Journal of Biomechanics 93 (2019) 111–117

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Influence of low muscle activation levels on the ankle torque and muscle shear modulus during plantar flexor stretching

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ARTICLE INFO

Article history:
Accepted 20 June 2019

Keywords:
Stretching
Plantarflexors
Electromyography
Torque
Shear modulus

ABSTRACT

During stretching studies, surface electromyography (sEMG) is used to ensure the passive state of the muscle, for the characterization of passive muscle mechanical properties. Different thresholds (1%, 2% or 5% of maximal) are indifferently used to set "passive state". This study aimed to investigate the effects of a slight activity on the joint and muscle mechanical properties during stretching.

The joint torque and muscle shear modulus of the *triceps surae* muscles were measured in fifteen healthy volunteers during ankle dorsiflexions: (i) in a "fully relaxed" state, (ii) during active conditions where participants were asked to produce an sEMG amplitude of 1%, 2% or 5% of their maximal sEMG amplitude of the *triceps surae*. The 1% condition was the only that did not result in significant differences in joint torque or shear modulus compared to the relaxed condition. In the 2% condition, increases in joint torque were found at 80% of the maximal angle in dorsiflexion, and in the shear modulus of *gastrocnemius medialis* and *gastrocnemius lateralis* at the maximal angle in dorsiflexion. During the 5% condition, joint torque and the shear modulus of *gastrocnemius medialis* were higher than during relaxed condition at angles larger than 40% of maximal angle in dorsiflexion. The results provide new insights on the thresholds that should be considered for the design of stretching studies. A threshold of 1% seems much more appropriate than a 2% or 5% threshold in healthy participants. Further studies are required to define similar thresholds for patients.

1. Introduction

Passive mechanical properties are important components of muscle function (Proske and Morgan, 1999; Azizi, 2014) and are related to muscle extensibility (Gajdosik, 2001). Several studies have shown the importance of measuring these properties to understand acute or chronic adaptations after trauma (Shelbourne et al., 1996), injury (Diong et al., 2012), immobilization (de Boer et al., 2007), or training (Fouré et al., 2013). In humans, the majority of studies evaluating these properties measure the passive joint [for review, see Magnusson (1998); Gajdosik (2001)], muscle fascicle length [using B-mode ultrasound, (Herbert et al., 2002)] or muscle shear modulus using shear wave elastography (Maïsetti et al., 2012). To accurately reflect passive mechanical properties, these approaches require the muscle to be fully inactive during stretch.

To ensure that muscles are in a passive state, it is important to monitor muscle activation to detect any unwanted activity that might influence the passive joint torque, the muscle fascicle length or the muscle shear modulus. For that purpose, myoelectrical activity is often measured through surface electromyography (sEMG) (McNair and Portero, 2005). The amplitude of the sEMG signal is then normalized to measurements during isometric maximal voluntary contractions (MVC) (McNair and Portero, 2005). Most of previous studies considered muscle to be passive when the sEMG amplitude remained below a given threshold (Gajdosik, 2001). However, a wide range of thresholds has been used to qualify such a passive state: 1% (McNair et al., 2001), 2% (Nordez et al., 2010), 5% (Gajdosik et al., 2005), up to 10% of the maximal activation (Halbertsma et al., 1999; Bar-On et al., 2018).

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https://doi.org/10.1016/j.jbiomech.2019.06.018
0021-9290/

As some of these thresholds were relatively high, it is likely that low activation was present during a supposedly passive stretch. To our knowledge, only Foo et al. (2019) reported that hamstring activity might decrease the range of motion of the hip joint during a straight leg raise test (−0.6° for every increase of 1% in myoelectrical activation). The effects of such low muscle activation on muscle mechanical properties are still misunderstood. Even though activation as low as 1% of maximal activation are likely to have minimal influence, it is necessary to determine activation levels that significantly affect the estimation of muscle mechanical properties.

The present study was designed to quantify the effect of low muscle activation levels on the mechanical properties of ankle joint and *triceps surae* (TS) muscles. Ankle joint was chosen here because ankle joint torque and TS mechanical properties are often examined in stretching studies. To this end, joint torque and shear modulus of the three heads of TS were measured while the ankle was dorsiflexed by an ergometer. Participants were asked to produce low muscle activation using a visual feedback of sEMG amplitude (detailed below in protocol; at 1%, 2% and 5% of maximal sEMG amplitude) or to stay fully passive [named "relaxed" condition] where participants were asked to stay as relaxed as possible during stretching.

