Journal of Applied Biomechanics, (Ahead of Print)
https://doi.org/10.1123/jab.2019-0021

Human Kinetics
ORIGINAL RESEARCH

Muscular Coordination of Single-Leg Hop Landing in Uninjured and Anterior Cruciate Ligament-Reconstructed Individuals

Prasanna Sritharan,[1] Luke G. Perraton,[2] Mario A. Munoz,[3] Peter Pivonka,[4] and Adam L. Bryant[3]
[1]La Trobe University; [2]Monash University; [3]University of Melbourne; [4]Queensland University of Technology

This study compared lower-limb muscle function, defined as the contributions of muscles to center-of-mass support and braking, during a single-leg hopping task in anterior cruciate ligament-reconstructed (ACLR) individuals and uninjured controls. In total, 65 ACLR individuals and 32 controls underwent a standardized anticipated single-leg forward hop. Kinematics and ground reaction force data were input into musculoskeletal models to calculate muscle forces and to quantify muscle function by decomposing the vertical (support) and fore-aft (braking) ground reaction force components into contributions by individual lower-limb muscles. Four major muscles, the vasti, soleus, gluteus medius, and gluteus maximus, were primarily involved in support and braking in both ACLR and uninjured groups. However, although the ACLR group demonstrated lower peak forces for these muscles (all $P\text{s} < .001$, except gluteus maximus, $P = .767$), magnitude differences in these muscles' contributions to support and braking were not significant. ACLR individuals demonstrated higher erector spinae ($P = .012$) and hamstrings forces ($P = .085$) to maintain a straighter, stiffer landing posture with more forward lumbar flexion. This altered landing posture may have enabled the ACLR group to achieve similar muscle function to controls, despite muscle force deficits. Our findings may benefit rehabilitation and the development of interventions to enable faster and safer return to sport.

Keywords: muscle contributions, muscle function, musculoskeletal modeling, postural control, ACL injury

Anterior cruciate ligament (ACL) rupture is one of the most common and serious injuries occurring in competitive sports.[1] Despite advancements in reconstructive surgery and rehabilitative practices, only 65% of ACL-reconstructed (ACLR) individuals return to preinjury levels of competitive sport.[2] Increasing the rate of postsurgery participation in sport represents a challenge to rehabilitation research and may be garnered through improvements in clinical assessment.

The single-leg hop for distance is a well-studied and routinely employed task in the functional evaluation of ACLR individuals.[3] Landing from a single-leg hop demands a highly specific neuromuscular control strategy to arrest forward momentum of the body center of mass while preventing the stance limb from collapsing under the body's weight,[4,5] known as center-of-mass braking and support, respectively. During landing, individuals with ACLR have shown considerable biomechanical adaptations at the trunk, hip, knee, and ankle.[4,6–8] In particular, reduced peak knee flexion angle,[6] reduced peak knee extension torque,[7] and greater trunk flexion[8] compared with controls have been reported. Neuromuscular changes, such as quadriceps force deficits and activation failure, are common in individuals with ACLR.[9,10] Furthermore, adaptations have been observed, such as elevated levels of quadriceps–hamstrings muscle co-contractions to increase joint stiffness,[11,12] which may stabilize the knee and help prevent injury.[13,14]

Sritharan is with the La Trobe Sports & Exercise Medicine Research Centre, La Trobe University, Bundoora, VIC, Australia. Perraton is with the Department of Physiotherapy, Monash University, Clayton, VIC, Australia. Munoz is with the School of Mathematics & Statistics, University of Melbourne, Melbourne, VIC, Australia. Pivonka is with the School of Chemistry, Physics and Mechanical Engineering, Queensland University of Technology, Brisbane, VIC, Australia. Bryant is with the Centre for Health, Exercise and Sports Medicine, University of Melbourne, Melbourne, VIC, Australia. Sritharan (p.sritharan@latrobe.edu.au) is corresponding author.

Due to dynamic coupling, localized changes in kinematics and muscle forces can influence sagittal-plane control of the body center of mass[15] and therefore the successful performance of the landing task. As far as we are aware, quantification of muscular performance in arresting center of mass motion during landing in single-leg hopping tasks has not yet been undertaken in uninjured individuals; hence, the subsequent impact of ACLR-specific adaptations is unknown. As the ground reaction force (GRF) represents a close approximation to the resultant force acting on the center of mass in many tasks, from Newton's Second Law of Motion (*resultant force* = *mass* × *acceleration*), the GRF can be used to directly characterize the resultant center-of-mass accelerations. Thus, by decomposing the GRF into contributions by individual muscles, gravity, and inertia,[16,17] it is possible to quantify the muscular contributions to the spatiotemporal coordination of the center of mass—one definition of "muscle function[16]"—during dynamic activities. Understanding how muscle function differs between uninjured and ACLR individuals may help guide rehabilitation and the development of novel interventions to enable faster and safer return to sport.

Muscles are the largest contributors to the GRF,[16] and gait studies of healthy individuals have shown that each muscle of the lower limb contributes to center-of-mass modulation in a highly coordinated and predictable way.[16,18,19] In walking, the gluteus maximus, gluteus medius, vasti, soleus, and gastrocnemius are the key muscles responsible for vertical support[16] and forward progression[19] of the center of mass. Subsequent analyses of running[20] and stair ambulation[21] have shown that the pattern of muscle contributions to GRF, that is, timing, direction, and magnitude, is highly specific to the task undertaken, and that musculoskeletal pathologies can alter these coordination patterns.[22,23]

The aims of this study were 2-fold: (1) to quantify muscle function during landing in a standardized single-leg hopping task for uninjured individuals and therefore to identify which muscles were most important in modulation of the center of mass during this task, and (2) to compare muscle function in ACLR individuals undertaking the same hop-landing task with those of the uninjured controls. We hypothesized that (1) gluteus maximus, gluteus medius, vasti, soleus, and gastrocnemius would be the primary muscles for support and braking in hop landings, but that their coordination patterns would necessarily differ from those found for other tasks, and (2) ACLR individuals would demonstrate diminished contributions to support and braking from the vasti due to quadriceps force deficits, but elevated contributions from the gluteus maximus, gluteus medius, soleus, and gastrocnemius as compensation.

Methods

Participant Recruitment

Previously reported experimental data for 65 ACLR individuals and 32 uninjured controls were used for this study[24] with participant characteristics provided in Table 1. Ethical approval (ID: 1136167) for the study was provided by the Behavioural and Social Sciences Human Ethics subcommittee at the University of Melbourne, and all participants provided written informed consent. ACLR individuals had undergone single-bundle primary ACLR surgery performed by one of 2 experienced orthopedic surgeons using ipsilateral semitendinosus–gracilis tendon graft 12 to 24 months prior to testing. Eligible individuals were aged between 18 and 50 years at the time of surgery, participating in physical activity 1 to 3 times per week.

