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The effectiveness of two novel techniques in establishing the mechanical and contractile responses of biceps femoris

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Abstract

Portable tensiomyography (TMG) and myotonometry (MMT) devices have been developed to measure mechanical and contractile properties of skeletal muscle. The aim of this study was to explore the sensitivity of the aforementioned techniques in detecting a change in passive mechanical properties of the biceps femoris (BF) muscle as a result of change in knee joint angle (i.e. muscle length). BF responses were assessed in 16 young participants (23.4 ± 4.9 years), at three knee joint angles (0°, 45° and 90°), for maximal isometric torque (MIT) along with myo-electrical activity. Contractile and mechanical properties were measured in a relaxed state. Inter-day reliability of the TMG and MMT was also assessed. MIT changed significantly ($p < 0.01$) across the three angles, so did stiffness and other parameters measured with MMT ($p < 0.01$). Conversely, TMG could detect changes only at two knee angles (0° and 45°, $p < 0.01$), when there is enough tension in the muscle. Reliability was overall insufficient for TMG whilst absolute reliability was excellent (coefficient of variation < 5%) for MMT. The ability of MMT more than TMG to detect an inherent change in stiffness can be conceivably exploited in a number of clinical/therapeutic applications that have to do with unnatural changes in passive muscle stiffness.

Keywords: force–length relationship, passive muscular tension, rate of torque development, neuromuscular efficiency

(Some figures in this article are in colour only in the electronic version)

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1. Introduction

Notoriously, active tension in human skeletal muscle is associated with cross-bridge interactions between myosin and actin and the concomitant ability to generate force (Rassier *et al* 1999), whereas passive tension has been attributed to connective tissue elements between muscle fibres, along with sarcolemma and sarcoplasm (Whitehead *et al* 2001). The length of human skeletal muscle affects both the ability to generate force (e.g. Rassier *et al* 1999), and its passive mechanical properties (Hoang *et al* 2007). Likewise, factors other than length can alter passive mechanical properties of skeletal muscle. A number of different studies have examined changes in passive muscle tension as a result of eccentric exercise (Hoang *et al* 2007, Whitehead *et al* 2001), stretching manoeuvres (Magnusson 1998, Reisman *et al* 2009), massage (Huang *et al* 2010), or neuromuscular disorders (Alhusaini *et al* 2010).

Research looking at mechanical properties of skeletal muscle is generally carried out in laboratories with heavy and expensive equipment (e.g. isokinetic and ultrasound machines). Two new *in vivo*, non-invasive methods, tensiomyography (TMG) and myotonometry (MMT), have been developed in the last 15 years to study mechanical and contractile muscular properties with the additional advantage of being portable and relatively inexpensive. TMG has been used to measure muscle action characteristics (contraction time and displacement) (Dahmane *et al* 2001), muscle tone (Valencic and Knez 1997) and muscle fibre type (Dahmane *et al* 2001, 2005, Simunic *et al* 2011). With this device, an electrical stimulus is applied percutaneously, and the consequent displacement caused by the muscle contraction is measured by a digital transducer pressed perpendicularly above the muscle belly. This method has been reported to be a useful tool in injury prevention and for the detection of muscle imbalances and asymmetries (Tous-Fajardo *et al* 2010). MMT has been adopted to characterize the viscoelastic properties of skeletal muscle as tissue displacement when a mechanical perturbation is applied to the muscle in a relaxed state (Hein and Vain 1998). The frequency and decrement of damping oscillations which are then obtained can be used to calculate muscle stiffness (Korhonen *et al* 2005, Viir *et al* 2006, Bizzini and Mannion 2003). Potential interesting applications of this method include the effect of massage, stretching, relaxation therapy, muscle relaxants and disuse on muscle stiffness (Bizzini and Mannion 2003).

