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Abstract:	In vivo assessment of the force-generating capacity of m. abductor hallucis (AbH) is problematic due to its combined abduction-flexion action and the inability of some individuals to voluntarily activate the muscle. This study investigated direct muscle electrical stimulation as a method to assess isometric force production in AbH about the 1st metatarsal phalangeal joint (1MPJ) at different muscle-tendon lengths, with the aim of identifying an optimal angle for force production. A 7s stimulation train was delivered at 20Hz pulse frequency and sub-maximal (150% motor threshold) intensity to the AbH of the left foot in 16 participants whilst seated, and with the Hallux suspended from a force transducer in 0°,5°,10°,15° and 20° 1MPJ dorsal flexion. Reflective markers positioned on the foot and force transducer were tracked with 5 optical cameras to continuously record the force profile and calculate the external 1MPJ joint flexion moment at each joint configuration. A parabolic relationship was found between AbH force production and 1MPJ configuration. The highest 1MPJ joint moments induced by electrical stimulation were found between 10° and 15° of Hallux dorsal flexion. However, the joint angle ($p<0.001$; $\eta 2=0.86$) changed significantly across all but one 1MPJ configurations tested during the stimulation-evoked contraction, resulting in a significant change in the corresponding external moment arm ($p<0.001$; $\eta 2=0.83$). Therefore, the changes in joint geometry during contraction should be accounted for to prevent an underestimation of the resulting joint moment. We conclude that direct muscle electrical stimulation combined with dynamometry offers a robust method for standardised assessment of AbH sub-maximal isometric force production.

Direct muscle electrical stimulation as a method for the *in vivo* assessment of force production in *m*. abductor hallucis

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Abstract

In vivo assessment of the force-generating capacity of *m*. abductor hallucis (AbH) is problematic due to its combined abduction-flexion action and the inability of some individuals to voluntarily activate the muscle. This study investigated direct muscle electrical stimulation as a method to assess isometric force production in AbH about the 1st metatarsal phalangeal joint (1MPJ) at different muscle-tendon lengths, with the aim of identifying an optimal angle for force production. A 7s stimulation train was delivered at 20Hz pulse frequency and sub-maximal (150% motor threshold) intensity to the AbH of the left foot in 16 participants whilst seated, and with the Hallux suspended from a force transducer in 0°,5°,10°,15° and 20° 1MPJ dorsal flexion. Reflective markers positioned on the foot and force transducer were tracked with 5 optical cameras to continuously record the force profile and calculate the external 1MPJ joint flexion moment at each joint configuration. A parabolic relationship was found between AbH force production and 1MPJ configuration. The highest 1MPJ joint moments induced by electrical stimulation were found between 10° and 15° of Hallux dorsal flexion. However, the joint angle ($p<0.001$; $\eta^2=0.86$) changed significantly across all but one 1MPJ configurations tested during the stimulation-evoked contraction, resulting in a significant change in the corresponding external moment arm ($p<0.001$; $\eta^2=0.83$). Therefore, the changes in joint geometry during contraction should be accounted for to prevent an underestimation of the resulting joint moment. We conclude that direct muscle electrical stimulation combined with dynamometry offers a robust method for standardised assessment of AbH sub-maximal isometric force production.

Introduction

M. abductor hallucis (AbH) is one of the strongest intrinsic foot muscles (Kura et al., 1997; Tosovic et al., 2012). Its low fibre-to-muscle length ratio predicates itself to force production in order to stabilise the 1st metatarsal phalangeal joint (1MPJ) for postural control (Fiolkowski et al., 2003; Kelly et al., 2012; Kelly et al., 2015) and forward progression during gait (Kelly et al., 2015; Farris et al., 2019). The muscle's capacity to generate force required for abduction-flexion of the Hallux is dependent on its complex multipennate fibre arrangement at its sites of origin (Tosovic et al., 2012). In Hallux Valgus, an insidious forefoot deformity that affects ~20% of adults aged 18 to 65 and ~35% over the age of 65 (Nix et al., 2010), this capacity is diminished (Arinci Incel et al., 2003) due to the inferior rotation of the AbH tendon under the proximal phalanx (Perera et al., 2011), which correspondingly alters the mechanical properties of the muscle (Stewart et al., 2013). With increasing severity of the deformity dysfunction of the muscle ensues (Eustace et al., 1994), leading to atrophy (Stewart et al., 2013) and an offloading of the Hallux and medial forefoot during gait (Galica et al., 2013; Shih et al., 2014). The consequence is an impaired gait pattern, particularly a higher than normal internal knee abduction moment (Shih et al., 2014); and postural instability, which in the elderly increases the likelihood of falling (Menz and Lord, 2005).

