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Biceps femoris fascicle length during passive stretching

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ABSTRACT

The purpose of this study was to quantify the relative changes in fascicle (FL) and muscle-tendon unit (*L*MTU) length of the long head of the biceps femoris (BFlh) at different combinations of hip and knee joint positions. Fourteen participants performed passive knee extension trials from 0°, 45° and 90° of hip flexion. FL, *L*MTU, pennation angle (PA) and effective FL (FL multiplied by the cosine of the PA) of the BFlh were quantified using ultrasonography (US). Three-way analysis of variance designs indicated that at each hip angle, FL and *L*MTU increased and PA decreased from 90° to 0° of knee flexion. Increasing hip flexion angle from 0° to 90° led to a higher FL and *L*MTU and a lower PA ($p < .05$). The average lengthening of the *L*MTU and effective FL was 28.00 ± 1.82% and 85.88 ± 21.92%, respectively. The average effective FL change accounted for 51.36 ± 7.39% of *L*MTU change. The relationship between effective FL and *L*MTU was almost linear with a slope equal to 0.49 ± 0.06 ($r^2 = 0.52$ to 0.97). To achieve greater lengthening of the fascicles of the BFlh, passive stretch with the hip flexed at least 45° and the knee reaching full extension is necessary.

1. Introduction

Stretching exercises aim to increase joint range of motion and to reduce passive resistance to stretch [Stojanovic and Ostojic, 2011]. Hamstring muscle injuries are commonly sustained in sports and have been attributed, among others, to excessive elongation of the biceps femoris long head (BFlh) [Thelen et al, 2005]. For this reason, stretching exercises are extensively used as a part of regular physical conditioning and rehabilitation programs of hamstring injuries [Heiderscheit et al., 2010; Umegaki et al., 2015]. Despite the numerous research studies on the effects of muscle stretching, evidence on the mechanisms responsible for the role of muscle-tendon unit (MTU) properties for stretching remains inconclusive [Abellaneda et al., 2009; Weppler and Magnusson, 2010].

Hamstring responses to stretch depend on several factors, including the way hamstrings operate when they are elongated. It has been reported that the shear modulus of the BFlh increases at least twice as the hip was flexed from 70 to 110° and 3 times as the knee was extended from 110° to 20° of knee flexion [Le Sant et al., 2015]. This indicates an increase of passive muscle stiffness as a response to lengthening. Nevertheless, in this study [Le Sant et al., 2015] the magnitude of MTU lengthening and its components was not examined. Previous studies have found that whole MTU length (*L*MTU) of the BFlh changes by 10.6% during passive knee extension (from 70° till 20° of knee flexion) from a hip flexion angle of 120–135° [Magnusson et al., 2000] and by 14.7% [Kumazaki et al., 2012] as the knee was extended from 90 to 0° from the prone position. Further, the BFlh fascicles exhibit a greater relative stretch compared with the remainder of the hamstrings [Kumazaki et al., 2012]. However, the changes in fascicle length (FL) as the whole MTU is elongated were not examined.

Research has examined the contribution of muscle and tendon to the passive mechanical properties of the MTU. Some studies in rabbit soleus, human tibialis anterior and human gastrocnemius muscle–tendon units [Herbert et al., 2002; Herbert and Crosbie, 1997; Hoang et al., 2007; Kwah et al., 2012] reported that the tendon contributes between half and three-quarters of the total compliance of the relaxed muscle–tendon units. Others, however, have found a higher contribution of the fascicles [Abellaneda et al., 2009] or an equal contribution of the tendon and the fascicles [Morse et al., 2008] to *L*MTU change. Research has shown an increase in passive semimembranosus *L*MTU and FL as the hip (with the knee fully extended) was flexed from approximately 20° to 80° [Diong et al., 2012]. Although the semimembranosus and BFlh display similar architecture, they have different proximal and distal tendon attachments and line of muscle path [Kellis et al., 2009] and therefore their mechanical properties may differ. A few studies have examined the FL of the BFlh [Bennett et al., 2014; Diong et al., 2012; Kellis et al., 2009; Timmins et al., 2016, 2015] but only one study measured FL at various joint positions [Chleboun et al., 2001]. Particularly, BFlh FL increased by approximately 1/3 as hip flexion angle changed from 0° to 90° as well as when knee flexion angle changed from 90° to 0° while the opposite was found for the pennation angle (PA) [Chleboun et al., 2001]. However, these data were collected from

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various static joint positions while simultaneous changes in whole *L*MTU and FL during passive stretching were not examined.

