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Effects of hip and head position on ankle range of motion, ankle passive torque, and passive gastrocnemius tension

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Ankle joint range of motion (ROM) is notably influenced by the position of the hip joint. However, this result remains unexplained. Thus, the aim of this study was to test if the ankle passive torque and gastrocnemius muscle tension are affected by the hip and the head positions. The torque and the muscle shear elastic modulus (measured by elastography to estimate muscle tension) were collected in nine participants during passive ankle dorsiflexions performed in four conditions (by combining hip flexion at 90 or 150°, and head flexed or neutral). Ankle maximum dorsiflexion angle significantly decreased by flexing the hip from 150 to 90° ($P < 0.001$; mean difference $17.7 \pm 2.5°$), but no effect of the head position was observed ($P > 0.05$). Maximal passive torque and shear elastic modulus were higher with the hip flexed at 90° ($P < 0.001$). During submaximal ROM, no effects of the head and hip positioning ($P > 0.05$) were found for both torque and shear elastic modulus at a given common ankle angle among conditions. Shifts in maximal ankle angle due to hip angle manipulation are not related neither to changes in passive torque nor tension of the gastrocnemius. Further studies should be addressed to better understand the functional role of peripheral nerves and fasciae in the ankle ROM limits.

Understanding the factors limiting range of motion (ROM) of a joint is a topic of continued interest for both researchers and clinicians. Commonly, the muscle passive tension is often estimated by using a passive torque–angle measurement (Magnusson, 1998; Gajdosik, 2001; McNair & Portero, 2005; Weppler & Magnusson, 2010). Both passive tension developed by muscle-tendon units and the sensation of this tension (i.e., discomfort) are the main limits of the maximal tolerable ROM available in a joint during a passive stretching maneuver (for review, see Weppler & Magnusson, 2010).

It is well known that the maximal dorsiflexion angle of the ankle and passive torque are affected when knee angle is changed due to a change in the gastrocnemius length (e.g., Hoang et al., 2005; Nordez et al., 2010). In respect to hip and ankle interactions, biomechanical models typically consider that these joints do not have common actuators because no muscle-tendon complex crosses both joints (Klein Horsman et al., 2007; Standring, 2008). Therefore, a change in hip angle should not affect both the passive muscle tension of ankle plantar flexors and the maximal dorsiflexion angle. "However, it has been shown that when the knee is fully extended, flexing the hip notably decreased passive maximum dorsiflexion angle of the ankle in comparison to a neutral hip position (Mitchell et al., 2008)." Two main hypotheses have been proposed to explain this finding. Firstly, a transmission of tension via lower limb fascial connections should imply an increased tension in triceps surae muscles during ankle dorsiflexion when the hip is flexed (Boyd et al., 2009). The second hypothesis is that there is a change in tension in the sciatic nerve tract. In regard to this hypothesis, there is evidence of sciatic nerve movement that is altered by trunk flexion (Ellis et al., 2012). Furthermore, Johnson and Chiarello (1997) have found a decrease in knee ROM when the head was flexed, and they proposed the tensioning of the peripheral nerves tracts as a potential mechanism. There does not appear to be similar studies that investigated the effect of cervical position on ankle ROM.

Joint torque has been extensively used to indirectly measure passive muscle mechanical properties (Magnusson, 1998; Gajdosik, 2001; McNair & Portero, 2005; Weppler & Magnusson, 2010). However, it could be argued that this parameter is more related to the resistance of the "global" musculo-articular complex to motion, and involves several anatomic structures crossing the joint (Riemann et al., 2001). Recently, it was reported *in vitro* (Koo et al., 2013) and *in vivo* (Maisetti et al., 2012) that the measurement of the shear elastic modulus using an elastographic technique (i.e., supersonic shear imaging) is strongly related to the localized passive muscle tension during stretching. Briefly, supersonic shear imaging relies on measuring the speed of internal propagation of shear waves generated by an acoustic radiation force to estimate the shear elastic modulus of a localized area in different soft tissues (Bercoff et al., 2004). Thus, the shear elastic modulus measurement could be used to determine whether gastrocnemius tension is changed due to a change in hip angle positioning.