The functional assessment of AbH for diminished capacity or adaptation in response to exercise is constrained because of its combined abduction-flexion action and the inability of persons with Hallux Valgus to perform an isolated voluntary contraction of the muscle (Arinci-Incel et al., 2003; Stewart et al., 2013). A toe flexor maximal voluntary contraction protocol (Goldmann and Brüggemann, 2012; Kurihara et al., 2014; Latey et al., 2018; Yamauchi and Koyama, 2019a) is inappropriate, not least because of the recruitment of other intrinsic and extrinsic toe flexor muscles, but also because AbH activation during this movement may account for less than half of its maximal capacity (Yamauchi and Koyama, 2019b). Given the superficial location of AbH, ultrasonography has been widely used to assess the muscle morphology in the Hallux Valgus foot (Stewart et al., 2013; Aiyer et al., 2015; Lobo et al., 2016;

Mickle and Nester, 2018) since it is associated with muscle strength (Mickle et al., 2013). Whilst this might be true, ultrasonography does not provide an insight into the functional capacity of a muscle; therefore an alternative solution for direct assessment of isolated AbH force generating capacity is required.

Direct muscle electrical stimulation has been successfully used to assess the *in vivo* isometric functional capacity of upper (Leeham and Dowling, 1995) and lower extremity muscles (Koh and Herzog, 1995; Maganaris, 2001; Wüst et al., 2008; De Monte and Arampatzis, 2009). Despite the different motor unit recruitment patterns between voluntary and evoked contractions (Bickel et al., 2011), electrical stimulation provides a means to evoke a sustained tetanic contraction in AbH and isolate its mechanical action (James et al., 2018).

Presently, the optimum muscle-tendon length for AbH to produce force is uncertain. This can be identified *in vivo* by constructing a joint moment – angle relationship curve. Previous work on *m*. tibialis anterior (Koh and Herzog, 1995; Maganaris, 2001), *m*. soleus (Maganaris, 2001), triceps surae (De Monte and Arampatzis, 2009) and *m*. biceps brachii (Leedham and Dowling, 1995) has demonstrated that this relationship curve can be constructed by combining electrical stimulation with dynamometry. For AbH, this curve can reveal the relationship between the external joint moment acting about 1MPJ (in response to direct muscle electrical stimulation) and the range of angles over which 1MPJ operates. The resultant curve depicts the functional capacity of the muscle and allows for identification of a 1MPJ angle about which AbH can produce its greatest force.

However, both voluntary and evoked muscle activation alter the joint axis of rotation in relation to the axis of the dynamometer, which thereby alters the moment arm of the reaction force acting about the joint (Arampatzis et al., 2004; Arampatzis et al., 2005). This leads to a misrepresentation of the joint moments measured via dynamometry against those calculated using inverse dynamics. Indeed, previous studies that have accounted for the change in joint axis during muscle contraction have shown that the differences between the measured and the calculated joint moments can reach as high as 23% at the ankle (Arampatzis et al., 2005) and 17% at the knee (Arampatzis et al., 2004).

Therefore, the purpose of this study was to investigate whether direct muscle electrical stimulation combined with dynamometry can be used as a method for the *in vivo* assessment of AbH force production in healthy participants. This was performed at a sub-maximal stimulation intensity and at different muscle-tendon lengths in order to identify the optimal 1MPJ angle for force production. There were two hypotheses: 1) an optimal 1MPJ configuration for AbH force production will exist when the joint is positioned further into dorsal flexion; and 2) the stimulation-induced contraction will affect the 1MPJ axis of rotation and alter the corresponding moment arm.