Table 1 Participant Characteristics for ACLR and Control Groups

Quantity	ACLR (n = 65)	Control (n = 32)	*P*
Age, y	28.2 (6.4)	25.2 (4.8)	*.022*
Sex,[a] females	24 (37%)	17 (53%)	.129
Height, m	1.76 (0.1)	1.70 (0.1)	*.010*
Body mass, kg	78.6 (14.7)	68.0 (11.4)	*.001*
Body mass index, kg/m^2	25.4 (3.3)	23.3 (2.7)	*.003*
Time between injury and ACLR,[b] mo	3 (4.2)	–	–
Time between ACLR and assessment,[b] mo	18 (3.0)	–	–
Level I/II sports participation at testing time[a]	45 (69%)	22 (72%)	.801
Limb dominance,[a] right leg	32 (49%)	19 (59%)	.336
Tegner Activity Scale, /10	6.0 (1.8)	5.8 (2.1)	.675
Cincinnati Knee Rating Scale,[b,c] %	86.5 (12.5)	100.0 (0.0)	*<.001*
Single-leg hop-for-distance score,[b,c] %	96.0 (12.1)	101.7 (6.5)	*<.001*
Time of peak knee flexion angle, s	0.16	0.15	.090
Distance hopped, m	0.88 (0.06)	0.85 (0.06)	*.047*

Abbreviation: ACLR, anterior cruciate ligament reconstruction.

Note: Unless indicated otherwise below, values are presented as mean (SD) with *P* values calculated using Student *t* tests at a significance level of $\alpha = .05$. Statistically significant group differences are presented in italics.

[a]Values presented as number of subjects (% of subjects), groups compared with chi-square tests. [b]Values presented as median (interquartile range). [c]Groups compared using Mann–Whitney *U* tests.

Control participants were physically active (1–3 times per week) and had no history of hip, knee, or ankle injury. Functional performance was quantified with a single-leg hop-for-distance test prior to undertaking the main hopping task. Physical activity level and patient-reported knee function were quantified with the Tegner Activity Scale[25] and the Cincinnati Knee Rating Scale.[26]

Experiments and Data Collection

Each participant completed 5 trials of an anticipated, submaximal effort, single-leg forward hop task.[27] For convenience of data collection and to reduce burden on participants, only the ACLR limb was assessed. Participants walked forward 3 steps at a cadence of 100 beats per minute in time with a metronome to standardize center-of-mass velocity prior to landing. The participant then performed a single forward hop, taking off, and landing with the same leg. ACLR individuals performed the hop on their affected leg while controls used their right leg only. The distance each participant was required to hop was standardized to be equal to the participant's leg length. The specific take-off and landing points for each participant were marked on the floor using colored tape. Participants were barefoot and folded their arms across their chest throughout the task. Landings were defined as successful if participants were able to land on their mark and maintain balance after landing without shuffling, moving their arms or touching the ground with their contralateral leg. The trial was deemed valid if balance was maintained for at least 2 seconds after foot strike. Previous testing in our laboratory found the kinematic and kinetic variables derived from this task to be highly reliable ($ICC_{3,3}$: .76–.96).[27]

For each participant, 40 retroreflective markers were placed on specific anatomical landmarks on the torso, pelvis, and both lower limbs using the protocol of Schache and Baker.[28] Prior to undertaking the single-leg hop trials, a static standing trial was first recorded to determine anatomical measurements for subsequent musculoskeletal modeling. The single-leg hop trials were then undertaken as described above. The spatial trajectories of the retroreflective markers were collected using a 12-camera Vicon motion analysis system (Oxford Metrics, Oxford, United Kingdom) at 120 Hz. The 3-dimensional GRF was recorded using a single force plate (AMTI, Watertown, MA). Kinematic and GRF data were filtered at 15 Hz using a fourth-order, zero-lag, recursive Butterworth filter. Electromyographic (EMG) data were collected at 2400 Hz using Noraxon and silver/silver-chloride surface electrodes (Myotronics, Kent, WA), placed over the bellies of the vastus lateralis, vastus medialis, rectus femoris, lateral and medial hamstrings, gluteus medius, and medial gastrocnemius of the ACLR individuals. EMG data were not collected for controls. To facilitate qualitative comparison with modeled muscle forces, for each trial, EMG data for each muscle were de-trended, de-spiked, full-wave rectified, and then normalized to the peak value of the waveform.

Musculoskeletal Modeling

A musculoskeletal modeling pipeline was implemented using OpenSim[29] and MATLAB (The MathWorks Inc, MA). For each subject, a generic 3-dimensional 23-degree-of-freedom 92-muscle whole-body model was scaled based on anthropometric measurements from the static trial.[29] The hips were modeled as ball joints, while knees, ankles, and subtalar joints were modeled as hinge joints. The head, arms, and torso were lumped into a single body that articulated with the pelvis via a ball joint. The metatarsophalangeal joints, modeled as hinge joints, were locked at their reference position.

For each trial, inverse kinematics was used to calculate joint angles by minimizing the sum of squares of the distances between virtual markers on the model and measured marker trajectories.[30] Joint torques were calculated using inverse dynamics. Static optimization was used to estimate muscle forces by minimizing the sum of squares of activations subject to constraints imposed by each muscle's force–length–velocity characteristics.[31]

Muscle function was quantified using a form of induced accelerations analysis. Briefly, the vertical and fore-aft components of the GRF were decomposed into contributions by muscles, gravity, and inertia using a pseudoinverse-based approach, for which a full mathematical treatment and validation is provided by Lin et al.[17] Briefly, to represent the distributed GRF on the foot, 5 foot-ground contact points were defined on the foot segment of the model. At each time step, each foot-ground contact point was assigned a weight indicating if they were fully constrained (1), unconstrained (0), or partially constrained (between 0 and 1), from the instantaneous phase of ground contact and the location of the GRF center of pressure.

Individual muscle forces, gravitational forces, and inertial forces were successively applied to the model in isolation, and the force induced at each foot-ground contact point as well as the induced joint accelerations were simultaneously calculated. Specifically, at each time step, for each muscle in the model (or gravitational forces, or inertial forces), an overdetermined system of equations was constructed,[17] with general form:

$$\tilde{\mathbf{W}}\begin{bmatrix} \mathbf{M} & -\mathbf{E} \\ \mathbf{W}\cdot\mathbf{N} & 0 \\ 0 & \mathbf{I} \end{bmatrix}\begin{Bmatrix} \ddot{\mathbf{q}}^{\alpha} \\ \mathbf{f}^{\alpha} \end{Bmatrix} = \tilde{\mathbf{W}}\begin{Bmatrix} \mathbf{F}^{\alpha} \\ \mathbf{W}\cdot\mathbf{Z} \\ 0 \end{Bmatrix} \tag{1}$$

where α is the current muscle (or gravitational forces, or inertial forces) of interest, $\tilde{\mathbf{W}}$ is a matrix of constraint weights, $\mathbf{W}$ is a matrix of weights between 0 and 1 for each foot-ground contact points as described above, $\mathbf{N}$ and $\mathbf{Z}$ are matrices of coefficients associated with the accelerations of the foot-ground contact points, $\mathbf{M}$ is the system mass matrix, $\mathbf{E}$ is a matrix of partial velocities, $\mathbf{F}^{\alpha}$ is the applied muscle force (or gravitational forces, or inertial forces), $\ddot{\mathbf{q}}^{\alpha}$ and $\mathbf{f}^{\alpha}$ are, respectively, the vectors of unknown generalized accelerations and GRFs at each foot-ground contact point induced by α. Rewriting Equation 1 more succinctly:

$$\mathbf{A}\begin{Bmatrix} \ddot{\mathbf{q}}^{\alpha} \\ \mathbf{f}^{\alpha} \end{Bmatrix} = \mathbf{b}. \tag{2}$$