Therefore, the purpose of this study was to simultaneously measure the ankle torque and gastrocnemius medialis (GM) shear elastic modulus during passive ankle dorsiflexion movement through a combination of hip and head position tests. Based on previous results that show notable changes in ankle ROM of the ankle with changes in hip angle (Mitchell et al., 2008), we hypothesized that gastrocnemius elastic modulus and/or ankle joint torque could be affected by the hip positioning.

Methods

Participants

Nine healthy men (age: 25 ± 3 years, height: 181 ± 6 cm, weight: 73 ± 8 kg) volunteered to participate in this study and signed an informed consent form. None of the participants reported any known current or ongoing musculo-skeletal lower limb and spine injuries, neuromuscular diseases, or orthopedic-related problems. The local ethics committee approved the study, and all the procedures were undertaken in accordance with the Declaration of Helsinki.

Equipment

Ergometer

An isokinetic dynamometer (Biodex 3 Medical, Shirley, New York, USA) was utilized to dorsiflex the ankle joint, and provide simultaneous measures of ankle angle and torque. The participants were seated in the Biodex chair with the knee fully extended (0°). The back of the chair was then adjusted to alter the hip joint position. Testing positions were determined utilizing a manual goniometer (MSD, Londerzeel, Belgium). The neutral position of the ankle was defined as 0° (i.e., foot was positioned perpendicular to the leg).

Supersonic shear imaging

An Aixplorer ultrasound scanner (version 7; Supersonic Imagine, Aix-en- Provence, France) coupled with a linear transducer array (4–15 MHz Super Linear 15-4, Vermon, Tours, France) was used in shear ware elastography mode (musculo-skeletal preset) as previously described (Bercoff et al., 2004). Assuming a linear elastic behavior, the muscle shear elastic modulus was calculated as follows (1):

$$\mu = \rho V s^2 \tag{1}$$

where ρ is the density of soft tissues (1000 kg/m^3) and Vs is the shear wave speed. The region of interest for probe location was the gastrocnemius muscle belly, and this was identified by an experienced examiner. The transducer was held statically over the GM muscle belly (i.e., mid-distance between the muscle-tendon junctions) with a custom-made cast positioned perpendicularly to the skin and along the muscle fascicle orientation (Blazevich et al., 2006; Maisetti et al., 2012; Hug et al., 2013). Appropriate probe alignment was achieved when several fascicles could be traced without interruption across the image. The transducer location was not changed within the test session and therefore it is unlikely that the skin pressure induced from probe fixation could affect the trend of the muscle stiffness results during passive gastrocnemius stretching (Maisetti et al., 2012). The maps of the shear elastic modulus were obtained at 1 Hz with a spatial resolution of 1 × 1 mm.

Electromyography

To rule out any active muscle involvement, electromyography (EMG) activity was monitored during passive stretching. A pair of a conductive adhesive hydrogel surface EMG electrodes (Kendall™ 100 Series Foam Electrodes, Covidien, Massachusetts, USA) with an interelectrode distance of 20 mm (center to center) was placed over the GM, gastrocnemius lateralis, soleus, and tibialis anterior. The electrodes were located according to the recommendations of SENIAM (Surface EMG for Non-Invasive Assessment of Muscles) as described by Hermens et al. (2000). Skin was shaved and cleaned with alcohol to minimize impedance before applying the electrodes. Raw EMG signals were amplified close to the electrodes (gain 375, bandwidth 8–500 Hz) and digitized at a sampling rate of 1 kHz (ME6000, Mega Electronics Ltd., Kuopio, Finland). The EMG was monitored by an examiner during all trials.

