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Research article

DOMINANCE-INDUCED MODIFICATIONS ON MAXIMAL FORCE AND NEURAL ACTIVATION OF THE ANKLE MUSCLES

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Abstract. *The purpose of the present empirical study was to assess the differences in dominance between lower limb maximal voluntary contraction (MVC) and neural activation. Twenty active and right-leg dominant (age: 31.3 ± 9.5 years, height: 178.2 ± 7.6 cm, weight: 76.5 ± 11.0 kg) participants performed 3 maximal dorsal flexions (DF) and plantar flexions (PF) at 3 ankle angles (75°, 90°: anatomical position, and 105°) which corresponded to short, intermediate and long lengths for DF muscles and the opposite for PF muscles. Electromyography (EMG) was used to assess ankle muscle activity (tibialis anterior, gastrocnemius medialis and soleus). The results showed non-significant differences between lower limb MVC force and EMG-muscle activation. However, a significant main effect of the angle was observed. During DF, the MVC force was greater ($p < 0.05$) at 90° and 105° than at 75° for both legs and during PF, the MVC force was greater ($p < 0.05$) at 75° and 105° than at 90° for both legs. Moreover, during DF and PF, the EMG-muscle activation was greater ($p < 0.05$) at 105° than at 75° an 90° for both legs. The results indicate that dominance was not associated with different levels of force and neural activation during maximal voluntary contraction with the ankle muscles. It is concluded that dominance does not have an impact on maximal strength and neural activation of the ankle muscles and any mechanisms that contribute to the dominance effect were not evident within this experimental protocol.*

Key words: *Leg dominance, maximal voluntary isometric contraction, electromyographic muscle activation, dominant leg, non-dominant leg, force, lower limb*

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1. INTRODUCTION

Leg dominance is an often-discussed factor amongst both healthy and injured athletes. The term of dominancy is defined as the preferred use of one side of the body, which shows superiority in movement performances than the other side of the body (Hebbal, & Mysorekar, 2003). Empirical evidence supports the claim that maximal force in a right-handed person's upper limbs appears significantly higher for the dominant (right) than for the non-dominant hand (Li et al., 2015; Mitchell et al., 2017). Indeed, previous studies concluded that these differences were due to the cerebral cortex of the so-called dominant (left) hemisphere which controls most voluntary movements on the opposite side of the body (Adamo, Scotland, & Martin, 2012; Martin, & Adamo, 2011).

Previous studies have shown that more than 90% of the population is right-handed (Debbarma, & Mehta, 2018). Considering the dominance of the lower extremities, the dominance of the right leg is manifested in 60% to 82% of the population (Taylor et al., 2007; Zouhal et al., 2018). Several studies revealed differences in many biomechanical parameters of the lower limbs such as the vertical jump, moving time, reaction time, and torque around a joint (Ball, 2011; Kobayashi et al., 2013; Rumpf et al., 2014; Sinsurin et al., 2017; Zouhal et al., 2018). These asymmetries between limbs in healthy individuals could lead to injuries (Knapik et al., 1991). Although, it is reported that a leg's dominance asymmetries could often disappear, in case the subjects could perform at much higher frequencies or higher level of force ability (Carpes et al., 2010). Therefore, taking into consideration the abovementioned, the scientific question that arises is whether a leg's dominancy can affect the maximal lower limb force. In the classic study of Knapik et al. (1991) it is reported that asymmetry up to 15% seems to be the upper limit for healthy limb function. More recently, the asymmetry between legs in dorsiflexion and plantar flexion maximal forces was confirmed (Valderrabano et al., 2007). Finally, other studies demonstrated an asymmetry in plantar flexors maximal force up to 45%, rather than the recommended 15% (Furlong, & Harrison, 2015). This means that such observed individual differences (Smak et al., 1999) could be highly dependent on the used test (Jones, & Bampouras, 2010), and due to a variety of factors, such as neural control (Adam et al., 1998; Fort-Vanmeerhaeghe et al., 2015), sport one side preference (Ball, 2011; Nunome, Ikegami, Kozakai, Apriantono, & Sano, 2006), and injury (Croisier, 2004).