The Moore–Penrose pseudoinverse was then used to find a least-squares approximation for the unknown-induced accelerations and forces on the foot-ground contact points from the overdetermined system of equations given by Equation 2:

$$\begin{Bmatrix} \ddot{\mathbf{q}}^{\alpha} \\ \mathbf{f}^{\alpha} \end{Bmatrix} = \mathbf{A}^{+}\mathbf{b}, \tag{3}$$

where A^{+} is the Moore–Penrose pseudoinverse of $\mathbf{A}$.

Data Analysis

The vertical and fore-aft components of the GRF were defined as *support* and *braking,* respectively. For each valid trial, temporal data were trimmed to 2 seconds, the minimum time allowed to successfully complete a landing, and defined as the *total time to stabilization*. The *landing phase* was defined as the time period from foot strike to peak knee flexion angle. All subsequent time until 2 seconds elapsed was defined as the *stabilization phase*. Quadriceps–hamstrings co-contraction levels were quantified by calculating the co-contraction index in both groups using a modified form of the equation provided by Rudolph et al[32]:

$$\mathrm{CCI}_i = \frac{\text{lower } a_i}{\text{higher } a_i} \times (\text{lower } a_i + \text{higher } a_i), \tag{4}$$

where $\mathrm{CCI_i}$ is the quadriceps–hamstrings co-contraction index at time step *I*, lower a_i and higher a_i are the instantaneous activation levels (between 0 and 1) of the less-activated muscle and more-activated muscle, respectively. We calculated (1) the

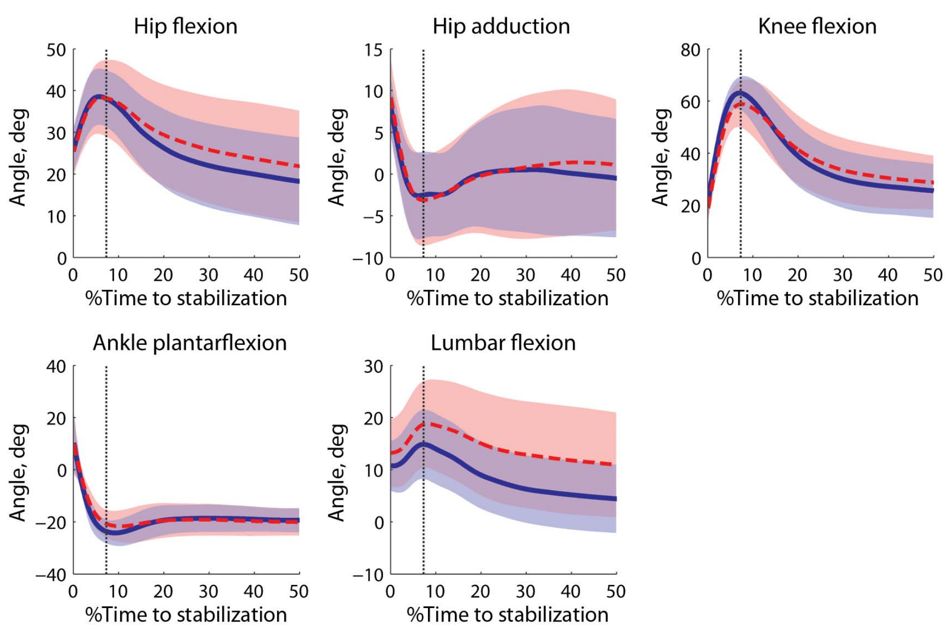

Figure 1 — Mean and SD for joint angles calculated for the ACLR (dashed line) and uninjured control (solid line) groups. The vertical dashed line represents the approximate end of landing phase for both groups. Positive values represent hip flexion, hip adduction, knee flexion, ankle plantar flexion, and lumbar flexion, respectively. ACLR indicates anterior cruciate ligament reconstruction.

co-contraction index at the time instant of peak vertical GRF, (2) the average co-contraction index during the landing phase, (3) the average co-contraction index during the stabilization phase, and (4) the average co-contraction index for the entire landing task.

To compare muscle function between the groups, muscle contributions to the GRF were normalized to body weights (BWs). At the time instant of peak GRF, we also calculated the *percentage* of contribution by each muscle to the total peak GRF. We undertook this for each GRF component—fore-aft and vertical—separately.

Means and SDs were calculated for participant characteristics, joint angles, joint torques, muscle forces, and individual contributions to the GRF. Male–female ratio, sports participation level, and limb dominance were compared with chi-square tests. Cincinnati Knee Rating Scale and hop-for-distance scores, which were positively skewed, were compared with Mann–Whitney U tests. Differences between the control and ACLR groups for participant characteristics and the peak values of kinematics, joint torques, muscle forces, co-contraction index, and normalized contributions to GRF were compared using Student t tests. Significance levels were set at $\alpha = .05$ for all analyses.

Results

Anterior cruciate ligament-reconstructed individuals landed with lower peak ankle dorsiflexion and peak knee flexion angle than did uninjured controls, but their knees and hips were more flexed through stabilization (Figure 1). ACLR individuals landed with more forward lumbar flexion through both phases. Statistically significant differences occurred for peak ankle dorsiflexion, knee flexion, and lumbar flexion angles only (Table 2).

Statistically significant differences occurred at the peaks of all reported joint torques (Table 2). The magnitudes of peak hip flexion and lumbar flexion torques were higher in ACLR individuals throughout both phases, while all other reported peak torques were lower in the ACLR group. Time histories of net joint torques are provided as Supplementary Figure S1.

The pattern of ACLR muscle forces agreed well with the pattern of EMG (Figure 2). Peak soleus, vasti, rectus femoris, and gluteus medius muscle forces were significantly lower in the ACLR group (Table 2). Peak gluteus maximus forces were slightly lower in the ACLR group but not significantly different. Nevertheless, gluteus maximus forces were higher in the ACLR group through the stabilization phase (Figure 2). Hamstrings and erector spinae muscle forces were higher in the ACLR group throughout both phases, but only peak erector spinae forces were significantly different (Table 2).

The ACLR group demonstrated significantly higher co-contraction index at the time instant of peak vertical GRF, as well as significantly higher time-averaged co-contraction index for the landing phase, stabilization phase, and for the entire landing task, compared with controls (Table 2).

