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Three-dimensional geometry of the human biceps femoris long head measured in vivo using magnetic resonance imaging

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ABSTRACT

Background: The human biceps femoris long head is susceptible to injury, especially when sprinting. The potential mechanical action of this muscle at a critical stage in the stride cycle was evaluated by calculating three-dimensional lines-of-action and moment arms about the hip and knee joints in vivo.
Methods: Axial magnetic resonance images of the right lower-limb (pelvis to proximal tibia) were recorded from four participants under two conditions: a reference pose, with the lower-limb in the anatomical position and the hamstrings relaxed; and a terminal swing pose, with the hip and knee joints flexed to mimic the lower-limb orientation during the terminal swing phase of sprinting and the hamstrings isometrically activated. Images were used to segment biceps femoris long head and the relevant bones. The musculotendon path and joint coordinate systems were defined from which lines-of-action and moment arms were computed.
Findings: Biceps femoris long head displayed hip extensor and adductor moment arms as well as knee flexor, abductor and external-rotator moment arms. Sagittal-plane moment arms were largest, whereas transverse-plane moment arms were smallest. Moment arms remained consistent in polarity across all participants and testing conditions, except in the transverse-plane about the hip. For the terminal swing pose compared to the reference pose, sagittal-plane moment arms for biceps femoris long head increased by 19.9% to 48.9% about the hip and 42.3% to 93.9% about the knee.
Interpretation: Biceps femoris long head has the potential to cause hip extension and adduction as well as knee flexion during the terminal swing phase of sprinting.

1. Introduction

Hamstring muscle strain-type injuries are common in many sports, including soccer (Ekstrand et al., 2011), rugby (Fuller et al., 2008) and Australian Rules football (Orchard and Seward, 2002). Such injuries tend to occur while sprinting at or close to maximal speeds (Askling et al., 2007; Brooks et al., 2006; Gabbe et al., 2006; Verrall et al., 2003; Woods et al., 2004). When sprinting, the hamstrings are highly activated during the stance and terminal swing phases of the stride cycle (Higashihara et al., 2010; Jonhagen et al., 1996; Kyröläinen et al., 1999; Mann et al., 1986; Mero and Komi, 1987). It has been proposed that injury risk is likely to be greatest during terminal swing when the hamstrings are contracting eccentrically (Chumanov et al., 2011, 2012; Schache et al., 2012). Numerous studies have shown that hamstring muscle strain-type injuries typically involve the biceps femoris long head (BF^{LH}) (Askling et al., 2007; Connell et al., 2004; Koulouris and Connell, 2003; Verrall et al., 2003). An understanding of the geometrical properties that determine the mechanical action of BF^{LH} about the hip and knee joints during the terminal swing phase of the stride cycle is therefore of interest.

For a given muscle spanning a joint, the magnitude of the joint torque generated by the muscle's force depends on the muscle's moment arm. Since the moment arm of a muscle is a function of its musculotendon path, the mechanical action of a muscle may be described by both the line-of-action and moment arm of the muscle. Many experimental studies have attempted to measure these geometrical properties for BF^{LH} at the hip and/or knee joints (Buford et al., 1997, 2001; Dostal et al., 1986; Duda et al., 1996; Herzog and Read, 1993; Kellis and Baltzopoulos, 1999; Nemeth and Ohlsen, 1985; Pohtilla, 1969; Spoor and van Leeuwen, 1992; Visser et al., 1990; Wretenberg et al., 1996). However, there are some notable limitations associated with studies conducted to date. First, results are often based on measurements obtained from elderly cadaveric specimens (Buford et al., 1997, 2001; Dostal et al., 1986; Duda et al., 1996; Herzog and Read, 1993; Pohtilla, 1969; Spoor and van Leeuwen, 1992), and thus may not necessarily be directly applicable to young living people. Second, some studies have only evaluated the mechanical action of BF^{LH} about a single joint (Buford et al., 1997, 2001; Dostal et al., 1986; Kellis and Baltzopoulos, 1999; Nemeth and Ohlsen,

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1985; Pohtilla, 1969; Spoor and van Leeuwen, 1992; Wretenberg et al., 1996), or just a single plane of joint motion (Buford et al., 1997, 2001; Herzog and Read, 1993; Kellis and Baltzopoulos, 1999; Nemeth and Ohlsen, 1985; Pohtilla, 1969; Spoor and van Leeuwen, 1992; Visser et al., 1990). To our knowledge, no study has measured the three-dimensional lines-of-action and moment arms for BF^{LH} about the hip and knee joints in vivo. Does BF^{LH} function solely as a hip extensor and knee flexor during the terminal swing phase of the stride cycle or does it have the potential to cause rotation about the hip and knee joints in the frontal and transverse planes also? This question cannot be answered based on available experimental data in the literature.

In the present study, magnetic resonance (MR) imaging was used to measure the three-dimensional lines-of-action and moment arms of BF^{LH} about the hip and knee joints in vivo. The specific aim was to quantify these geometric properties for a pose where the hip and knee joints were flexed to mimic the orientation of the lower-limb during the terminal swing phase of the stride cycle. The lines-of-action and moment arms of BF^{LH} about the hip and knee joints were also quantified for a reference pose (the anatomical position) for comparative purposes.

