Published in final edited form as:
Clin Biomech (Bristol, Avon). 2010 February ; 25(2): 142. doi:10.1016/j.clinbiomech.2009.10.005.

LIMITED HIP AND KNEE FLEXION DURING LANDING IS ASSOCIATED WITH INCREASED FRONTAL PLANE KNEE MOTION AND MOMENTS

Christine D. Pollard, PhD, PT[1], **Susan M. Sigward, PhD, PT, ATC**[1], and **Christopher M. Powers, PhD, PT**[1]

[1] Jacquelin Perry Musculoskeletal Biomechanics Research Laboratory, University of Southern California, Los Angeles, CA, USA

Abstract

Background—It has been proposed that female athletes who limit knee and hip flexion during athletic tasks rely more on the passive restraints in the frontal plane to deceleration their body center of mass. This biomechanical pattern is thought to increase the risk for anterior cruciate ligament injury. To date, the relationship between sagittal plane kinematics and frontal plane knee motion and moments has not been explored.

Methods—Subjects consisted of fifty-eight female club soccer players (age range: 11 to 20 years) with no history of knee injury. Kinematics, ground reaction forces, and surface electromyography were collected while each subject performed a drop landing task. Subjects were divided into two groups based on combined sagittal plane knee and hip flexion angles during the deceleration phase of landing (high flexion and low flexion).

Findings—Subjects in the low flexion group demonstrated increased knee valgus angles ($P = 0.02$, effect size 0.27), increased knee adductor moments ($P = 0.03$, effect size 0.24), decreased energy absorption at the knee and hip ($P = 0.02$, effect size 0.25; and $P < 0.001$, effect size 0.59), and increased vastus lateralis EMG when compared to subjects in the high flexion group ($P = 0.005$, effect size 0.35).

Interpretation—Female athletes with limited sagittal plane motion during landing exhibit a biomechanical profile that may put these individuals at greater risk for anterior cruciate ligament injury.

Keywords

ACL; Injury Prevention; Joint Moments; Stiff Landing

INTRODUCTION

Females are 4 to 6 times more likely to tear their anterior cruciate ligament (ACL) than their male counterparts (Arendt and Dick, 1995). Over the past decade, considerable attention has focused on understanding why females incur this disproportionate number of ACL injuries. What is particularly perplexing is the fact that the majority of ACL tears are non-contact

Address correspondence to: Christine D. Pollard, PhD, PT, Division of Biokinesiology and Physical Therapy, University of Southern California, 1540 East Alcazar Street, CHP-155, Los Angeles, California, 90033, USA, Phone:(323) 442-2454, Fax: (323) 442-1515, cpollard@usc.edu.

(Boden et al., 2000); suggesting that females are interacting with their environment in a way that may be putting them at greater risk for injury.

In an effort to better understand and prevent ACL injuries, numerous studies have examined gender differences in lower extremity mechanics during athletic tasks. Such investigations have consistently reported that lower extremity mechanics differ between males and females. In particular, females have been shown to perform athletic maneuvers with decreased knee flexion (Malinzak et al., 2001; Lephart et al., 2002), hip flexion (Salci et al., 2004; Pollard et al., 2007), increased quadriceps activation (Malinzak et al., 2001; Sigward and Powers, 2006), increased knee valgus angles (McLean et al., 1999; Malinzak et al., 2001; Ford et al., 2003; Pappas et al., 2007; Kernozek et al, 2008), and increased internal knee adductor moments (McLean el al., 2005; Sigward and Powers, 2006), when compared to males. Taken together, this biomechanical profile is thought to put females at an increased risk for ACL injury. For example, it has been demonstrated that contraction of the quadriceps muscle at relatively small knee flexion angles strains the ACL (Renstrom et al., 1986). Furthermore, Hewett et al. (2005) have reported that increased internal knee adductor moments predict ACL injury.