Understanding the mechanisms that influence the function of ankle plantar flexors (e. g. soleus, gastrocnemius medialis, tibialis anterior) is important for exercise and therapy since it has been shown that these muscles are very often injured in the field of sports (Fong et al., 2007). Further, the tibialis anterior muscle is researched in patients after CNS traumas (Merletti et al., 1978) and a history of falls (Perry et al., 2007; Skelton et al., 2002). However, an ideal method to determine leg dominance in relation to task performance is still lacking. Therefore, the aim of the present study was to investigate the effect of dominance on the lower limb's maximal and subsequent neural activation during dorsiflexion and plantar flexion. A secondary purpose was to examine the influence of joint angle on strength and activation of these muscles.

2. MATERIAL AND METHODS

2.1. Participants

Twenty active and right-leg dominant males (age: 31.3 ± 9.5 years, height: 178.2 ± 7.6 cm, weight: 76.5 ± 11.0 kg), participated in the study. They reported no cardiovascular and neurological disorders or injury to the legs and were asked to refrain from taking alcohol, medication, or participating in strenuous activity 24 h prior to testing. Approval (ERC-013/2020) was obtained from the Aristotle University Ethics Committee on Human Research in accordance with the Declaration of Helsinki.

2.2. Experimental procedure

In order to assess lower limb dominancy, all participants voluntarily completed the self-reported Waterloo Footedness Questionnaire-Revised for leg dominance (WFQ-R; Van Melick, Meddeler, Hoogeboom, Nijhuis-van der Sanden, & van Cingel, 2017), and provided written informed consent, approved by the Institutional University Review Board and in accordance with the Declaration of Helsinki. All volunteers were asked to be available for three laboratory sessions. At the first session, participants were familiarized with the testing procedures, protocols, and performance of MVC. The second session was used to obtain baseline measurements, which included anthropometric characteristics, signed informed consent, and leg dominancy determining. Force and EMG recordings were assessed during the second and third session; the two limbs were tested randomly.

2.3. Experimental setup

The experimental setting consisted of an adapted ankle ergometer (OT Bioelettronica, Torino, Italy), attached by a dynamometer and two adjustable belts. All participants sat comfortably upon an adjustable electrical table, with their dominant/non-dominant leg placed in the ledge of the dynamometer, whereas the foot was tightened with straps (~ 2 cm wide). The hip position was adjusted at an angle of 110° and the knee joint angle adjusted at 120° (180°: full extension). To examine the muscle-tendon length effect, 3 ankle angles were selected, 75°, 90° (anatomical position) and 105°, corresponding to short, intermediate and long length for the dorsiflexors (tibialis anterior), and inversely for the plantar flexors (gastrocnemii and soleus). Two digital bipolar goniometers with a single degree of freedom (MLTS700, AD Instruments) were used to continuously measure the knee and ankle angles. The foot was fixed with straps to the adjustable base that was continuously connected with the calibrated cell (CCT Transducer, Model TF 022., Toronto, Italy). The fixing feet straps were placed over the distal third of the metatarsal bones and immediately in the front part of the ankle. After familiarization with the setup and a standardized warm-up (5 submaximal isometric contractions of 20-40% MVC), the participants performed 4 maximal isometric contractions of 5s with the dorsiflexors and the plantar flexors in the 3 ankle angle positions. Time (up to 2 min) was provided for rest between trials. The applied force was measured and displayed on a 50-inch monitor located at eye level ~ 1.5 m in front of the participant (Figure 1).