The peak measured vertical and fore-aft GRF components were not different between the control and ACLR groups when normalized to BW (Table 3). In both groups, vasti, soleus, gluteus maximus, and gluteus medius contributed positively to support, while only vasti and soleus provided substantial braking force through both landing and stabilization phases (Figure 3). Hamstrings and gastrocnemius contributions were very small, while rectus femoris and erector spinae muscles contributed almost nothing to both components of the GRF (<0.01 BW). Gravity and inertial force contributions to the GRF were also virtually zero throughout landing and are not considered further.

Table 2 Comparison of Mean Peak Values for Joint Angles, Joint Torques, Muscle Forces, and Quadriceps–Hamstrings Co-Contraction Index

Quantity	Control	ACLR	P
Joint angles, deg			
Hip flexion	38 (7)	38 (9)	.733
Hip adduction	−3 (5)	−3 (5)	.317
Knee flexion	63 (6)	58 (9)	*<.001*
Ankle plantar flexion	−24 (4)	−21 (6)	*<.001*
Lumbar flexion	15 (7)	19 (8)	*<.001*
Joint torques, %BW*HT			
Hip flexion	−4.2 (2.0)	−5.4 (2.1)	*<.001*
Hip adduction	9.3 (1.8)	7.9 (2.0)	*<.001*
Knee flexion	−18.5 (2.8)	−16.2 (3.3)	*<.001*
Ankle plantar flexion	10.2 (3.2)	9.1 (2.8)	*<.001*
Lumbar flexion	−5.0 (2.6)	−6.5 (3.0)	*<.001*
Muscle forces, BW			
Soleus	4.03 (1.19)	3.62 (1.04)	*<.001*
Gastrocnemius	0.55 (0.34)	0.49 (0.27)	.059
Vasti	5.49 (1.10)	5.10 (1.13)	*<.001*
Rectus femoris	0.95 (0.74)	0.48 (0.48)	*<.001*
Hamstrings	0.65 (0.48)	0.73 (0.49)	.085
Gluteus maximus	1.40 (0.53)	1.38 (0.56)	.767
Gluteus medius	3.50 (0.80)	2.91 (0.68)	*<.001*
Erector spinae	1.32 (0.88)	1.54 (0.93)	*.012*
Co-contraction index			
At time of peak vertical GRF	0.28 (0.19)	0.37 (0.24)	*<.001*
Average for landing phase	0.19 (0.07)	0.22 (0.08)	*.001*
Average for stabilization phase	0.07 (0.04)	0.12 (0.10)	*<.001*
Average for entire task	0.08 (0.04)	0.13 (0.09)	*<.001*

Abbreviations: %BW*HT, percentage of body weight multiplied by subject height; ACLR, anterior cruciate ligament reconstruction; BW, body weight; GRF, ground reaction force. Note: Values presented as mean (SD). P values calculated using Student t tests at a significance level of $\alpha = .05$. Statistically significant group differences are presented in italics. Co-contraction index is the quadriceps–hamstrings co-contraction index as defined by Equation 4 in the "Methods" section, which is itself based on that described by Rudolph et al.[32]

Group differences in the peak magnitude of muscle contributions to both support and braking were very small when normalized to BW but some differences reached statistical significance (Table 3). Specifically, the vasti, gluteus maximus, and hamstrings contributions were significantly higher in the ACLR group for both support and braking, but the magnitudes of differences were not material. Similarly, between-group differences for the percentage of contribution by individual muscles to both the peak fore-aft and peak vertical GRF components were very small, although some differences did reach statistical significance (Table 3).

Discussion

The objectives of this study were (1) to quantify muscle function during landing in a standardized single-leg hopping task for uninjured individuals and therefore to identify which muscles were most important in modulation of the center of mass during this task and (2) to compare muscle function in ACLR individuals