2. Methods

2.1. Participants

A sample size calculation for the study was based on analyse of variance (ANOVA repeated measures, within-between interactions) for testing the association at 5% level between the conditions, angles, and muscle shear modulus. Using an effect size of $\eta_p^2 = 0.14$ and a power of 0.95, 12 participants would be required to detect significant difference [Gpower 3.1.9.2; Faul et al. (2007)]. Fifteen healthy participants (male/female: 12/3; age: 24.1 ± 4.3 years; height: 177 ± 7 cm; body mass: 65.5 ± 14.9 kg) volunteered to participate in this study. The study received Institutional Ethics Committee approval (CPP-MIP-015). The procedures conformed to the Declaration of Helsinki.

2.2. Equipment

2.2.1. Ergometer

An isokinetic dynamometer (Biodex 3 Medical, Shirley, New York, USA) was used. The ankle joint was carefully aligned with the center of rotation of the ergometer by and experienced user (Bruening et al., 2008). 0° of ankle dorsiflexion was defined as the foot sole is perpendicular to the tibia, and negative angles relative to plantarflexion (PF). Additional thermoformable customed-device and adhesive non-elastic taping to ergometer Velcro-straps were realized to avoid foot or heel motion. Ankle angles and joint torque were recorded using an external 16-bit analog/digital converter (1 kHz, 16–35 PowerLab ADInstruments Inc., Colorado Springs, USA).

2.2.2. Surface electromyography (sEMG)

The myoelectrical activity of the *gastrocnemius medialis* and *gastrocnemius lateralis* (GM and GL, respectively), the *soleus* (SOL) and the *tibialis anterior* (TA) were recorded using hydrogel adhesive surface electrodes (KendallTM 100 foam-series, Covidien, Mansfield, USA). The skin was prepared according to the SENIAM guidelines (Hermens et al., 2000). Due to overlapping probe locations on the skin, ultrasound imaging ensured that the electrodes were positioned appropriately: longitudinally with respect to the muscle fascicle alignment, and away from neighboring muscles (Lacourpaille et al., 2017). sEMG signals were amplified with a biosignal amplifier (g.BSamp 0201a, Guger Technologies, Schiedlberg, Austria; gain = 1000) and digitized together with mechanical signals using the same acquisition system (see Ergometer) at a sampling rate of 1 kHz. For each muscle, the sEMG signal was filtered (band-pass: 20–400 Hz and band-stop: 50 Hz) in order to remove unwanted noise associated with artifacts. The signals were quantified in terms of root mean square (EMG-RMS) using a moving average window (300 ms width, LabChart v7.0, Inc., Colorado Springs, USA). The EMG-RMS signals for each muscle (GM, GL, SOL, and TA) was normalized to the maximal EMG-RMS value recorded during MVC. The EMG-RMS of TS was provided in real time on a computer screen (Fig. 1). The quality of the sEMG signals was evaluated by calculating (i) the baseline noise in % of the maximal value reached during MVC; (ii) a signal-to-noise ratio (SNR, Matlab® function *snr*) (Table 1).

2.2.3. Shear wave elastography (SWE)

An ultrasound scanner (Supersonic Imagine, v. 6.1, Aix-en-Provence, France) coupled with a linear transducer [2–10 MHz for SOL or 4–15 MHz for GM and GL, Supersonic Imagine, Aix-en-Provence, France] was used in shear wave elastography mode (MSK preset, penetration mode, spatial smoothing 5/9, persistence off, scale: 0–600 kPa). The shear modulus (μ) is related to the shear wave velocity (Vs) measured using elastography:

$$\mu = \rho Vs^2 \quad (1)$$

where ρ the density of the tissue (1000 kg m^{-3} for muscle).

The shear modulus has been shown to have strong linear relationship with Young's modulus (R2 between 0.916 and 0.988) (Eby et al., 2013; Koo et al., 2013) and thus, can be used as an index of changes in muscle stiffness (Hug et al., 2015). The transducer was aligned along the longitudinal axis of the leg (i.e., the direction of shortening/lengthening) and perpendicular to the skin so that the image plane perpendicularly intersects the muscle aponeurosis (Le Sant et al., 2017). Previous results from our team showed very good reliability of SWE for measuring the shear modulus of *triceps surae* muscles during stretching (Le Sant et al., 2017). A transistor-transistor logic pulse (duration < 50 ms) was sent by the ultrasound scanner at each shear modulus measurement (i.e. each second) that allowed to synchronize shear modulus measurements with the ankle angle, passive torque and sEMG signals.