2. Methods

2.1. Participants

Four healthy adult male participants were recruited with an average age of 26.3 (SD 7.5) years, height 173.0 (SD 4.2) cm and body mass 62.8 (SD 8.7) kg. Participants provided written informed consent and prior approval was obtained from the Human Research Ethics Advisory Group at The University of Melbourne.

2.2. MR image acquisition

MR images were obtained using a Siemens 1.5 T Magnetom Espree system. Axial scans of the right lower-limb were performed from the iliac spines to the proximal tibia under two testing conditions: a reference pose and a terminal swing pose. For the reference pose, the participant lay in a supine position with their right hip and knee joints fully extended and their leg muscles relaxed. For the terminal swing pose, the participant lay in a supine position, but with their trunk inclined and their right leg elevated (via firm MR-compatible cushions) in order to flex the right hip and knee joints (Fig. 1). The objective was to simulate the orientation of the lower-limb at the point of peak musculotendon unit lengthening stretch for the biarticular hamstrings during the terminal swing phase of the stride cycle (Fig. 2), based upon previously reported data (Thelen et al., 2005). The resulting joint angles were measured using a standard hand-held goniometer and ranged from 45° to 50° hip flexion and 30° to 35° knee flexion for all participants. Images were recorded for the terminal swing pose with the hamstrings activated. Three submaximal isometric hamstring contractions were performed, with each contraction held for ~90 s duration. For each contraction, participants self-selected an activation intensity that they were capable of comfortably maintaining for the entire duration of the contraction (i.e., 90 s). Rest time of between 1 and 2 min duration was provided between repeated contractions. The hamstrings were imaged while being activated for the terminal swing pose because previous research has found that moment arm magnitudes can differ when a muscle is contracted compared to when it is relaxed (Maganaris et al., 1998, 1999).

Proton density sequences were acquired from each participant for both testing conditions. For the reference pose, two series were performed using the following parameters: TR = 1200 ms; TE = 39 ms; FA = 150°; FOV = 400 mm; slice thickness = 0.9 mm; slices per scan = 160; and base resolution = 384. For the terminal swing pose, three series were performed with the following parameters: TR = 2590 ms; TE = 32 ms; FA = 180°; FOV = 210 mm; slice thickness = 5.0 mm; slices per scan = 25; and base resolution = 256. The imaging parameters for the terminal swing pose ensured that the scan time was no longer than 90 s per series.

2.3. MR image processing

Meshes of the pelvis, right femur and right BF^{LH} (including muscle and proximal/distal tendon tissues) were developed from the MR images using a commercially available software package (AMIRA, Visage Imaging, Inc., San Diego, CA, USA). For each participant and for both testing conditions, meshes were produced by manually segmenting each image. An experienced musculoskeletal radiologist (GK) helped to distinguish muscle, tendon and bony anatomy on the MR images. The geometrical properties of the muculotendon unit and bones were then computed from the resulting meshes using MATLAB (MathWorks, Natick, MA, USA). Three-dimensional coordinate systems associated with the hip and knee joints (Fig. 3) were defined from the bony anatomy in a similar manner to previously described conventions (Cappozzo et al., 1995; Eckhoff et al., 2005; Fernandez et al., 2008) (see Appendix A for details).

2.4. Calculation of musculotendon lines-of-action

The muscle centroid of BF^{LH} was calculated for each axial slice. The centroid coordinates were then reconstructed to create the

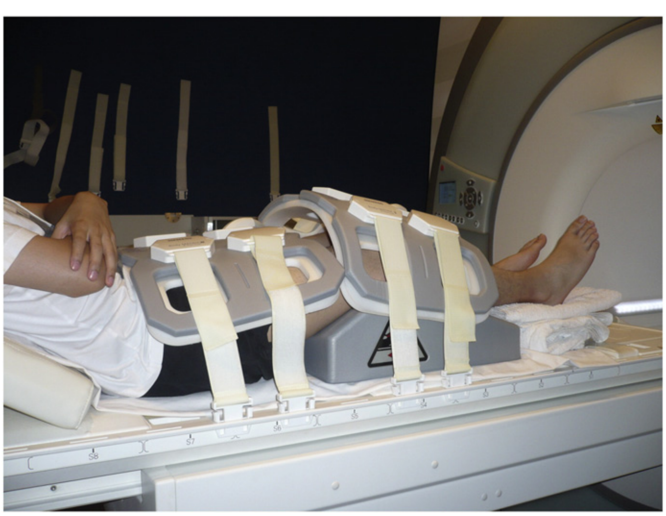

Fig. 1. The experimental set up for a participant in the MR scanner with their right lower-limb orientated in the terminal swing pose.

Fig. 2. The point of peak musculotendon unit lengthening stretch for the BF^{LH} during the terminal swing phase of the stride cycle..