Protocol

The subjects were familiarized with the experimental setup. They were previously instructed not to practice vigorous exercise 48 h before the session time. The passive ankle dorsiflexion tests were performed by combinations of hip flexion at 90 or 150° with head flexion (HF) or without head flexion (HN), see Fig. 1. Thus, a total of four tests (i.e., 90 HF, 90 HN, 150 HF, 150 HN) were performed by each participant in a randomized order. A 5-min rest interval separated each test. Flexion of the head was defined as the

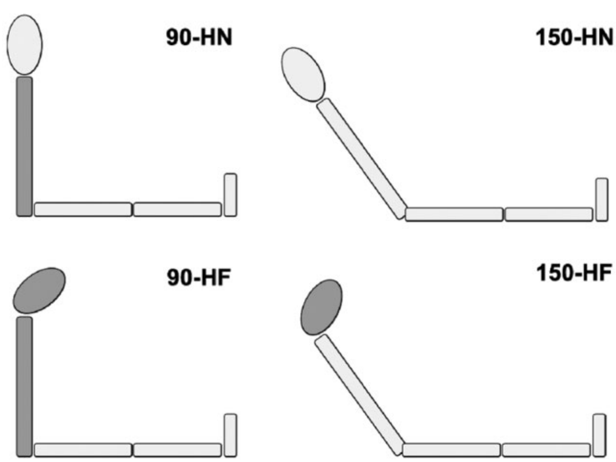

Fig. 1. Test conditions used in this study.
Legend: 90 HN, 90° of hip angle and head in neutral position; 90 HF, 90° of hip angle and head in flexion position; 150 HN, 150° of hip angle and head in neutral position; 150 HF, 150° of hip angle and head in flexion position.

maximum passive angle tolerated by participants and it was fixed in that position by a custom-made cast. At the beginning of each test session, five slow loading/unloading cycles (5°/s) were performed for conditioning purposes (Nordez et al., 2008). Thereafter, starting from 40° of plantar flexion, participant's plantar flexors were passively stretched (2°/s) until their maximal dorsiflexion angle was attained. The maximum perceived plantar flexor muscle stretch that participants could tolerate (i.e., onset of pain) was considered the criterion of maximal dorsiflexion ankle angle. At this point, the subjects pushed a button that immediately stopped the ankle passive motion imposed by the dynamometer. Three repetitions were done for each test condition, and the third repetition was used for all analyses. Participants were blindfolded to ensure no visual perception of stretching during testing. They were instructed to stay as relaxed as possible through each trial. Thigh and pelvis were firmly fastened with straps in all tests. EMG feedback was provided to the examiner during testing. At the end of the protocol, in order to normalize EMG-RMS (root mean square), three maximal isometric voluntary contractions were performed in plantar flexion and dorsiflexion at ankle neutral position, with 1 min of rest in-between each contraction.

Data analysis

Data were processed using standardized Matlab (The Mathworks, Natick, Massachusetts, USA) scripts. Each focused ultrasound push beam (generated within the probe) used to produce the shear waves generates a sound signal. Thus, for each test repetition, the timing of each shear elastic modulus measurement among ankle passive motion was therefore detected using a microphone (MB Quart K800, frequency response: 40–18 000 Hz, sensitivity: 16,25 mV/Pa) coupled to the transducer, and the last timing was used to synchronize all collected data. Ankle torque was gravity corrected and calculated every second to match shear elastic modulus measurements. The RMS of electromyographic signals (RMS-EMG, averaged over 300 ms windows) was also calculated for each second. The shear elastic modulus was quantified as an average of the region of interest chosen as the biggest region without detectable artifacts defined as a localized abnormal value of the shear elastic modulus that was not present on the previous image (Bouillard et al., 2012). Maximum ankle dorsiflexion angle varied between participants. Thus, it was expressed as a percentage of the maximum ROM achieved among the four test conditions for each subject.