Although reliability for TMG (Tous-Fajardo *et al* 2010) and MMT (Bizzini and Mannion 2003) has been established, only limited research is available looking at muscle function by using the described methods (e.g. Gavronski *et al* 2007, Pisot *et al* 2008, Carrasco *et al* 2011). Accordingly, the main aims of this study were, firstly, to explore the sensitivity of TMG and MMT in detecting contractile and mechanical responses of the biceps femoris (BF) muscle to alterations in knee joint angle (i.e. muscle length), and secondly, to ascertain the inter-day reliability of the measured variables. BF, because of its anatomical position, is a muscle easily measurable with both TMG (Pisot *et al* 2008) and MMT (Bizzini and Mannion 2003) techniques. Furthermore, a hamstring muscle complex is a major contributor during dynamic activity. As such, it is one of the most likely to get injured (e.g. Petersen *et al* 2010) and a significant correlation has been demonstrated between passive hamstring stiffness and symptoms of muscle damage (McHugh *et al* 1999).

It was hypothesized that TMG and MMT variables would change as a result of knee joint angle alteration. Based on the limited literature available on the same or different musculatures (Bizzini and Mannion 2003, Carrasco *et al* 2011), it was also hypothesized that the TMG and MMT variables would show acceptance to good reliability.

2. Methods

2.1. Research design

In this descriptive study, participants were tested, at the knee joint angles of 0° (leg fully extended), 45° and 90°, for unilateral dominant leg isometric maximal voluntary contraction, surface electromyography (sEMG), and contractile and mechanical properties of the BF, respectively. The latter were measured using TMG and MMT techniques. The order with which TMG and MMT were administered was randomized, so was the order of knee joint angles. For estimation of the reliability of the TMG and MMT measured variables, 10 of the subjects were re-tested after 2 days.

2.2. Participants

Sixteen healthy young individuals (14 males and 2 females; mean ± SD of age, stature and body mass were 23.4 ± 4.9 years, 1.76 ± 0.08 m and 74.2 ± 8.6 kg, respectively) volunteered to participate in this study and gave written informed consent. They were university students involved in different sports. Participants arrived at the laboratory, and were kept at a temperature of 20 °C (±2) in rested conditions. They were asked to refrain from strenuous physical exercise the previous 24 h, and from drinking coffee, tea and other potential diuretic beverages in the 3 h preceding the testing sessions. Only subjects who had not suffered a recent soft-tissue injury, exercise-induced muscle damage to the leg or did not reported other significant health issues were recruited for the study, which was approved by the local Ethics Committee in accordance with the Declaration of Helsinki of 1975.

2.3. Testing procedures

2.3.1. Maximal isometric voluntary contraction. Participants were secured in the prone position on a padded table of an isokinetic dynamometer (Biodex, System 3, Medical Systems, USA), firmly strapped at the hip. The thigh was aligned horizontally to the upper body, so that the hip was kept at an angle of 180°. The rotational axis of the dynamometer was aligned to the lateral femoral epicondyle of the subject. The lower leg was strapped 2 cm above the lateral malleolus to the machine lever arm, which was adjusted according to the length of the leg. Participants were familiarized with the task by performing a few sub-maximal practice contractions. At a verbal signal, subjects were required to exert a force against the dynamometer lever as strongly and as quickly as possible for 3–4 s, while strong verbal encouragement was provided. They performed three maximal voluntary contractions at each angle. Thirty seconds and 1 min of rest was allowed in between the three contractions and the three angles, respectively. The three knee joint angles were controlled through the isokinetic dynamometer's software and double-checked with a manual goniometer.

2.3.2. Electromyography. sEMG was recorded from the BF muscle whilst performing the MVC tasks. The skin was carefully prepared (shaved, abraded and cleaned with alcohol), and Ag/AgCl bipolar electrodes (Biopac, Santa Barbara, CA, USA) were placed midway between the ischial tuberosity and the lateral epicondyle of the tibia (SENIAM guidelines, Freriks *et al* 1999) with an interelectrode distance of 20 mm. The sEMG data were amplified with a gain of 1000 and automatically anti-aliased by the hardware (Biopac Systems) using a 1–500 Hz band-pass filter.