Methods

Sixteen healthy volunteers (12M/4F, mean ± standard deviation [SD]: 25.6 ± 5.8 years, 78.8 ± 13.7 kg, 1.7 ± 0.1 m) provided written informed consent to participate in the study that had received prior local ethical approval (SAS1806a) and was compliant with the Declaration of Helsinki (2013). Prior to participation all volunteers completed a health screen questionnaire and reported good health and absence of lower extremity injuries, underlying pathologies and neurological problems.

Participants visited the laboratory twice: for familiarisation and for the main testing session. As part of the familiarisation visit, optimisation procedures for direct muscle electrical stimulation were performed and included AbH motor point area location and motor threshold determination. The navicular tuberosity served as the reference point to drawing a 7x4cm matrix on the skin overlying the target muscle (James et al., 2018). A single square-wave (500µs) pulse was delivered systematically over each point of the matrix at 10mA intensity using a constant-current stimulator (DS7A, Digitimer, UK) and a custom-made pen-type cathode with the anode fixed over the 1MPJ. The largest twitch force recorded by a uniaxial force transducer (range: 0-250N; RDP Electronics Ltd., UK), calibrated for measuring low forces and mounted to the experimental apparatus above the foot (Figure 1), was used to identify the motor point area of the muscle. Then, five 1ms pulses were delivered to this location at 20Hz pulse frequency (Jones et al., 1979) and increasing current intensity, starting at 0.5mA with increments of 0.5mA. AbH motor threshold was accepted when the stimulus intensity evoked a twitch force that exceeded the baseline force level, which was measured within a 1s window starting 1.5s prior to stimulus onset, by >2SD. These procedures were repeated at the start of the main trial to verify the motor point area and the motor threshold. Following verification, a 7s train of 1ms pulses was delivered to AbH at low-frequency (20Hz), at an intensity of 150% motor threshold (James et al., 2018), and at the following 5 sagittal plane 1MPJ angle configurations: neutral (0°) and 5°, 10°, 15° and 20° dorsal flexion. Angle 0° was always measured first in order to associate a representative force with neutral position. Thereafter, the order of testing in the remaining 4 joint configurations (5°- 20°) was randomised following a Latin-square design.

During the main tests participants were seated in a custom-made apparatus with their left foot securely fixed at the ankle and forefoot and positioned at 35º ankle plantar flexion with respect to foot flat (Figure 1A; Goldmann and Brüggemann, 2012). The Hallux was covered with a polymer gel support and secured to the uni-axial force transducer by way of a semi-rigid thermoplastic cable that encapsulated the proximal phalanx, immediately distal to the 1MPJ (Figure 1B). Five optical-based cameras (Oqus-3+, Qualisys AB, Sweden) were used to track the locations of 4mm retro-reflective passive markers placed on the navicular tuberosity, the anode overlying 1MPJ, and the interphalangeal joint of the Hallux (Figure 1B). The tracking first identified the starting 1MPJ configuration for each investigated angle and secondly, monitored this continuously throughout each 7s train of electrical stimulation. The line of pull from the force transducer was described by two markers placed in a vertical arrangement on its rigid shaft (Figure 1B). The external moment arm (*r*) was then defined as the perpendicular distance between the line of pull and the 1MPJ axis of rotation. The force data (500Hz) was synchronously recorded with the raw marker positions (50Hz) through an A/D convertor (Qualisys AB, Sweden) and imported into Spike2 software (v7.12, Cambridge Electronic Design Ltd., UK) for analysis. Waveforms for 1MPJ angle (°) and the external moment arm (m) were generated, and along with the force recording, were smoothed using a moving average function with a time constant of 0.1s. Then, the external joint flexion moment at each 1MPJ configuration was calculated using the standard equation: $\mathrm{M} = \mathrm{F} \bullet \mathrm{r}\,(\sin\theta)$; where θrepresents the sagittal plane angle formed by the line projected to the floor from markers d and e (Figure 1) and the horizontal distance from 1MPJ to the line of pull.

