Elevated gastrocnemius forces compensate for decreased hamstrings forces during the weight-acceptance phase of single-leg jump landing: implications for anterior cruciate ligament injury risk

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A B S T R A C T

Approximately 320,000 anterior cruciate ligament (ACL) injuries in the United States each year are non-contact injuries, with many occurring during a single-leg jump landing. To reduce ACL injury risk, one option is to improve muscle strength and/or the activation of muscles crossing the knee under elevated external loading. This study’s purpose was to characterize the relative force production of the muscles supporting the knee during the weight-acceptance (WA) phase of single-leg jump landing and investigate the gastrocnemius forces compared to the hamstrings forces. Amateur male Western Australian Rules Football players completed a single-leg jump landing protocol and six participants were randomly chosen for further modeling and simulation. A three-dimensional, 14-segment, 37 degree-of-freedom, 92 muscle-tendon actuated model was created for each participant in OpenSim. Computed muscle control was used to generate 12 muscle-driven simulations, 2 trials per participant, of the WA phase of single-leg jump landing. A one-way ANOVA and Tukey post-hoc analysis showed both the quadriceps and gastrocnemius muscle force estimates were significantly greater than the hamstrings (p < 0.001). Elevated gastrocnemius forces corresponded with increased joint compression and lower ACL forces. The elevated quadriceps and gastrocnemius forces during landing may represent a generalized muscle strategy to increase knee joint stiffness, protecting the knee and ACL from external knee loading and injury risk. These results contribute to our understanding of how muscle's function during single-leg jump landing and should serve as the foundation for novel muscle-targeted training intervention programs aimed to reduce ACL injuries in sport.

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1. Introduction

Over 400,000 anterior cruciate ligament (ACL) injuries occur annually in the United States (Utturkar et al., 2013) costing the health care system approximately $1.5 billion (Boden et al., 2000; Kao et al., 1995). Approximately 80% of ACL injuries are non-contact, with most occurring during single-leg jump landing or sidestepping sports tasks (Cochrane et al., 2007; Koga et al., 2010; Krosshaug et al., 2007). During single-leg jump landing with the knee near full extension, the application of externally applied translational forces coupled with valgus and internal rotation knee moments elevate the forces on the ACL to injurious thresholds (> 2160 N) (Markolf et al., 1995; McLean et al., 2005, 2008; Walla et al., 1985; Woo et al., 1991). Most injury preventative training protocols focus on reducing externally applied knee loads and/or increasing the support of muscles crossing the knee when loading is elevated to mitigate ACL strain and injury risk. With ACL injury rates appearing to increase 50% over the past decade (Donnelly et al., 2012a), it appears injury prevention research is not effectively translating into injury prevention among heterogeneous community-level athletic populations.

The roles muscles play in supporting the knee during landing are not well understood. The primary function of the neuromuscular system during landing is to generate a support moment, keeping the center of mass (CoM) upright. A secondary proposed function is the co-contraction of the quadriceps and hamstrings muscles, which is believed to be essential to protecting the knee during dynamic movements, specifically with regard to ACL injury prevention. However, recent literature has shown that the gastrocnemius muscles may play an elevated role in supporting the knee during landing because hamstrings, as well as the gastrocnemii...
and soleus muscles, can potentially reduce ACL injury risk (Boden et al., 2010; Hewett et al., 2007; Mokhtarzadeh et al., 2013; Podraza and White, 2010). Furthermore, moderate hamstrings activation, compared to quadriiceps activation, has been linked to elevated knee valgus and internal rotation moments which are often predictors of ACL injury risk (Donnelly et al., 2012a; Hewett et al., 2005; Wojtys et al., 2002). Thus, it is possible that elevated gastrocnemii force could function to replace and/or work in conjunction with the hamstrings to reduce ACL injury risk during dynamic sports tasks.

Surface electromyography (sEMG) has been used to estimate muscle activation, where muscle force and function during sports tasks is then inferred (Besier et al., 2003; Lloyd and Buchanan, 2001; Wikstrom et al., 2008). Yet, sEMG measurements do not account for muscle architecture, force-length-velocity relationships or muscle moment arm geometry during dynamic movements. Therefore a gap exists in estimating muscle forces, and more importantly function, during dynamic sports tasks. Muscle-actuated, forward dynamic simulation is an in silico computational tool bridging this gap, providing valuable insights into the roles individual muscles play during dynamic movements (Seth et al., 2011; Thelen and Anderson, 2006; Thelen et al., 2003). This tool has been used to analyze muscle force contributions to dynamic movements such as walking, cycling, running, sidestepping and landing tasks, and in combination with sEMG may be used to investigate single-leg jump landing (Arnold et al., 2007; Hamer et al., 2010; Laughlin et al., 2011; Thelen et al., 2003; Weinhandl et al., 2013).