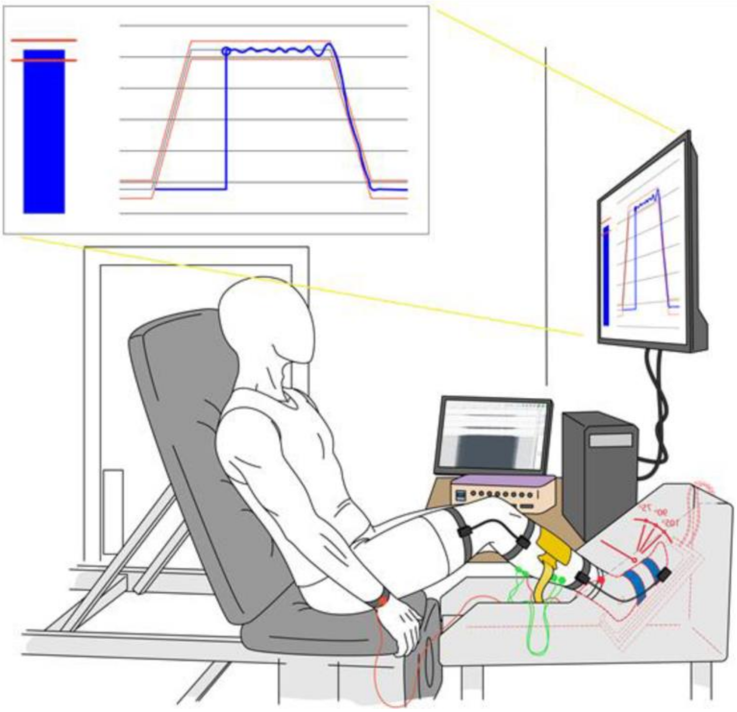

Fig. 1 The experimental setup consisted of an adapted ankle ergometer (OT Bioelettronica, Torino, IT). The force exerted by the dorsiflexor muscles of each leg was measured with a force transducer attached under the foot. High-density electromyography (HDsEMG) signals were recorded from the tibialis anterior muscle of each leg with a semi-resistant adhesive grid (yellow pad). Surface EMG recordings were also obtained from a pair of surface electrodes placed over the soleus and gastrocnemius medialis (green wires). The reference electrodes were placed at the wrist for the bipolar recordings and at the ankle for the grid (red wires). One goniometer was placed over each of the knee and the ankle joints to measure thejoint angle. Visual feedback was provided of the target force (red lines) and the applied force (blue lines) during the ramp-up, plateau, and ramp-down phases (middle screen) and on a moment-to-moment basis (right side of the screen). The display covered about 80% of the screen (Petrović et al., 2021)

2.4. Data processing and analysis

The dorsiflexion and plantar flexion forces were measured with a force transducer (200 kg; 2,001 mv/V, S/N 11406; TF022, CCT transducers) attached to the plate below the foot. The EMG signals from the left and right tibialis anterior were recorded using a semi-resistant adhesive 64 channel grid (5 × 13, 8 mm inter-electrode distance). The EMG signals from the left and right soleus and medial gastrocnemius were taken using bipolar surface electrodes (Thought Technology Ltd, CA) with an inter-electrode distance of 1 cm. According to SENIAM recommendations, the electrodes for soleus were placed at 2/3 of the line between the medial condyle of the femur to the medial malleolus, and the electrodes for the medial gastrocnemius were placed over the most prominent bulge of the muscle (Hermens et al., 2000). The signals were acquired, amplified, band-pass filtered (-3dB bandwidth, 10-500 Hz), and digitized using a 12-bit A/D converter, using Quattrocento (OT Bioelettronica, IT) with a sampling frequency of 2048 Hz. The force applied by the foot was sampled at the same rate and synchronized with EMG recordings. The participants received verbal encouragement during all MVC trials. The greatest peak force was taken as maximum and used for further analysis. To quantify the activation of each muscle the Root Mean Square (RMS) of EMG recordings was computed during three seconds at the highest level. The RMS values of the plantar flexor muscles were summed to compute the overall RMS.

To determine the level of bilateral asymmetry, between the dominant and non-dominant leg, the absolute bilateral asymmetry index (ASI) was calculated: ASI (%) = [|XD - XND| / 0,5 * (XD + XND)] * 100 (Karamanidis et al., 2003), in which X is the measure of interest and D and ND refer to the dominant and non-dominant leg, respectively.