To assess the effect of stimulation-induced contraction on the 1MPJ axis of rotation, the maximal force and average values of the joint angle and the corresponding external moment arm were calculated (from two 3s-epoch observation windows) prior to (relaxed condition) and 1s into (contracted condition) each 7s-stimulation train for each 1MPJ configuration. Using the force registered within the selected 3s-epoch during the evoked contraction, the maximal external joint moment (N•m) was calculated twice for each 1MPJ configuration – first, using the external moment arm calculated from the 3s-epoch during the contracted condition (corrected joint moment), and second – using the external moment arm calculated from the 3s-epoch during the relaxed condition (uncorrected joint moment).

Individual values (n=16) for 1MPJ angle and the external moment arm were normally distributed (Shapiro-Wilk, SPSS v.21, IBM, USA); therefore, a two-way repeated measures ANOVA, with condition (relaxed vs contracted) and 1MPJ configuration (0°, 5°, 10°, 15°, 20°) as the within-subject factors, was performed to assess for main and interaction effects of condition with effect size (η^2). Multiple comparisons were made using a Bonferroni correction factor and statistical significances were accepted when $p \leq 0.05$.

Individual values (n=16) for the external joint moments were not normally distributed, even after Log transformation; therefore, non-parametric Wilcoxon sign-rank tests were performed to compare the uncorrected vs corrected joint moments at the corresponding 1MPJ configuration. Hence, the cut-off for accepting statistical significance here was increased to $p \leq 0.01$ to account for the multiple comparisons. To address the primary hypothesis of this study, the corrected external joint moments were statistically analysed using a Friedman, followed again by Wilcoxon Signed-Rank tests to assess for an optimal 1MPJ configuration for force production. Statistical significances were accepted when $p \leq 0.01$.

Results

The average electrical stimulation intensity delivered to participants to evoke a contraction at 150% motor threshold was 4.8 ± 2.2 mA. In 50% of participants, this was delivered to the motor point located 1cm posterior and 4cm distal to the navicular tuberosity. All other participants' motor points were within 1cm of this location.

Significant interaction ($p<0.001$; $\eta^2=0.73$) and main effects of condition ($p<0.001$; $\eta^2=0.86$) and joint configuration ($p<0.001$; $\eta^2=0.99$) were found for 1MPJ angle (Figure 2A). Post-hoc tests identified that 1MPJ dorsal flexion occurred during electrical stimulation to a significantly different degree between the relaxed and contracted conditions at all investigated 1MPJ configurations apart from 20° of dorsal flexion.

As a result of the change in joint angle, a significant difference ($p<0.001$; $\eta^2=0.83$, main effect of condition) was found in the external moment arm between relaxed and contracted conditions (Figure 2B). Specifically, during contraction the moment arm increased on average by up to 2mm. Thus, the corrected external joint moment was significantly greater than the uncorrected moment at all 1MPJ configurations (all $p \leq 0.001$) (Table 1; Figure 2D).

A significant main effect of 1MPJ configuration was found in the corrected external 1MPJ joint moment–angle relationship ($p<0.01$), which fits a parabolic-like curve (Figure 2C). The external joint moments at 10° and 15° 1MPJ dorsal flexion were significantly higher compared to 0°

(both $p<0.01$) and 5° ($p<0.01$, $p<0.05$, respectively), but not 20° (Table 1, Figure 2C). The external joint moment at 20° 1MPJ dorsal flexion was significantly higher than 0° ($p<0.05$).

Discussion

Dysfunction of *m.* abductor hallucis underlies common foot pathologies such as Hallux Valgus; thus, a robust method is required to evaluate functional improvements in this muscle in response to training, conservative treatment and/or surgery. A toe flexor maximal voluntary contraction protocol (Goldmann and Brüggemann, 2012; Yamauchi and Koyama, 2019a) is inadequate in this sense because of different intrinsic and extrinsic foot muscle synergies that are available for this movement (Yamauchi and Koyama, 2019b). Therefore, the present study aimed to investigate direct muscle electrical stimulation as a method to evaluate the *in vivo* force production of AbH. The study's hypotheses are supported with the following main findings: i) the highest 1MPJ external joint moments are produced at 10° and 15° of 1MPJ dorsal flexion; and ii) significant 1MPJ rotation occurs during AbH contraction, which increases the external moment arm and, if not accounted for, leads to a significant underestimation of the calculated joint moment.

