SIMULATION OF LANDING MOVEMENT IN DOWNHILL SKIING USING A PLANAR MUSCULOSKELETAL MODEL

1Dieter Heinrich, 2Antonie J. van den Bogert and 1Werner Nachbauer
1Department of Sport Science, University of Innsbruck, Innsbruck, Austria; email: dieter.heinrich@uibk.ac.at
2Orchard Kinetics LLC, Cleveland, USA

INTRODUCTION
Musculoskeletal models have many applications such as predictive simulations of gait and running, planning of surgical treatments or simulations of sports injuries.

In alpine skiing the high incidence of anterior cruciate ligament (ACL) injuries is well reported. These injuries are typically associated with a twisting fall or during a landing movement from a jump in downhill skiing [1]. In the latter case the skier is typically observed in a backward off-balance situation during the flight and after ground contact and tries to regain balance by a forceful contraction of the quadriceps muscles. Only few studies such as [2] investigated this particular injury situation and further studies are necessary for a better understanding and to analyze possible risk factors. The purpose of this paper was to (a) develop a musculoskeletal model of an alpine skier capable of simulating a landing movement in downhill skiing and (b) to investigate modifications of equipment on the load of the anterior cruciate ligament in a possible injury situation.

METHODS
A planar, nine degree of freedom model of an alpine skier was developed. The skier model consisted of seven segments, three segments for each leg (thigh, shank and boot) and one segment representing the head, arms and torso. The ski boots were rigidly mounted on the skis. The ski boot was represented as passive joint moment acting at the ankle joint. Contact between each ski and snow was modeled by three contact elements located at the tail, middle and tip of the ski. The force normal to the snow surface was modeled as: find trajectories \( \mathbf{x}(t) \) and \( \mathbf{u}(t) \) that minimize a cost function \( J \) subject to constraints due to system dynamics. System dynamics was given in implicit form:

\[
\mathbf{f}(\mathbf{x}, \mathbf{x}, \mathbf{u}) = 0
\]

The cost function for the optimal control problem was:

\[
J = \frac{1}{N_{\text{dof}}} \int_0^T \left[ \sum_{i=1}^{N_{\text{dof}}} \left( q_i(t) - q_{i, \text{data}}(t) \right)^2 / \mathbf{S}_d \right] + \frac{w_{\text{effort}}}{N_{\text{mus}}} \int_0^T \sum_{i=1}^{N_{\text{mus}}} \mu_i(t)^2
\]

The first term represents the deviation of the simulation with respect to experimental data scaled by a factor \( \mathbf{S}_d \). Here we used kinematic data from a competitive skier, collected during the 1994 Winter Olympics. The same data was also used in [2]. The second term, weighted with \( w_{\text{effort}} \), encourages the simulation to track the measured kinematics with low muscle excitation, penalizing co-contraction and high spikes.

The optimal control problem was transformed into a constrained Nonlinear Programming Problem (NLP) using direct collocation [3]. Unknowns were the states and controls at each node, \( 66N \) unknowns in total for \( N \) time nodes. System dynamics was discretized using the implicit midpoint formula resulting in \( 50N \) nonlinear constraints. The NLP was solved using IPOPT, an interior point optimization solver, and a mesh refinement strategy. Solving the NLP first on a coarse mesh \((N=6)\), the number of time nodes was gradually increased until the final mesh with \( N = 180 \) nodes. The weight factor \( w_{\text{effort}} = 10 \) was found to give good results. Accuracy of results was assessed by comparing the results obtained with the mesh refinement strategy.

Initial states and controls obtained by solving the tracking problem were modified and used as input data to a forward dynamics simulation to create a possible injury situation. The initial orientation of the trunk was perturbed counterclockwise to induce a slightly off-balance position followed by a recovery movement by maximally activating the quadriceps muscles and the m. iliopsoas. It was assumed that the hamstrings did not contribute to this recovery.

Knee ligament forces (ACL respectively PCL) were calculated incorporating the resultant knee joint forces projected onto the tibial plateau, the forces in the quadriceps and hamstring muscles, and the orientation of the knee ligament forces, hamstrings and patella tendon with respect to the tibial plateau [4]. A posterior tibial plateau slope \( \beta = 9^\circ \) was assumed. As application, ski stiffness was changed to investigate modifications of equipment on the load of the ACL.
RESULTS AND DISCUSSION
The solution of the optimal control problem aiming at tracking kinematic data of an alpine skier at 180 time nodes during a landing movement was obtained by a mesh refinement strategy. Starting with 5 intervals (6 time nodes) the number of intervals was gradually increased by a factor 1.21 until the final mesh with 180 time nodes (objective function $J = 0.676 \text{ (fit)} + 0.298 \text{ (effort)}$ at the final mesh with $N = 180$ nodes). Most of the simulation time was spent in the last two iterations corresponding to the highest number of nodes. Predicted optimal muscle stimulation patterns were almost indistinguishable at the 100-node and 180-node solution.

Figure 1 shows a stick diagram of a simulated possible injury situation. The initial orientation of the trunk segment was perturbed counterclockwise by an angle $\phi = 8^\circ$ with respect to the optimized solution found by solving the tracking problem. In the simulation the skier contacted the ground at $t = 0.5$ s and made a recovery motion movement by maximally activating the quadriceps muscles and the m. iliopsoas.

Peak ACL forces were encountered during the recovery motion at $t = 0.81$ s and were substantially higher compared to the tracked non-injury situation. Peak forces were 1907 N in the right ACL (Figure 2) and 1724 N in the left ACL. Since the ACL typically tears at loads greater than 2000 N [5] this situation can be considered as a close to injury situation. At these peaks approximately 62% of the load of the ACL was generated by muscles forces (quadriceps and hamstrings) and approximately 38% by external forces.

Simulations with modified ski stiffness showed that increased ski stiffness resulted into higher peak ACL forces during the recovery motion. Setting the stiffness parameter $k$ at the ski ends (tail and tip) to 2000 N/m, 3000 N/m, 3500 N/m and 4000 N/m, peak ACL forces in the right knee amounted to 1819 N, 1983 N, 2047 N and 2096 N, respectively. During simulations with a slightly increased perturbation of the initial trunk orientation ($\phi = 9^\circ$) even a higher peak ACL force was observed in the right knee (2215 N) revealing that the peak ACL forces during the recovery attempt of the simulation are more sensitive to a perturbed initial landing position than a modification in ski stiffness.

CONCLUSIONS
A planar musculoskeletal model of an alpine skier was applied to simulate a non-injury landing movement in downhill skiing. Possible injury situations with peak ACL forces up to 2215 N were investigated where the skier was in a backward off-balance situation during the flight and after ground contact and tried to regain balance by a forceful contraction of the quadriceps muscles. Lower ski stiffness, as an example of modified equipment, showed to decrease injury risk by decreasing peak ACL forces in the knee. Based on simulations with perturbed initial trunk orientation it is concluded that a proper preparation of the skier before a jump is important to avoid a marked off-balance situation that in combination with a recovery motion might take the ACL at risk.

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REFERENCES