2.3. Protocol

Participants first performed a familiarization session. They laid prone on the ergometer with the ankle to be tested firmly strapped to the footplate of the device. First, the participants performed 3 isometric MVCs (separated by a rest of 90 s) in PF and in DF, respectively. Second, the maximal angle in DF (max DF) was defined. For that purpose, the ankle was moved by the ergometer at 2°/s, and participants pressed a button when they felt their maximum tolerable stretch limit (onset of pain in the posterior region of the leg). The highest angle obtained during three trials was kept as their maximal DF. Three conditioning cycles (from 40° of PF to max DF, 2°/s) were then performed (Nordez et al., 2008). Third, participants underwent stretching at 2°/s for each condition [1%, 2%, 5% of maximal EMG-RMS(TS) and "relaxed" conditions]. The peak EMG-RMS(TS) from maximal plantarflexion was used to calculate submaximal levels. Participants were asked to maintain a given muscle activity level during the ankle motion, using feedback provided in real time on a screen (Fig. 1). The familiarization continued until satisfactory, i.e. if the averaged muscle activity [EMG-RMS(TS)] corresponded to the targeted level ± ~0.5%. All

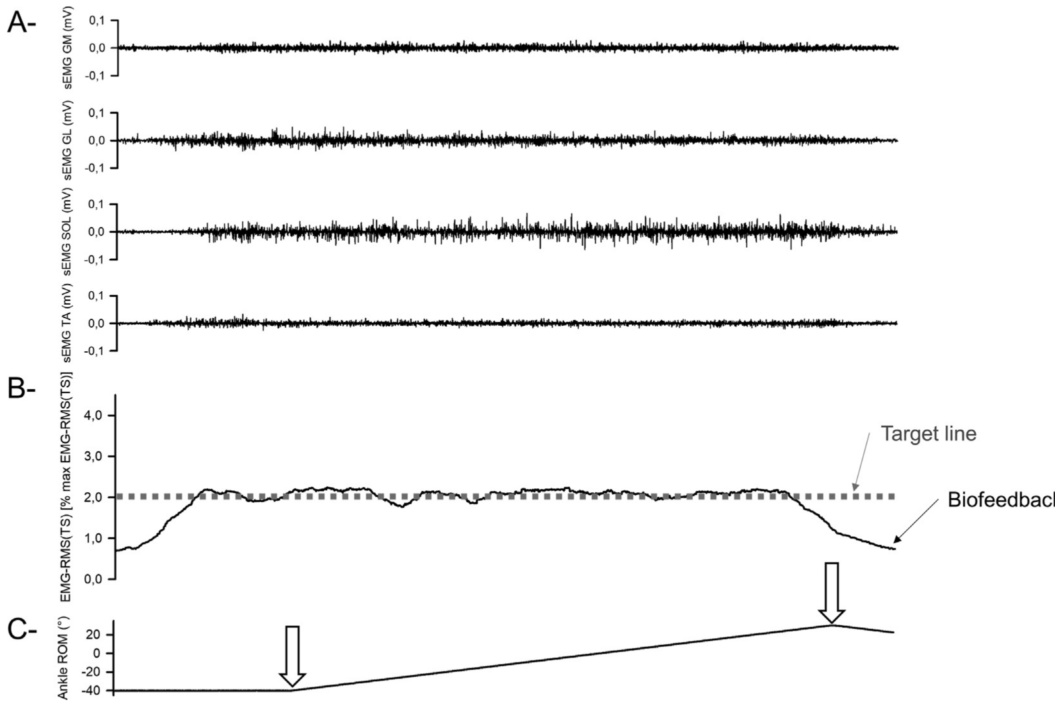

Fig. 1. Typical example of biofeedback for one participant. (A) The EMG-RMS(TS) was calculated via the sEMG of *triceps surae* muscles (mean of EMG-RMS of *triceps surae* muscles). (B) The EMG-RMS(TS) was provided as a biofeedback to the participant in real-time. The participant was asked surimpose the biofeedback line to the target line (grey dashed line). (C) The muscle activity was maintained during the ankle motion (from 40° of plantarflexion to the maximal ROM in dorsiflexion, 2°/s, white arrows). Negative angles relate to ROM in plantarflexion, while positive angles relate to ROM in dorsiflexion. GM: *gastrocnemius medialis*; GL: *gastrocnemius lateralis*; ROM: range of motion; EMG-RMS: root mean square; sEMG: surface electromyography; SOL: *soleus*; TA: *tibialis anterior*; TS: *triceps surae.*

Table 1
Quality of the myolectrical activity recorded for each muscle during data collection.

Muscle	EMG-RMS (baseline)/ EMG-RMS (MVC) (%)	Signal to noise ratio (dB)
GM	0.35 ± 0.11	37.8 ± 3.1
GL	0.32 ± 0.15	37.6 ± 4.3
SOL	0.44 ± 0.18	33.1 ± 3.7
TA	0.33 ± 0.22	42.7 ± 3.7

Data are shown as mean ± standard deviation.
Legend: GM: gastrocnemius medialis; GL: gastrocnemius lateralis; SOL: soleus; TA: tibialis anterior; EMG-RMS: root mean square.

the participants were able to perform the task and completed the second session dedicated to data collection.