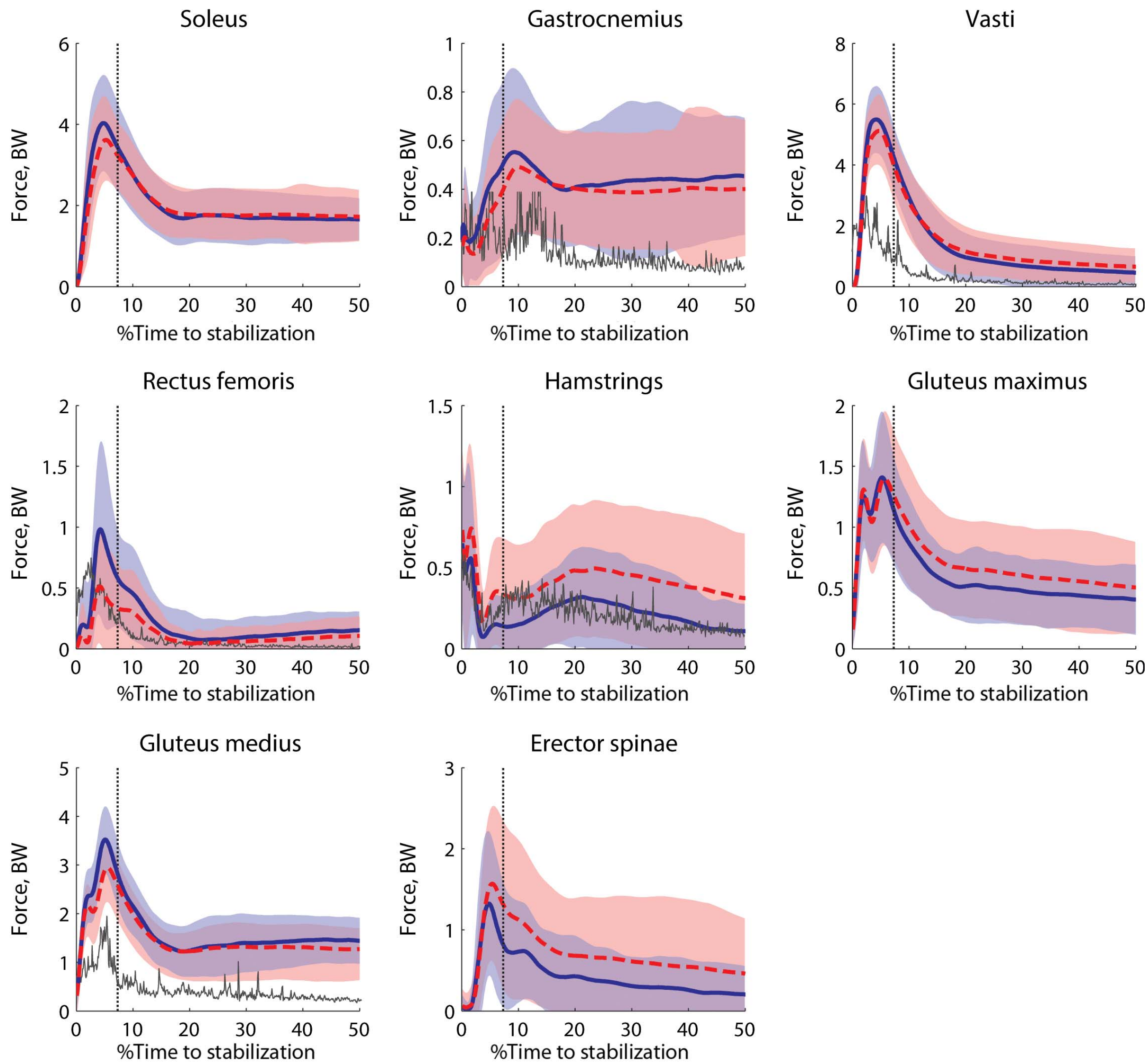

Figure 2 — Mean and SD for the major lower-limb muscle forces calculated using static optimization for the ACLR (dashed line) and uninjured control (solid line) groups. Mean peak-normalized EMG profiles for the ACLR group (thin gray) are presented for qualitative comparison with muscle forces. EMG was not taken for controls. The vertical dashed line represents the approximate end of landing phase for both groups. ACLR indicates anterior cruciate ligament reconstruction; BW, body weight; EMG, electromyography.

undertaking the same hop-landing task with those of the uninjured controls. Our hypotheses were both disproved: (1) only 4 of the hypothesized 5 key muscles—vasti, soleus, gluteus medius, and gluteus maximus—were important in center-of-mass modulation, with the gastrocnemius contributing little; and (2) differences in muscle function between ACLR and control groups were not significant, despite some differences reaching statistical significance.

In both control and ACLR groups, we found that 4 major uniarticular muscles were principally involved in controlling the center of mass during both landing and stabilization phases: vasti, soleus, gluteus medius, and gluteus maximus. Of these, the vasti was the largest contributor to the sagittal-plane GRF throughout both the landing and stabilization phases (about 38% of peak vertical GRF and 50% of peak fore-aft GRF in both uninjured and ACLR groups), followed by the soleus (about 25% of peak vertical GRF and 32% of peak fore-aft GRF in both groups). During landing, these 4 uniarticular muscles of the lower limb acted eccentrically across the joints they span to decelerate the body in the sagittal plane during the landing phase,[33] before acting concentrically to raise the center of mass during the stabilization phase. In both phases, all 4 muscles were essential for support, while in braking, only the vasti and soleus contributions were notable. Modeling studies of walking,[16,19] running,[20,34] and stair ambulation[21] have shown that the same 4 uniarticular muscles identified in the present study—the vasti, soleus, gluteus medius, and gluteus maximus—are also major modulators of support and fore-aft progression in these tasks; however, their relative contributions differ from hop landing. For example, in running, although the vasti provides the most support throughout stance, all 4 muscles play a roughly equal role in fore-aft progression, with gluteus medius and gluteus maximus more dominant in early stance, while the soleus dominates in late stance.[20,34] Thus our findings reinforce the task dependency of muscle function during dynamic activities.

The contributions to support and braking by the major biarticular muscles of the lower limb—the gastrocnemius, hamstrings, and rectus femoris—were of negligible magnitude in both groups.

Table 3 Contributions of Muscles at the Time Instants Corresponding to Peak Values of Support (Vertical GRF) and Braking (Fore-Aft GRF) for ACLR and Control Groups

	Support (vertical GRF)			Braking (fore-aft GRF)		
Quantity	**Control**	**ACLR**	***P***	**Control**	**ACLR**	***P***
Contribution to peak GRF, BW						
Total GRF	2.40 (0.45)	2.41 (0.36)	.760	0.77 (0.17)	0.75 (0.13)	.472
Gastrocnemius	0.08 (0.06)	0.09 (0.06)	.081	0.02 (0.02)	−0.02 (0.02)	.825
Vasti	0.91 (0.34)	0.97 (0.27)	*.042*	0.39 (0.10)	0.41 (0.11)	*.033*
Hamstrings	0.03 (0.04)	0.05 (0.05)	*.003*	−0.08 (0.04)	−0.09 (0.04)	*.011*
Gluteus maximus	0.31 (0.18)	0.34 (0.17)	*.034*	0.03 (0.03)	0.04 (0.03)	*<.001*
Gluteus medius	0.23 (0.12)	0.25 (0.13)	.147	−0.03 (0.03)	−0.03 (0.03)	.758
Soleus	0.62 (0.40)	0.59 (0.40)	.471	0.25 (0.10)	0.25 (0.10)	.962
Proportion of peak GRF (% of total GRF)						
Total GRF	100	100	–	100	100	–
Gastrocnemius	3 (3)	4 (3)	.087	2 (2)	2 (2)	.939
Vasti	38 (14)	40 (11)	.072	52 (13)	56 (14)	*.007*
Hamstrings	1 (2)	2 (2)	*.004*	−11 (6)	−12 (5)	*.003*
Gluteus maximus	13 (8)	14 (7)	.052	4 (4)	6 (5)	*<.001*
Gluteus medius	10 (5)	10 (5)	.208	−7 (4)	−6 (4)	*.022*
Soleus	26 (17)	25 (17)	.460	32 (13)	33 (14)	.726

Abbreviations: ACLR, anterior cruciate ligament reconstruction; BW, body weight; GRF, ground reaction force. Note: Values are presented as mean (SD). Contributions to peak GRF are reported in units of BW. Proportion of peak GRF is reported as percentage of total peak GRF. *P* values calculated using Student *t* tests at a significance level of $\alpha = .05$. Statistically significant group differences are presented in italics. Positive values for support and braking represent forces directed upward and backward, respectively. Rectus femoris and erector spinae contributions were almost zero throughout.

In particular, the negligible contributions by the gastrocnemius are notable, as this muscle plays a prominent role in support and progression during walking,[16,19] running,[20,34] and stair ambulation,[21] where it acts together with the soleus to generate large propulsive forces across the ankle. Our finding of a lesser role of the gastrocnemius in hop landing is supported by an EMG study of drop landing that found diminished medial gastrocnemius activity immediately after impact.[35] That study found that the gastrocnemius was in fact the most strongly activated lower-limb muscle *prior* to foot strike,[35] most likely to increase ankle-joint stiffness in preparation for landing.[36] However, this large preactivation diminished rapidly after impact, with soleus subsequently becoming more highly activated.[35] Thus the gastrocnemius may be a more important muscle in preparing for and maintaining a comfortable and protective landing posture, than support and braking of the center of mass during landing tasks.