Although females exhibit sagittal and frontal plane mechanics that are thought to contribute to ACL injury, the underlying reasons for this movement pattern are not known. It has been hypothesized that because of poor strength of the sagittal plane musculature, females limit the amount of knee and hip flexion during dynamic tasks, and instead, rely more on their passive restraints in the frontal plane (i.e. ligaments) to decelerate the body center of mass. Increased frontal plane loading at the knee is of particular concern as knee valgus angles and adductor moments have been found to be predictive of ACL injury (Hewett et al., 2006). Hewett et al. (2002) has describes this movement strategy as “ligament dominance.” Despite the fact that this theory suggests a relationship exists between sagittal plane kinematics and frontal plane loading at the knee, no study has confirmed this association. For example, do female athletes who demonstrate decreased knee and hip flexion during landing, often referred to as a “stiff” landing pattern, also exhibit higher frontal plane moments at the knee?

The purpose of this study was two-fold. First, we examined whether individuals who demonstrate a low flexion (i.e. low knee and hip flexion) landing pattern exhibit increased frontal plane moments and angles at the knee compared to those who exhibit a high flexion (i.e. high knee and hip flexion) landing pattern. Second, we examined the biomechanical and neuromuscular characteristics of both groups (i.e. sagittal and frontal plane kinematics, kinetics at the hip and knee, as well as quadriceps and hamstrings muscle activation patterns). By examining differences in lower extremity mechanics between individuals that demonstrate different landing strategies (i.e. low flexion versus high flexion) we can better understand the relevance of the low flexion landing pattern to ACL injury; moreover, we will gain insight into the potential cause for these varied movement strategies.

METHODS

Subjects

Subjects consisted of 58 healthy female club soccer players ranging in age from 11 to 20 years (average age 13.5 yrs). Individuals were excluded from the study if they reported any of the following: 1) history of previous ACL injury or repair 2) previous injury that resulted in ligamentous laxity at the ankle, hip or knee or, 3) presence of any medical or neurological condition that would impair their ability to perform a landing task; or, 4) previous participation in an ACL injury prevention training program.

Procedures and Instrumentation

Testing took place at the Musculoskeletal Biomechanics Research Laboratory at the University of Southern California. All procedures were explained to each subject and informed consent was obtained as approved by the Institutional Review Board for the University of Southern California Health Sciences Campus.

Kinematic data were collected using an eight-camera, three dimensional motion analysis system (Vicon, Oxford Metrics LTD, Oxford, England) at a sampling frequency of 250 Hz. The cameras were interfaced to a microcomputer and placed around a force plate embedded within the floor (Advanced Mechanical Technologies, Inc., Newton, MA, USA). The force plate (1500 Hz) was interfaced to the same microcomputer that was used for kinematic data collection via an analog to digital converter, allowing for synchronization of kinematic and kinetic data.

Reflective markers (14 mm spheres) were placed bilaterally over the following anatomical landmarks: the 1st and 5th metatarsal heads, medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanters, iliac crests, and a single marker on the joint space between the fifth lumbar and the first sacral spinous processes. In addition, triads of rigid reflective tracking markers were securely placed bilaterally on the lateral surfaces of the subject's thigh, leg and heel counter of the shoe. To control for the potential influence of varying footwear, subjects were fitted with same style of cross-training shoe (New Balance Inc., Boston, MA, USA).

Electromyographic (EMG) activity of the medial hamstrings, lateral hamstrings and the vastus lateralis of the dominant lower extremity (the lower extremity used to kick a soccer ball) were recorded at 1500 Hz using the same analog to digital converter used for the force plate. Pre-amplified bipolar, grounded, surface electrodes (Motion Control, Salt Lake City, UT, USA) were placed over the muscles in a manner similar to that described by Basmajian and Deluca (1985). Electrodes were connected to an EMG receiver unit, which was carried in a small pack on the subjects' back. EMG data were normalized to the highest magnitude of EMG activity acquired during a maximum voluntary isometric contraction (MVIC). The MVIC test for the quadriceps was performed with the subject pushing against a fixed resistance in the seated position (hip and knee flexed to 90° and 60°, respectively). The MVIC for the medial and lateral hamstrings was performed in supine with the hip and knee flexed to 30°. A strap was secured to the table and placed around the subject's hips which allowed them to perform a single leg bridge (i.e. resisting hip extension with the knee flexed) against a fixed resistance. Subjects performed one MVIC for each muscle group tested, and each contraction was held for six seconds.