Statistical analysis

The data were processed using IBM SPSS Statistics 20.0 (IBM Corporation, Armonk, New York, USA) software. Descriptive statistics were reported as the mean and standard deviation (mean ± SD). Data were tested for normality with the Shapiro–Wilk test, and no violations were noted. Three two-way repeated measures analyses of variance (ANOVAs) (2 hip angles × 2 head angles) were used to compare absolute maximal values of ankle dorsiflexion angle, passive torque, and shear elastic modulus across test conditions. The changes in shear elastic modulus and passive torque at the largest common ankle among the four tests were compared using two two-way repeated measures ANOVAs (2 hip angles × 2 head angles). For instance, if maximum dorsiflexion angles of the 90-HF, 90-HN, 150-HF, and 150-HN tests are 20, 25, 35, and 40°, respectively, the largest common ankle angle will be 20° or 50% if it expresses as the percentage of maximum dorsiflexion ankle angle achieved among the four tests. The partial eta square ($_p\eta^2$) values were reported as measures of effect size, with moderate and large effects considered for $_p\eta^2 = 0.07$ and $_p\eta^2 \geq 0.14$, respectively (Cohen, 1988). In the present study, moderate to large effect sizes ($_p\eta^2$) were obtained across the ANOVAs performed. Because EMG-RMS did not follow a normal distribution, Friedman test was used for the mean rank comparisons between the four tested conditions for each muscle in different percentages of maximum dorsiflexion ankle angle: 20%, 40%, 60% and the highest common percentage of maximal dorsiflexion ankle angle among tests conditions (i.e., 64%). Statistical significance was set at 0.05.

Results

A typical example of raw data (torque–angle and shear elastic modulus–angle relationships) obtained for one subject is depicted in Fig. 2.

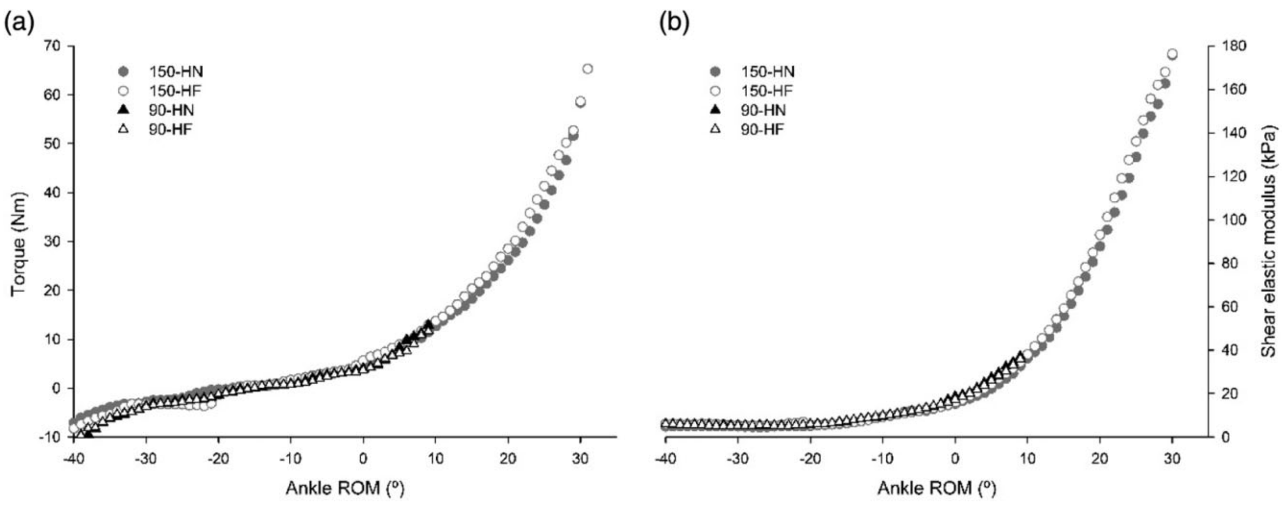

Fig. 2. Typical example. (a) Typical joint torque–angle curves during passive ankle dorsiflexion for each test condition. (b) Typical muscle shear elastic modulus–angle curves during passive ankle dorsiflexion for each test condition.
Legend: 90 HN, hip angle at 90° and head in neutral position; 90 HF, hip angle at 90° and head in flexion position; 150 HN, hip angle at 150° and head in neutral position; 150 HF, hip angle at 150° and head in flexion position. ROM is expressed in degrees (°).