Overall, between-group differences in sagittal-plane muscle function were not material (Figure 3 and Table 3) despite considerable deficits in the peak muscle forces generated by the vasti, soleus, gluteus medius, and gluteus maximus for the ACLR group (Figure 2 and Table 2). All muscles contribute to the GRF through dynamic coupling[16] and the configuration of the lower limb is a principal factor that determines how a torque produced by a muscle acting across a joint induces accelerations elsewhere in the kinematic chain.[15] Thus, it is possible that the straighter and stiffer lower-limb postural adaptation shown by ACLR individuals (Figure 1)—primarily an adaptation to quadriceps weakness[37] and a protective mechanism for the knee[14]—may have a secondary benefit in achieving similar center-of-mass modulation with reduced muscle forces by allowing better transmission of muscle-induced joint accelerations through the lower limb to the ground. Similar kinematic adaptations that improve lower-limb muscle function were previously described for gait in children with cerebral palsy[22] and adults with tibiofemoral osteoarthritis.[23]

Our findings suggest that the elevated hamstrings and erector spinae muscle forces in the ACLR group were more likely to be associated with altered postural demands, than with support and braking. The hamstrings and erector spinae muscle forces were higher in the ACLR group than uninjured controls throughout both landing and stabilization phases (Figure 2 and Table 2, significantly different for the erector spinae only). Yet, the hamstrings and erector spinae contributed very little to the sagittal-plane GRF in both groups, therefore they played a minor role in overall center-of-mass modulation. In our models, the larger erector spinae muscle force was likely required to maintain greater forward lumbar flexion in the ACLR group (Figure 2). Any associated induced forward pelvic tilt could be constrained by activating the hip extensors. Preferentially activating the hamstrings for this purpose could, hypothetically, enable the gluteus maximus and gluteus medius to be more usefully engaged in braking and support. In this way, the elevated quadriceps–hamstrings co-contractions observed in ACLR individuals in our present work and previous studies,[11] understood to be a protective mechanism for the compromised knee,[5,14] may also be partially attributed to the altered postural demands. Greater lumbar flexion angle throughout the landing task (Figure 1) is an important adaptation in ACLR individuals, being the primary postural mechanism by which lower-limb torque demand is shifted away from the compromised knee.[38] This also ensures a lower center-of-mass height above ground to more reliably achieve successful landings, by compensating for a straighter lower limb which itself would tend to push the center-of-mass upward.[37]

Our findings have important clinical implications. A straighter and stiffer landing posture likely comes at increased risk of reinjury. Prilutsky and Zatsiorsky[33] found that muscular coordination of all the lower-limb muscles is important in dissipating impact energy. During landing, the biarticular muscles transfer mechanical energy from distal to proximal limbs, where it is dissipated through eccentric action of the

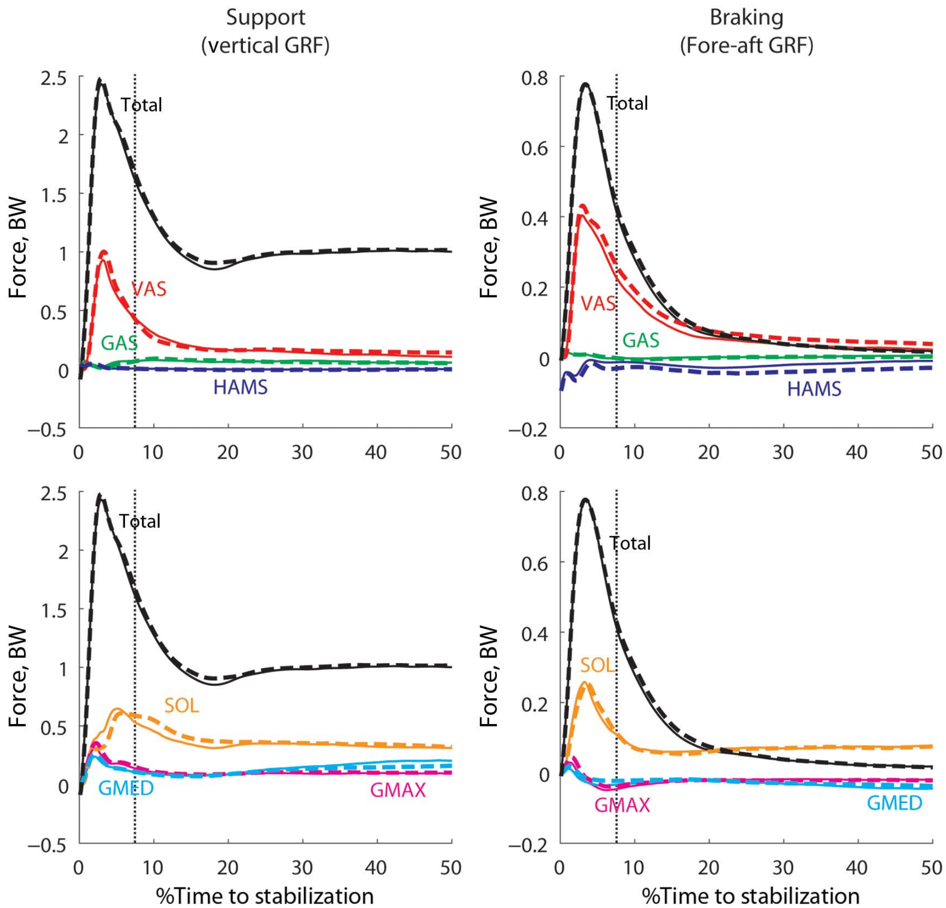

Figure 3 — Pattern of contributions by the major knee-spanning muscles (top row) and non-knee-spanning muscles (bottom row) of the lower limb to support (left) and braking (right) for the ACLR (dashed line) and uninjured control (solid line) groups. The vertical dashed line represents the approximate end of landing phase for both groups. Positive values for support and braking represent forces directed upward and backward, respectively. Rectus femoris and erector spinae contributions were almost zero throughout. ACLR indicates anterior cruciate ligament reconstruction; BW, body weight; GRF, ground reaction force; GAS, gastrocnemius; GMAX, gluteus maximus; GMED, gluteus medius; HAMS, hamstrings; SOL, soleus; VAS, vasti.

uniarticular muscles.[33] Thus a straighter, stiffer lower-limb landing configuration may be less able to dissipate the energy of impact through the muscles, instead directing it through the skeletal system. Tsai and Powers[12] found that this altered landing configuration increased axial hip, knee, and ankle joint loading, with implications for risk of ACL reinjury and future osteoarthritis. Thus, although we found that muscle function in support and braking is relatively unaffected by the altered lower-limb landing posture adopted by ACLR individuals, a retraining approach that focuses on improving overall kinematics to a more-flexed and less-stiff configuration may benefit landing comfort and mitigate reinjury risk. Additionally, based on our finding of larger vasti muscle forces in the uninjured group, we can infer that when a more-flexed and less-stiff landing posture is adopted, larger quadriceps forces may necessarily be required for effective support and braking. This therefore reinforces the current clinical focus on improving quadriceps strength, which has been associated with greater knee flexion angles and torques during single-leg landing tasks,[39] as part of a post-ACLR rehabilitation program.

From our findings, we derive an important constraint on the use of the GRF decomposition for quantifying muscle function. Since the GRF represents the body's interaction with its environment, GRF decomposition can therefore naturally only identify muscles that modulate interaction with the environment, but not muscles that modulate internal phenomena, such as posture. Other forms of analysis, such as muscle-induced joint/segment motion[40] or simply inspecting results (as in our present study), may be required to better quantify and explain these internal phenomena.