Each subject performed 3 trials of the drop landing task. Subjects began this maneuver from a standing position on a 36 cm platform. They were instructed to step off the platform (leading with their dominant limb) and land with their right foot on one force plate and their left foot on an adjacent force plate and then proceed to immediately jump as high as they could after landing. Subjects were not given any verbal cues on landing or jumping technique. Practice trials were allowed for the subjects to become familiar with the procedures and instrumentation.

Data Analysis

Coordinate data were digitized in Vicon Workstation software (Workstation, Oxford Metrics LTD, Oxford, England). Consistent with previous studies evaluating dynamic tasks, kinematic data were filtered using a fourth-order zero-lag Butterworth 12-Hz low-pass filter.(Ford et al., 2007; Schmitz et al., 2007) Visual3D™ software (C-Motion, Inc., Rockville, MD, USA) was used to quantify three dimensional knee and hip kinematics. Joint kinematics were calculated using a joint coordinate system approach (Grood and Suntay, 1983) and were reported relative to a static standing trial.

Kinematics, ground reaction forces and anthropometrics (Dempster, 1955) were used to calculate joint moments at the knee and hip using inverse dynamics equations in Visual 3D™ software (Bresler and Frankel, 1950). Kinetic data were normalized to body mass. The joint moments referred to in this investigation are the internal resultant moments. Sagittal plane joint power was computed for the hip and knee as the scalar product of angular velocity and net joint moment. Total energy absorbed at the hip and knee was calculated by integrating the respective power-time curves during the deceleration phase of landing.

EMG signals were transmitted to an analog to digital converter using an 8-channel hardwired EMG unit. Differential amplifiers were used to reject the common noise and amplify the remaining signal (gain = 2000). EMG signals were then band pass filtered (20–500Hz) and a 60Hz notch filter was applied. The data were subsequently full wave rectified and a moving average smoothing algorithm (75-millisecond window) was used to generate a linear envelope. EMG signals were expressed as a percentage of EMG obtained during the MVIC. EMG signal processing and smoothing were performed using Motion Lab Systems software (Motion Lab Systems, Baton Rouge, LA, USA).

The landing cycle was identified as the period from foot contact to toe-off, as determined by the force plate recordings. For the purposes of this study, only the deceleration phase of the drop land task was examined as this timeframe corresponds to the period of greatest knee loading (Boden et al., 2000). The deceleration phase was defined as foot contact to peak knee flexion. Variables of interest during the deceleration phase of landing included peak knee flexion angle, peak hip flexion angle, peak knee valgus angle, average knee adductor moment, average knee extensor moment, average hip extensor moment, energy absorbed at the knee and hip, and average EMG for the vastus lateralis as well as the medial and lateral hamstrings of the dominant limb.

To further explore the relationship between the average knee extensor and hip extensor moments, we examined the knee to hip extensor moment ratio (knee to hip extensor moment ratio = average knee extensor moment/average hip extensor moment). Using this ratio, a value greater than “1” would indicate increased knee extensor moments compared to hip extensor moments while a value of less than “1” would indicate increased hip extensor moments compared to knee extensor moments. A similar ratio was used to explore the relationship between the energy absorbed at the knee and hip (knee to hip energy absorption ratio = knee energy absorption/hip energy absorption).

Statistical Analysis

To compare lower extremity mechanics between subjects who utilized a high flexion landing versus a low flexion landing pattern, subjects were divided into groups based on their combined peak knee and hip flexion angles during the landing task. Subjects who exhibited a combined hip and knee flexion value above the group average (170°) were assigned to the high flexion landing group (N=27) and those who exhibited a value below the group average were assigned to the low flexion landing group (N=31).

Differences between groups were evaluated using independent sample t-tests. Statistical analyses were performed using SPSS statistical software (Chicago, IL, USA) and a significance level of $P \leq 0.05$.

RESULTS

Means, standard deviations, P-values and effects sizes for the dependent variables are reported in Table 1. On average, the combined knee and hip flexion angles for subjects in the low flexion landing group was 154° (SD: 13°). The average combined knee and hip flexion angles for the subjects assigned to the high flexion landing group was 190° (SD: 16°). Subjects in the low flexion landing group performed the land in 14° less knee flexion and 23° less hip flexion.

