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A model-based approach to predict neuromuscular control patterns that minimize ACL forces during jump landing

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ABSTRACT
Jump landing is a common situation leading to knee injuries involving the anterior cruciate ligament (ACL) in sports. Although neuromuscular control is considered as a key injury risk factor, there is a lack of knowledge regarding optimum control strategies that reduce ACL forces during jump landing. In the present study, a musculoskeletal model-based computational approach is presented that allows identifying neuromuscular control patterns that minimize ACL forces during jump landing. The approach is demonstrated for a jump landing maneuver in downhill skiing, which is one out of three main injury mechanisms in competitive skiing.

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KEYWORDS
ACL injuries; musculoskeletal simulation; optimal control; muscle activation

1. Introduction

Knee injuries involving the anterior cruciate ligament (ACL) are one of the most common and severe injuries in sports (Griffin et al. 2006; Alentorn-Geli et al. 2009). ACL injuries imply high surgical costs, a long and intensive rehabilitation time and long-term consequences for the athletes such as an increased risk of osteoarthritis or a second ACL injury after ACL reconstruction (Lohmander et al. 2007; Yu and Garrett 2007; Paterno et al. 2014).

Inciting events/typical situations, which result in ACL injuries, are non-contact in nature and involve sudden deceleration tasks such as jump landing maneuvers with or without a change of direction (Griffin et al. 2006; Carlson et al. 2016). Video analysis studies provide important information regarding the kinematics of the athlete during actual ACL injury cases and possible risk factors. In a recent review of Carlson et al. (2016), for example, the landing position of the athlete was stated to influence the likelihood of ACL injury to a much greater extent than inherent risk factors. However, video analysis studies are restricted to kinematic analyses and do not provide information about the underlying joint moments or the muscle forces corresponding to the kinematics of the athlete.

Muscle forces are well known to affect the loading of the ACL and impaired neuromuscular control has been identified as an important modifiable risk factor of ACL injury (Griffin et al. 2000; Hewett et al. 2005, 2010; Hurd et al. 2006; Taylor et al. 2011; Letafatkar et al. 2015). Specifically, impaired neuromuscular control of the muscles spanning the knee joint has been associated with insufficient and/or delayed recruitment of the hamstrings (McLean et al. 2010; Saunders et al. 2014) and elevated activation of the quadriceps muscles(Griffin et al. 2000; Chappell et al. 2007; Yu and Garrett 2007; Hewett et al. 2010). While high quadriceps activation in combination with a small knee flexion angle induces an anterior directed shear force on the tibia straining the ACL, the hamstrings induce a posterior directed shear force on the tibia protecting the ACL. Muscles spanning the ankle and hip joints were also proposed to affect the loading of the ACL during jump landing. In particular, *in vitro* (Elias et al. 2003) as well as *in vivo* experimental studies (Mokhtarzadeh et al. 2013) pointed to the capability of the soleus to act as an agonist for the ACL (i.e. protecting the ACL) and of the gastrocnemius to act as an ACL antagonist. Reports of reduced ACL loading through a less erect posture during landing tasks (Shimokochi et al. 2013) also suggest that the

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activation patterns of the hip muscles might affect injury risk. Elucidation of neuromuscular control strategies resulting in low ACL forces during injury-prone or high-risk situations might be an important step toward a better understanding of the corresponding injury mechanism and the development of dedicated neuromuscular training programs.

Computer simulation incorporating a model of the human musculoskeletal system is a powerful tool to analyze the relationship between joint kinematics, joint moments, muscle forces, neuromuscular control patterns and knee joint loading (Pandy and Andriacchi 2010). In addition, optimization methods can be applied to identify optimum movement and neuromuscular control patterns that minimize certain quantities contributing to knee joint loading. Musculoskeletal simulation models, for example, were successfully applied to identify optimized movement and control patterns during walking and sidestepping such as to minimize the axial knee joint contact force during walking (Miller et al. 2013), valgus knee loading during sidestepping (Donnelly et al. 2012) or the external knee adduction moment during gait (Miller et al. 2015). To our knowledge, however, optimized control strategies that minimize ACL forces during jump landing maneuvers have not been investigated. Based on the results of previous studies one might infer that ACL injury risk during jump landing might be reduced through altered muscle activation patterns such as decreased quadriceps activation and/or increased hamstrings activation and/or increased soleus activation. However, it is unknown whether these effects are additive, whether there are compensatory effects of other muscles, which of these changes have the highest effect on peak ACL force and how changed activation patterns might affect the corresponding kinematics of the athlete during the landing movement.