This study has some limitations. First, due to the cross-sectional study design, it was not possible to determine whether muscle force deficits resulted from ACL injury and/or surgery, or were preexisting; and whether these deficits are prospectively associated with knee function or sports participation. In addition, we did not exclude individuals with full-thickness chondral lesions or meniscal pathology and, as such, the relationship between concomitant chondral or meniscal injuries and muscle force deficits is unclear. Second, by implementing a hinge knee in our models, we did not include the effects of anteroposterior knee-joint laxity in our muscle force calculations. Knee-joint laxity is an important clinical feature in some ACLR individuals,[41] which may influence the level of coactivity in the knee-spanning muscles. Thus, although our models predicted increased quadriceps–hamstrings co-contraction in the ACLR group (Table 2), we would expect this level of co-contraction to be underestimated. Underestimation of co-contraction may also be partially attributed to our use of static optimization to calculate muscle forces. Nevertheless, static optimization produced patterns of muscle forces in the ACLR group that agreed well with measured EMG (Figure 2), and in the uninjured group, muscle forces agreed with both muscle forces and EMG reported for single-leg hop landings by Mokhtarzadeh et al,[42] and as well as drop landings by Iida et al.[35] Importantly, the study by Mokhtarzadeh et al[42] established that static optimization is a valid and robust technique for estimating muscle forces during the landing phase of a single-leg hop task. Therefore, we are confident that the stated limitations in muscle force calculations should not have significant impact on our findings.

Four major muscles—the vasti, soleus, gluteus medius, and gluteus maximus—were principally involved in support and braking during single-leg hop landing. Both groups achieved similar muscle function despite force deficits in these 4 muscles in the ACLR group. Postural adaptations in the ACLR group required more hamstrings and erector spinae activity, but these muscles did not impact center-of-mass modulation. Our findings provide a new perspective on understanding the complex neuromuscular interactions in biomechanical tasks associated with the evaluation of ACLR individuals for a successful return to sport.

Acknowledgments

L.G.P. was a recipient of a National Health and Medical Research Council (NHMRC) postgraduate scholarship. A.L.B. is a recipient of an NHMRC Career Development Fellowship (R.D. Wright Biomedical, 1053521). A.L.B. and P.P. gratefully acknowledge financial support from the Australian Defence Health Foundation (Medical Research Booster Grant). The authors have no conflicts of interest to disclose.

References

1. Mall NA, Chalmers PN, Moric M, et al. Incidence and trends of anterior cruciate ligament reconstruction in the United States. *Am J Sports Med*. 2014;42(10):2363–2370. PubMed ID: 25086064 doi:10.1177/0363546514542796
2. Ardern CL, Taylor NF, Feller JA, Webster KE. Fifty-five per cent return to competitive sport following anterior cruciate ligament reconstruction surgery: an updated systematic review and meta-analysis including aspects of physical functioning and contextual factors. *Br J Sports Med*. 2014;48(21):1543–1552. PubMed ID: 25157180 doi:10.1136/bjsports-2013-093398
3. Nawasreh Z, Logerstedt D, Cummer K, Axe M, Risberg MA, Snyder-Mackler L. Functional performance 6 months after ACL reconstruction can predict return to participation in the same preinjury activity level 12 and 24 months after surgery. *Br J Sports Med*. 2018;52(6):375. PubMed ID: 28954801 doi:10.1136/bjsports-2016-097095
4. Johnston PT, McClelland JA, Webster KE. Lower limb biomechanics during single-leg landings following anterior cruciate ligament reconstruction: a systematic review and meta-analysis. *Sports Med*. 2018;48(9):2103–2126. PubMed ID: 29949109 doi:10.1007/s40279-018-0942-0
5. Bryant AL, Newton RU, Steele J. Successful feed-forward strategies following ACL injury and reconstruction. *J Electromyogr Kinesiol*. 2009;19(5):988–997. PubMed ID: 18656383 doi:10.1016/j.jelekin.2008.06.001
6. Xergia SA, Pappas E, Zampeli F, Georgiou S, Georgoulis AD. Asymmetries in functional hop tests, lower extremity kinematics, and isokinetic strength persist 6 to 9 months following anterior cruciate ligament reconstruction. *J Orthop Sports Phys Ther*. 2013;43(3):154–162. PubMed ID: 23322072 doi:10.2519/jospt.2013.3967
7. Holsgaard-Larsen A, Jensen C, Mortensen NH, Aagaard P. Concurrent assessments of lower limb loading patterns, mechanical muscle strength and functional performance in ACL-patients—a cross-sectional study. *Knee*. 2014;21(1):66–73. PubMed ID: 23835518 doi:10.1016/j.knee.2013.06.002
8. Markstrom JL, Tengman E, Hager CK. ACL-reconstructed and ACL-deficient individuals show differentiated trunk, hip, and knee kinematics during vertical hops more than 20 years post-injury. *Knee Surg Sports Traumatol Arthrosc*. 2018;26(2):358–367. PubMed ID: 28337590
9. Lewek M, Rudolph K, Axe M, Snyder-Mackler L. The effect of insufficient quadriceps strength on gait after anterior cruciate ligament reconstruction. *Clin Biomech*. 2002;17(1):56–63. doi:10.1016/S0268-0033(01)00097-3
10. Williams GN, Buchanan TS, Barrance PJ, Axe MJ, Snyder-Mackler L. Quadriceps weakness, atrophy, and activation failure in predicted noncopers after anterior cruciate ligament injury. *Am J Sports Med*. 2005;33(3):402–407. PubMed ID: 15716256 doi:10.1177/0363546504268042
11. Bryant AL, Clark RA, Pua YH. Morphology of hamstring torque-time curves following ACL injury and reconstruction: mechanisms and implications. *J Orthop Res*. 2011;29(6):907–914. PubMed ID: 21259335 doi:10.1002/jor.21306
12. Tsai LC, Powers CM. Increased hip and knee flexion during landing decreases tibiofemoral compressive forces in women who have undergone anterior cruciate ligament reconstruction. *Am J Sports Med*. 2013;41(2):423–429. PubMed ID: 23271006 doi:10.1177/0363546512471184
13. Sharifi M, Shirazi-Adl A, Marouane H. Computation of the role of kinetics, kinematics, posterior tibial slope and muscle cocontraction on the stability of ACL-deficient knee joint at heel strike—towards identification of copers from non-copers. *J Biomech*. 2018;77:171–182. PubMed ID: 30033382 doi:10.1016/j.jbiomech.2018.07.003
14. Swanik CB, Lephart SM, Swanik KA, Stone DA, Fu FH. Neuromuscular dynamic restraint in women with anterior cruciate ligament injuries. *Clin Orthop Relat Res*. 2004(425):189–199.
15. Zajac FE, Gordon ME. Determining muscle's force and action in multi-articular movement. *Exerc Sport Sci Rev*. 1989;17(1):187–230. PubMed ID: 2676547.
16. Anderson FC, Pandy MG. Individual muscle contributions to support in normal walking. *Gait Posture*. 2003;17(2):159–169. PubMed ID: 12633777 doi:10.1016/S0966-6362(02)00073-5
17. Lin YC, Kim HJ, Pandy MG. A computationally efficient method for assessing muscle function during human locomotion. *Int J Numer Meth Biomed Eng*. 2010;27(3):436–449.
18. Pandy MG, Lin YC, Kim HJ. Muscle coordination of mediolateral balance in normal walking. *J Biomech*. 2010;43(11):2055–2064. PubMed ID: 20451911 doi:10.1016/j.jbiomech.2010.04.010
19. Liu MQ, Anderson FC, Pandy MG, Delp SL. Muscles that support the body also modulate forward progression during walking. *J Biomech*. 2006;39(14):2623–2630. PubMed ID: 16216251 doi:10.1016/j.jbiomech.2005.08.017
20. Hamner SR, Seth A, Delp SL. Muscle contributions to propulsion and support during running. *J Biomech*. 2010;43(14):2709–2716. PubMed ID: 20691972 doi:10.1016/j.jbiomech.2010.06.025
21. Lin YC, Fok LA, Schache AG, Pandy MG. Muscle coordination of support, progression and balance during stair ambulation. *J Biomech*. 2015;48(2):340–347. PubMed ID: 25498364 doi:10.1016/j.jbiomech.2014.11.019
22. Correa TA, Pandy MG. On the potential of lower limb muscles to accelerate the body's centre of mass during walking. *Comput Methods Biomech Biomed Engin*. 2013;16(9):1013–1021. PubMed ID: 22372586 doi:10.1080/10255842.2011.650634
23. Sritharan P, Lin YC, Richardson SE, Crossley KM, Birmingham TB, Pandy MG. Lower-limb muscle function during gait in varus mal-aligned osteoarthritis patients. *J Orthop Res*. 2018;36(8), 2157–2166.