The second session was performed between one and seven days after the familiarization session. The procedures were repeated: MVCs, maximal ROM assessments, and conditioning. Then, the participants were asked to match a target during ankle dorsiflexions (40° of PF to max DF, 2°/s) at either 1%, 2%, 5% of maximal EMG-RMS(TS) or during the "relaxed" condition (random order). For each condition, three dorsiflexions were performed such that the shear modulus of each muscle of the TS was measured (random order). The transducer was placed over the mid muscle region for GM and GL, and over the distal region for SOL, as defined in Le Sant et al. (2017). A total of twelve dorsiflexions were realized (4 conditions × 3 muscles) during which the joint torque, shear modulus of one muscle and muscle activity levels of four muscles were measured.

2.4. Data analysis

Data were processed using Matlab® scripts (The MathWorks Inc., Natick, USA) as previously described in details (Le Sant et al., 2017). Shear modulus values were averaged over the largest region of interest (ROI) that avoided aponeurosis and artefacts (average surface of 92 mm^2, 89 mm^2 and 86 mm^2 for the GM, GL and SOL). Angle and joint torque were filtered (Butterworth low-pass filter, cutoff 10 Hz) and torque corrected for gravity. Maximal EMG-RMS for each muscle corresponded to the peak value reached during MVC. For each trial, the EMG-RMS of each muscle (moving windows width: 300 ms) and joint torque were calculated for each shear modulus measurement (ie. at 1 Hz, timing determined by using the transistor-transistor logic pulse sent by the ultrasound scanner). Because the max DF angle differed among individuals, the angle was normalized to the maximal ROM (i.e., 0% as 40° PF, and 100% as the max DF). Then, EMG-RMS, passive torque and shear modulus values were calculated every 5% (i.e., 21 values) using linear interpolation.

2.5. Statistics

The data were processed using Statistica® (v.10, Statsoft Inc., Tulsa, USA). Data passed the Shapiro-Wilk normality test. For each condition, the averaged value of the 3 trials was used for joint torque and RMS-EMG. The changes in the measured EMG-RMS across conditions were assessed using a repeated-measures ANOVA (4 conditions × 4 angles × 3 muscles). The four conditions were "relaxed", 1%, 2%, and 5% of max EMG-RMS(TS). The four angles corresponded to the angle reached by each participant at 0%, 40%, 80% and 100% (i.e., max DF) of ankle ROM. Another repeated measures ANOVA (4 conditions × 4 angles) was used to assess changes in EMG-RMS(TA). These analyses were realized to confirm that the conditions were well achieved (ie, targeted level, without any potential influences of TA). In addition, we wanted to assess whether conditions affect in the same way or not the various muscles depending on joint angles. Two repeated-measures ANOVAs were conducted to assess the impact of the condition on joint torque (4 conditions × 4 angles) and on shear modulus (4 conditions × 4 angles × 3 muscles [GM, GL and SOL]). We hypothesized an increase in joint torque and muscle shear modulus at a same ankle angle with the increasing level of muscle activity. Partial eta square (η_p^2) was reported as effect-size estimation. Small, medium and large effects were considered for $\eta_p^2 = 0.01$, $\eta_p^2 = 0.07$ and $\eta_p^2 \geq 0.14$, respectively (Cohen, 1988). To account for hidden multiplicity in exploratory multiway ANOVA, a Benjamini-Hochberg procedure was used as described in Cramer et al. (2016). The p-values from the ANOVAS were sorted in ranks from smallest to largest (7 ranks for RMS-EMG, 3 ranks for torque, 7 ranks for shear modulus, respectively) and were judged significant if inferior to the corresponding Benjamini-Hochberg critical value (Benjamini and Hochberg, 1995). Post-hoc analyses were performed using Bonferroni correction. In order to better describe the small differences between the "relaxed" and 1% conditions, Bland-Altman plots were generated to study the agreements for joint torque, shear modulus and muscle activity (Bland and Altman, 1986). The statistical significance was set at $p < 0.05$. Data are reported as mean ± standard deviation (SD).

3. Results

The averaged ± SD angles were −40.0 ± 0.4° PF (0%), −10.0 ± 2.6° PF (40%), 20.2 ± 5.3° DF (80%) and 35.3 ± 6.6° DF (100%), respectively (Figs. 1–4).

3.1. Effects of condition on muscle activation

The averaged EMG-RMS of the TS were 0.4 ± 0.1% (range 0.2–0.7%), 1.2 ± 0.2% (range 0.9–1.6%), 2.1 ± 0.3% (range 1.7–2.7%) and 4.8% ± 0.7% (range 4.3–6.7%) of max EMG-RMS(TS) for the "relaxed", 1%, 2% and 5% conditions, respectively (Fig. 2).

