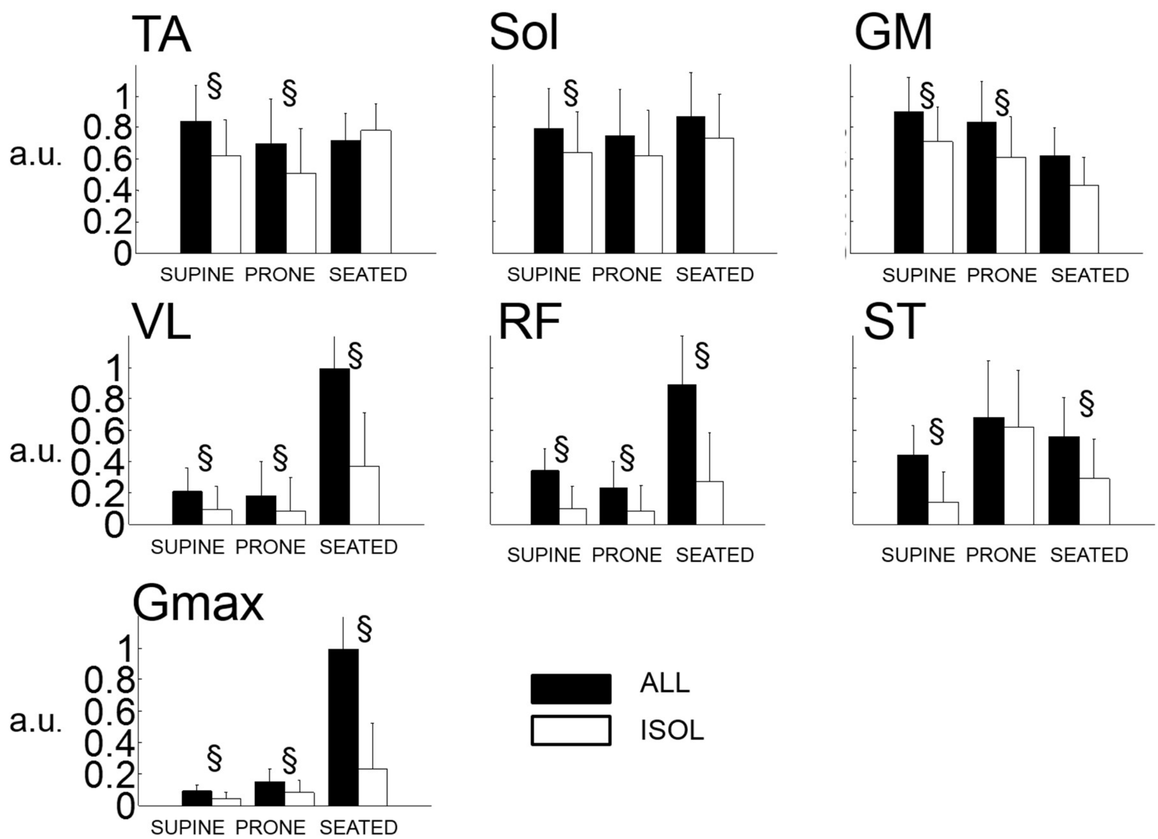

Figure 4. **Normalized EMG activities**. Bars represent the mean and error-bars one SD. § indicates a significant difference (*p*<.05) between ALL and ISOL.