24. Perraton L, Clark R, Crossley K, et al. Impaired voluntary quadriceps force control following anterior cruciate ligament reconstruction: relationship with knee function. *Knee Surg Sports Traumatol Arthrosc*. 2016;25(5):1424–1431. PubMed ID: 26745965 doi:10.1007/s00167-015-3937-5
25. Tegner Y, Lysholm J. Rating systems in the evaluation of knee ligament injuries. *Clin Orthop Relat Res*. 1985(198):43–49.
26. Noyes FR, Barber-Westin SD. Surgical reconstruction of severe chronic posterolateral complex injuries of the knee using allograft tissues. *Am J Sports Med*. 1995;23(1):2–12. PubMed ID: 7726346 doi:10.1177/036354659502300102
27. Culvenor AG, Perraton L, Guermazi A, et al. Knee kinematics and kinetics are associated with early patellofemoral osteoarthritis following anterior cruciate ligament reconstruction. *Osteoarthr Cartilage*. 2016;24(9):1548–1553. PubMed ID: 27188685 doi:10.1016/j.joca.2016.05.010
28. Schache AG, Baker R. On the expression of joint moments during gait. *Gait Posture*. 2007;25(3):440–452. PubMed ID: 17011192 doi:10.1016/j.gaitpost.2006.05.018
29. Delp SL, Anderson FC, Arnold AS, et al. OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE Trans Biomed Eng*. 2007;54(11):1940–1950. PubMed ID: 18018689 doi:10.1109/TBME.2007.901024
30. Lu TW, O'Connor JJ. Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. *J Biomech*. 1999;32(2):129–134. PubMed ID: 10052917 doi:10.1016/S0021-9290(98)00158-4
31. Anderson FC, Pandy MG. Static and dynamic optimization solutions for gait are practically equivalent. *J Biomech*. 2001;34(2):153–161. PubMed ID: 11165278 doi:10.1016/S0021-9290(00)00155-X
32. Rudolph KS, Axe MJ, Snyder-Mackler L. Dynamic stability after ACL injury: who can hop? *Knee Surg Sports Traumatol Arthrosc*. 2000;8(5):262–269. PubMed ID: 11061293 doi:10.1007/s001670000130
33. Prilutsky BI, Zatsiorsky VM. Tendon action of two-joint muscles: transfer of mechanical energy between joints during jumping, landing, and running. *J Biomech*. 1994;27(1):25–34. PubMed ID: 8106533 doi:10.1016/0021-9290(94)90029-9
34. Pandy MG, Andriacchi TP. Muscle and joint function in human locomotion. *Annu Rev Biomed Eng*. 2010;12(1):401–433. PubMed ID: 20617942 doi:10.1146/annurev-bioeng-070909-105259
35. Iida Y, Kanehisa H, Inaba Y, Nakazawa K. Activity modulations of trunk and lower limb muscles during impact-absorbing landing. *J Electromyogr Kinesiol*. 2011;21(4):602–609. PubMed ID: 21549617 doi:10.1016/j.jelekin.2011.04.001
36. Neptune RR, Wright IC, van den Bogert AJ. Muscle coordination and function during cutting movements. *Med Sci Sports Exerc*. 1999;31(2):294–302. doi:10.1097/00005768-199902000-00014
37. Oberlander KD, Bruggemann GP, Hoher J, Karamanidis K. Altered landing mechanics in ACL-reconstructed patients. *Med Sci Sports Exerc*. 2013;45(3):506–513.
38. Orishimo KF, Kremenic IJ, Mullaney MJ, McHugh MP, Nicholas SJ. Adaptations in single-leg hop biomechanics following anterior cruciate ligament reconstruction. *Knee Surg Sports Traumatol Arthrosc*. 2010;18(11):1587–1593. PubMed ID: 20549185 doi:10.1007/s00167-010-1185-2
39. Lepley LK, Wojtys EM, Palmieri-Smith RM. Combination of eccentric exercise and neuromuscular electrical stimulation to improve biomechanical limb symmetry after anterior cruciate ligament reconstruction. *Clin Biomech*. 2015;30(7):738–747. doi:10.1016/j.clinbiomech.2015.04.011
40. Zajac FE, Neptune RR, Kautz SA. Biomechanics and muscle coordination of human walking. Part I: introduction to concepts, power transfer, dynamics and simulations. *Gait Posture*. 2002;16(3):215–232. PubMed ID: 12443946 doi:10.1016/S0966-6362(02)00068-1
41. Keays SL, Bullock-Saxton JE, Keays AC, Newcombe PA, Bullock MI. A 6-year follow-up of the effect of graft site on strength, stability, range of motion, function, and joint degeneration after anterior cruciate ligament reconstruction: patellar tendon versus semitendinosus and Gracilis tendon graft. *Am J Sports Med*. 2007;35(5):729–739. PubMed ID: 17322130 doi:10.1177/0363546506298277
42. Mokhtarzadeh H, Perraton L, Fok L, et al. A comparison of optimisation methods and knee joint degrees of freedom on muscle force predictions during single-leg hop landings. *J Biomech*. 2014;47(12):2863–2868. PubMed ID: 25129166 doi:10.1016/j.jbiomech.2014.07.027