Table 1. Average of absolute values of torque and shear elastic modulus at endpoint of maximum stretching and maximum ROM accepted by participants in each test condition

	Test condition			
	90 HN	90 HF	150 HN	150 HF
Maximum dorsiflexion angle of the ankle (°)	19.11 ± 7.20	18.89 ± 6.92	36.62 ± 6.45	36.71 ± 6.31
Peak torque (Nm)	30.31 ± 13.42	30.10 ± 18.39	69.53 ± 14.02	67.37 ± 12.46
Shear elastic modulus (kPa)	71.46 ± 24.50	70.36 ± 31.54	140.41 ± 38.97	143.26 ± 41.85

Values are presented as mean ± SD. Legend: 90 HN, hip angle at 90° and head in neutral position; 90 HF, hip angle at 90° and head in flexion position; 150 HN, hip angle at 150° and head in neutral position; 150 HF, hip angle at 150° and head in flexion position. ROM, range of motion; 0° = neutral position of the ankle joint (foot positioned perpendicular to the leg).

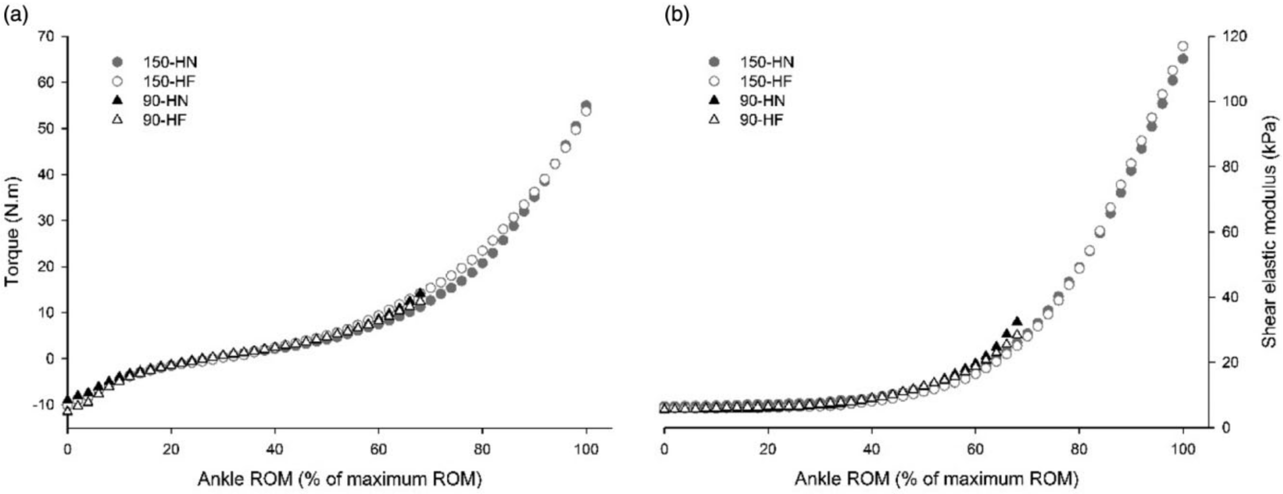

Fig. 3. Average results. (a) Average joint torque–angle curves during passive ankle dorsiflexion for each test condition. (b) Average muscle shear elastic modulus–angle curves during passive ankle dorsiflexion for each test condition.
Legend: 90 HN, hip angle at 90° and head in neutral position; 90 HF, hip angle at 90° and head in flexion position; 150 HN, hip angle at 150° and head in neutral position; 150 HF, hip angle at 150° and head in flexion position. ROM is expressed as a percentage of the maximum ROM achieved in the overall four test conditions by each subject. *No significant differences ($P < 0.05$) were observed for every common ROM between conditions.

Absolute maximum ankle angle, shear elastic modulus, torque, and EMG

A significant main effect of hip angle was observed for the following dependent variables: maximal ankle angle (mean difference = 17.67 ± 2.48, $P < 0.001$; ${}_p\eta^2 = 0.89$), peak torque (mean difference = 38.25 ± 4.65 Nm, $P < 0.001$; ${}_p\eta^2 = 0.95$), and peak shear elastic modulus (+69.00 ± 14.29 kPa, $P = 0.001$; ${}_p\eta^2 = 0.90$) (Table 1). No main effects or interactions associated with head position were observed for all dependent variables ($P > 0.05$; ${}_p\eta^2 > 0.14$). No main effects or interactions were found for EMG-RMS of each muscle (mean %MVC EMG signals across muscles were less than 2% and did not vary across the entire ROM).

