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Original research

Associations between lower limb muscle activation strategies and resultant multi-planar knee kinetics during single leg landings

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ABSTRACT

Objectives: Anterior cruciate ligament injury prevention programs purportedly improve knee joint loading through beneficial modification of lower limb neuromuscular control strategies and joint biomechanics, but little is known about how these factors relate during single-legged landings. Thus, we examined the relationship between explicit lower limb muscular pre-activity patterns and knee joint biomechanics elicited during such landings.

Design: Randomized controlled trial.

Methods: Thirty-five female athletes had 3D knee joint biomechanics and lower limb EMG data recorded during a series of single-leg landings. Regression analysis assessed the relationship between pre-activity of vastus lateralis, lateral hamstring and rectus femoris with peak knee flexion angle and moment, and external anterior tibial shear force. Vastus lateralis, lateral hamstring and vastus lateralis:lateral hasmtring co-contraction assessed the relationship with knee abduction angle and moment.

Results: Greater pre-activity of rectus femoris predicted increased peak anterior tibial shear force ($R^2 = 0.235$, $b = 2.41$ and $P = 0.003$) and reduced knee flexion moment ($R^2 = 0.131$, $b = -0.591$, and $P = 0.032$), while greater lateral hamstring predicted decreased peak knee flexion angle ($R^2 = 0.113$, $b = 8.96$ and $P = 0.048$). No EMG pre-activity parameters were predictors ($P > 0.05$) for knee abduction angle and moment.

Conclusions: Current outcomes suggest reducing reliance on quadriceps activation may be beneficial during single-legged landings. It also, however, may be required for adequate joint stability during such maneuvers. Further research is needed to determine if inadequate hamstring activation, rather than elevated quadriceps activation, leads to hazardous loading during single-legged landings.

1. Introduction

Developing lower limb neuromechanical profiles that limit non-contact anterior cruciate ligament (ACL) injury is paramount, as the short and long-term disability of this traumatic injury has been well documented.[1] During a non-contact ACL injury episode, which account for up to 80 percent of all ACL injuries,[2] ligamentous rupture typically occurs during rapid deceleration followed by landing and/or pivoting where a person, him or herself, generates the forces and/or moments at the knee. This injurious ACL loading pattern is thought to stem from high-risk lower limb landing neuromechanics, or rather landing on a single-leg[2] while the knee is at or near full extension with excessive quadriceps and/or insufficient hamstrings activation that cannot prevent hazardous knee abduction or anterior shear loading.[3]

Recent experimental evidence suggests significant associations between neuromuscular control strategies and knee biomechanics.[4–6] Specifically, increased vastus lateralis pre-activity predicted increased knee abduction motion,[5] while reactive vastus medialis and medial hamstring co-contraction accounted for a significant portion of peak knee abduction moment.[6] Additionally, vastus lateralis re-activity has been identified as a significant predictor of an externally applied proximal anterior tibial shear force.[4] The initial experimental evidence suggests neuromuscular control strategies and knee joint biomechanics are linked. It is unclear, however, how quadriceps and hamstring pre-activity relate to explicit knee kinematics and kinetics during risky, single-legged landings.

Neuromuscular control maintains dynamic joint stability, with both pre- and re-activity patterns regulating muscle stiffness.[7] A non-contact ACL injury appears to occur from 17 to 50 ms after initial ground contact during landing,[2] suggesting there is insufficient time for mechanosensory feedback (i.e. reflex coordinated

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muscle activity) to protect against injury. This may indicate that muscle pre-activity might be necessary for dynamic joint stability. Thus, understanding the pre-activity neuromuscular parameters relating to high-risk knee joint biomechanical parameters may provide a platform to improve current ACL injury prevention program effectiveness. Since the current ACL injury prevention model is comprised of a lengthy series of training modalities that limit participant compliance,[8] the knowledge of which neuromuscular parameters to target explicitly may shorten these protocols. Shorter protocols may improve compliance, and coincide with greater effectiveness and a reduction in the injury rate. With that in mind, this study aimed to examine relations between explicit lower limb muscular pre-activity patterns and knee joint biomechanics elicited during a single-leg land and cut maneuver. We hypothesized that higher levels of vastus lateralis activation would be associated with a higher peak stance knee flexion moment, abduction angle and moment, and externally applied anterior tibial shear force, while higher levels of lateral hamstring activation would be associated with a lower peak stance knee flexion moment and anterior tibial shear force, as well as, higher knee flexion angle.

2. Methods

A power analysis of preliminary data indicated 26 participants were needed to achieve 80% statistical power. Thus, thirty-five female athletes (15.1 ± 1.2 years, 1.65 ± 0.05 m and 57.9 ± 9.3 kg) currently participating on athletic teams involved in high-risk activities (e.g. basketball, field hockey, and soccer) were recruited for participation. Participants were excluded if they had: (1) a history of previous knee injury or surgery, (2) pain in the lower extremity prior to testing or training, (3) any recent injury to the lower extremity (previous 6 months), and/or (4) currently pregnant. Before testing, all participants had leg dominance assessed and defined as the leg which they could kick a ball the furthest.[9] The University Institutional Ethics Committee granted research approval, and all participants provided written consent after being informed of the possible risks and benefits associated with the experimental procedures before participating. For participants under the age of 18, parents or legal guardians were also informed of the possible risks and benefits, and provided written consent before their daughters participated.

