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Prophylactic knee bracing alters lower-limb muscle forces during a double-leg drop landing

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ABSTRACT

Anterior cruciate ligament (ACL) injury can be a painful, debilitating and costly consequence of participating in sporting activities. Prophylactic knee bracing aims to reduce the number and severity of ACL injury, which commonly occurs during landing maneuvers and is more prevalent in female athletes, but a consensus on the effectiveness of prophylactic knee braces has not been established. The lower-limb muscles are believed to play an important role in stabilizing the knee joint. The purpose of this study was to investigate the changes in lower-limb muscle function with prophylactic knee bracing in male and female athletes during landing. Fifteen recreational athletes performed double-leg drop landing tasks from 0.30 m and 0.60 m with and without a prophylactic knee brace. Motion analysis data were used to create subject-specific musculoskeletal models in OpenSim. Static optimization was performed to calculate the lower-limb muscle forces. A linear mixed model determined that the hamstrings and vasti muscles produced significantly greater flexion and extension torques, respectively, and greater peak muscle forces with bracing. No differences in the timings of peak muscle forces were observed. These findings suggest that prophylactic knee bracing may help to provide stability to the knee joint by increasing the active stiffness of the hamstrings and vasti muscles later in the landing phase rather than by altering the timing of muscle forces. Further studies are necessary to quantify whether prophylactic knee bracing can reduce the load placed on the ACL during intense dynamic movements.

1. Introduction

Prophylactic knee braces are designed to prevent knee injuries during athletic activities, including anterior cruciate ligament (ACL) rupture, which results in painful, costly, and long-term effects. Non-contact ACL injuries commonly occur during high-risk maneuvers, such as rapid changing of direction or landing from a jump, and have rapidly increased over the past decade (Donnelly et al., 2012). Research has thus focused on knee injury prevention techniques, but not all programs aimed at reducing externally applied knee loads and avoiding non-contact injury by building strength and altering neuromuscular patterns are effective in reducing ACL injury rates (Noyes and Barber Westin, 2012). Prophylactic knee braces were introduced to prevent ACL and medial collateral ligament (MCL) injuries during athletic activities and can be purchased off-the-shelf. However, a conclusion has not yet been reached on the efficacy of prophylactic knee braces, with some studies even suggesting increases in the number of injuries sustained due to bracing (Grace et al., 1988; Sitler et al., 1990; Taft et al., 1985; Teitz et al., 1987).

Research has focused on the role of the lower-limb muscles in supporting the knee during dynamic movements and preventing ACL injury. Co-contraction of the hamstrings and quadriceps, which function as agonists and antagonists, respectively, to the ACL, is widely believed to provide stabilization of the knee joint (Baratta et al., 1988; Solomonow et al., 1987). The hamstrings provide the counterbalancing force against the anterior tibial translation induced by the quadriceps (Draganich and Vahey, 1990; Liu and Maitland, 2000; Yanagawa et al., 2002). In addition to the simultaneous and coordinated activation of the hamstrings and quadriceps, other lower-limb muscles, such as the triceps surae complex, activate during landing and may also play a role in stabilizing the knee (Elias et al., 2015; Fleming et al., 2001; Mokhtarzadeh et al., 2013; Morgan et al., 2014; Podraza and White, 2010). Delays in muscle activation (for instance, caused by

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fatigue) can lead to a lack of muscular protection and is proposed as an inciting event for ACL failure (Hashemi et al., 2011). ACL injury occurs shortly after initial contact, which may be too early for the muscles to alter ACL loading (Laughlin et al., 2011); however, maximum force generated by the muscles can help mitigate ACL forces throughout the landing phase (Kernozek and Ragan, 2008). Furthermore, neuromechanical imbalances partially explain why female athletes have a two to eight times greater incidence of ACL injury compared to their male counterparts (Agel et al., 2005; Arendt and Dick, 1995; Lindenfeld et al., 1994).

While motion studies have determined changes in landing kinematics and kinetics with prophylactic knee braces (Lin et al., 2008; Yu et al., 2004), few studies have investigated the neuromuscular mechanics of prophylactic knee bracing (Hangalur et al., 2015). Older studies have solely used surface electromyography (EMG) to find differences in muscle activity due to bracing during other dynamic activities such as running and side-step cutting (Branch et al., 1989; Osternig and Robertson, 1993). Furthermore, these studies have tested different types of knee braces, including a unilaterally-hinged knee brace (Osternig and Robertson, 1993), bilaterally-hinged rigid knee braces with and without a constraint to knee extension (Hangalur et al., 2015; Lin et al., 2008; Yu et al., 2004), and a functional knee brace (Branch et al., 1989), which is primarily designed for an ACL-deficient athlete. While the mechanisms of different knee braces and the types of braces used across studies vary, understanding the effects of prophylactic knee bracing, in particular, on muscle function can provide further insight into the efficacy of knee bracing in preventing injury.

Musculoskeletal modeling techniques have been developed to predict muscle force and activation during various tasks with the aim of better understanding the role of individual muscles (Erdemir et al., 2007). Motion analysis studies, coupled with computational methods, have proven to be a valuable tool in investigating the interaction between muscle and ground reaction forces in relation to ACL loading during landing (Kar and Quesada, 2012, 2013; Kernozek and Ragan, 2008; Laughlin et al., 2011; Mokhtarzadeh et al., 2013; Morgan et al., 2014; Pflum et al., 2004). The resulting muscle force temporal patterns are typically compared to EMG activation timing as an estimate of validity. The purpose of this study was to investigate the changes in lower-limb muscle function, in terms of the magnitude and timing of peak muscle forces, with prophylactic knee bracing in male and female athletes during landing from two heights. We hypothesized that the differences between the brace and no brace conditions would be the same at the higher height compared to the lower height, and also that the differences in bracing conditions would be the same in female athletes compared to their male counterparts. We further hypothesized that knee bracing would result in significant changes in the magnitude and timing of peak lower-limb muscle forces.