The purpose of the present study was to develop a computational approach aiming to identify neuromuscular control patterns that minimize ACL forces during jump landing. In this work, we focused on jump landing in competitive downhill skiing, which was reported as a high-risk situation and one out of three main injury situations in competitive alpine skiing (Bere et al. 2011) with landing heights approaching 2 m (Schindelwig et al. 2015). First, we evaluated a commonly assumed muscle coordination strategy minimizing the sum of squared muscle activation patterns. Second, we calculated an optimized control strategy aiming at minimizing the force in the ACL during the jump landing maneuver.

2. Methods

2.1. Musculoskeletal model

A two-dimensional sagittal plane 25 degrees of freedom musculoskeletal model of an alpine skier with two racing skis was used (Heinrich et al. 2018) to simulate jump landing in downhill skiing. The model of the skier consisted of seven rigid segments. A trunk segment was used to represent the head, arms and torso of the skier and three segments were used for each lower extremity (i.e. thigh, shank and foot). The restraining effect of the ski boot at the ankle joint was described by a passive moment and a nonlinear hysteresis curve (Eberle et al. 2017). The mass of the ski boot was incorporated in the foot segment. The model of the racing ski, with a nominal length of 2.11 m, consisted of a chain of nine rigid segments. Bending stiffness and damping were incorporated using passive spring-damper elements. The contact between the skis and the snow was described by a penetration force and Coulomb friction acting on every segment of the two skis (Heinrich et al. 2014).

The motion of the skier was actuated by 16 three-element Hill-type muscles, eight for each lower extremity: iliopsoas (Ili), glutei (Glu), hamstrings (Ham), rectus femoris (RF), vasti (Vas), gastrocnemius (Gas), soleus (Sol) and tibialis anterior (TA). Muscle contraction dynamics were modeled in accordance with McLean et al. (2003) and incorporated the muscles' force-length-velocity properties. Muscle activation dynamics were described as a first-order process (He et al. 1991) and modeled the activation of each muscle in response to a given muscle excitation pattern as input. A linear relationship was assumed between joint angles and muscle-tendon length (McLean et al. 2003) and muscle parameters were taken from Gerritsen et al. (1996). Based on the relationship between joint angles and muscle-tendon length, the muscle's moment arms and corresponding joint moments were derived using the principle of virtual work (van den Bogert et al. 2011). Muscle parameters and moment arms are provided as Supplementary online material.

The dynamics of the musculoskeletal skier model – multibody dynamics, contraction and activation dynamics – were formulated in the following compact implicit form

$$f(x(t), \dot{x}(t), u(t)) = 0 \tag{1}$$

which offered efficient solution methods for the jump landing simulations (van den Bogert et al. 2011). In Equation (1), the vector $x(t)$ represents the state variables consisting of the generalized coordinates $q(t)$ and velocities $\dot{q}(t)$ of the skier model, the muscle activations $a(t)$ and the length of the contractile elements of the muscles $L_{CE}(t)$. The vector $u(t)$ represents the muscle excitation patterns, which served as control variables of the skier model.

2.2. Baseline landing simulation

We computed a baseline landing simulation tracking experimental kinematic data of a professional downhill skier, who performed a safe jump landing maneuver in competitive downhill skiing (Nachbauer et al. 1996). Specifically, the landing simulation was formulated as optimal control problem (Heinrich et al. 2018). The task of the optimal control problem was to find the values of the state variables $x(t)$ and control variables $u(t)$ of the musculoskeletal skier-ski model such that a given objective function J_{ref}

$$J_{ref} = J_1 + J_2 \tag{2}$$

is minimized, while subjected to the following constraints

$$f(x(t), \dot{x}(t), u(t)) = 0, \qquad t_0 \le t \le t_1 \tag{3a}$$

$$0 \le u(t) \le 1, \qquad t_0 \le t \le t_1 \tag{3b}$$

$$x_{hip}(t) = x_{hip,\,data}(t), \qquad t = 0 \tag{3c}$$

The objective function J_{ref} consisted of two terms J_1 and J_2. The first term J_1 was used to minimize the differences between the experimental data $q_{i,\,data}(t)$ and the corresponding generalized coordinates $q_i(t)$ of the skier model in a weighted least square sense. The experimental data consisted of hip, knee and ankle joint angles, and the position and orientation of the trunk segment of a professional downhill skier, who performed a landing maneuver during a World Cup competition and were taken from the study of Nachbauer et al. (1996). The differences were scaled by factors $1/\sigma_i$ and averaged across the number of experimental kinematic variables $n = 9$ and the simulation duration $t_1 - t_0$. The simulation time was defined from $t_0 = -0.1$s before ground contact until the time of peak knee flexion $t_1 = 0.27$s after ground contact. J_1 had the following form

$$J_1 = \frac{1}{9(t_1 - t_0)} \sum_{i=1}^{9} \int_{t_0}^{t_1} \left(\frac{q_i(t) - q_{i,\,data}(t)}{\sigma_i} \right)^2 dt. \tag{4}$$