Torque measurements of maximum isometric voluntary contractions have been shown to misrepresent the actual joint moments produced about the ankle (Arampatzis et al., 2005) and knee (Arampatzis et al., 2004) by as much as 23% and 17%, respectively. This was due to unavoidable relative movement of the joint axis in relation to the axis of the dynamometer during contraction, caused by the non-rigidity of the leg-measurement system. Similarly, in the present study, contraction-induced movement of the 1MPJ axis increased the external moment arm leading to an underestimation of the external joint moments by as much as 30%. The reason for this higher underestimation, when compared to the aforementioned studies, is likely due to the greater non-rigidity of our toe-dynamometer system. The important implication from this main finding of the present study is that any study wishing to replicate the present protocol needs to account for this non-rigidity, and the ensuing change in the external moment arm during contraction, to prevent significant underestimation of the resulting 1MPJ joint moment.

The joint moment–angle relationship determines the optimal muscle-tendon length for force production, and also, broadly identifies the operating region of a muscle or muscle group on the ‘hypothetical’ force–length (F–L) relationship curve (Leedham and Dowling, 1995; Maganaris, 2001; Kubo et al., 2006; De Monte and Arampatzis, 2009; Hahn et al., 2011). Using this approach, the ankle plantarflexors have been indicated to operate on the ascending limb of the F–L curve (Maganaris, 2001; De Monte and Arampatzis, 2009; Hahn et al., 2011), whereas the knee extensors do so around the curve’s plateau region (Karamanidis and Arampatzis, 2005; Kubo et al., 2006). The present findings imply that AbH may operate on both the ascending and descending limbs of the F–L curve, as demonstrated by the identified parabolic-like joint moment–angle relationship (Figure 2C). This potentially highlights the functional importance of this muscle within the foot; that it is able to generate maximal force within its normal operating length according to the specific demands placed on it (Rubenson et al., 2012).

AbH contributes to forefoot stiffness during the terminal phase of gait, and without its influence, the ankle joint is unable to generate sufficient mechanical power for propulsion (Farris et al., 2019). During this phase, the metatarsal-phalangeal joints extend from neutral to as high as 70° of dorsal flexion, with the largest sagittal plane joint moment occurring at around 50° (Farris et al., 2019). However, this includes the contributions of *m*. flexor hallucis longus and *m*. flexor digitorum longus, both of which are extrinsic foot muscles. Negating the influence of these, whilst still considering all intrinsic (foot) toe flexor muscles lowers the optimal metatarsal-phalangeal joints’ angle for force production to approximately 35° dorsal flexion (Goldmann and Brüggemann, 2012). In the present study, the optimal 1MPJ angle for isolated AbH force production, using a combination of direct muscle electrical stimulation and toe dynamometry, appears to reside between 10° to 15°. Based on the muscle’s joint moment–angle relationship at 1MPJ, there is no reason to anticipate that this optimum angle increases thereafter.

Therefore, the present study puts forward a practical protocol for the *in vivo* assessment of AbH functional capacity. This is particularly noteworthy for sufferers of Hallux Valgus deformity who have a diminished capacity in this muscle (Eustace et al., 1994). Having a robust diagnostic tool at hand can help inform an earlier stage intervention of conservative therapy to offset the insidious nature of the condition. The protocol overcomes an important limitation relating to the nature of assessing muscle functional capacity and force production. Commonly, individuals are required to maximally activate the target muscle, but for some muscles, particularly AbH (Arinci Incel et al., 2003; Boon and Harper, 2003; Stewart et al., 2013), this is not easily achieved. Stimulation-evoked muscle contraction on the other hand overcomes this limitation and standardises the force generation at a given intensity. However, our electrostimulation paradigm was delivered at a sub-maximal current intensity primarily to avoid participant discomfort; therefore, the maximal force-generating capacity of AbH is unlikely to have been revealed here. Future work will quantify the contribution of our current paradigm intensity to total AbH force generating capacity.