2.3.3. Tensiomyography. Response contraction time and amplitude throughout maximal passive twitch contractions were recorded, from the BF muscle, using TMG. With this

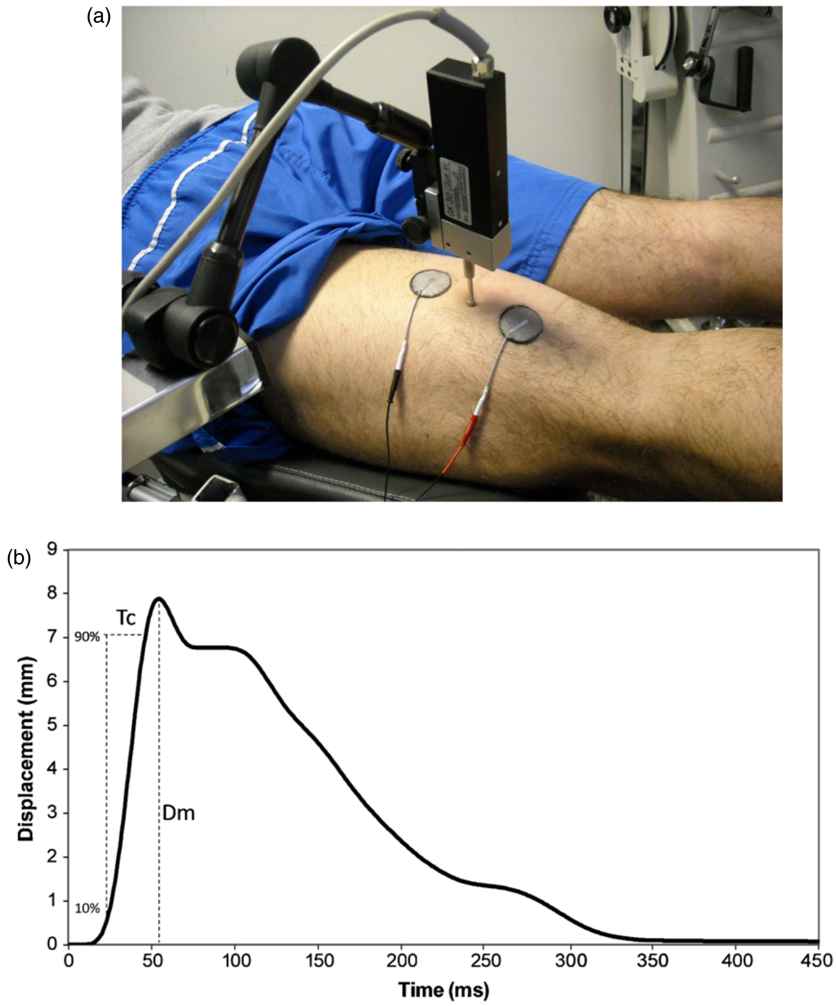

Figure 1. (a) Device and setup for TMG measurement. (b) Typical displacement/time signal recorded as a result of an electrical stimulation. Tc = contraction time (ms) expressed as the response from 10% to 90% of peak contraction; Dm = total displacement (mm) of the muscle.