The second term was used to resolve muscle redundancy. Especially we considered a commonly assumed muscle coordination strategy (Laughlin et al. 2011; Mokhtarzadeh et al. 2013, 2017) corresponding to the sum of squared muscle activation patterns $a_j(t)$. The sum was averaged across the number of muscles $m = 16$ and the simulation duration $t_1 - t_0$. J_2 was scaled by a factor w_{mus} set to 10 according to previous simulations (Heinrich et al. 2014). J_2 had the following form

$$J_2 = \frac{w_{mus}}{16(t_1 - t_0)} \sum_{j=1}^{16} \int_{t_0}^{t_1} a_j(t)^2 dt. \tag{5}$$

The constraints of the optimal control problem referred to the dynamics of the skier model (3a) and lower and upper bounds of 0 and 1 on the muscle excitation patterns (3b), because muscle excitations are always limited to unity bound constraints (Erdemir et al. 2007) Additionally, the initial position of the hip $x_{hip}(0)$ was constraint (3c) to match the tracking data at the beginning of the simulations $x_{hip,\,data}(0)$. Otherwise, the jump height of the skier would be reduced in the simulation in favor of minimizing the sum of squared muscle activations (Heinrich et al. 2018).

The optimal control problem was solved numerically using the method of direct collocation and transformed into a constrained nonlinear programming (NLP) problem (van den Bogert et al. 2011). The NLP problem was solved using IPOPT 3.11.7 (Wächter and Biegler 2006), an interior point optimization solver. In the optimization, the state and control variables of the model were iteratively adjusted such the objective function is minimized and the imposed constraints are satisfied.

2.3. Knee loading

ACL tensile forces were calculated using a data-driven knee model (Kernozek and Ragan 2008; Laughlin et al. 2011; Weinhandl et al. 2014).

First, the resultant axial F_{ax} and anterior–posterior F_{ap} knee reaction forces were calculated from the simulated landing movement, gravity and the snow contact forces using a standard inverse dynamics analysis Winter (2009).

Second, the tibiofemoral contact force F_{tf} was computed from the resultant axial knee reaction force F_{ax} according to the following equation:

$$F_{ax} = F_{tf} \cos(\phi_{tf}) - F_q \cos(\phi_q) - F_h \cos(\phi_h) - F_g \cos(\phi_g) \tag{6}$$

where F_q, F_h and F_g denote the muscle forces of the quadriceps, hamstrings and gastrocnemius. The line of action of the quadriceps ϕ_q and hamstrings ϕ_h were defined as functions of knee flexion angle using the data of (Herzog and Read 1993). The line of action of the gastrocnemius ϕ_g was assumed to be

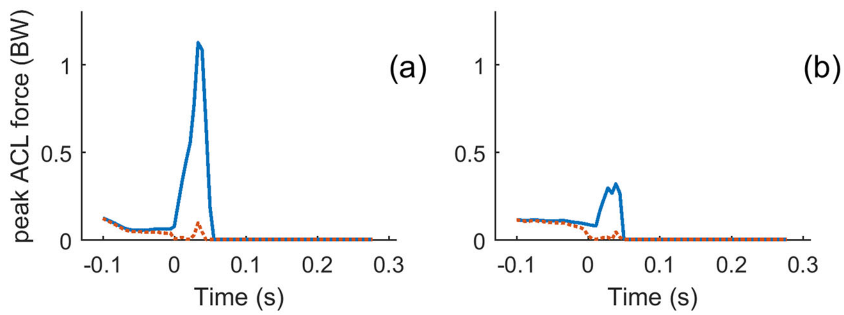

Figure 1. Peak ACL force in the right leg (a) and left leg (b) corresponding to the baseline simulation (solid lines) and the simulation with minimized ACL forces and the weighting factor w set to 1 (dashed lines). Time 0 denotes the initial ground contact.

parallel to the tibia throughout the range of motion of the knee (Kernozek and Ragan 2008; Laughlin et al. 2011). A tibial slope angle of $\phi_{tf} = 9^\circ$ was assumed (Matsuda et al. 1999).

Third, anterior–posterior ligamentous shear force F_{lig} was calculated from the resultant anterior–posterior knee reaction forces F_{ap}:

$$F_{ap} = F_{tf}\sin(\phi_{tf}) + F_q\sin(\phi_q) - F_h\sin(\phi_h) + F_g\sin(\phi_g) + F_{lig} \tag{7}$$

Finally, the ACL force F_{ACL} was calculated based on the anterior–posterior shear force F_{lig} and cadaveric knee measurements (Markolf et al. 1990, 1995):

$$F_{ACL} = \frac{F_{100}(\theta_k) - F_0(\theta_k)}{100} F_{lig} + F_0(\theta_k) \tag{8}$$

where F_{100} and F_0 are ACL forces when 100 N and 0 N of anterior tibial force were applied, respectively, at the knee flexion angle θ_k.