This study used dynamic simulation, with motion capture data, to investigate the role lower limb muscles crossing the knee play in mitigating ACL injury risk during single-leg jump landing. The objective of this work was to characterize the force production of the muscles supporting the knee during the weight-acceptance (WA) phase of single-leg jump landing. Here, support is defined as increasing joint stiffness, and mitigating ACL forces. It is hypothesized that the gastrocnemii and quadriceps will produce force to elevate joint compression, which will prevent anterior tibial translation (ATT). With this information, our understanding of muscle function in single-leg jump landing will increase so researchers are better informed on which muscles to target in developing preventative ACL injury training protocols to reduce ACL injury risk.

2. Methods

2.1. Experimental protocol and data collection

Thirty-four amateur male Western Australian Rules Football players were recruited to perform a single-leg jump landing experimental protocol (Donnelly et al., 2012b). Six participants (age 20.5 ± 1.9 years, height 1.9 ± 0.1 m, mass 88.3 ± 5.5 kg) were randomly selected from this cohort. Two trials per participant for a total of 12 experimental trials were chosen for further subject-specific modeling and dynamic simulation analysis. Participants were instructed to jump from their preferred leg (right leg for all) and, in flight, grab an Australian rules football randomly swung medially, laterally or held central to the participants approach direction (Dempsey et al., 2012). The ball height was approximately 90% of each participant’s maximal vertical jump height. Participants were instructed to land on the force platform with their preferred leg. Of the 12 jump landing trials analyzed, eight trials were assessed when the ball was swung laterally, three trials when the ball was swung medially and one central. All experimental procedures were approved by the Human Research Ethics Committee and all participants provided informed consent prior to data collection.

Fifty-six upper- and lower-body retro-reflective markers were utilized to capture kinematic trajectories (Dempsey et al., 2012). Marker trajectories were recorded at 250 Hz using a 12-camera Vicon MX motion capture system (Vicon Peak, Oxford Metrics Ltd., UK) (Dempsey et al., 2007; Donnelly et al., 2012b). Ground reaction force (GRF) data were synchronously recorded at 2,000 Hz using an AMTI (Advanced Mechanical Technology Inc., Watertown, MA) 12 × 1.2 m force platform. Both the kinematic and GRF data were low-pass filtered using a zero phase-shift, fourth-order Butterworth filter with a cutoff frequency of 20 Hz in Workstation (ViconPeak, Oxford Metrics Ltd., UK). The sEMG data were synchronously collected at 2,000 Hz for six muscles: medial and lateral vasti, gastrocnemii and hamstrings. The raw experimental sEMG data were band-pass filtered using a zero phase-shift, fourth-order Butterworth filter with a band-pass filter cutoff frequencies of 30 and 500 Hz, full wave rectified and then low-pass filtered using a zero phase-shift, fourth-order Butterworth filter at a cutoff frequency of 6 Hz to create linear envelopes. The peak muscle activation from each muscle recorded during the protocol was used to normalize each muscle’s maximum sEMG signal.

2.2. Subject-specific models and simulations

Six three-dimensional, 14-segment, 37 degree-of-freedom (DoF), 92 muscle-tendon actuated subject-specific models were created in OpenSim 1.3.1 (Delp et al., 1990) to generate simulations of each participant performing the landing task (Fig. 1). The details of this model have been described previously (Donnelly et al., 2012b). The 92 muscle-tendon units actuated the lower extremities and lower back joints, while the arms were actuated by torque actuators described previously (Hammer et al., 2010). The maximum isometric force of each muscle was increased by 60% compared to the generic OpenSim model based on research by Arnold et al. (2010). The model included a 5 DoF knee that rotated about all three planes. Sagittal and transverse plane translations were modeled as a function of knee angle (Donnelly et al., 2012b, Delp et al., 1990). The knee was actuated by muscles and ideal torque actuators (+ 50 Nm), the values are consistent with previous literature that provided the resistance supplied by the knee ligaments and articular surface that help support the knee in the frontal plane (Seedhom et al., 1972; Zhao et al., 2007). An ACL was included in the model and scaled to each participant. The ligament model was described in a previous study (Xu et al., 2014). Subject-specific joint centerst were derived using functional knee and hip joint methods (Besier et al., 2003), custom biomechanical models in MATLAB (MATLAB 7.8, The MathWorks, Inc., Natick Massachusetts, USA) and Vicon Bodybuilder (Dempsey et al., 2007). The resulting joint centers, marker trajectories and GRF data were then exported to OpenSim. Segment lengths were scaled to each participant’s specific joint centers and segment masses to each participant’s total body mass.