2.5. Statistical analysis

All statistical analyses were performed using the software package SPSS software (version 25, IBM, Chicago). The normality of the data was assessed with the one-sample Kolmogorov-Smirnov test in SPSS. To determine the differences between the lower limb MVC force and EMG-muscle activation, and the differences between ankle angles (DF 75°, 90°, 105° and PF 75°, 90°, 105°) in MVC force, a two-way repeated-measures ANOVA was used. Moreover, a Paired-Samples T Test on each dependent variable (MVC force and EMG) at different angles (DF 75°, 90°, 105° and PF 75°, 90°, 105°) was conducted as a follow-up test. In all measurements, statistical significance was set at the level of $p < 0.05$. Results are reported as mean and standard deviation (SD).

3. RESULTS

3.1. Legs Dominance

3.1.1. Maximal Voluntary Contraction Force

The analysis revealed a non-significant mean difference on MVC force, among the two independent variables, DF_{MVC} and PF_{MVC} as presented in Table 1. The non-significant differences between mean scores of dominances (DL and NL) and MVC force means of multiple angle groups implies that leg dominance had no effect on the multiple angle group's MVC force. Hence, it seems that MVC force is probably formed independently of leg dominance. Therefore, other factors need to be examined in order to investigate the exact leg dominance mechanism concerning MVC force.

Table 1 Differences in dorsal and plantar flexion between lower limb MVC force means for multiple angle groups

Variable	F	Effect - df	Error - df	p-value	ηp^2
DF_{MVC} 75, 90, 105	0.256	1	38	0.616	0.007
PF_{MVC} 75, 90, 105	0. 240	1	38	0.627	0.006

Legend: DF - dorsal flexion, PF - plantar flexion, MVC – maximum voluntary contraction. The significance levels: $p < 0.05$.

Furthermore, a Paired-Samples T Test, on each of the dependent variables (MVC force of DF 75°, 90°, 105° and MVC force of PF 75°, 90°, 105°) was conducted. As can be seen in Figure 2 A-B, all the results were non-statistically significant, $p > 0.05$. Therefore, the analysis reveals that there are similar levels of MVC force for dorsal- and plantar flexion at either 75°, 90°, or 105°, between the dominant and non-dominant leg of the sample.

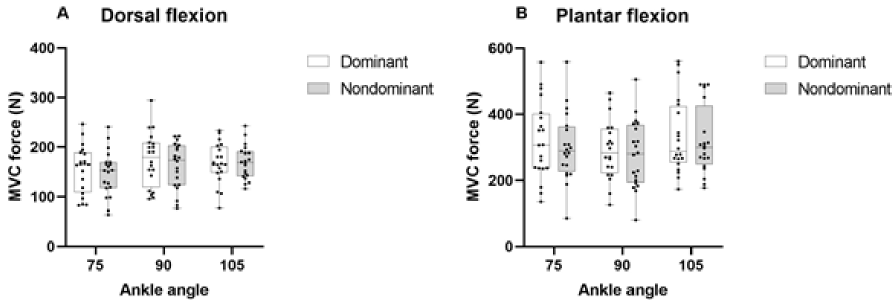

Fig. 2 A. Maximal voluntary contraction force developed between the dominant (open) and non-dominant (grey) leg during dorsiflexion at an ankle angle of 75°, 90°, and 105°. B. Maximal voluntary contraction force developed between dominant (open) and non-dominant (grey) leg during plantar flexion at an ankle angle of 75°, 90°, and 105°

Table 2 Differences in dorsal and plantar flexion between lower limb EMG-muscle activation means for multiple angle groups

Variable	F	Effect - df	Error - df	p-value	ηp^2
DF_{EMG} 75, 90, 105	0.241	1	38	0.904	0.006
PF_{EMG} 75, 90, 105	0.572	1	38	0.415	0.015

Legend: DF - dorsal flexion, PF - plantar flexion, EMG - electromyography.
The significance levels: $p < 0.05$.