technique, oscillations of the muscle belly are generated in response to a percutaneous electrical stimulation. These oscillations were recorded at the skin surface using a spring-loaded displacement sensor (Digital-optical comparator, RLS Ltd, Slovenia) positioned perpendicularly directly above the centre point of the BF to enable sensitive recording of mechanical displacement of the underlying muscle tissue. This displacement was recorded by a host online computer at a sampling rate of 500 Hz. Two stimulating adhesive electrodes (3 cm $\times$ 3 cm) were placed symmetrically 5 cm distally and proximally from the position of the sensor (figure 1(a)). Passive twitch contractions were stimulated at each of the three considered knee angles. A single square wave monophasic maximal 1 ms pulse was applied to elicit a twitch response of the muscle that was recorded by the displacement sensor. To obtain maximal mechanical response the stimulation was increased by 10 mA at a frequency of 10 s intervals to minimize the effects of fatigue and potentiation. The stimulation was increased gradually until no further displacement of the muscle belly could be produced. This maximal response was typically achieved between 40 and 70 mA. If required the measuring position was slightly adjusted to obtain the greatest response. The two main TMG derived measures were (1) contraction time (Tc, in ms) expressed as the response from 10% to 90% of peak contraction, and (2) the displacement (Dm, in mm) of the muscle from the total distance of peak amplitude (figure 1(b)). The positions of the sensor and the electrodes were marked with a semi-permanent marker pen in order to replicate the positioning for the subsequent day used for the reliability measures.

2.3.4. Myotonometry. Muscle stiffness of BF was measured using the Myometer (Myoton-3, Müomeetria AS, Tallinn, Estonia). The device incorporates a probe and an acceleration sensor. The probe was kept perpendicular with its end touching the muscle belly in the same position as the TMG sensor (figure 2(a)). A mechanical impact (with a duration of 15 ms, a force of 0.3–0.4 N and a local deformation in the order of a few millimetres) was delivered to the muscle and its damped natural oscillation recorded by means of the acceleration sensor (figure 2(b)), which samples at 3200 Hz, with an accuracy of ±5%. Stiffness (S, N m^{-1}) was calculated as a ratio between the force applied and the muscle deformation:

$$S = \frac{f}{\Delta l}$$

$$f = ma_1$$

where f is the force applied, m is the mass of the probe, a_1 is the positive peak of the damped acceleration (figure 2(b)), Δl is the muscle deformation, i.e. the negative muscle displacement calculated by double integrating the acceleration signal.

From the oscillation (figure 2(b)), frequency (F, Hz) and the logarithmic decrement of dampening (D) were also calculated as

$$F = \frac{l}{\Delta t}$$

and

$$D = ln\frac{a_1}{a_2}$$

respectively.

Five consecutive measurements were taken in each condition and the average value was used for later analysis.

2.4. Data recording and reduction

Torque and sEMG signals were synchronized, sampled at 2000 Hz, and stored on a PC using a 16 bit A/D converter data acquisition system (Biopac Systems, Inc. Goleta, CA, USA).

The torque signal was off-line low-pass filtered (fourth-order, Butterworth filter, cut-off frequency 15 Hz) by a digital fourth-order zero lag Butterworth filter. In each condition, the trial with the highest peak torque was considered for later analysis. A 250 ms duration epoch was used centred around peak torque and its average value was labelled maximal isometric torque (MIT).

The sEMG signal was off-line high-pass filtered (fourth-order, Butterworth filter, cut-off frequency 20 Hz), followed by the application of a moving root-mean-square filter with a time constant of 100 ms. The average sEMG value of the same 250 ms epoch centred around peak torque was used to calculate neuromuscular efficiency (NME) (see below).

(a)

(b)