Unfortunately a comparison between voluntary and evoked AbH joint moment – angle relationship curves was not possible because of the inability of even healthy individuals to perform a true isolated AbH muscle action (Arinci Incel et al., 2003; Boon and Harper, 2003). A limitation of this study therefore is the uncertainty of how much the evoked joint moment – angle relationship curve differs from one constructed by voluntary contraction. To the best of our knowledge, only one study has directly compared this between voluntary and evoked muscle (*m*. tibialis anterior) responses (Koh and Herzog, 1995). Koh and Herzog (1995) found no differences in the normalised shape or amplitude of their curves when dorsiflexion MVC was compared to the force evoked by tetanic 20Hz and 40Hz direct muscle electrical stimulation. This gives us confidence that our protocol for functional assessment of AbH is trustworthy and has practical virtue; and whilst it may not capture all of the abduction force generated by AbH, unpublished work from our laboratory confirms that the majority of the force produced by AbH in response to our electrostimulation paradigm occurs in the sagittal plane (~85%).

In conclusion, the highest external joint moment produced by *m.* abductor hallucis in response to sub-maximal electrical stimulation occurs when the 1st metatarsal phalangeal joint is positioned between 10° and 15° of dorsal flexion. This joint moment however can be significantly underestimated if the changes in joint geometry during muscle contraction are not taken into account. Therefore, a robust and standardised approach for *in vivo* assessment of AbH force-generating capacity has been proposed. This method has practical implications for evaluation of the mechanical properties of this essential muscle within the foot as well as for determining the efficacy of strengthening and rehabilitation interventions.

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Disclosure of interest

No conflicts of interest are declared by the authors.

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Table 1. Mean (± SD, *n* = 16) uncorrected vs corrected external joint moments (N•m) at each 1MPJ angle configuration. † significantly different to the respective uncorrected joint moment at p≤0.001 level; [a] significantly different to the corrected joint moment at 0°; [b] significantly different to the corrected joint moment at 5°; * significantly different at p≤0.05 level; ** significantly different at p≤0.01level.

Figure 1. Experimental set-up and foot-hallux arrangement. **A)** Participant position on the custom-built apparatus with the left foot fixed to the foot platform and the ankle positioned at 35° plantarflexion. **B)** Sagittal plane view of the experimental foot in the neutral configuration (0°), the Hallux suspended from the uniaxial force transducer, and the retro-reflective marker placements at the: navicular tuberosity (a); first metatarsophalangeal joint (1MPJ, b); interphalangeal joint (c); and the proximal (d) and distal (e) shaft of the uniaxial force transducer. *r* represents the external moment arm length calculated as the perpendicular distance from the 1MPJ to the force line of pull $(\tan^{-1}\left(\frac{\Delta y}{\Delta x}\right) \text{ from markers d \& e})$ along the x-axis. **C)** The experimental foot positioned in 10° 1MPJ dorsal flexion with respect to the neutral configuration (0°). **D)** Coronal plane view of the foot and Hallux arrangement. The antereoposterior axis of 1MPJ coincides with the end of the foot platform to achieve Hallux suspension from the uni-axial force transducer.

Figure 2. Mean (±SD, n = 16) participant responses for: **A)** 1MPJ angle (°) during relaxed and contracted conditions; **B)** the external moment arm (m) during relaxed and contracted conditions; **C)** the corrected external joint flexion moment (N•m) at each 1MPJ angle configuration (x-error bars reflect the standard deviation values from y-axis in panel **A**); and **D)** comparison of the uncorrected (filled circles) vs corrected (unfilled circles) external joint moments. [a] indicates significantly different to 0°; [b] significantly different to 5°; * significantly different at $p \leq 0.05$ level; ** significantly different at $p \leq 0.01$ level; † significantly different between conditions at the respective 1MPJ configuration at $p \leq 0.001$ level.

