THE EFFECTS OF SINGLE-LEG LANDING TECHNIQUE ON ACL LOADING

[1]Walter Laughlin, [1]Joshua Weinhandl, [2]Thomas Kernozek, and [1]Kristian O'Connor

[1]University of Wisconsin-Milwaukee, Milwaukee, WI, USA
[2]University of Wisconsin-LaCrosse, LaCrosse, WI, USA
email: laughlin@uwm.edu, Web: www.chs.uwm.edu/neuromechanics

INTRODUCTION

Anterior cruciate ligament (ACL) injury is one of the most debilitating and costly lower extremity injuries experienced by athletes [1]. It has been shown that over 70% of ACL ruptures occur in a non-contact situation [2], specifically during closed chain movements requiring rapid decelerations of the body's center of mass [2] such as; cutting, pivoting, stopping, or landing.

In an attempt to reduce ACL injury rates during landing tasks, neuromuscular and proprioceptive training programs have been developed to reduce an athlete's risk of injury. Decreases in ACL injury rates associated with these intervention programs have been attributed to a number of biomechanical changes during landing. In the sagittal plane, trained athletes show an increase in knee flexion angle at initial contact (IC) and throughout the range of motion of the knee [3]. This altered lower extremity position has been suggested to decrease ACL injury risk. However, no study has compared the change in ACL force that occurs as a result of sagittal plane knee mechanics. Therefore, the purpose of this study was to determine the effects of lower limb landing technique on ACL forces during single-leg drop landings.

METHODS

Eight physically active females (average mass = 63 kg), free from musculoskeletal injury volunteered to perform soft and stiff landings from a 37 cm box. The order in which these trials were performed was counterbalanced between subjects.

Three-dimensional kinematic data were collected using a ten-camera Motion Analysis Eagle system (200 Hz), and force data were collected with an AMTI force platform (1000 Hz). Surface electromyography (EMG) of the medial and lateral hamstrings and quadriceps were recorded using at 1000 Hz. Kinematic and kinetic data were used as inputs into a subject specific three-dimensional musculoskeletal model (Figure 1) [4,5,6]. The model then used a computed muscle control (CMC) algorithm to calculate muscle forces. These outputs were then used in a sagittal plane model of the knee [7] to calculate ACL force.

Dependent t-tests were assessed to determine potential differences in ACL force between the two landing conditions. Other variables of interest included hip and knee kinematics in the sagittal plane at IC as well as 200 ms post IC.

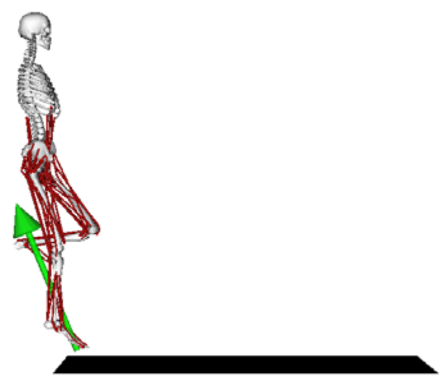

Figure 1. 19 degree-of-freedom, 92 actuator musculoskeletal model.

RESULTS AND DISCUSSION

The hip and knee kinematics (Figure 2) at IC and 200 ms post IC were different between the two landing conditions (Table 1). There was also a difference in the peak ACL force between the two landing conditions (Table 1). The stiff landing resulted in a 23% greater peak ACL force. In both landing conditions the peak ACL force occurred at ~10 to 20 ms post IC and dissipated to zero by 60 ms post IC (Figure 2).

Instructing subjects to land in a “soft” or “stiff” manner led to expected differences in maximum flexion angles and initial contact angle. The timing of the peak ACL force supports the theory that body position at contact is a critical determinant of ACL loading. This result suggests that a simple instructional cue may substantially decrease ACL loading. While the absolute forces were substantially lower than those believed to rupture the ACL (stiff = ~600 N), this estimation of ACL loading only accounts for sagittal plane mechanisms and does not account for other contributors to tension on the ACL. Reducing the sagittal plane contribution may serve to protect the ACL from excessive out of plane loading and reduce injury risk.