With respect to the frontal plane variables of interest, subjects in the low flexion landing group demonstrated average knee adductor moments that were 2.2 times greater than subjects in the high flexion landing group. In addition, when compared to subjects in the high flexion landing group subjects in the low flexion landing group exhibited greater peak knee valgus angles.

In regards to the sagittal plane, significant group differences were identified at both the knee and hip. When compared to subjects who exhibited a high flexion landing pattern, individuals in the low flexion landing group exhibited a 10% increase in average knee extensor moments and a 23% decrease in average hip extensor moments. Knee to hip extensor moment ratio was 66% greater in the low flexion landing group when compared to the high flexion landing group. (Figure 1) In regard to energy absorption, subjects in the low flexion landing group exhibited decreased energy absorbed at the knee and hip of 9% and 30%, respectively. Knee to hip energy absorption ratio was 52% greater low flexion landing group when compared to the high flexion landing group. (Figure 2) Finally, individuals in the low flexion landing group exhibited a 35% increase in average vastus lateralis muscle activation; however, no group differences were observed in average medial or lateral hamstring activation.

DISCUSSION

The results of our study demonstrate that individuals who utilize a low flexion landing pattern demonstrate greater frontal plane loading at the knee during a drop landing task. This is illustrated by the fact that females who utilized a low flexion landing pattern exhibited greater knee valgus angles and knee adductor moments. Our findings support the theory that females who limit motion in the sagittal plane employ a strategy of reliance on passive restraints in the frontal plane to control the deceleration of the body center of mass.

Apart from differences in the frontal plane, females in the low flexion landing group exhibited distinct differences in the sagittal plane. More specifically, females who utilized a low flexion landing pattern exhibited decreased energy absorption at the knee and hip, increased knee extensor moments, increased vastus lateralis muscle activation, and decreased hip extensor moments. In contrast, individuals who utilized a high flexion landing pattern demonstrated the opposite biomechanical pattern (i.e. increased energy absorbed at the knee and hip, decreased knee extensor moments, decreased vastus lateralis muscle activation, and increased hip extensor moments). Taken together, the sagittal plane profile exhibited by the subjects in the low flexion group is suggestive of a strategy that emphasizes use of the knee extensors over the hip extensors to attenuate impact forces. This is evident by the fact that these individuals exhibited a high knee to hip extensor moment ratio (2.5) as well as a high knee to hip energy absorption ratio (3.5). In contrast, females in the high flexion group attenuated impact forces through a more equal utilization of the knee and hip extensors (i.e. knee to hip extensor moment ratio of 1.5; knee to hip energy absorption ratio of 2.3).

Previous work by Devita and Skelly (1992) examining sagittal plane lower extremity joint energetics has shown that a soft landing strategy requires increased use of the hip musculature. These authors identified the critical role of the hip extensor moment in modifying landing stiffness. In particular, they reported that the hip and knee extensors each absorbed 50% more energy during a soft land as compared to a stiff land. As illustrated in Figure 3A, individuals in the low flexion landing group limited sagittal plane loading at the hip, and in turn, relied on frontal plane knee as a secondary means to decelerate the body center of mass. In contrast, females in the high flexion landing group demonstrated more effective sagittal plane control, and minimal frontal plane loading at the knee (Figure 3B).

While it was evident that the females evaluated in the current study displayed two distinct landing patterns, the reason for the varied strategies are not known. One possible theory is that hip extensor weakness may contribute to a low flexion landing strategy. For example, if the hip extensors are unable to share the control of the body center of mass during landing, individuals may compensate by adopting an over-reliance on their quadriceps. Although hip muscle strength was not quantified as part of this study, it appears that those individuals who are able to limit frontal plane knee loading did so by engaging their hip extensors. Interestingly, the increase in hip extensor moments observed in the high flexion landing group was not accompanied by an increase in hamstring activation. This suggests that the gluteus maximus may have played a greater role in the high flexion landing strategy.