Passive torque–angle relationship

No hip and head position effects or interactions were observed throughout comparable ankle ROM (Fig. 3(a); $P > 0.05$; $p\eta^2 > 0.07$ for all ankle angles until 64% of maximal dorsiflexion).

Muscle shear elastic modulus–angle relationship

Averaged curves of the shear elastic modulus–angle relationships across subjects are shown in Fig. 3. No hip and head effects were observed among comparable ankle angles until the highest common ankle angle among performed tests ($P > 0.05$; ${}_p\eta^2 = 0.08$ for head effects and ${}_p\eta^2 > 0.14$ for hip effects and interactions).

Discussion

This study describes the effect of the hip and the head positions on the ankle joint dorsiflexion ROM limit, the ankle torque, and the GM passive tension estimated using the shear elastic modulus measurement. No previous study has simultaneously investigated these parameters. The main finding was that maximum dorsiflexion angle of the ankle was strongly affected by the hip angle position, while ankle torque and passive tension of the GM were unchanged for an equivalent ankle angle. In addition, the head position did not affect the ankle dorsiflexion ROM.

A dramatic decrease in maximum dorsiflexion angle of the ankle (> 50%) was found when hip angle was changed from 150 to 90°. This result is in accordance with the study of Mitchell et al. (2008). They have demonstrated that the hip flexion with the knee fully extended produced significant deficits in ankle joint ROM (ΔROM 22.3°, 95% CI 18.0–26.7). Other studies also support these results. For instance, Gajdosik et al. (1985) reported a decrease (10.1 ± 5.1°) in hip maximum ROM when performing a passive straight leg raise (SLR) with the ankle in a dorsiflexed position compared with an SLR performed with the ankle in plantar flexion. Boyd et al. (2009) have observed quite similar results (decrease in 10.1 ± 9.7°) for the same tests. However, none of these previous studies have investigated other important variables (e.g., passive torque).

Our results clearly demonstrate that both the ankle passive torque and the GM shear elastic modulus obtained for a given common ankle angle are unaffected by the hip and the head positions (Fig. 3). While several structures contribute to the passive torque, elastographic measurements performed in the present study were focused on one targeted muscle and provide an indirect estimation of its passive muscle tension (Maisetti et al., 2012). Thus, our results suggest that ROM changes were not caused by alterations in GM muscle tension. However, the specific influence of other minor plantar flexor muscles should be examined in the future. In addition, the peak torque and the maximal shear elastic modulus obtained at the maximal ROM were significantly higher when the hip was positioned at 150° (Table 1). Thus, the participants were able to achieve much more tension in the plantar flexor muscles with the hip extended compared with the hip flexed because in the first condition (hip extended) they tolerated higher dorsiflexion ROM.

Our results also suggest that there is no force transmission between hip-related structures and the gastrocnemius muscle. Therefore, other anatomical structures that cross both the hip and the ankle joints that do not influence ankle torque should explain the decrease in maximal dorsiflexion angle and stretch tolerance when the hip is flexed.

Fascia and peripheral nerves are passive and continuous anatomical structures that cross both hip and ankle joints, and their contributions to the limitations of the maximal dorsiflexion ankle angle when the hip is flexed have not been investigated. The architecture of the superficial and deep fascia system of the lower limb, between ankle and hip joints, has been documented (Gerlach & Lierse, 1990). However, the role of fascial tissue on force transmission between non-mechanically-related joints and muscle-tendon complexes during passive stretching remains unknown. Carvalhais et al. (2013) noted that passive torque is increased when fascial tissue is indirectly tensioned by moving a non-related joint. This suggests that the tension of the fascial tissue may affect the passive torque and muscle tension. In the current study, the lack of changes in torque and shear elastic modulus for a given ankle angle suggests that fasciae were not involved notably in the changes in dorsiflexion ROM across test conditions.