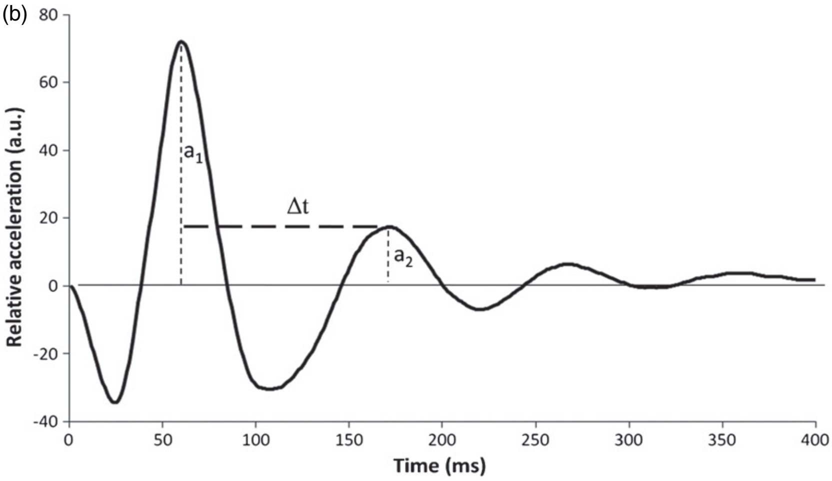

Figure 2. (a) Device and setup for MMT measurement. (b) Typical damped acceleration curve. a_1 = first acceleration peak; a_2 = second acceleration peak; Δt = time window between the first and the second acceleration peak.

2.4.1. Rate of torque development. The isometric rate of torque development (RTD) was calculated using the average slope of the filtered torque profile from 0 to 100 and 0 to 300 ms (RTD100 and RTD300). Based on the methods previously described by Aagaard *et al* (2002), the onset of muscle contraction was defined as the point at which the torque curve exceeded the baseline level by 3 Nm. Baseline torque was computed by taking the average reading over 0.5 s, starting 1 s before the onset of muscle contraction.

2.4.2. Neuromuscular efficiency. This was defined as the ratio between MIT and the average sEMG value of the same 250 ms epoch centred around peak torque and measured in Nm mV^{-1}. As previously explained (Rainoldi *et al* 2008) NME estimates the amount of torque produced per unit of EMG amplitude, with higher ratios being indicative of superior efficiency.

Table 1. Variables measured during isometric maximal voluntary contraction of BF at three different knee angles.

Angle (deg)	MIT (Nm)	RTD100 (Nm s^{-1})	RTD300 (Nm s^{-1})	sEMG (mV)	NME (Nm mV^{-1})
0	100.3 (20.9)[a,b]	571.7 (212.7)[b]	280.6 (59.0)[a,b]	0.690 (0.210)[b]	157.0 (53.2)
45	82.0 (16.8)[b]	482.9 (195.1)[b]	224.7 (49.3)[b]	0.637 (0.248)[b]	138.0 (38.2)
90	57.4 (15.0)	366.7 (154.4)	138.9 (39.4)	0.449 (0.212)	150.2 (64.1)

[a] Significantly different from 45° ($p < 0.01$);
[b] Significantly different from 90° ($p < 0.01$).
MIT = maximal isometric torque; RTD100 = rate of torque development measured during the first 100 or 300 ms of contraction; sEMG = surface electromyography; NME = neuromuscular efficiency.

Table 2. Tensiomyographic and myotonometric variables of BF measured in a relaxed state at three knee angles.

	Tensiomyography		Myotonometry		
Angle (deg)	Tc (ms)	Dm (mm)	Stiffness (N m^{-1})	Frequency (Hz)	Log decrement
0	26.20 (15.03)[a,b]	2.379 (1.314)[a]	328.3 (31.4)[a,b]	15.8 (2.0)[a,b]	1.95 (0.16)[a,b]
45	42.36 (16.05)	4.109 (1.991)[b]	300.3 (25.3)[b]	14.5 (1.3)	2.06 (0.16)[b]
90	43.92 (20.76)	2.281 (1.842)	280.4 (33.9)	14.0 (1.6)	2.13 (0.20)

[a] Significantly different from 45° ($p < 0.01$);
[b] Significantly different from 90° ($p < 0.01$).
Tc = contraction time; Dm = displacement.

2.5. Statistical analysis

Results are presented as mean and standard deviation (SD). A one-way ANOVA for repeated measures was used to compare the measured variables over the three knee angles (0°, 45° and 90°). When a significant effect was found, a Tukey *post hoc* test was performed to identify where differences occurred. A p value < 0.05 was regarded as significant.