Regardless of the cause of the different landing strategies exhibited by subjects in our study, the consequence of the low flexion landing pattern is an increase in sagittal and frontal plane loading at the knee. This is relevant to ACL injury since in-vitro studies have demonstrated that increased quadriceps contraction at small knee flexion angles (Renstrom et al., 1986) as well as the combination of knee valgus loading and anterior shear (Markolf et al., 1995) result in increased ACL strain. Although the exact mechanism of the ACL tear continues to be debated (i.e. frontal plane versus sagittal plane), our results suggest that a low flexion landing strategy results in potentially abnormal loading in both planes.

In addition to providing evidence for the relationship between sagittal plane kinematics and frontal plane knee angles and moments, our study also provides support for the underlying framework of numerous ACL injury prevention programs which train females to land with greater sagittal plane motion. While programs that emphasize hip and knee flexion have been shown to decrease non-contact ACL injury (Hewett et al., 1999; Mandelbaum et al., 2005), the underlying mechanism for this "protective effect" has not been identified. Based on the results of the current investigation, it is possible that ACL injury prevention programs that emphasize sagittal plane shock absorption may decrease frontal plane loading at the knee. Further research would be necessary to fully test this hypothesis.

Care must be taken in inferring injury risk based on our findings as subjects were divided into 2 distinct groups based on the distribution of the combined hip and knee flexion angles during landing. As such, subjects falling below the mean (i.e. those assigned to the low flexion group), were considered to have "sub-optimal" landing mechanics. However, care must be taken in assuming that all subjects in the low flexion group would be at increased risk for experiencing an ACL tear. It is likely that only a percentage of subjects assigned to the low flexion group would truly be considered to be "at risk" (i.e. the lower quartile). A prospective study design would be needed to assess the relationship between an individual's landing strategy and ACL injury risk.

CONCLUSION

During the deceleration phase of landing, females who exhibited low hip and knee flexion exhibited increased knee valgus angles and knee adductor moments compared to those who demonstrated high hip and knee flexion. In addition, females who were limited in their sagittal plane motion also exhibited higher knee extensor moments and vastus lateralis activation. We propose that this biomechanical pattern may increase the risk for ACL injury. Furthermore, we hypothesize that decreased hip extensor strength may play a role in this movement strategies.

Acknowledgments

This study was funded by the National Institutes of Health (R01 AR053073-02).

References

Arendt EA, Dick R. Knee injury patterns among men and women in collegiate basketball and soccer. NCAA data and review of literature. Am J Sports Med 1995;23:694–701. [PubMed: 8600737]

Basmajian, JV.; Deluca, CJ. Muscles Alive: Their Functions Revealed by Electromyography. 5. Williams & Wilkins; Baltimore: 1985.

Boden B, Dean G, Feagin J, Garrett W. Mechanisms of anterior cruciate ligament njury. Orthopedics 2000;23:573–578. [PubMed: 10875418]

Bresler B, Frankel JP. The forces and moments in the leg during level walking. Transactions of the American Society of Mechanical Engineering 1950;72:27–36.

Devita P, Skelly WA. Effect of landing stiffness on joint kinetics and energetics in the lower extremity. Med Sci Sports Exerc 1992;24:108–115. [PubMed: 1548984]

Ford KR, Myer GD, Hewett TE. Valgus knee motion during landing in high school female and male basketball players. Med Sci Sports Exerc 2003;35:1745–1750. [PubMed: 14523314]

Ford KR, Myer GD, Hewett TE. Reliability of landing 3D motion analysis: Implications for longitudinal analyses. Med Sci Sports Exerc 2007;39:2021–2028. [PubMed: 17986911]

Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. J Biomech Eng 1983;105:136–144. [PubMed: 6865355]

Hewett TE, Riccobene JV, Lindenfeld TN. The effect of neuromuscular training on the incidence of knee injury in female athletes: A prospective study. Am J Sports Med 1999;24:699–706. [PubMed: 10569353]

Hewett TE, Paterno MV, Myer GD. Strategies for enhancing proprioception and neuromuscular control of the knee. Clin Orthop Relat Res 2002;402:76–94. [PubMed: 12218474]

Hewett TE, Myer GD, Ford KR, Heidt RSJ, Colosimo AJ, McLean SG, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes. Am J Sports Med 2005;33:492–501. [PubMed: 15722287]