2.4. Landing simulation minimizing ACL forces

Following the baseline landing simulation, we aimed at computing a landing simulation with neuromuscular control patterns of the skier that result in minimized ACL forces. To this aim, we modified the muscle coordination strategy of the baseline landing simulation. Especially, the optimal control problem of the baseline simulation was reformulated adding a term J_{acl} with a weighting factor w to the objective function J_{ref}

$$J = J_{ref} + wJ_{acl} \tag{9}$$

with

$$J_{acl} = \frac{1}{2}\int_{t_c}^{t_c+\Delta t} \left(\frac{F^R_{ACL}(t)}{c}\right)^2 + \left(\frac{F^L_{ACL}(t)}{c}\right)^2 dt. \tag{10}$$

In J_{acl}, the integrand represented the sum of squared ACL forces in the right knee (F^R_{ACL}) and left knee (F^R_{ACL}), respectively, normalized by a constant c. We set c to 1000 N such that the terms in the objective function are of the same order. Since it was reported that peak ACL forces typically occur within the first 50 ms after initial ground contact (Krosshaug et al. 2007; Kernozek and Ragan 2008; Koga et al. 2010; Laughlin et al. 2011; Mokhtarzadeh et al. 2013), the sum of squared ACL forces was also restricted to the first $\Delta t = 50$ ms after the time of initial ground contact t_c. The weighting factor w was set to 1 after conducting a parameter study. In the parameter study, we increased the weighting factor w gradually and observed that for values higher than 1 ACL forces were not further decreased during the landing simulation. The detailed results of the parameter study are provided as Supplementary online material.

The reformulated optimal control problem was subjected to the same constraints (3a), (3b) and (3c) as in the baseline simulation. Additionally, we assumed a fourth constraint such that the initial posture of the skier was constraint to match the initial posture of the skier in the baseline simulation. As a consequence, the changes in ACL forces were caused by changes in the neuromuscular control patterns during the landing maneuver only and without altering the initial posture of the skier. The reformulated optimal control problem was solved numerically in analogy to the baseline landing simulation using the method of direct collocation.

2.5. Data analyses

For each landing simulation, the peak ACL force was computed based on the two-dimensional data-driven knee model. At the time of peak ACL force the joint angles, the trunk lean angle of the skier with respect to the vertical axis, the joint moments at the hip, knee and ankle as well as the muscle forces and neuromuscular activation patterns were determined. Positive moments referred to hip extension, knee extension and ankle plantarflexion moments.