As expected, there was a main effect of condition ($F(3,42) = 417.96$, $p < 0.001$, $\eta_p^2 = 0.97$) indicating an increase in EMG-RMS among TS muscles with the increase in activation level target. Main effects of angle ($F(3,42) = 6.62$, $p < 0.001$, $\eta_p^2 = 0.32$), and muscle ($F(2,28) = 6.13$, $p = 0.006$, $\eta_p^2 = 0.30$) were also found. Condition × muscle ($F(6,84) = 5.12$, $p < 0.001$, $\eta_p^2 = 0.27$), condition × angle ($F(9,126) = 3.39$, $p < 0.001$, $\eta_p^2 = 0.19$), muscle × angle ($F(6,84) = 23.65$, $p < 0.001$, $\eta_p^2 = 0.63$), and condition × muscle × angle ($F(18,252) = 21.12$, $p < 0.001$, $\eta_p^2 = 0.59$) interactions were also significant. Post-hoc analyses revealed an increase in EMG-RMS(SOL) at 80% and at max DF during the 2% and 5% conditions (all p-values < 0.001) and a decrease in EMG-RMS(GM) at max DF ($p < 0.001$) and in EMG-RMS(GL) during the 5% condition (all p-values < 0.005). Regarding the TA, there were main effects of condition ($F(3,42) = 32.62$, $p < 0.001$, $\eta_p^2 = 0.70$) and angle ($F(3,42) = 4.09$, $p = 0.012$, $\eta_p^2 = 0.23$), but the interaction

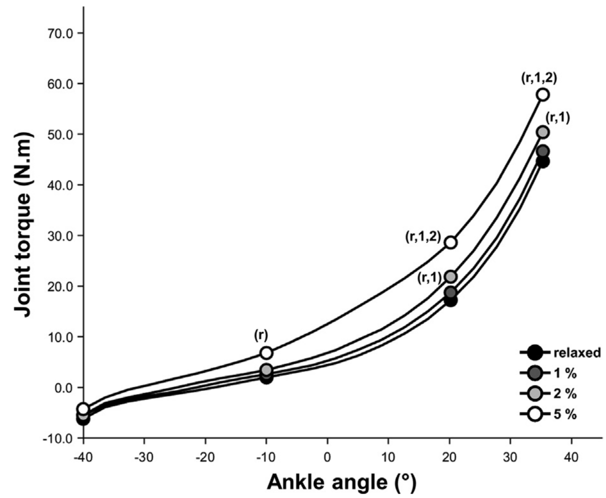

Fig. 3. Averaged joint torque-ankle angle relationships for each tested condition ("relaxed", 1%, 2% and 5% of maximal muscle activity). Standard deviations bars were omitted for clarity. Negative angles relate to range of motion in plantarflexion, while positive angles relate to range of motion in dorsiflexion. Significant differences ($p < 0.05$) are reported at the tested angles relative to the other conditions: (r) "relaxed", (1) 1%, (2) 2%, and (5) 5% conditions, respectively.