Absolute and relative reliability of the TMG and MMT variables were analysed through the coefficient of variation (CV) and the intra-class correlation coefficient (ICC), respectively. The CV is defined as $(s/mean)\cdot 100$ where s is the SD and *mean* is the mean of the change scores of the measure (Atkinson and Nevill 1998). The ICC is defined as $(V\text{–}v)/V$, where V is the between-subject variance averaged over the two trials analysed, and v is the square of the standard error of measurement (Weir 2005). Statistical analysis was carried out using Statistica v. 9.1 (StatSoft, Bedford, England, UK).

3. Results

In table 1 the results from the maximal voluntary isometric contraction are shown. A significant difference in MIT ($p < 0.01$) and RTD300 is ($p < 0.01$) demonstrated across the three knee angles. sEMG and RTD100 show a significant difference between 0° and 90° ($p < 0.01$) and between 45° and 90° ($p < 0.01$). No significant differences were detected across the three knee angles for NME ($p = 0.20$).

Table 3. Reliability of tensiomyographic and myotonometric variables.

Angle (deg)	Tensiomyography (Dm) ICC (95% CI)	Tensiomyography (Dm) CV (95% CI)	Tensiomyography (Tc) ICC (95% CI)	Tensiomyography (Tc) CV (95% CI)	Myotonometry (Stiffness) ICC (95% CI)	Myotonometry (Stiffness) CV (95% CI)
0	0.82 (0.42–0.95)	19.8 (11.9–27.6)	0.82 (0.42–0.95)	16.5 (8.9–24.0)	0.73 (0.22–0.92)	3.6 (2.5–4.7)
45	0.57 (−0.05–0.87)	19.7 (11.2–28.1)	0.62 (0.03–0.89)	20.5 (10.7–30.3)	0.54 (−0.09–0.86)	3.7 (2.1–5.3)
90	−0.57 (−0.87–0.05)	43.1 (24.4–61.7)	−0.40 (17.0–49.7)	33.3 (−0.81–0.27)	0.71 (0.13–0.92)	4.9 (3.7–6.0)

Tc = contraction time; Dm = displacement; ICC = intraclass correlation coefficient; CV = coefficient of variation.

The results from TMG and MMT are documented in table 2. Tc measured at 0° is significantly different from 45° and 90° ($p < 0.01$) whereas no significant differences were documented between 45° and 90° ($p = 0.96$). Dm measured at 0° is significantly different from 45° ($p < 0.01$); nevertheless, Dm at 90° turned out to be non-significantly different from 0° ($p = 0.98$). S and D show a significant difference across the three knee angles ($p < 0.01$) whereas F measured at 0° is significantly different from 45° ($p < 0.01$) and 90° ($p < 0.01$).

Reliability results are summarized in table 3. Stiffness exhibited a moderate ICC (0.54–0.73) but an excellent CV (3.6–4.9%). Tc and Dm showed a moderate to good ICC for the 0° and 45° knee joint position, whereas the CV resulted quite high (16.5 to 20.5%). In particular, the 90° position yielded very poor reliability indexes.

4. Discussion

The aim of this investigation was to ascertain the sensitivity of TMG and MMT in detecting contractile and mechanical changes of BF muscle as a result of a change in its length (i.e. change in knee joint flexion). The main results of this study reveal that the parameters measured with MMT (S, F and D) are significantly altered following a variation in knee angle, whilst the variables measured with TMG (Tc and Dm) change sensibly between 0° and 45° but the device seems unable to detect meaningful changes at 90°. Furthermore, MMT presents a good level of inter-day reliability whereas TMG's is insufficient.