3.1.2. Electromyographic Muscle Activation

The analysis revealed a non-significant mean difference on EMG-muscle activation, among the two independent variables, DF_{EMG} and $PF_{EMG,}$ as presented in Table 2.

Table 3 The % absolute asymmetry index (ASI) of the sample

Variables	ASI%
DF 75°	17.9 ± 16.2
DF 90°	25.8 ± 18.7
DF 105°	17.3 ± 13.9
PF 75°	18.4 ± 19.1
PF 90°	21.5 ± 15.4
PF 105°	17.1 ± 13.9

Legend: DF - dorsal flexion, PF - plantar flexion.

A Paired-Samples T Test, on each of the dependent variables (EMG of DF 75°, 90°, 105° and EMG of PF 75°, 90°, 105°) was calculated. All the results were non-statistically significant, $p > 0.05$. Therefore, the analysis reveals that there are similar levels of EMG for dorsal- and plantar flexion at either 75°, 90°, or 105°, between the dominant and non-dominant leg of the sample as presented in Figure 3 A-B.

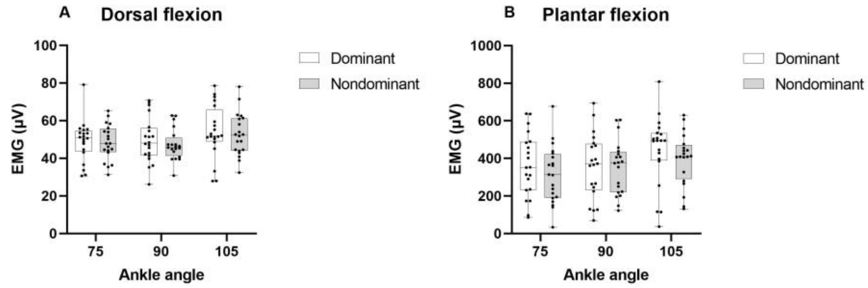

Fig. 3 A. Electromyography muscle activation developed between the dominant (open) and non-dominant (grey) leg during dorsiflexion at an ankle angle of 75°, 90°, and 105°. B. Electromyography muscle activation developed between the dominant (open) and non-dominant (grey) leg during plantar flexion at an ankle angle of 75°, 90°, and 105°

Specifically, the results showed that the EMG-muscle activation levels were similar between the dominant and non-dominant leg of the sample, although it seemed that the EMG-muscle activation of the non-dominant leg weakened in strength throughout repetitions, compared to the dominant leg. Therefore, further long-term studies on EMG-muscle activation need to be done, so as to explore the exact leg dominance mechanism concerning EMG-muscle activation over time.

3.1.3. Absolute symmetry index

The asymmetries observed during MVC of DF and PF ranged from 17.9 ± 16.2 to 25.8 ± 18.7 and from 17.1 ± 13.9 to 21.5 ± 15.4, respectively, as can be observed in Table 3. The ASI data do not indicate the appearance of asymmetry between the lower limbs.

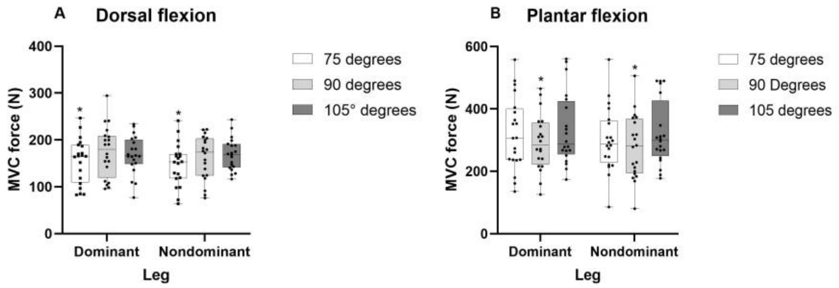

Fig. 4 A: Maximal voluntary contraction force developed during dorsiflexion at an ankle angle of 75° (open), 90° (grey), and 105° (dark grey), *: significantly less than 90° and 105°. B. Maximal voluntary contraction force developed during plantar flexion at an ankle angle of 75° (open), 90° (grey), and 105° (dark grey), *: significantly less than 75° and 105°

Table 4 Differences in dorsal and plantar flexion between three angles (75°, 90° and 105°) in MVC force

Variable	F	Effect - df	Error - df	p-value	ηp^2
DF_{MVC}	6.090	2	76	0.004	0.138
PF_{MVC}	9.752	2	76	0.000	0.204

Legend: DF - dorsal flexion, PF - plantar flexion, MVC – maximum voluntary contraction. The significance levels: $p < 0.05$.