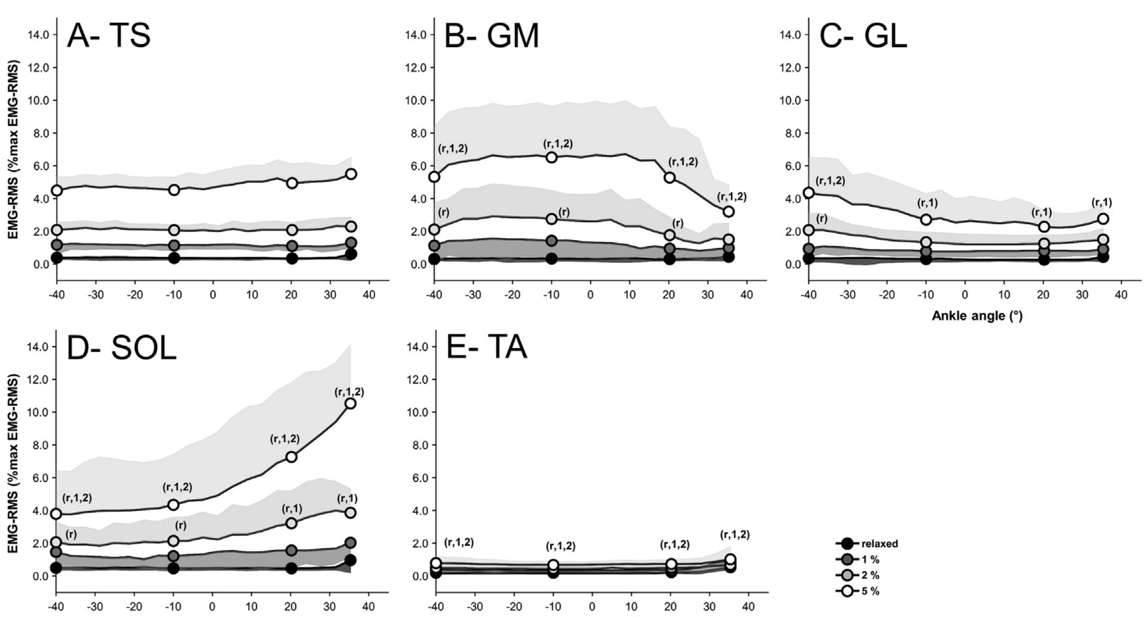

Fig. 2. Averaged (standard deviation in shaded) EMG-RMS-ankle angle relationships for each condition ["relaxed", 1%, 2% and 5% of maximal muscle activity (max EMG-RMS)] for *triceps surae* (A), *gastrocnemius medialis* (B), *gastrocnemius lateralis* (C), *soleus* (D), and *tibialis anterior* (E). Negative angles relate to range of motion in plantarflexion, while positive angles relate to range of motion in dorsiflexion. Significant differences ($p < 0.05$) are reported at the tested angles relative to the other conditions: (r) "relaxed", (1) 1%, (2) 2%, and (5) 5% conditions, respectively.

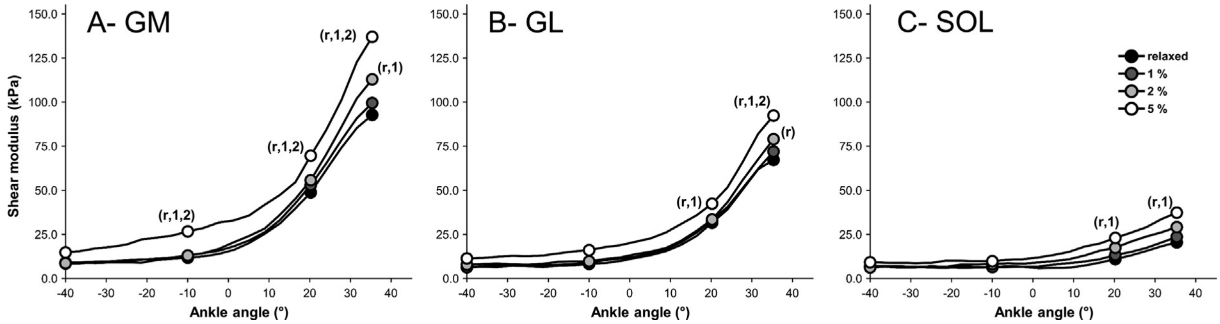

Fig. 4. Averaged shear modulus-ankle angle relationships for each tested condition ("relaxed", 1%, 2% and 5% of maximal muscle activity) for *gastrocnemius medialis* (A), *gastrocnemius lateralis* (B), and *soleus* (C). Standard deviations bars were omitted for clarity. Negative angles relate to range of motion in plantarflexion, while positive angles relate to range of motion in dorsiflexion. Significant differences ($p < 0.05$) are reported at the tested angles relative to the other conditions: (r) "relaxed", (1) 1%, (2) 2%, and (5) 5% conditions, respectively.

condition × angle was not significant ($F(9,126) = 0.78$, $p = 0.64$). EMG-RMS(TA) was significantly higher during 5% condition than during the other conditions (all p-values < 0.018). During the 2% condition, EMG-RMS(TA) was higher at max DF than at 40% and 80% ($p < 0.001$ and $p = 0.032$, respectively). During the 5% condition, EMG-RMS(TA) was higher at max DF (all p-values < 0.008) (Fig. 2).

In addition, regardless the muscle, EMG-RMS was not different between the "relaxed" and the 1% conditions (all p-values > 0.11). The mean difference between "relaxed" and 1% EMG-RMS was 0.77 ± 0.11% for the GM, 0.51 ± 0.05% for the GL, 0.97 ± 0.13% for the SOL, and 0.10 ± 0.32% for the TA (Supplementary File 1).

3.2. Effects of muscle activation on joint torque

A significant main effect of condition ($F(3,42) = 56.56$, $p < 0.001$, $\eta^2_p = 0.80$) and angle ($F(3,42) = 268.97$, $p < 0.001$, $\eta^2_p = 0.94$) were found. There was also a significant interaction condition × angle ($F(9,126) = 26.23$, $p < 0.001$, $\eta^2_p = 0.65$). Compared to the "relaxed" condition, joint torque was not different for the 1% condition (all p-values > 0.41). The mean difference between "relaxed" and 1% torque values was 1.2 ± 1.9 N m (Supplementary File 1). When compared to the relaxed condition, joint torque was significantly larger in the 5% condition at 40%, 80% and max DF angles, and in the 2% condition at 80% and max DF angles (all p-values < 0.001) (Fig. 3). At max DF, joint torque was 44.5 ± 12.0 N m, 46.5 ± 11.8 N m, 50.3 ± 12.4 N m, and 57.7 ± 12.6 N m, for the "relaxed", 1%, 2%, and 5% conditions, respectively (Fig. 3). When compared to the "relaxed" condition, the peak torque was increased by 33 ± 24% for the 5% condition, and by 14 ± 10% for the 2% condition.