Kernozek TW, Torry MR, Iwasaki M. Gender differences in lower extremity landing mechanics caused by neuromuscular fatigue. Am J Sports Med 2008;36:554–565. [PubMed: 18006677]

Landry SC, McKean KA, Hubley-Kozey CL, Stanish WD, Deluzio KJ. Neuromuscular and lower limb biomechanical differences exist between male and female elite adolescent soccer players during an unanticipated side-cut maneuver. Am J Sports Med 2007;35:1888–1900. [PubMed: 17921416]

Lephart SM, Ferris CM, Riemann BL, Myers JB, Fu FH. Gender differences in strength and lower extremity kinematics during landing. Clin Orthop Relat Res 2002;401:162–169. [PubMed: 12151893]

Malinzak RA, Colby SM, Kirkendal DT, Yu B, Garrett WE. A comparison of knee joint motion patterns between men and women in selected athletic tasks. Clin Biomech 2001;16:438–445.

Mandelbaum BR, Silvers HJ, Watanabe DS, Knarr JF, Thomas SD, Griffin LY, et al. Effectiveness of a neuromuscular and proprioceptive training program in preventing anterior cruciate ligament injuries in female athletes. Am J Sports Med 2005;33:1003–1010. [PubMed: 15888716]

Markolf KL, Burchfield DM, Shapiro MM, Shepard MF, Finerman GA, Slauterbeck JL. Combined knee loading states that generate high anterior cruciate ligament forces. J Orthop Res 1995;13:930–935. [PubMed: 8544031]

McLean SG, Neal RJ, Myers PT, Walters MR. Knee joint kinematics during the sidestep cutting maneuver: potential for injury in women. Med Sci Sports Exerc 1999;31:959–968. [PubMed: 10416556]

McLean SG, Huang X, van den Bogert AJ. Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: Implications for ACL injury. Clin Biomech 2005;20:863–870.

Pappas E, Hagins M, Sheikhzadeh A, Nordin M, Rose D. Biomechanical differences between unilateral and bilateral landings from a jump: gender differences. Clin J Sport Med 2007;17:263–268. [PubMed: 17620779]

Pollard CD, Sigward SM, Powers CM. Gender differences in hip joint kinematics and kinetics during side-step cutting maneuver. Clin J Sport Med 2007;17:38–42. [PubMed: 17304004]

Renstrom P, Arms SW, Stanwyck TS, Johnson RJ, Pope MH. Strain within the anterior cruciate ligament during hamstring and quadriceps activity. Am J Sports Med 1986;14:83–87. [PubMed: 3752352]

Salci Y, Kentel BB, Heycan C, Akin S, Korkusuz F. Comparison of landing maneuvers between male and female college volleyball players. Clin Biomech 2004;19:622–628.

Schmitz RJ, Kulas AS, Perrin DH, Riemann BL, Shultz SJ. Sex differences in lower extremity biomechanics during single leg landings. Clin Biomech 2007;22:681–688.

Sigward SM, Powers CM. The Influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting. Clin Biomech 2006;21:41–48.