Inverse kinematics (IK) was used to derive simulated joint angles from the experimental marker data recorded during the jump landing. Residual reduction analysis (RRA) was used to create simulations that were dynamically consistent with the experimentally recorded GRFs (Delp et al., 2007; Donnelly et al., 2012b). Muscle forces were estimated for the WA phase of single-leg jump landing using computed muscle control (CMC). CMC is an algorithm that utilizes static optimization, forward dynamics and feedback control to estimate individual muscle forces during dynamic movements (Thelen and Anderson, 2006; Thelen et al., 2003). Static optimization and forward dynamic analyses are used to compute the muscle forces that drive the Hill-type muscle model to replicate the experimental joint motion. The proportional-derivative feedback control is implemented to ensure the simulated joint motion tracks the experimental joint motion.

The WA phase was defined as the time from the initial contact to the end of peak loading in the vertical GRF profile (Dempsey et al., 2007). The WA phase of single-leg jump landing was analyzed as this phase is when the ACL is at the greatest risk for injury (Dempsey et al., 2007; Donnelly et al., 2012a).

2.3. Muscle force estimates during single-leg jump landing

Muscle force estimates for nine muscles crossing the knee and the soleus were analyzed to determine their contribution during the WA phase of single-leg jump landing. The mean normalized maximum muscle forces for nine muscles (vastus medialis, vastus lateralis, vastus intermedius, rectus femoris, biceps femoris, semitendinosus, semimembranosus, medial gastrocnemius, lateral gastrocnemius) and the soleus were analyzed individually and in groups of functional relevance (i.e., quadriceps, hamstrings and gastrocnemii). One-way ANOVAs were conducted to compare the mean maximum individual and group muscle force estimates. A Tukey post-hoc analysis was performed to determine if differences observed in the one-way ANOVA analysis were significant (α = 0.05).

2.4. ACL force calculation during single-leg jump landing

The ACL force during single-leg jump landing was calculated as a function of the change in the gastrocnemii forces and the length of the mean ACL length. The cohort mean ACL length was 35.3 ± 2.3 mm and the linear elastic stiffness was 240 N/mm based on Woo et al. (1991). An ANOVA of the participant’s maximum ACL forces identified two distinct populations – a high- and low-risk group whose forces exceeded or fell below 2,000 N, respectively. A one-way ANOVA was conducted to compare the muscle group means for the aforementioned risk groups while a Tukey post-hoc analysis was conducted to determine the significance of the observed differences between the two groups (α = 0.05). Joint reaction force analysis was performed and a one-way ANOVA was conducted to compare the means of the joint compressive and shear forces for the high- and low-risk groups.
3. Results

Gastrocnemii and quadriceps forces were higher than hamstrings forces during the WA phase of single-leg jump landing. No differences were observed in individual muscle force production between the participants and trials by conducting a one-way ANOVA that compared the means of the maximum individual muscle forces based on the swing direction. Thus, all 12 trials were analyzed together. The individual muscle forces for the nine muscles crossing the knee were normalized by their individual maximum isometric force values used during the simulation and plotted as such to determine their relative force contribution (Fig. 2); however, their non-normalized forces were compared for the one-way ANOVA (Table 1). The largest muscle force estimates during the WA phase of single-leg jump landing in decreasing order were the quadriceps (1,748 ± 286 N), gastrocnemii (1,331 ± 516 N) and hamstrings (475 ± 232 N) (Table 2). The maximum force production between these muscle groups were significantly different \((p < 0.001)\) with the post-hoc analysis showing the quadriceps muscles produced significantly greater force than both the gastrocnemii \((p < 0.001)\) and hamstrings \((p < 0.001)\) muscles and mean maximum gastrocnemii muscle force estimates were significantly greater than the hamstrings \((p < 0.001)\).