3.3. Effects of muscle activation on shear modulus

The main effects of condition ($F(3,42) = 55.01$, $p < 0.001$, $\eta^2_p = 0.80$), muscle ($F(2,28) = 67.05$, $p < 0.001$, $\eta^2_p = 0.83$) and angle ($F(3,42) = 256.90$, $p < 0.001$, $\eta^2_p = 0.95$) were found on shear modulus (Fig. 4). There were also significant interactions: condition × muscle ($F(6,84) = 4.96$, $p < 0.001$, $\eta^2_p = 0.26$), condition × angle ($F(9,126) = 13.30$, $p < 0.001$, $\eta^2_p = 0.58$), muscle × angle ($F(6,84) = 65.64$, $p < 0.001$, $\eta^2_p = 0.82$) and condition × muscle × angle ($F(18,252) = 2.62$, $p < 0.001$, $\eta^2_p = 0.16$). When the 2% condition was compared to the relaxed condition, the only significant differences in shear modulus were found for GM and GL at max DF ($p < 0.001$) (Fig. 4). When the 5% condition was compared to the relaxed condition, significant differences were found at 40%, 80% and max DF for GM (all p-values < 0.001), and at 80% and max DF for GL and SOL (all p-values < 0.002), respectively. When compared to the "relaxed" condition, the peak shear modulus was increased by 26 ± 20%, 19 ± 9% and 49 ± 41% for the 2% condition, and by 55 ± 40%, 38 ± 13% and 100 ± 87% for the 5% condition, for GM, GL and SOL, respectively. No significant differences were found for the shear modulus between "relaxed" and 1% conditions (all p-values = 1.000). The mean difference between the "relaxed" and 1% shear modulus was 2.9 ± 0.9 kPa for the GM, 1.7 ± 0.6 kPa for the GL, and 1.4 ± 0.7 kPa for the SOL (Supplementary File 1).

4. Discussion

This study describes the effects of muscle activation levels that are usually considered 'negligible' during passive stretching on both joint torque and muscle shear modulus. The main finding was that muscle activation higher than 1% affects the joint torque and shear modulus, which increase with ankle dorsiflexion. In the 5% condition, the effect was evident at low angles (i.e., 40% of maximal DF) (Figs. 3 and 4).

Measuring myoelectrical activity during stretching studies is used to confirm the absence of unwanted muscle activation (McNair and Portero, 2005). This is crucial because muscle activation, albeit small, may affect muscle mechanical properties. For that purpose, participants are trained to stay "as relaxed as possible" and their spontaneous myoelectrical activity is measured during joint rotation (McNair and Portero, 2005). According to these studies, intrinsic stiffness and muscle activation should both contribute to joint torque in the last degrees of ankle's ROM towards DF. As a result of the study conducted by McNair et al. (2001), 80% of the max DF is commonly used to limit this unwanted activity. Beyond this ankle angle, myoelectrical activity was shown to increase significantly (McNair et al., 2001). In our study, a significant increase in the EMG-RMS of TS at 80% of max DF was only found during the 2% and 5% conditions. The results also revealed that it is possible for participants to maintain a very low level of myoelectrical activity among TS during stretching [0.4 ± 0.1% of max EMG-RMS(TS), during the "relaxed condition" Fig. 2]. We think that the familiarization session was helpful to obtain low EMG values during the stretching in the relaxed condition.

Regarding the shear modulus, the values reached at max DF were the highest for GM when compared to GL and SOL, whatever the tested condition. This result is in accordance with previous studies that reported between-muscle differences in shear modulus among plantar flexors under 'relaxed' conditions (Hirata et al., 2015; Hirata et al., 2016; Le Sant et al., 2017, 2019). Considering that the GM cross-sectional area is larger than that of GL (Fukunaga et al., 1992), passive force developed by the GM muscle during stretch is likely much larger than that produced by the GL muscle. However, considering a lower shear modulus for the SOL compared to the GM and GL and a higher volume, it is more difficult to estimate the contribution to the passive torque of the SOL using the methods of this study.