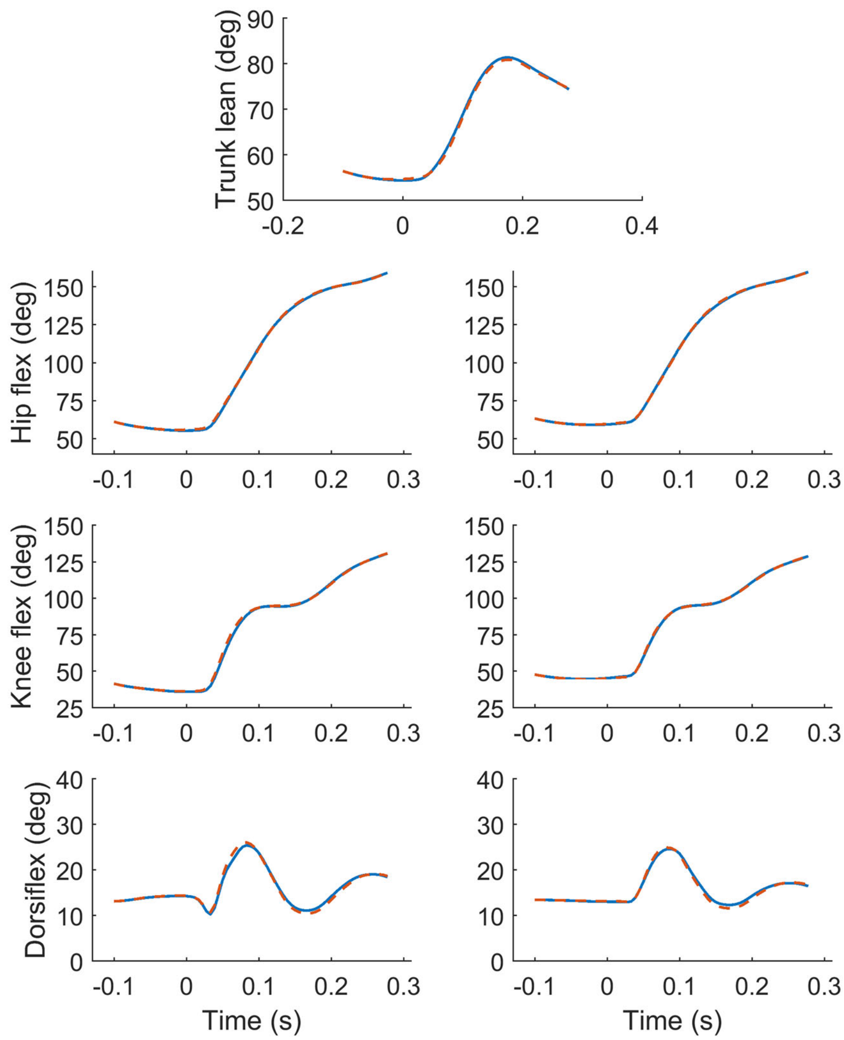

Figure 2. Trunk lean and joint angles in the right leg (first column) and left leg (second column) corresponding to the baseline simulation (solid lines) and the simulation with minimized ACL forces and the weighting factor *w* set to 1 (dashed lines). Time 0 denotes the initial ground contact.

3. Results

3.1. Baseline simulation

In the baseline simulation, the skier landed from a landing height of 1.29 m in a slightly asymmetric position. Comparing the baseline simulation to the experimental data, root mean square differences (RMSD) were in the range from 2.4 to 5.0° for the joint angles and trunk lean of the skier (Supplementary online material). These differences were close to the noise in the measurements determined by a residual analysis in the range from 2.5 to 5.0° (Winter 2009). Initial ground contact occurred at the right leg and peak normal ground reaction force was 6.92 BW 33 ms after initial contact. At the same time, peak ACL force amounted to 1.13 BW at the right knee; at the left knee, peak ACL was 0.32 BW and substantially lower than in the right knee (Figure 1).

At the time of peak ACL force in the right leg, the kinematics of the skier was characterized by a trunk lean angle of 54.9°, hip and knee flexion angles of 58.4 and 40.1° as well as 10.3° dorsiflexion at the ankle (Figure 2); in the left hip and knee flexion angles of 61.5° and 47.3° and ankle dorsiflexion of 13.1° were observed.

Joint moments at the hip, knee and ankle joint amounted to 0.90, 1.62 and −0.85 Nm/kg in the right leg and to 0.61, 0.62 and 0.51 Nm/kg in the left leg (Figure 3). Muscle activation patterns were characterized primarily by activation of the iliopsoas prior to ground contact as well as activation of the hamstrings, vasti, glutei and soleus during the landing phase after ground contract (Figure 4).

3.2. Minimizing ACL forces

In the landing simulation minimizing ACL forces, the peak forces in the ACL of the right and left knee decreased considerably from 1.13 BW to 0.13 BW