Table 1
Mean (± one standard error of mean) peak torque (N-m/kg) produced by the major muscle groups spanning each joint for no brace and brace conditions. Positive values indicate extension/plantarflexion, while negative values indicate flexion/dorsiflexion. Muscle symbols appearing in the table are: GMAX (gluteus maximus), ILPSO (iliacus and psoas major), HAMS (biceps femoris long head, biceps femoris short head, semimembranosus, and semitendinosus), RF (rectus femoris), VAS (vastus medialis, vastus intermedius, and vastus lateralis), GAS (medial and lateral compartments of gastrocnemius), SOL (soleus), and TA (tibialis anterior).

Muscle Group	No Brace	Brace	*P*-value
Hip			
GMAX	0.99 ± 0.07	1.12 ± 0.07	0.015*
ILPSO	-0.33 ± 0.03	-0.34 ± 0.03	0.674
HAMS	0.77 ± 0.07	0.97 ± 0.07	0.002*
RF	-0.74 ± 0.09	-0.86 ± 0.09	0.091
Knee			
HAMS	-0.45 ± 0.04	-0.57 ± 0.04	0.004*
RF	0.74 ± 0.08	0.84 ± 0.08	0.124
VAS	1.70 ± 0.06	1.85 ± 0.06	0.004*
GAS	-0.29 ± 0.02	-0.31 ± 0.02	0.275
Ankle			
GAS	0.62 ± 0.04	0.67 ± 0.04	0.124
SOL	1.52 ± 0.06	1.50 ± 0.06	0.675
TA	-0.014 ± 0.002	-0.014 ± 0.002	0.878

* Denotes significance for $p \leq 0.05$.

2. Methods

2.1. Participant recruitment and preparation

Fifteen participants (8 female, 7 male) who regularly participated in a landing sport (e.g. basketball, netball, soccer), with a mean[SD] age of 23.3[3.6] years, height of 171.7[5.9] cm, and mass of 68.4[9.1] kg, provided written informed consent to participate in this study. All testing procedures were approved by the relevant Human Research Ethics Committees (Melbourne University Ethics ID: 1034932). Participants were free from lower limb injury or disease. Prior to testing in the Gait Analysis Laboratory at Victoria University (Melbourne, Australia), each participant was fitted with commercial knee braces (K300 MX, POD Orthotic Pty Ltd., Torquay, Australia) on both legs. This knee brace consists of upper and lower rigid polymer frames connected by medial and lateral hinges that incorporate synthetic "ligaments." These ligaments are believed to possess similar properties to those of the native knee ligaments; however, the mechanism of this feature and brace overall is not fully understood. Proper fit was determined by measuring the width of the participant's knees using the calipers and instructions provided from the manufacturer such that the braces fit tightly but not uncomfortably. A total of 50 retro-reflective markers (0.014 m diameter) were mounted on the participant's trunk, thigh, shank, and feet using a custom marker set (Dorn et al., 2012; Schache et al., 2011). Calibration markers were affixed to the participant's medial and lateral femoral condyles and medial and lateral malleoli in order to define joint centers. The medial femoral and medial malleoli markers were removed for the no brace trials; the lateral femoral markers were further removed for the braced trials. Pairs of pre-gelled Ag/AgCl surface electromyography (EMG) electrodes were mounted on the skin over the muscle bellies of biceps femoris, semitendinosus, vastus medialis, and vastus lateralis.

2.2. Instrumentation

Three-dimensional (3D) kinematic data were acquired using a ten-camera motion analysis system (500 Hz; VICON, MX-T 40 S cameras, Oxford Metrics LTF, Oxford, UK), while ground reaction forces (GRFs) were simultaneously collected with two synchronized force plates (1000 Hz; AMTI, USA) embedded into the floor. Muscle EMG data was sampled using a telemetered system (1000 Hz, Noraxon Telemyo X, Noraxon USA Inc., Scottsdale, AZ, USA).

2.3. Drop landing protocol

A static trial with the participant in a normal standing position over the force plates was first captured. After a standardized warm-up, each participant was instructed to perform a double-leg landing by stepping off the platform with his or her dominant leg, defined as the preferred limb for kicking a ball, and landing barefoot onto the force plates using a natural landing strategy. Data were collected from heights of 0.30 m and 0.60 m, respectively. A trial was considered successful if the participant stepped off the platform without an upward or forward jumping motion and landed with a stable posture with his or her feet on the force plates. The double-leg landing tasks were repeated while the participant wore the knee brace. Up to five trials were collected at each height (0.30 m/0.60 m) and brace condition (brace/no brace) combination.

2.4. Data pre-processing

The raw kinematic marker, GRF, and muscle EMG data from each trial were extracted using a freely available Gait-Extract toolbox (Dorn, 2008). Markers were labeled in VICON Nexus (Version V1.7, VICON, Oxford Metrics LTF, Oxford, UK). Marker trajectories and GRF data were smoothed with 20 Hz fourth-order Butterworth filters (Bisseling and Hof, 2006). A Teager-Kaiser energy (TKE) filter was applied to the raw EMG signal to better determine the muscle onset and offset times (Li et al., 2007), which was used to quantitatively determine the relative overlap of predicted muscle activations and EMG activity (Akbarshahi et al., 2014). Raw EMG data were also high-pass filtered using a fourth order, zero-lag recursive Butterworth filter with a cut-off

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