It has been extensively shown that that the BFlh displays the highest injury incidence amongst sprint-type injuries [Opar et al., 2012] while there is evidence that this particular muscle is extensively stretched during sprinting [Fiorentino et al., 2012]. The impact of stretching on BFlh morphology is currently unclear as some studies [E Lima et al., 2014] reported no changes in BFlh FL after 8 weeks of stretching while others have reported a significant decline in passive stiffness after acute stretching [Freitas and Mil-Homens, 2015]. Interestingly, however, the changes in *L*MTU, tendon and FLs as this particular muscle is stretched are unclear. Such information may be useful to understand the BFlh responses to stretching exercises which combine various hip and knee joint positions. Therefore, the purpose of this study was to assess the FL, pennation angle and MTU changes of the BFlh as the knee was extended from various hip positions using real-time ultrasound (US).

2. Methods

A total of 14 males (age 21.6 ± 0.4 years; mass 80.1 ± 2.8 kg; height 1.76 ± 0.02 m) volunteered to participate in this study after signing written informed consent. The participants were healthy and they had no injury of the lower limbs including history of hamstring strain or any other muscle or ligamentous injury of the knee. The participants were physically active but they did not engage in a specific sport or exercise program during the measurement period. The procedures conformed with the Declaration of Helsinki and were approved by the University ethics review committee.

The tests were performed on a Cybex (Humac Norm, CSMI, MA, USA) dynamometer with the subject in the prone position. A customized chair was placed on top of the dynamometer chair to allow passive knee joint motion from various hip joint positions, such that the axis of rotation of the dynamometer was aligned with the lateral femoral condyle. The hip joint angle was checked using a standard analog goniometer (Lafayette Gollehon, Model 01135, Lafayette Instrument Company, Lafayett, USA) at each initial testing position. A twin – axis goniometer (Model TSD 130B, Biopac Systems, Inc., Goleta, CA, USA) was used to record knee angular position (0° = full knee extension). The goniometry signal was fed through BNC connectors (Model CLB107, Biopac Systems) to a 12-bit analog-to-digital converter sampling at a rate of 1000 Hz per channel using the Acknowledge (version 3.9.1, Biopac Systems) software. Ultrasonic images from the BFlh were recorded using an ultrasonic apparatus (SSD-3500, ALOKA, Japan with an electronic linear array probe of 10 MHz wave frequency and a length of 6 cm. Dynamometer torque and knee joint angular position were simultaneously recorded at 1000 Hz. The video capturing module of the system software allowed simultaneous recording of the ultrasound video images at a rate of 30 Hz.

The electromyography (EMG) signal from the BFlh was collected using a pair of bipolar bar surface electrodes (inter-electrode distance 1 cm, TSD 150B, Biopac System Inc., Goleta, CA, USA) which were positioned on the BFlh muscle belly, proximally to the US probe. The skin was shaved and cleaned with alcohol wipes. The ground electrode was placed over a bony land mark on the lateral epicondyle. The EMG signal was amplified (gain × 1000) with an input impedance of 10 MΩ and a common rejection ratio of 130 dB. The signal was filtered using a band pass filter (low 15 Hz and high 450 Hz, full-wave rectified) and the root mean square (RMS) was displayed on line during the passive trial.