A threshold such as 5% of EMG-RMS(TS) is often used to evaluate elderly patients (Gajdosik et al., 2004, 2005) or people with neurological conditions (Kalkman et al., 2018; Le Sant et al., 2019). Such an activity level could have significantly influenced the results of these studies (Figs. 3 and 4). Our results indicate that using such a threshold could induce an overestimation of 64% and 33% for the muscle shear modulus and passive torque, respectively. However, because of structural differences with young populations (e.g., muscle volume, intra-muscular connective/adipose composition) it is difficult to estimate how much a 5% of EMG-RMS(TS) might have influenced the shear modulus and joint torque among elderly/patient populations. In addition, it remains difficult to be sure that patients reached maximal activity levels during MVC. It should have influenced the activity levels measured during stretching in previous studies with patients (Kalkman et al., 2018; Le Sant et al., 2019). Thus, further studies should be conducted in patients to assess the effects of unwanted EMG activity on passive torque and muscle behaviour during stretching. The use of motor blocks could help isolate the passive contribution to joint stiffness (Deltombe et al., 2004; Bollens et al., 2011). Few studies have assessed the impact of this approach on biomechanical parameters (Bleyenheuft et al., 2008; Buffenoir et al., 2013). The aforementioned studies showed that the passive resistance of the ankle joint was decreased after a selective block conducted at the superior *soleus* nerve. However, since they used sinusoidal perturbations applied to the ankle joint at different frequencies, it remains difficult to interpret those results in the context of classical stretching studies. Future studies, assessing the impact of nerve blocs of the TS could help to understand the impact of muscle activation 'at rest' (such as spastic dystonia) on the mechanical properties of the TS during stretching.

This study was designed to measure the effects of minimal activity on ankle joint and TS muscle mechanical properties during stretching. As described, a feedback EMG-RMS(TS) was used, based on an average EMG-RMS for the three heads of the TS, and an amplitude-normalization procedure (i.e., maximal EMG-RMS during MVC). This procedure supposes a high sensibility of sEMG to detect very low levels of muscle activity. Other techniques such as high-density EMG or intramuscular EMG using fine wires might have been more direct approaches of measurement. We chose our experimental set-up because it is used in classical stretching studies. The high maximal/baseline ratios, SNR values and Bland-Altman results make us confident that our procedures were appropriate, in particular for the lowest conditions ("relaxed" *versus* 1%). Participants reached and maintained the targeted levels during submaximal conditions (1%, 2% an 5%) using a biofeedback of normalized EMG-RMS(TS) provided in real time (Fig. 1). While the mean values were in accordance with the targets of the feedback, some inter-individual variability was present (average of ±0.32% in regard of the targeted level) but, the ranges of sEMG activity levels reached clearly showed that there was no overlap between conditions. However, the contributions of GM, GL and SOL to the EMG-RMS(TS) were not constant, in particular during the 5% condition (Fig. 2). It seems that a specific coordination between muscles occurred with a decrease in the activity of the *gastrocnemii* combined with an increase in the activity of SOL. These changes in muscle activity for such low levels were not expected to occur. Their mechanical consequences on joint torque are difficult to evaluate, but did not lead to visible changes in the shear modulus-angle relationships. However, this point would not alter the main conclusions of the present paper and the observations for the most active conditions (e.g., 5% of maximal) compared to the "relaxed" condition. As previously described in the method, the joint axis of rotation was carefully aligned with the axis of rotation of the ergometer. To our knowledge, the influence of a possible misalignment on the joint torque measured during passive stretching has not be quantified in the literature. However, considering that the foot was very firmly strapped in the present study, and movement of the foot are generally lower in passive conditions compared to maximal isometric contractions, we think that the effect of foot motion was small. It was confirmed in a previous study (Ates et al., 2018) that found a mean difference in ankle angle of 1.4° (Confidence Interval 95% 1.3°; 1.6°) between the ergometer and 3 optoelectronic system during a pilot-testing (n = 6, range of −20°PF to 20°DF). Therefore, this effect was probably small and marginally influenced our results. Due to potential various joint configuration (e.g., muscle volume, lever arm...) and neuromechanical properties (eg EMG-force relationships), it is difficult to extrapolate the results of the present study to other the joints and muscles. Therefore, further studies are needed to examine the effects of muscle activity during stretching on other joints (e.g., knee torque) with different structure/architecture.

5. Conclusion

This study showed that joint stiffness and muscle shear modulus (in particular GM) were higher if participants activated their TS muscle at 5% of its maximal voluntary, activation (measured by sEMG), during slow dorsiflexions of the ankle. A lower level of EMG amplitude (2%) had significant influence on joint torque and shear modulus at the end of the range of dorsiflexion. These results provide new insights that might be useful for the design of future stretching studies, as 5% has classically been used as a threshold to describe a "passive state". A threshold of 1% seems more appropriate for healthy participants. Further studies are required to set acceptable EMG thresholds for patients.

Author contributions

Conceived and design the study: GLS, RG, FH & AN. Performed experiments: GLS. Analyzed and interpreted data: GLS, RG & AN. Edited manuscript: GLS, RG, FH & AN.

Declaration of Competing Interest

No conflicts of interest, financial or otherwise, are declared by the authors.

Acknowledgements

The authors thank Valentin Doguet, PhD (University of Nantes) for technical support in data collection. The studies were supported by a grant from the University of Nantes (interdisciplinary program), and the Région des Pays de la Loire (QUETE project). François Hug is supported by a fellowship from the Institut Universitaire de France (IUF). The funders had no role in design, data collection and analysis, manuscript redaction and publication.

Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2019.06.018.

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