3.2. Muscle length

3.2.1. Influence of muscle length on MVC force

The analysis revealed a significant influence of the ankle angle on MVC force for the dorsiflexors and the plantar flexors as presented in Table 4.

Table 5 Differences in dorsal and plantar flexion between three angles (75°, 90° and 105°) in EMG-muscle activation

Variable	F	Effect - df	Error - df	p-value	ηp^2
DF_{EMG}	5.455	2	76	0.006	0.126
PF_{EMG}	7.793	2	76	0.002	0.170

Legend: DF - dorsal flexion, PF - plantar flexion, EMG - electromyography. The significance levels: $p < 0.05$.

Furthermore, the MVC force for the dorsiflexors was at 75º (DL: 155.7 ± 47.8; NL: 147.9 ± 45.7 N) significantly less than 90º (p = .003; DL: 174.7 ± 54.5; NL: 161.9 ± 46.8 N) and 105º (p = .005; DL: 167.9 ± 40.5; NL: 169.1 ± 33.9 N), as can be seen in Figure 4A, and for plantar flexors at 90º (DL: 291.9 ± 90.0; NL: 278.2 ± 102.6 N) significantly less than 75º (p = .006; DL: 321.0 ± 116.6; NL: 300.4 ± 103.6 N) and 105º (p = .000; DL: 333.0 ± 115.7; NL: 321.0 ± 103.9 N), as can be seen in Figure 4 B.

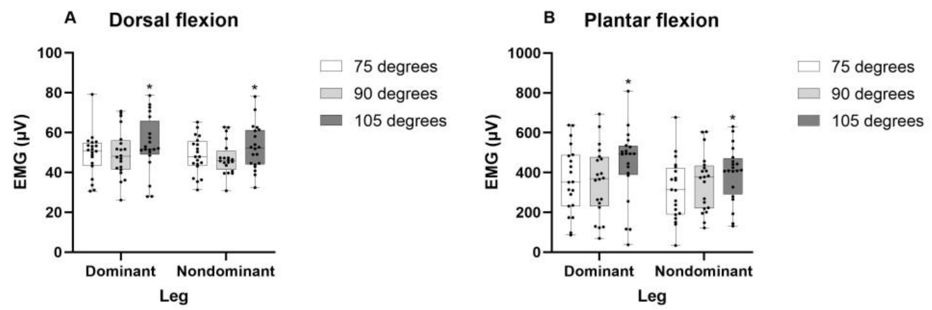

Fig. 5 A: Electromyography muscle activation developed during dorsiflexion at an ankle angle of 75° (open), 90° (grey), and 105° (dark grey), *: significantly less than 75° and 90°. B. Electromyography muscle activation developed during plantar flexion at an ankle angle of 75° (open), 90° (grey), and 105° (dark grey), *: significantly less than 75° and 90°

3.2.2. Influence of muscle length on EMG-muscle activation

The analysis revealed a significant influence of the ankle angle on EMG-muscle activation for the dorsiflexors and the plantar flexors as presented in Table 5.

Furthermore, EMG-muscle activation for the dorsiflexors was at 105º (DL: 54.0 ± 14.2; NL: 52.6 ± 11.4 N) significantly higher than 75º (p = .013; DL: 49.1 ± 10.9; NL: 48.5 ± 9.3 N) and 90º (p = .003; DL: 49.6 ± 12.1; NL: 47.2 ± 8.3 N), as can be seen in Figure 5A, and for plantar flexors at 105º (DL: 441.5 ± 186.6; NL: 397.1 ± 139.1 N) significantly higher than 75º (p = .004; DL: 364.1 ± 168.7; NL: 312.4 ± 153.3 N) and 90º (p = .002; DL: 364.2 ± 174.3; NL: 351.0 ± 144.7 N), as can be seen in Figure 4 B.