The individual contributors to the calculation of the ACL force were the shear components of the net joint force, tibiofemoral contact force, and the patellar tendon and hamstrings forces. The loading difference at the instant of peak ACL force was primarily due to a decreased posterior shear contribution by the hamstrings and secondarily due to an increased anterior shear by the tibiofemoral contact force. An increased shortening velocity of the hamstrings immediately after contact depressed the hamstrings force contribution.

Table 1. Hip and knee kinematics, and peak ACL forces.

		Soft	Stiff	p-value
Hip (°)	IC	29.1 ± 9.1	19.8 ± 8.8	0.003*
	200ms	65.1 ± 7.4	33.8 ± 15.6	<0.001*
Knee (°)	IC	16.5 ± 5.5	10.0 ± 3.4	0.007*
	200ms	81.7 ± 8.4	47.9 ± 19.1	0.001*
Peak ACL Force (BW)		0.80 ± 0.40	0.97 ± 0.47	0.046*

* Significantly different from soft ($p<0.05$)

CONCLUSIONS

The results of this study suggest that, when instructed to land in a soft manner, females show different single-leg landing mechanics at the knee and the hip as well as a decrease in the peak ACL force. Although the estimates of ACL forces did not reach magnitudes necessary to rupture the ligament, these findings support current research which suggests that changes in sagittal plane knee mechanics during landings may reduce an athlete’s risk of ACL injury.

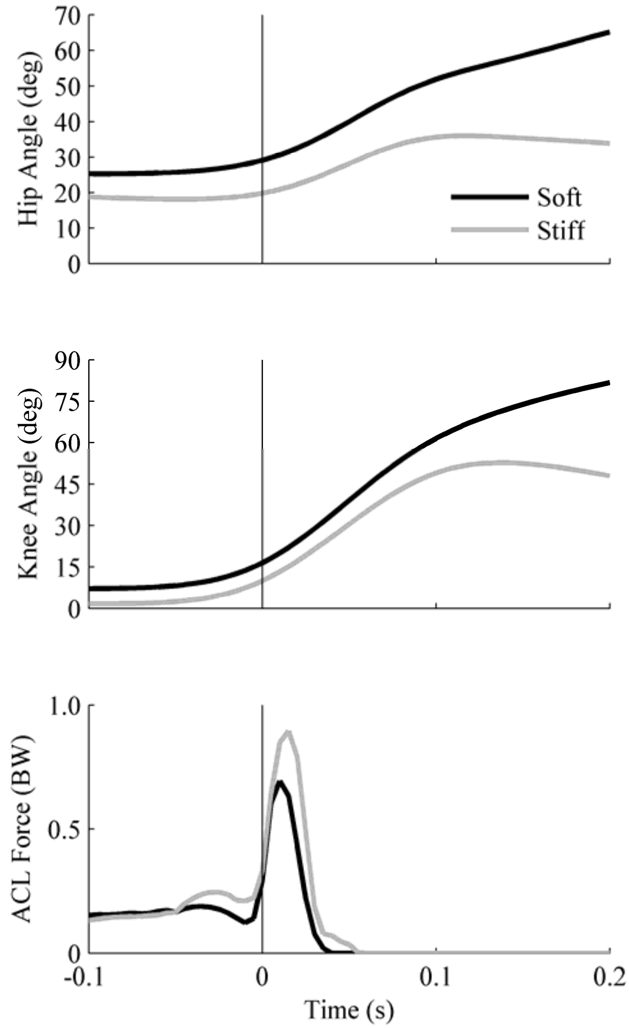

Figure 2. Hip and knee kinematics, and ACL forces. Zero time represents initial ground contact.

REFERENCES

1. Hewett, T. E. et al. *Am.J.Sports Med.*, **34,** 299-311, 2006.
2. Olsen, O. E. et al. *Am.J.Sports Med.*, **32**, 1002-12, 2004.
3. Mandlebaum, B.R. et al. *Am.J.Sports Med.*, **33**, 1003-10, 2005.
4. Thelen, D.G. & Anderson, F.C. *J Biomech*, **39**, 1107-15, 2006.
5. Delp, S.L. et al. *IEEE Trans.Biomed.Eng*, **37**, 757-67, 1990.
6. Thelen, D.G. et al. *J. Biomech*, **36**, 321-28, 2003.
7. Kernozek, T. W. & Ragan, R. J. *Clin Biomech*, **23**, 1279-86, 2008.

ACKNOWLEDGEMENTS

UWM College of Health Sciences Student Research Grant