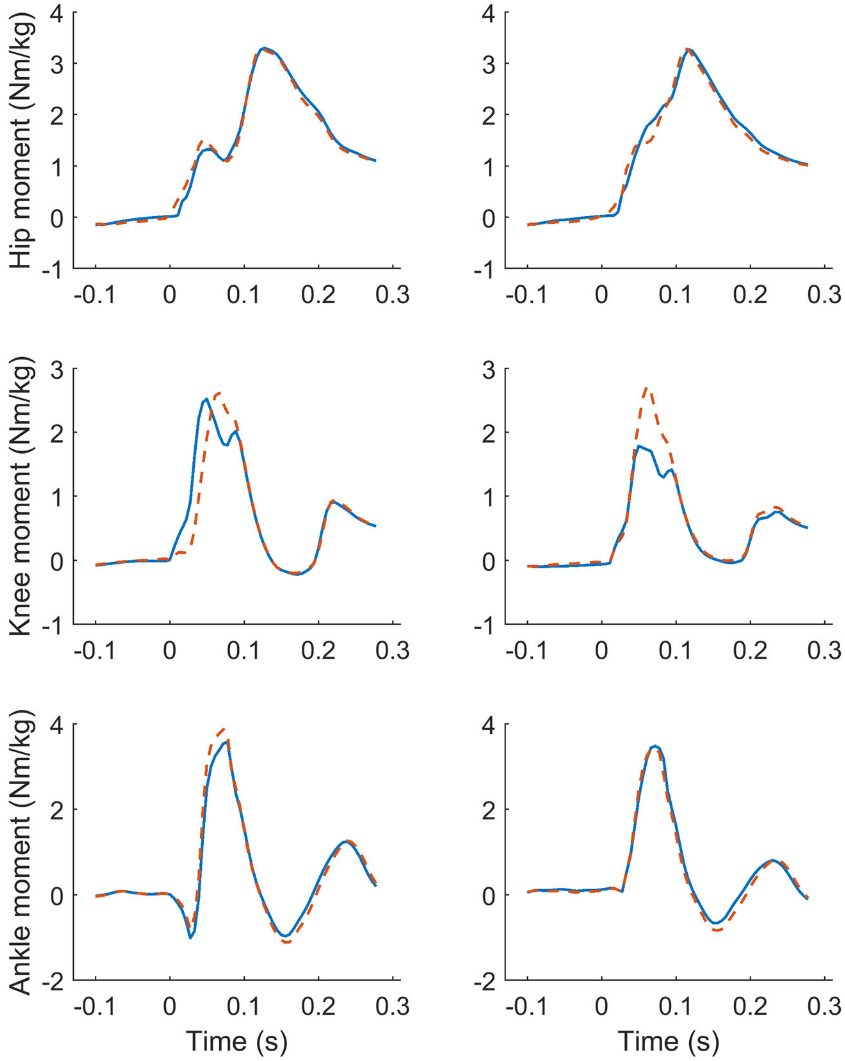

Figure 3. Joint moments in the right leg (first column) and left leg (second column) corresponding to the baseline simulation (solid lines) and the simulation with minimized ACL forces and the weighting factor *w* set to 1 (dashed lines). Time 0 denotes the initial ground contact. Positive values represent hip and knee extension and ankle plantarflexion moments.

and from 0.32 BW to 0.12 BW, respectively (Figure 1). Surprisingly, the reduction of the ACL forces was accompanied with only little changes in the kinematics. Specifically, the RMSDs were below 1° for all joint angles and trunk lean (Figure 2).

In the landing simulation with reduced peak ACL forces, the joint moments at the knee joints changed considerably while the joint moments at the hip and ankle joint remained similar (Figure 3). Especially, in the left knee, the moment rose and the peak moment was almost equal to the peak value in the right knee. Consequently, load was shifted to the left leg and well balanced between both legs.

Muscle activations also changed considerably and affected muscles crossing the knee as well as the hip and ankle joints in both legs (Figure 4). At the time of peak ACL force, the hamstrings and soleus were higher activated in the right leg; the vasti were lower activated and the peak activation was delayed compared to the baseline simulation. Prior to ground contact, the iliopsoas showed increased activation. In the left leg, the vasti were higher activated to counter the load shift from the right to the left leg. Similar to the right leg the hamstrings, soleus and iliopsoas were higher activated.

4. Discussion

The purpose of the present study was to develop a computational approach to predict neuromuscular control patterns resulting in minimized ligament forces during jump landing in downhill skiing. Our results showed that alterations of muscle activation patterns (amplitude and timing) of a subset