Participants had synchronous bilateral three dimensional (3D) lower limb (hip, knee and ankle) joint kinetic, kinematic, and surface electromyography data recorded during a series of single-leg jump landings. A force platform (AMTI OR6, Advanced Mechanical Technology Inc., Watertown, MA) embedded in the floor captured ground reaction force data, while eight high-speed (240 fps) cameras (MX-13, Vicon, Lake Forest, CA) recorded synchronous motion data during all landings. For each landing, the participants jumped from a distance equal to the length of their dominant limb from the front edge of the force platform, over a 17 cm box and performed a pre-defined (anticipated) landing.[10] Specifically, participants jumped forward, landed on the force platform on a single (dominant) limb and then aggressively jumped laterally to their non-dominant side. To complete the protocol, participants performed five successful landings.

For each landing, lower extremity joint rotations were quantified based on the 3D coordinates of thirty-one (14 mm diameter) reflective skin markers.[10] The markers were attached and secured to pre-determined anatomical landmarks by a single experimenter (TNB).

Following marker placement, a high-speed video recording of the athlete standing in a stationary (neutral) position was taken. A kinematic model comprised of seven skeletal segments (bilateral foot, shank and thigh segments and the pelvis) with 24 degrees of freedom was then defined using Visual 3D v3.99 software (C-Motion, Rockville, MD). The pelvis was defined with respect to the global (laboratory) coordinate system and assigned six (three translational and three rotational) degrees of freedom.[11] Hip,[12] knee,[13] and ankle[11] joint centers and associated orthogonal local segment coordinate systems were defined in accordance with previous literature and our own previous work.[11–13]

Synchronous 3D ground reaction force (GRF) data were collected at 1200 Hz during each landing and along with the 3D marker trajectories, low pass filtered with a fourth-order Butterworth filter at a cut-off frequency of 12 Hz.[10] The recorded 3D marker trajectories were subsequently processed by Visual 3D software to solve for the lower limb joint rotations at each time frame. Resultant hip and knee joint rotations were expressed relative to the athlete's static posture.[5,9] The filtered kinematic and GRF data were processed using conventional inverse dynamics analyses to obtain 3D intersegmental forces and moments at each lower limb joint. The segmental inertial properties were defined in accordance with the work of Dempster.[14] Hip and knee 3D intersegmental forces were transformed to respective distal segment reference frames (femoral and tibial) and anterior–posterior, medial–lateral and compression–distraction forces were calculated. The resultant intersegmental moments at the hip and knee were characterized as flexion–extension, abduction–adduction and internal–external rotation moments with respect to the cardanic axes of their respective joint coordinate systems.[9] All joint moments were normalized to participant body mass (kg) and height (m), while force values were normalized to body mass. Moments and forces were expressed externally for consistency with previous work in this area.[9,10] For example, an external abduction moment acted to abduct the knee joint. The kinematic and kinetic data were time-normalized to 100% of stance (heel strike to toe-off) and re-sampled at 1% increments ($N = 101$). Heel strike and toe-off defined as the instant GRF first fell below and exceeded 10 N.

During all landings, participants had lower extremity muscle activity quantified with surface electromyography (EMG) electrodes. Electromyographic data were recorded at 1200 Hz with a common-mode rejection rate of 92 dB and amplified using a 16-channel EMG system (Bagnoli, Delsys Inc., Boston, MA, gain 1000), which was synchronized with the force platforms via a motion capture system and stored by the data acquisition software (Nexus 1.6, Vicon, Lake Forest, CA) for later processing. Surface EMG electrodes (DE 2.1 single-differential, parallel-bar configuration, 99.9% Ag), with 10 mm inter-electrode distance were placed over the muscle bellies and in line with the muscle fibers of the lateral hamstrings (LH), vastus lateralis (VL), and rectus femoris (RF) muscles according to the guidelines of Delagi[15] to record muscle activity. Before attachment, the skin under each electrode site was lightly abraded and cleaned with alcohol. Prior to collecting the landing trials, EMG data was recorded during a two-second maximal voluntary isometric contraction (MVIC) for both the knee flexors and extensors. During all MVICs, participants were seated with their hip and knee angle approximately 90° and 45°, respectively, and were instructed to perform a maximal extension or flexion contraction into the resistance of the examiner.

Both the dynamic and MVIC EMG data were band-pass (10–500 Hz) filtered with a fourth order, zero lag Butterworth filter to attenuate movement artifacts and processed with a 50-millisecond root mean square (RMS) moving window. The dynamic EMG data was normalized to the MVIC activity of the respective muscle before calculating average RMS activity during the pre-activity (100 ms prior to ground contact) phase. Simultaneous RMS pre-activation of the vastus lateralis and lateral hamstrings (VL:LH) was used to calculate a muscle co-contraction ratio with a formula previously reported by Rudolph et al.[16]