The tension–length relationship of the BF muscle during maximal contractions has already been determined in other studies and our results are consistent with previous findings showing that the MIT, unlike other muscular groups (e.g. quadriceps), is obtained when the BF is at its maximal length (Mohamed *et al* 2002). This has been explained as the best compromise between the knee flexors maximum moment arm length and the advantageous positioning of the actin and myosin filaments in the muscle fibre (Mohamed *et al* 2002). Other authors presented similar results even when the same muscle was lengthened by means of a variation in hip rather than knee angle (Lunnen *et al* 1981). Interestingly, a significant decrease in torque when the knee angle was reduced from 0° to 90° was mirrored by a change in sEMG activity but not NME. This could suggest that the dependence of maximum muscle force generation on muscle length is mostly determined by the mechanical properties of muscle (Doheny *et al* 2008). To the best of our knowledge this is the first time that RTD for BF muscle is presented showing that its maximal value is achieved when the knee is fully extended and tends to be reduced when the knee flexes. A change in muscle–tendon length, brought about by a change in joint angle, affects both the passive tension generated by the connective tissue (in parallel and in series) and the position of the contractile elements of the muscle. This, in turn, determines the level of tension that can be generated (Mohamed *et al* 2002). MMT succeeded in establishing a difference in muscle tension as a result of a change in muscle length, with S significantly different from the other values at each knee angle. The F results recorded in the present study were considerably higher (15.8 versus 13.2) than those reported by Hein and Vain (1998). A similar outcome was found for S with our results notably higher (328 versus 258) than those described by Bizzini and Mannion (2003) for the same muscle group at the same knee angle. These discrepancies could be somewhat expected considering that, unlike the two cited studies, the subjects of our investigation were involved in sport on a regular basis. In two previous studies carried out with the MMT technique, stiffness characteristics of different lower limb muscles have been significantly related to mobility of both hip and knee joints (Hein and Vain 1998) and the stiffness level of tongue muscles managed to discriminate between healthy subjects and obstructive sleep apnea syndrome patients (Veldi *et al* 2004).

Unlike MMT, variables measured with TMG (i.e. Tc and Dm) increased significantly when the knee joint was altered from 0° to 45°; however, their value at 90° did not change accordingly. This is probably due to insufficient tension in the muscle to obtain a valid and reliable recording as can be observed by the dramatic increase in CV (table 3).

In the present study, average Tc at 0° (26.2 ms) was similar to that reported by Dahmane *et al* (2005) (29.7 ms) and Pisot *et al* (2008) (31.0 ms), with only a slight difference in knee angle (5° versus 0°) in the latter study. On the other hand, Dm was sensibly lower in this study (2.4 mm) compared to Pisot *et al* (2008) (5.3 mm). This could be explained again by a different population type used in our study that was involved in sport on a regular basis unlike the mentioned study where the participants where described as being just 'healthy'. Furthermore, many of the participants in our study regularly participated in resistance exercises which have been shown to increase general muscle specific tension (Erskine *et al* 2010). Therefore, it is possible that greater muscle specific tension resulted in lower Dm recorded in this study. Of the limited studies available carried out with the TMG technique, Tc and Dm were significantly altered following exposure to microgravity (Pisot *et al* 2008) in some of the four lower limb muscles analysed, which included BF. Furthermore, TMG discriminated the endurance characteristics of rectus femoris (duration of muscle response to submaximal electrical stimulation) in post-polyomelitic compared to healthy subjects (Grabljevec *et al* 2005). It has been advocated that a measure of muscle stiffness is obtained with both MMT (Bizzini and Mannion 2003) and the Dm parameter measured with TMG (Pisot *et al* 2008). Even though the latter is the result of an electrical stimulation, our data showed minimal electrically stimulated torque production ($<$10% of MIT) when obtaining the TMG measure. This is also confirmed by experiments conducted in other laboratories (Maffiuletti 2010). The ability of TMG and especially MMT to detect an inherent change in stiffness, as demonstrated in this study, can be conceivably exploited in a number of clinical/therapeutic applications. Eccentric exercise has been demonstrated to increase passive muscle tension (Hoang *et al* 2007, Whitehead *et al* 2001); therefore, the devices can be plausibly used to monitor the recovery process following the damage induced by eccentric contraction. Likewise, since most neuromuscular disorders are described with an increase in muscle stiffness (Svantesson *et al* 2000, Alhusaini *et al* 2010), TMG and MMT can be employed to determine the level of muscle stiffness and to monitor the impact of different interventions aiming at reducing muscle stiffness. The latter applications are even more valuable since the devices are easily transportable and can be conveniently used in the field as opposed to in the laboratory.