Figure 1

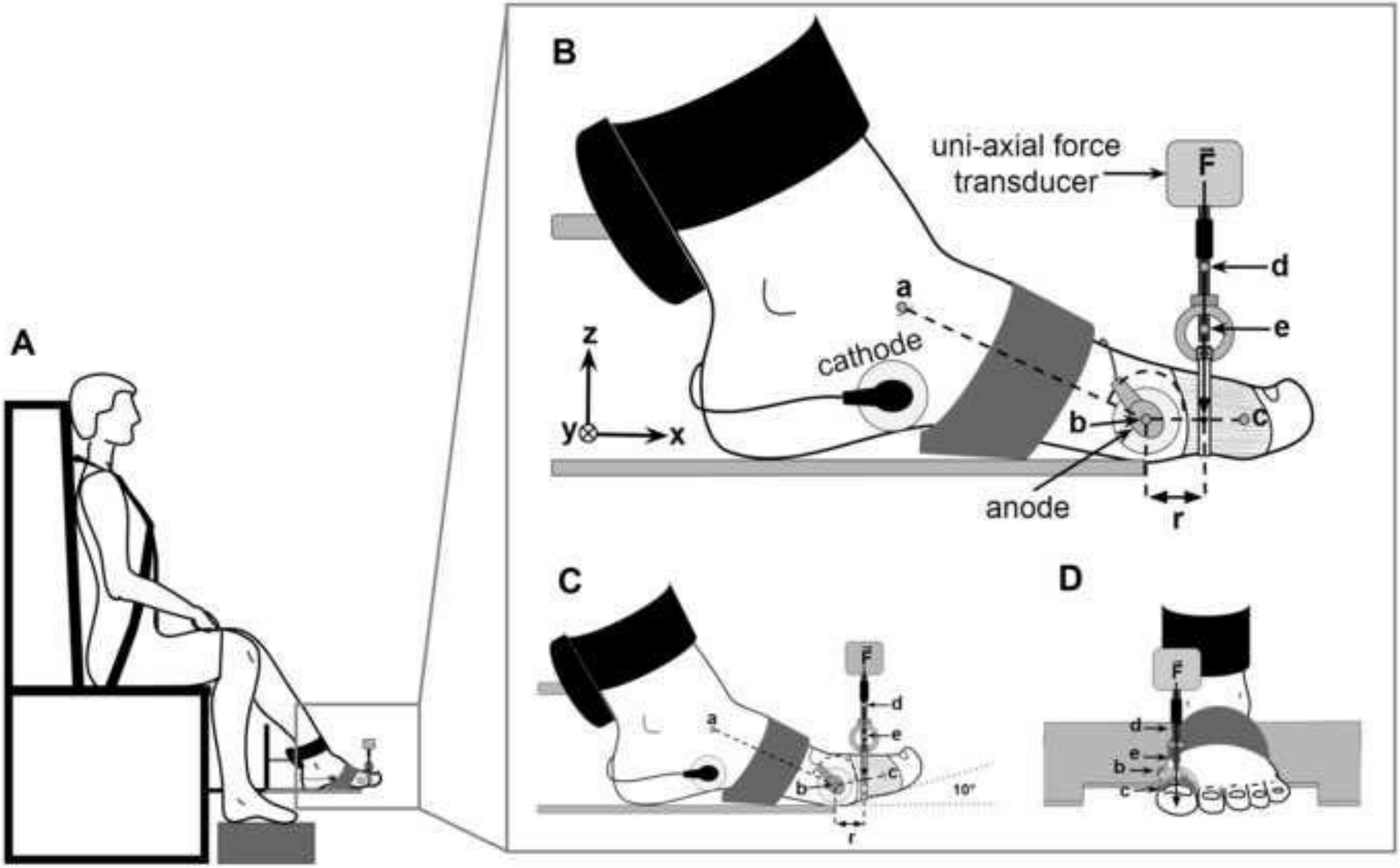
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Figure 2 Click here to access/download;Figure;Figure2.vReSubmit.ti