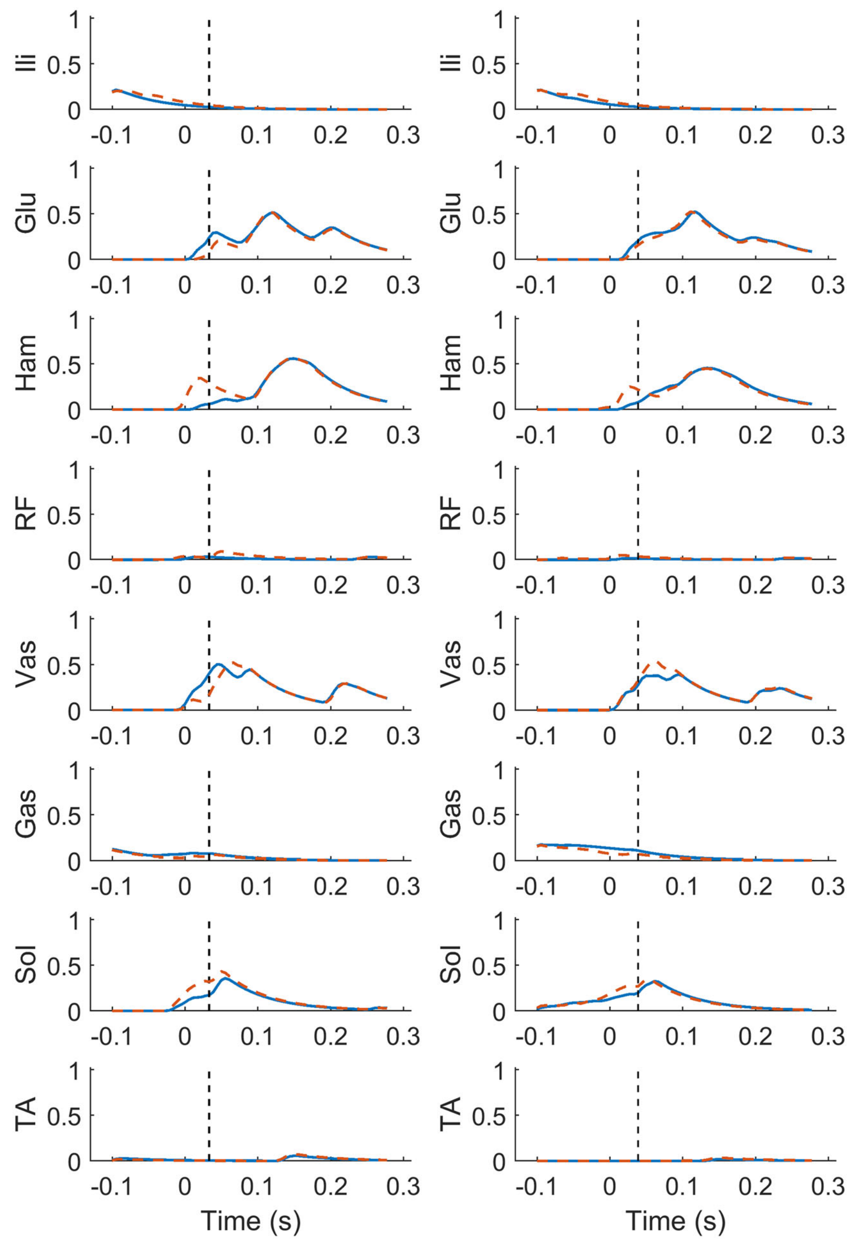

Figure 4. Muscle activation patterns in the right (first column) and left leg (right column) corresponding to the baseline simulation (solid lines) and the simulation with minimized ACL forces and the weighting factor *w* set to 1 (dashed lines). Time 0 denotes the initial ground contact and the vertical lines the time of peak ACL force in the baseline simulation.

of the muscles of both legs strongly affected the peak ACL force.

4.1. Baseline simulation

First, we computed a baseline simulation with a commonly assumed muscle coordination strategy minimizing the sum of squared muscle activation patterns during jump landing (Laughlin et al. 2011; Mokhtarzadeh et al. 2013, 2017). In the baseline simulation, the skier performed a jump landing maneuver from an equivalent landing height of 1.29 m. The resultant peak ACL force was 1.13 BW (880 N) and occurred after 33 ms of initial ground contact. Compared to simulation studies of single-leg landing maneuvers (Laughlin et al. 2011; Mokhtarzadeh et al. 2013) with peak ACL forces between 0.7 BW and 1.0 BW, respectively, the peak ACL force is higher in the present study. The difference might be explained by the higher landing height and the restricted plantarflexion at the ankle joint due to the ski boot. However, the peak force did not approach the failure load of 2167 N that has previously been suggested to cause ACL rupture (Woo et al. 1991). This was expected since the skier performed a safe jump landing maneuver. The timing of peak ACL force is consistent with previous simulation studies of landing maneuvers, where peak ACL forces were reported in the range of 1–40 ms (Pflum et al. 2004; Kernozek and Ragan 2008; Laughlin et al. 2011; Southard et al. 2012; Mokhtarzadeh et al. 2013); as well as video analysis of ACL injury cases, where the time of ACL injury was estimated within the first 50 ms (Krosshaug et al. 2007; Koga et al. 2010).

4.2. Minimizing peak ACL force

Following the baseline simulation, we used a novel approach to compute an optimized control strategy, aiming at minimizing the forces in the ACL during jump landing. Specifically, we reformulated the optimal control problem corresponding to the baseline simulation with an extended objective function that included the knee ligament forces (i.e. ACL forces). Solving the optimal control problem, the muscle excitation patterns were iteratively adjusted such that the ACL forces were additionally minimized during the impact phase of the landing movement where an ACL injury is most likely to occur (<50 ms). To our knowledge, the present method is the first computational approach to computing optimum neuromuscular control patterns with the aim to minimize ACL loading during jump landing. Previous landing simulation studies focused on the effects of movement interventions such as landing soft (Laughlin et al. 2011) or landing with increased knee and hip flexion (Southard et al. 2012) or the effect of leg dominance and landing height on ACL loading (Mokhtarzadeh et al. 2017).