4. DISCUSSION

The primary findings of the study indicated similar levels of lower limb MVC force and EMG-muscle activation between dominant and non-dominant ankle muscles. These results are in agreement with earlier studies which also reported similar MVC forces of both dominant and non-dominant limbs (De Ruiter et al., 2010). Another finding was that the maximal force of dorsiflexors and plantar flexors is dependent on the changes in muscle length (Anderson, & Pandy, 2001). We examined the force in three different lengths (short, intermediate, and long) corresponding to the 3 ankle angles. From the analyzed data we can conclude that, as the muscle length increases, the maximal force also increases.

The findings of the present study support previous studies which indicated the absence of leg difference during ankle joint isometric MVC (Yen et al., 2018), the first dorsal interosseous MVC (Adam et al., 1998), force-matching tasks (Yamaguchi et al., 2019), and bilateral quiet standing (Wang, & Newell, 2014). In contrast, our results disagree with other researchers who reported that the dominant leg is significantly stronger than the non-dominant (Dalleau, Belli, Bourdin, & Lacour, 1998). These authors commented that leg differences in strength could be related to the dominant leg's propulsive function, while the non-dominant leg is more adaptive in balance control. At the same line, Õunpuu, & Winter (1989) reported significantly greater plantar flexor activation in the dominant than the non-dominant limb, indicating a larger neural drive to the dominant plantar flexors to generate the necessary propulsive locomotion power. These differences were detected using repeated and dynamics tasks which can be due to the activation of other muscles as well. It is important to mention that the mean scores of EMG-muscle activation in muscles were somewhat higher for the dominant than for the non-dominant leg, which might suggest that the influence of dominance may vary between individuals. Others have also reported an influence of dominance on recorded EMG frequency for the tibialis anterior and gastrocnemius medialis (Valderrabano et al., 2007), which also does not agree with our findings. The difference between our study and that of Valderrabano et al., 2007 could be due to the different age of the participants (32-65 yr, Average 53), as recent studies have shown that dorsal flexion muscles become weaker with a present between-leg imbalance in the elderly (Perry et al., 2007; Skelton et al., 2002). Collectively, within the limitations of the study we were unable to observed leg dominance effects.

As shown in Table 3, the presented data are in line with previous findings of an asymmetry of 18.3% observed in plantar flexors, but not in dorsiflexors (Valderrabano et al., 2007). Also, large asymmetry was confirmed in plantar flexion with the ASI mean ranging from 11% to 25% (Furlong, & Harrison, 2015). The results of our data confirmed such a difference in plantar flexors and also in dorsiflexors (mean ASI 17.1 – 21.5%, and

17.3 – 25.8%, respectively). These results are greater than the 15% previously reported as upper limit of asymmetry for healthy limb function in upper limbs (Knapik et al., 1991), and similar to 20% suggested as lower limb upper limit of asymmetry (Maulder, 2013). These normal to high ASI found in present research are a lot smaller than those previously found ASI of 45% in healthy subjects (Furlong, & Harrison, 2015) and 60% in injured cyclists (Bertucci et al., 2012). The present results point to the constraints imposed during performance of the task and timely tracking appearance of asymmetry and implement rehabilitation to avoid injuries that may occur, such as deformation of the Achilles tendon (Bohm et al., 2015), or tibial stress fractures (Finestone et al., 1991). Finally, ASI was smaller in the non-dominant leg in all angles of dorsiflexion and plantar flexion, which can be explained by its everyday use where the non-dominant leg mainly supports and stabilizes body posture during movements (Sadeghi et al., 2000).