In regard to the effect of peripheral nerves, it seems improbable that a change in nerve tension may induce significant changes in the passive torque, but it may affect stretch tolerance. The biomechanical behavior of peripheral nerves has been studied (Topp & Boyd, 2006; Silva et al., 2014). Some studies have observed that both hip flexion (Coppieters et al., 2006; Ridehalgh et al., 2014) and ankle dorsiflexion (Coppieters et al., 2006; Alshami et al., 2008) substantially increased nerve deformation. In addition, Boyd et al. (2012) highlighted this rationale by showing a mean threefold reduction in the tibial nerve distal movement during ankle dorsiflexion when the hip was flexed. Furthermore, in a study involving cadavers, Borrelli et al. (2000) showed that intraneural pressure increased significantly when the hip was flexed from 0 to 45° of flexion and again when the hip was brought from 45 to 90° of flexion. The intraneural tissue fluid pressures measured within a localized section of the sciatic nerve appeared to exceed published critical thresholds for alterations of blood flow and neural function only when the hip was flexed to 90° and the knee was fully extended. Thus, the earlier stretching endpoint observed in our study when the hip was flexed with the knee fully extended could be explained by the increased mechanosensitivity of peripheral nerves. This is thought to be a normal protective response to the stresses applied to nerves during limb movement (Boyd et al., 2009). While this is a plausible possibility, there could also be other neural mechanisms that result in increased perception of tension or end range sensation. For instance, increased tension in a muscle group (e.g., muscles crossing hip and knee) might cause inhibition-excitation on sensory pathways through various interneurons.

In respect to the EMG results, these suggest that the changes in ROM observed were unlikely to have been influenced by local muscle activity. The low values of EMG activity in our study were in accordance with previous work (McNair et al., 2001, 2002; Nordez et al., 2010). More importantly, no hip and head effects were observed across ROM. In addition, the stretching repetitions performed in the present study should influence the passive torque and shear elastic modulus. It is classically considered that 5 min of rest is sufficient to counteract the effects of five cyclic stretching repetitions at slow velocity (Nordez et al., 2008, 2009). Moreover, the order of testing was randomized. Therefore, while the possibility of stretching effects can not be excluded, we think that it did not influence our results.

The present study also showed that the maximal ankle dorsiflexion angle, torque, and muscle tension of GM were unaffected by the head position, thus providing some evidence that the structure that limited ankle ROM was not strongly linked to the cervical spine. Similarly, Ellis et al. (2012) showed that minimal sciatic nerve excursion was evident during isolated cervical flexion.

In conclusion, our results suggest that less typically reported structures that cross both the hip and the ankle joint contribute to the ROM at the latter joint. Although our findings provide a better understanding of inter-joint biomechanical behavior, further investigations are required to explore the mechanisms behind this finding. In particular, the use of elastographic methods could provide interesting information about the role of peripheral nerve mechanics on joint motion.

Perspectives

Different hip angle positions notably affect the maximal ankle dorsiflexion angle, while the ankle torque and passive tension of the GM are unchanged for a given common ankle angle. The angle of the head did not cause any further change in ankle dorsiflexion ROM. This knowledge regarding the varied amount of ankle angle observed without any further changes in muscle tension and ankle torque should be important to aid in the design of lower limb manual therapy exercises, such as neurodynamic tests, and diagnosis maneuvers that simultaneously use ankle dorsiflexion and different hip angles (e.g., SLR test). Furthermore, the findings of this study suggest that both peripheral nerves and fasciae should be studied in the future to better understand their role in the ROM limits of the ankle joint. This study was limited to participants with no history of muscle-tendon, articular, or nerve pathologies. The same kind of experiment could be valuable to analyze the factors that affect the ROM limits in individuals with neuromuscular disorders.

Key words: Stretching, muscle, elastography, supersonic shear imaging, sciatic nerve, fascia, flexibility, neurodynamics, range of motion.

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