The mean ACL strain was 24.2 ± 0.01% and 13.3 ± 0.03% for the high- and low-risk groups, respectively. The mean ACL force for the high-risk group was 2,092 ± 73 N \((n = 4)\) and for the low-risk group 1,152 ± 206 N \((n = 8)\). The mean gastrocnemii muscle force was larger in the low-risk group (1,348 ± 549 N) compared to the high-risk group (1,296 ± 520 N). Conversely, the quadriceps and hamstrings produced greater forces when the ACL was at higher risk for injury (Table 3). The mean compressive force was larger in the low-risk group (11,074 ± 2,658) compared to the high-risk group (9,646 ± 2,269) while the mean shear force was larger in the high-risk group. The differences were not significant in either group (Table 3).

The mean deviation between experimental and muscle-actuated simulation kinematics was 3.5 ± 1.4° for all lower extremity joint angles during the WA phase of single-leg jump landing for all participants with a maximum of 9.9° for hip abduction (Fig. 3, Table 4). These deviations in simulated joint kinematics and external moments are needed to improve the dynamic consistency with experimentally recorded GRF. All simulations were dynamically consistent with low peak residual forces (5 N) and moments (8 Nm) at the pelvis. The CMC excitations used to drive the simulation were closely aligned with the experimentally measured...
sEMG activation data (Fig. 4). The consistency between the simulated joint kinematics, kinetics, and muscle excitations and the experimentally recorded data suggests the simulations represented the experimental task.

4. Discussion

The purpose of this study was to characterize the force production of the muscles supporting the knee during the WA phase of single-leg jump landing. Results showed that the quadriceps generated the greatest force followed by the gastrocnemii and then the hamstrings. This result was observed for both the high and low ACL injury risk groups. Additionally, the quadriceps reached maximum force earlier than the gastrocnemii and hamstrings. Future research for effectively designing preventative training protocols should consider targeting the strength and coordination of these muscle groups, particularly the quadriceps and gastrocnemii in the development of ACL injury prevention training protocols.

### Table 1
Mean maximum and minimum muscle force estimates for the individual muscles during the weight-acceptance phase of single-leg jump landing for 12 trials.

<table>
<thead>
<tr>
<th>Participant</th>
<th>Muscle</th>
<th>Value</th>
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<th>1b</th>
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<th>6a</th>
<th>6b</th>
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<td>1,426 ± 732^cd,e</td>
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<td>65</td>
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<td>69</td>
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Notes: ANOVA identified a significant difference for the maximum values of the individual muscles (p < 0.001; n = 12).
There are several possible biomechanical explanations for why each muscle group crossing the knee produced force differently when supporting the knee during single-leg jump landing. Force production by the quadriceps, gastrocnemii and hamstrings is likely utilized to improve joint kinematics and reduce the strain exerted on the ACL and other knee ligaments during single-leg landing (Podraza and White, 2010; Riemann and Lephart, 2002). Comparisons of the muscle forces between the high and low ACL risk groups showed that the gastrocnemii muscle forces were elevated when the ACL strain and subsequent ACL force was lower. The results showed that the force produced by the hamstrings was not enough to counterbalance the quadriceps force indirectly supporting the use of the gastrocnemii to help support the knee. The increased gastrocnemii muscle forces also corresponded with increased compressive forces in individuals with lower ACL forces. This indicates that when the ACL forces are lower, individuals adopt a strategy where the loads at the articular surface appear to be produced by the gastrocnemii muscles. Conversely, the load is distributed to the ACL when the gastrocnemii exhibit smaller muscle forces. These results are supported by research that found that joint compression via muscular contraction can limit ATT and valgus loading to help protect knee ligaments (Hewett et al., 2010; Mokhtarzadeh et al., 2013; Podraza and White, 2010; Rubenson et al., 2012).

The mean maximum soleus force produced by the participants during the single-leg jump landing task (3,339 ± 1,129 N) is consistent with peak isometric in vivo force measurements (3,469 ± 720 N) reported by Rubenson et al. (2012). Previous research has suggested that the force produced by the soleus muscle would add to the gastrocnemii-soleus complex force generating capacity, suggesting the role of the gastrocnemii in supporting the knee during single-leg landing may be underestimated (Boden et al., 2010; Mokhtarzadeh et al., 2013; Podraza and White, 2010; Rubenson et al., 2012).