Using the optimized control strategy the peak ACL force could be substantially reduced to 0.13 BW compared to 1.13 BW in the baseline simulation. The activations of the muscles attaching to the knee joint were adapted such that increased co-contraction was observed during the first 50 ms after initial ground contact and in particular at the time of peak ACL force. Especially, the hamstrings showed increased activation at both legs. The hamstrings act to counter anterior tibial translation and consequently induced a higher shear force at the knee joint in posterior direction, thus reducing ACL strain. This finding is in agreement with earlier studies to suggest that insufficient or slow hamstring reaction may result in inadequate knee stabilization during sporting tasks involving large external loads and increased risk of ACL tear (Blackburn et al. 2004; Saunders et al. 2014; Weinhandl et al. 2014). The activation of the quadriceps was carefully timed such that the knee joint moments were symmetric and consequently the impact load was uniformly balanced between both legs. These results are consistent with double versus single-leg landing studies from the same height, where lower peak ACL force was reported during double-leg landing (Kernozek and Ragan 2008; Mokhtarzadeh et al. 2013). In addition, reduced activation during the early impact phase and in particular the time of peak ACL force was observed, which is also in line with the state of knowledge that high quadriceps activation during eccentric contraction in combination with a small knee flexion angle is a major factor contributing to ACL injury (e.g. Griffin et al. 2000; Yu and Garrett 2007; Hewett et al. 2010).

Also, muscles crossing the ankle and hip joint showed altered activation patterns. At the ankle joint, in particular, the soleus was higher activated. Increased activation of the soleus as favorable activation pattern is consistent with the results of Mokhtarzadeh et al. (2013), who proposed a protective role of the soleus in ACL loading. At the hip joint, increased activation of the iliopsoas prior to ground contact was also observed. The increased activation of the iliopsoas might have been beneficial for the activation of the hamstrings during the impact phase. Since the iliopsoas acted as monoarticular hip flexor increased activation led to accelerated hip flexion and a slight forward rotation of the trunk prior to landing. This was compensated by increased activation of the hamstrings, which decelerated hip flexion and applied a higher shear force at the knee joint in posterior direction, thus reducing ACL strain.

In the simulation with decreased peak ACL force, the kinematics remained similar compared to the baseline simulation (Figure 3). Consequently, the reduction in peak ACL force was primarily caused by adaptation of the muscle activation patterns and was accompanied with only small changes in the movement of the skier. This result highlights that the muscle activation patterns are a key factor affecting the loading of the ACL and that athletes with almost the same kinematics might be exposed to either high or low ACL forces during the jump landing movement depending on the chosen control strategy. The inclusion of exercise regimes in ACL injury prevention to improve body position during landing was proposed by Shimokochi et al. (2013), who studied the influence of changing the sagittal plane body position during single-leg landings. Based on the present results, an additional focus on the control strategy during jump landing is suggested. In particular, increased activation of the monoarticular hip flexors prior to jump landing might be beneficial followed by increased activation of the hamstrings during the early impact phase. Corresponding neuromuscular training regimes might include plyometric, jump and balance training (Chimera et al. 2004; Nagano et al. 2011) as well as perturbation drills, which have shown to enhance hamstrings activity (Hurd et al. 2006; Letafatkar et al. 2015).

4.3. Limitations

Some limitations of the present study have to be mentioned. First, jump landing simulations and ACL forces were computed based on a two-dimensional, sagittal plane musculoskeletal model of an alpine skier with two skis. Consequently, knee joint loading due to valgus or internal and external rotations are not considered in the present study, which is known to affect knee ligament loading (Markolf et al. 1995; Weinhandl et al. 2014). However, analyzing injury cases during jump landing in World Cup downhill skiing (Bere et al. 2011), the sagittal plane was reported to be most important. Three-dimensional musculoskeletal models could allow for the role of frontal-plane and transverse-plane hip and knee movement and neuromuscular control strategies to be evaluated, which might be important during dynamic movement tasks with a change in direction such as sidestep cutting maneuvers (Weinhandl et al. 2014).

Second, the results of the present study are based on a single downhill skier performing a jump landing maneuver in competitive downhill skiing. Unfortunately, kinematic data of high-risk or injury-prone situations are rare especially during competitions. Recently, however, Eberle et al. (2019) proposed a computational approach to generate non-contact ACL injury-prone situations on a computer using kinematic data of non-injury situations and Monte Carlo simulation. Based on this approach injury-prone situations might be generated. With respect to these situations optimized neuromuscular control strategies aiming at low ACL forces might be investigated following the methodology in the present study. These results might provide further insight into the mechanisms, risk factors and preventive measures for ACL injuries.

4.4. Conclusion and implications

In conclusion, a computational approach was developed to compare control strategies and predict muscle activation patterns that minimize ACL forces during jump landing in downhill skiing. Using the optimized control strategy the peak ACL force could be substantially reduced from 1.13 BW in the baseline simulation to 0.13 BW. The reduction was primarily caused by altered muscle activation patterns (amplitude and timing) by a subset of the muscles of both legs and accompanied with only a small change in the kinematics of the skier (RMSD $< 1°$). The proposed computational approach might also be applied to determine optimized neuromuscular control patterns with respect to other injury-prone situations (e.g. sidestep cutting, pivoting or stop-jump maneuvers) or other movement tasks such as walking, running or stair climbing, which might be important during rehabilitation after ACL injury.

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Disclosure statement

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