The inter-day reliability was measured for both MMT and TMG: while for stiffness measured with MMT the index of relative reliability (i.e. ICC) is low to questionable, the absolute reliability (i.e. CV) is excellent, below 5% at the three angles analysed. In a previous study, looking at the inter-day reliability of MMT, BF exhibited a good ICC score; however, no CV was reported (Bizzini and Mannion 2003). It is however recognized that, whilst ICC reflects the consistency of individual ranks within a group and is affected by the level of heterogeneity of the sample, CV is unaffected by the range of measurements and allows us to compare reliability between different studies using different measurement tools (Atkinson and Nevill 1998). Tc and Dm measured with TMG showed an overall insufficient reliability. While ICC was acceptable at 0° and 45° of knee angle, CV was between 16% and 20% for both variables. Both reliability indexes were very poor at 90°. Although two recent studies found a good to excellent level of inter-rater (Tous-Fajardo *et al* 2010) and intra-session (Carrasco *et al* 2011) reliability for vastus medialis and rectus femoris, respectively, to the best of our knowledge this is the first time that inter-day reliability of measures taken with the TMG technique on the BF is assessed, and its low level could be due to the selection of muscle group. Smith and Hunter (2006) recorded TMG of the medial gastrocnemious on 21 healthy active males on four separate days and reported average CV of 8.7% and 17.5% for Tc and Dm, respectively. Hunter and coauthors (unpublished data) recorded TMG of the biceps brachii on 19 active males on seven separate days and reported average CV of 8.7% and 8.6% for Tc and Dm, respectively. The explanation for this inter-muscle group CV variation is unclear except for the fact that, based on the study of Arts *et al* (2010), the BF is likely to be thicker than the biceps brachii or medial gastrocnemious and this additional thickness may result in greater day-to-day variability due to altered electrical conducting properties of the muscle (Rutkove *et al* 2010) which may well affect TMG measurement. Finally, the limited number of subjects ($n = 10$) who have undergone the re-test measurement may have well affected the level of reliability.

BF passive muscle tension as measured by the isokinetic dynamometer has not been assessed and this is an acknowledged limitation of the present study. We deemed however that the positioning adopted for the TMG and MMT measurements would be unsuitable for a traditional measurement of passive tension because different factors (e.g. gravitational effect and level of tension in synergic and antagonist muscles) would affect the outcome. Yet, the concurrent measurement of passive muscle tension with a dynamometer, MMT and TMG is something to consider for future studies examining different musculatures.

5. Conclusions

An alteration in passive muscle length is mirrored by a change in its mechanical properties. The ability of MMT and—within the limitations outlined—TMG to detect a change in muscle tension at different angle joints seems to be very promising in all those clinical/therapeutic applications where an unnatural change in passive muscle stiffness occurs. Furthermore, these techniques have the advantage of employing portable devices that can be conveniently used in the field by sports medicine practitioners or conditioning coaches. The collection of the signal is automated with TMG, whereas the MMT device has to be operated manually and requires minimal training. Each measure is obtained in the order of a few seconds with both devices and this is very convenient in a clinical or field setting.

Eccentric exercise is notoriously followed by a rise in passive muscle tension which returns to the pre-exercise level in approximately one week (Hoang *et al* 2007). The devices can be conceivably used to monitor this mending phase, especially when external interventions are implemented to speed up the recovery process, as it happens in competitive sports. An additional important application in a clinical setting is related to the possibility of monitoring the effects of treatment on neuromuscular disorder (Svantesson *et al* 2000) and spasticity both in stroke (Sommerfeld *et al* 2004) and cerebral palsy (Ross and Engsberg 2002) patients.

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