The results of this study also showed the impact of muscle length on MVC dorsiflexors and plantar flexors strength (Table 4). These results confirm old (Bigland-Ritchie et al., 1992) and recent (Tsatsaki et al., 2021) research reports indicating that there is an influence of joint angles on dorsiflexion MVC force, as there were lower strength values at shorter lengths compared with longer length. Our results suggest a diminished force capacity of plantar flexors due to reduced excitation of the motor neuron pool when muscle length is decreased (Avancini et al., 2015; Miaki et al., 1999). EMG amplitudes are smaller for the gastrocnemii during plantar flexions performed with knee flexed compared to knee extended (Gandevia, & McKenzie, 1988). At a critical short length, the muscle is considered to be “actively insufficient”, and its motor neurons receive less net excitatory input (Kennedy, & Cresswell, 2001).

Overall, we found that dominance does not affect the maximal lower limb force and neural activation. Asymmetry between limbs may appear during submaximal contractions induced by the activation of the motor units (Adamo et al., 2012; Martin, & Adamo, 2011) but this issue remains open. The findings indicate that participants’ reports of preferred footedness were not associated with differences in the functional characteristics of the dorsiflexor and plantar flexor muscles during isometric contractions. The suggested 20% for normal between-limb function has to be analyzed further, and not be taken as the strict risk border line.

The disagreement between various findings on the effects of dominance on lower limb strength and activation suggests that further research is required to investigate the exact mechanism of leg dominance function. Our results suggest that dominance evaluation has limited importance for the evaluation of maximal voluntary contraction force and electromyography muscle activation, at least for these experimental settings.

5. Conclusion

In conclusion, the present study suggests that leg dominance has no effect on lower limb maximal voluntary contraction and neural activation. Within this study limitations, it appears that in healthy individuals there are no apparent effects of dominance on plantar flexor and dorsiflexor maximal strength. Large differences in strength between the legs may, therefore, accompany pathological conditions, although, clearly, more research is required in this direction.

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MODIFIKACIJE MAKSIMALNE SILE I NEURALNE AKTIVACIJE MIŠIĆA GLEŽNJA U ZAVISNOSTI OD DOMINANTNOG EKSTREMITETA

Cilj istraživanja bio je da se procene razlike između maksimalne voljne kontrakcije donjih ekstremiteta (MVC) i neuronske aktivacije na osnovu dominantne noge. Dvadeset aktivnih ispitanika sa dominantnom desnom nogom (starost: 31,3 ± 9,5 godina, visina: 178,2 ± 7,6 cm, težina: 76,5 ± 11,0 kg) izvelo je 3 maksimalne dorzalne fleksije (DF) i plantarne fleksije (PF) pod 3 ugla skočnog zgloba (75 °, 90 °: anatomski položaj i 105 °), što odgovara kratkim, srednjim i većim dužinama za DF mišiće, a suprotno za PF mišiće. Elektromiografija (EMG) je korišćena za procenu aktivnosti mišića skočnog zgloba (tibialis anterior, gastrocnemius medialis i soleus). Rezultati su pokazali neznatne razlike između MVC sile donjih ekstremiteta i aktivacije EMG mišića. Međutim, primećen je značajan glavni efekat ugla. Tokom DF, MVC sila je bila veća (p < 0,05) na 90° i 105° nego na 75° za obe noge i tokom PF, MVC sila je bila veća (p < 0,05) na 75° i 105° nego na 90° za obe noge. Štaviše, tokom DF i PF, aktivacija EMG mišića bila je veća (p < 0,05) na 105° nego na 75° i 90° za obe noge. Rezultati pokazuju da dominantna noga nije bila povezana sa različitim nivoom sile i neuronske aktivacije tokom maksimalne dobrovoljne kontrakcije mišića skočnog zgloba. Zaključeno je da dominantna noga nema uticaja na maksimalnu silu i neuralnu aktivaciju mišića skočnog zgloba i da bilo kakvi mehanizmi koji doprinose efektu dominacije ekstremiteta nisu evidentni na primeru ovog eksperimentalnog protokola.

Ključne reči: *Dominantna noga, maksimalna voljna kontrakcija, elektromiografija, ne-dominantna noga, sila, donji ekstremiteti*