The protocol included passive knee joint movement at a very slow angular velocity ($2^\circ\,s^{-1}$) when the hip joint was positioned at 0°, 45° and 90° of hip flexion (Fig. 1). Passive movement of the knee started from 90° to 45° of knee flexion, a 2-s pause, and then passive movement from 45° to 0° of knee flexion (full extension). Five passive knee joint movement repetitions in each combination of hip and knee joint positions. The subject was asked to remain completely relaxed throughout all trials. The limb gravitational torque was recorded from a static position of 30° of knee flexion to allow correction of the recorded torque for gravitational effects. The corrected torque was stored for further analysis. Passive trials accompanied by high bursts of EMG activation (> 2 standard deviations above baseline) were rejected and the trial was repeated until minimum EMG signal was detected. Following the passive test, two maximum voluntary trials (MVC) were performed with the hip in neutral position (0°) and the knee flexed by 45°. The EMG produced during the isometric MVC was measured during a 2-s epoch at the torque plateau. EMG activities during stretching were expressed as a percentage of maximum EMG during the MVC.

The *L*MTU was measured statically prior to the passive joint motion test (Fig. 2). Particularly, the hip and knee joint were fixed statically in each combination of hip and knee joint positions. In each testing position, the BFlh attachments on the fibular head (distal end) and ischial tuberosity (proximal origin) were scanned using US and they were marked on the skin. The US probe moved slowly from the distal towards the proximal end of the muscle and the muscle-tendon unit path was marked on the skin. The curved path between the two ends was measured using a flexible tape and it was considered as the whole *L*MTU. The differences in measured *L*MTU between the 9 combinations of hip and knee joint movement was considered as the change in *L*MTU.

The FL was determined from US images obtained during the dynamic passive knee joint motion tests (Fig. 1). Particularly, starting from the distal origin, the probe was positioned approximately at 35% of *L*MTU and the position was marked on the skin. This location allowed visualization of the most distal fascicles and intermediate tendon of the BFlh. Further, it was selected because distal fascicles are shorter [Kellis et al., 2012] and therefore easier to measure using US. A customized cast was used to ensure similar position of the probe during the passive motion. Furthermore, an echo absorptive marker was placed in the US field of view such that any random movement of the probe is recorded. US video images during the passive knee joint movement were further analysed.

US images were displayed simultaneously with the knee flexion angle and torque measurements throughout the passive joint movement. Based on angular position data, 1.5-s US video footages (~40–50 images) were obtained when the knee was positioned in each of the three knee flexion angles. In each video footage, using a video-based software (Max Traq Lite version 2.09, Innovision Systems, Inc., Columbiaville, Mich. U.S.A) the ends of two fascicles were manually digitized in the distal region of the BFlh, starting at the fascicle's superficial origin and ending at the fascicle's insertion onto the intermediate (deeper) aponeurosis (Fig. 3). When the FL exceeded the field of view, a straight line was drawn along the visible trajectory of the fascicle until it intersected with the corresponding straight line of the intermediate aponeurosis. Subsequently the FL was directly measured as the distance between the visible end and the intersection between the extended FL and intermediate tendinous lines [Ando et al., 2014]. The angle between the line marking the outlined fascicle and the deep aponeurosis was then measured giving the PA. The transducer was not removed during the knee joint test, but it was removed when the subject assumed a new hip joint testing position. Therefore, it is likely that measured fascicles are not the same at different hip joint positions; nevertheless, they all originated from the similar (distal) region of the muscle. An average of the lengths of three muscle fascicles and the whole video footage was used. The US had experience with musculoskeletal US in the hamstrings area for 10 years.

In a pilot study, 5 participants were retested one week after the main testing session using the same operator. The intraclass correlation coefficient (ICC) for the LMTU ranged from .84 to .96 with a standard error of measurement ranging from 1.02 to 4.92 mm (0.31 to 1.18%). The ICCs for FL measurements ranged from .84 to .95 and the standard error of measurement range was from 0.65 to 1.98 mm (.98 to 2.32%).

To examine the contribution of muscle fascicles to muscle-tendon compliance, the absolute length of muscle fascicles was multiplied by