The gastrocnemii are biarticular muscles that have multiple functions about the knee and ankle. Gastrocnemii’s primary function is to plantarflex the foot during landing, which contributes to the production of a support moment (Winter, 1980). Results presented here suggest its secondary function may be to co-contract with the quadriceps to elevate joint compression and protect the knee and ACL from external joint loading. Previous research has shown that the gastrocnemii muscles can act as a knee flexor while elevating joint compression (Kvist, 2004; O’Connor et al., 1990). Joint compression helps limit ATT via friction between the joints (Hsieh and Walker, 1976; Kvist, 2004; Myer et al., 2005). Joint compression is the result of muscular co-contraction and research supports the findings here that the quadriceps and gastrocnemii work synergistically to stabilize the knee through joint compression (Kvist and Gillquist, 2001). The ratio of compressive to shear force favors a compression mechanism, which limits ATT. Here elevated gastrocnemii forces were associated with a higher compressive to shear force ratio, thus it is possible that the gastrocnemii muscles may not be a strong contributor to elevated ATT. These results are further supported by studies where gastrocnemii muscle activity is significantly elevated compared to the hamstrings during jump landings (Fagenbaum and Darling, 2003; Nyland et al., 2010; Padua et al., 2005). Results here have shown mean gastrocnemii force was greater than the hamstrings across all 12 simulations irrespective of landing condition (i.e. swing direction). These findings suggest a generalized muscle force strategy may be used to generate a support moment to resist the fall of the CoM, while also compressing and supporting the knee and ACL from external knee loading and injury risk during single-leg landing (Boden et al., 2009).

Musculoskeletal modeling for biomechanical analysis is challenging. Often assumptions regarding model parameters have to be made. The model included a 5 DoF knee; however, the model did not include every knee ligament or articular surface, which can function to support the knee against frontal plane knee moments (Delp et al., 1990). Using OpenSim joints and actuators without implementation of complex contact model components, this limitation was addressed by allowing the knee to move in the frontal plane by including ideal torque actuators to represent the ligaments and articular surface that support the knee against external frontal plane knee moments. Since the simulated muscle excitations were similar to experimentally recorded excitations, it is unlikely that the inclusion of the ideal torque actuators have significantly affected the muscle force results in this study. Additionally, ideal torque actuators worked with muscles, not against the muscles, to help support the knee during landing as the model accurately tracked the frontal plane knee motion.

The model’s maximum isometric muscle forces were uniformly increased 60% to better represent the muscle architecture of a young healthy athletic adult male population (Arnold et al., 2010;}

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**Table 2**

Mean maximum and minimum muscle force estimates for the three muscle groups during the weight-acceptance phase of single-leg jump landing for 12 trials.

<table>
<thead>
<tr>
<th>Participant Muscle Force (N)</th>
<th>1a</th>
<th>1b</th>
<th>2a</th>
<th>2b</th>
<th>3a</th>
<th>3b</th>
<th>4a</th>
<th>4b</th>
<th>5a</th>
<th>5b</th>
<th>6a</th>
<th>6b</th>
<th>Mean ± SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Quadriceps</td>
<td>Max</td>
<td>1,897</td>
<td>1,521</td>
<td>1,839</td>
<td>1,797</td>
<td>1,733</td>
<td>2,134</td>
<td>1,946</td>
<td>1,762</td>
<td>1,273</td>
<td>1,345</td>
<td>1,507</td>
<td>1,734 ± 286</td>
</tr>
<tr>
<td></td>
<td>Min</td>
<td>261</td>
<td>309</td>
<td>334</td>
<td>274</td>
<td>115</td>
<td>394</td>
<td>238</td>
<td>196</td>
<td>115</td>
<td>119</td>
<td>645</td>
<td>114</td>
</tr>
<tr>
<td>Gastrocnemii</td>
<td>Max</td>
<td>1,553</td>
<td>1,680</td>
<td>1,013</td>
<td>2,366</td>
<td>607</td>
<td>1,278</td>
<td>1,457</td>
<td>1,377</td>
<td>1,694</td>
<td>852</td>
<td>867</td>
<td>745</td>
</tr>
<tr>
<td></td>
<td>Min</td>
<td>330</td>
<td>211</td>
<td>324</td>
<td>378</td>
<td>195</td>
<td>351</td>
<td>345</td>
<td>482</td>
<td>119</td>
<td>267</td>
<td>401</td>
<td>463</td>
</tr>
<tr>
<td>Hamstrings</td>
<td>Max</td>
<td>594</td>
<td>1,019</td>
<td>486</td>
<td>609</td>
<td>398</td>
<td>176</td>
<td>500</td>
<td>506</td>
<td>252</td>
<td>506</td>
<td>362</td>
<td>125</td>
</tr>
<tr>
<td></td>
<td>Min</td>
<td>69</td>
<td>145</td>
<td>64</td>
<td>46</td>
<td>36</td>
<td>58</td>
<td>63</td>
<td>81</td>
<td>125</td>
<td>97</td>
<td>7</td>
<td>63 ± 40</td>
</tr>
</tbody>
</table>

Notes: ANOVA identified a significant difference for the maximum values of the muscle groups (p < 0.001; n = 3).