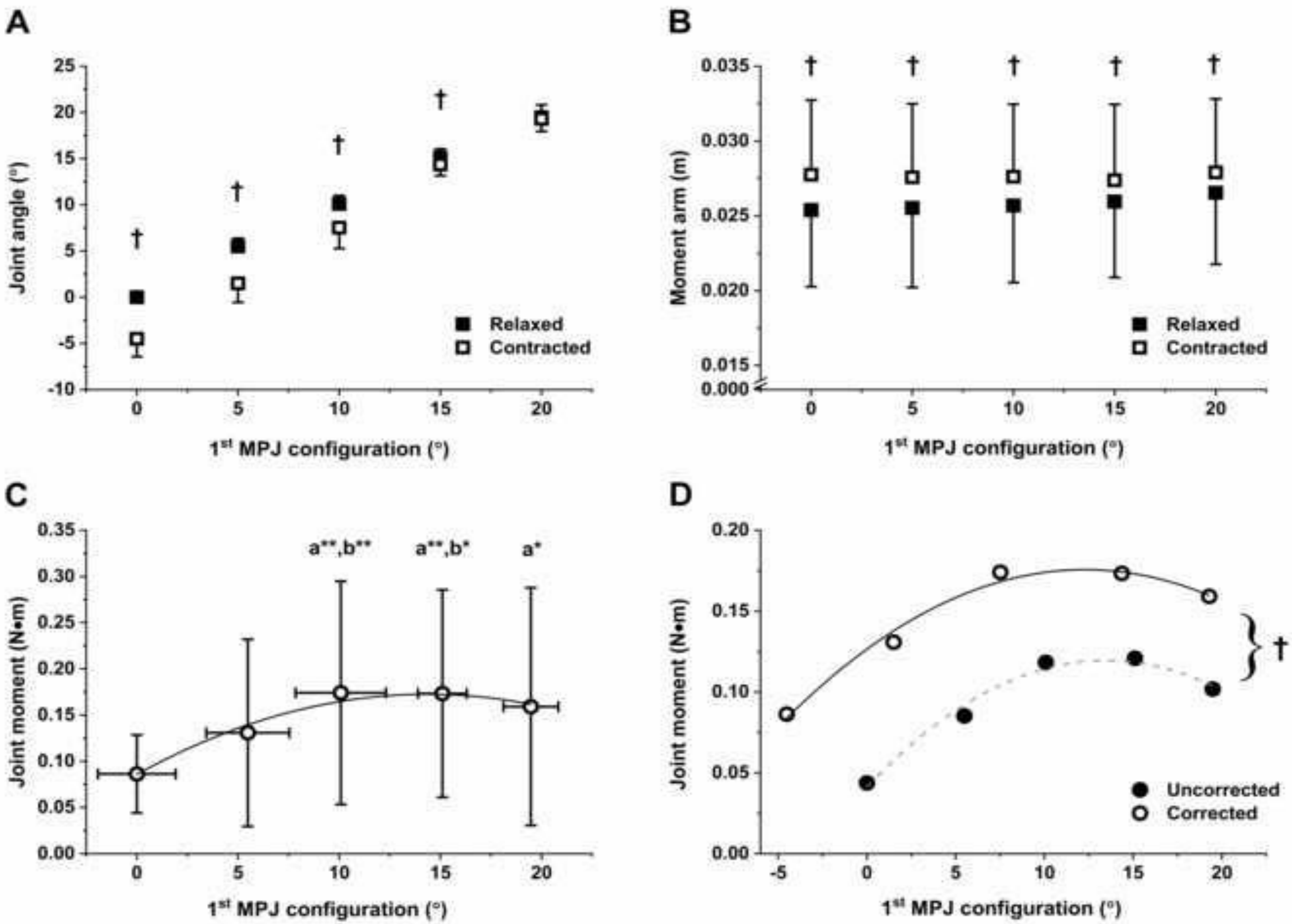

Table 1

	1st MPJ angle configurations				
	0°	**5°**	**10°**	**15°**	**20°**
Uncorrected (N•m)	0.04 (0.03)	0.09 (0.08)	0.12 (0.08)	0.12 (0.08)	0.10 (0.09)
Corrected (N•m)	0.09 † (0.04)	0.13 † (0.10)	0.17 †,a**,b** (0.12)	0.17 †,a**,b* (0.11)	0.16 †,a* (0.13)
Δ (%)	49.3	34.8	32.1	30.3	36.1

Conflict of Interest Statement

London South Bank University

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Author Declaration

We wish to confirm that there are no known conflicts of interest associated with this publication and there has been no significant financial support for this work that could have influenced its outcome.

We confirm that the manuscript has been read and approved by all named authors and that there are no other persons who satisfied the criteria for authorship but are not listed. We further confirm that the order of authors listed in the manuscript has been approved by all of us.

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