---

**Table 3**

<table>
<thead>
<tr>
<th>Muscle Group</th>
<th>Value 1a</th>
<th>Value 1b</th>
<th>Value 2a</th>
<th>Value 2b</th>
<th>Value 3a</th>
<th>Value 3b</th>
<th>Value 4a</th>
<th>Value 4b</th>
<th>Value 5a</th>
<th>Value 5b</th>
<th>Value 6a</th>
<th>Value 6b</th>
<th>Mean ± SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Quadriceps</td>
<td>Max</td>
<td>1,729 ± 289</td>
<td>1,778 ± 319</td>
<td>1,348 ± 549</td>
<td>1,296 ± 520</td>
<td>429 ± 146</td>
<td>568 ± 361</td>
<td>11,074 ± 2,658</td>
<td>9,646 ± 2,269</td>
<td>3,489 ± 1,556</td>
<td>2,858 ± 1,678</td>
<td>3,478 ± 286</td>
<td>4,238 ± 305</td>
</tr>
<tr>
<td>Gastrocnemii</td>
<td>Min</td>
<td>69</td>
<td>145</td>
<td>64</td>
<td>46</td>
<td>36</td>
<td>58</td>
<td>63</td>
<td>81</td>
<td>125</td>
<td>97</td>
<td>7</td>
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</tr>
</tbody>
</table>

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Lexell et al., 1988), since the baseline force values were derived from elderly cadavers (Delp et al., 1990). While these increases in maximum isometric muscle force were sufficient to facilitate the generation of accurate single-leg jump landing simulations, a more universal method for adjusting muscle forces for varying populations may be necessary and should be addressed in future research.

Despite these assumptions, the simulated kinematics, kinetics and muscle excitations were comparable against experimental kinematic, kinetic and muscle activation estimates and provided confidence that the results are representative of muscle forces during single-leg jump landing. Furthermore, the resulting ACL strain and forces were comparable with literature and the ACL forces were below the potential injury threshold of 2160 N defined by Woo et al. (1991).

Though females suffer ACL injuries at a disproportionately higher rate than males (Hewett et al., 2006), this study investigated male muscle force estimates during single-leg landing. Females tend to produce smaller knee flexor moments than men limiting their ability to counterbalance the quadriceps and reduce ATT may be the cause for higher injury rate (Hewett et al., 2006; Hewett et al., 1996). The males in this study demonstrated that elevated force production by the gastrocnemius-soleus complex could address this muscle imbalance and resist ATT, a finding observed in the literature (Mokhtarzadeh et al., 2013; Podraza and White, 2010). While female models would have different skeletal geometry and muscle strength, females would have the same goal for muscle force production during single-leg landing tasks; to generate a moment to support the CoM, stiffen the knee and mitigate ACL loading. Therefore, the overall findings of this study is the gastrocnemius muscle groups play an important role in mitigating ACL injury risk during single-leg jump landing and this information can serve as the foundation for novel muscle-targeted training intervention programs to reduce ACL injuries.

**Conflict of interest statement**

We do not have any financial or personal relationships with other people or organizations that could inappropriately influence our manuscript.
Fig. 4. Comparison of experimental surface electromyography (sEMG) and simulated muscle excitations during the weight-acceptance phase of single-leg jump landing for an example participant. Experimental unfiltered full wave rectified (gray area) and filtered (solid line) sEMG and simulated muscle excitations (dashed line) estimated during the weight-acceptance phase of single-leg jump landing. The experimental unfiltered full wave rectified (gray area) and filtered (solid line) sEMG data are individually normalized to the maximum recorded signal of each muscle over one of the landing trials. Simulated excitations (dashed line) are defined to be between 0 (no excitation) and 1 (full excitation).

Acknowledgments

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Appendix A. Supporting information

Supplementary data associated with this article can be found in the online version at http://dx.doi.org/10.1016/j.jbiomech.2014.08.016.

References