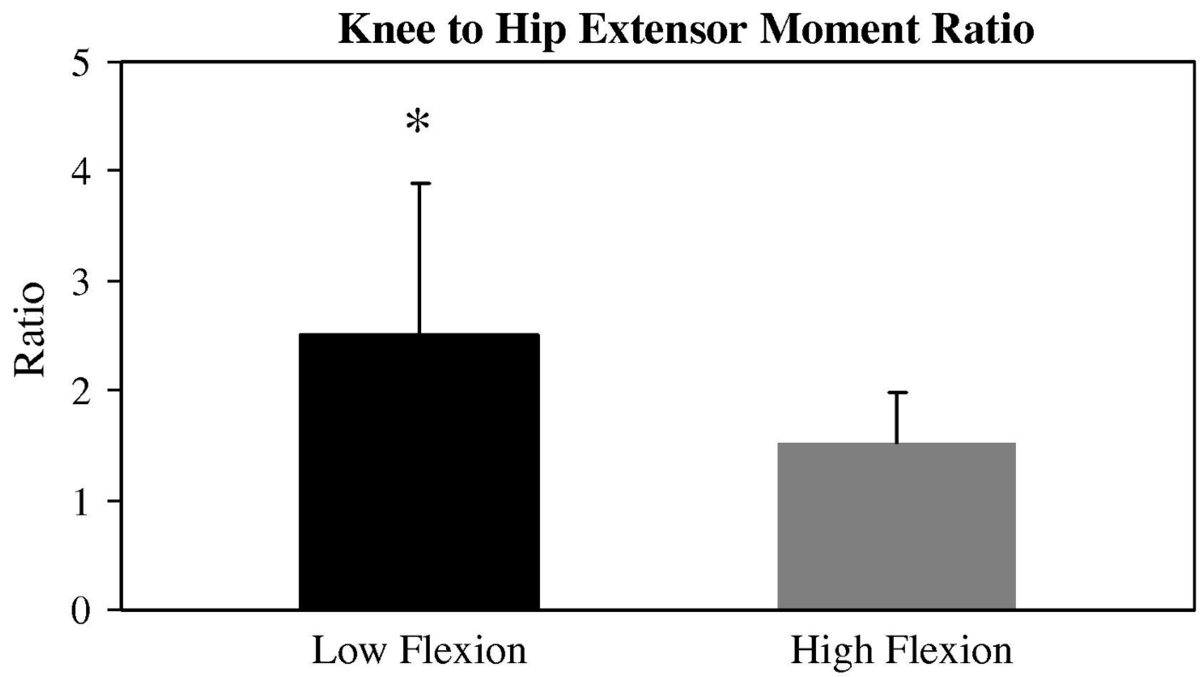

Figure 1.
Group differences in knee to hip extensor moment ratio during the deceleration phase of landing. *Indicates significant differences ($P \leq 0.05$).

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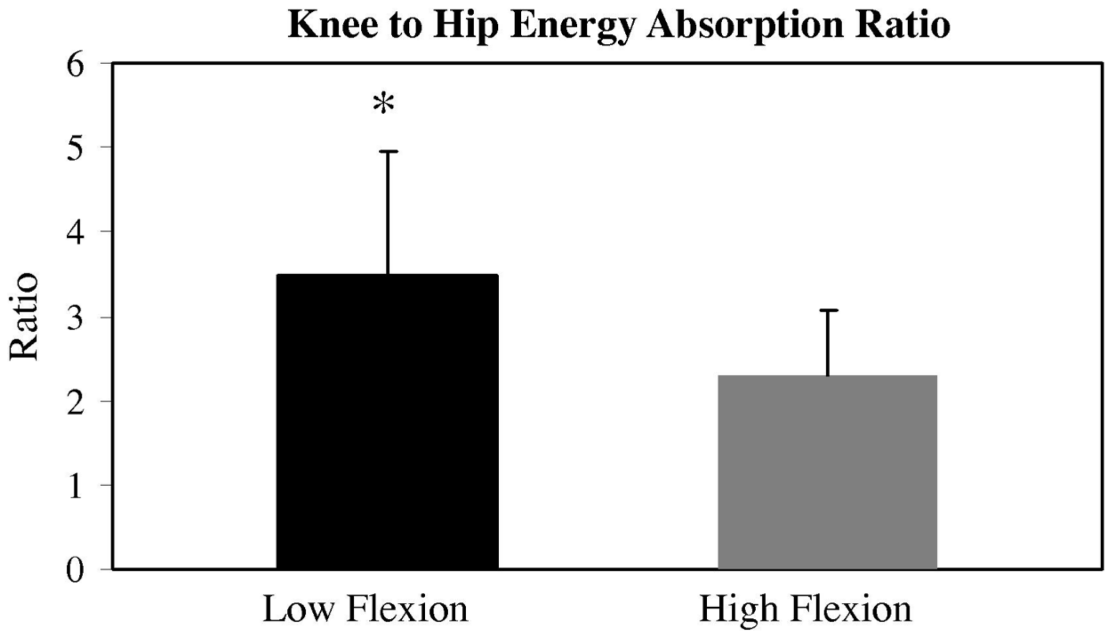

Figure 2.
Group differences in knee to hip extensor energy absorption ratio during the deceleration phase of landing. *Indicates significant differences ($P \leq 0.05$).

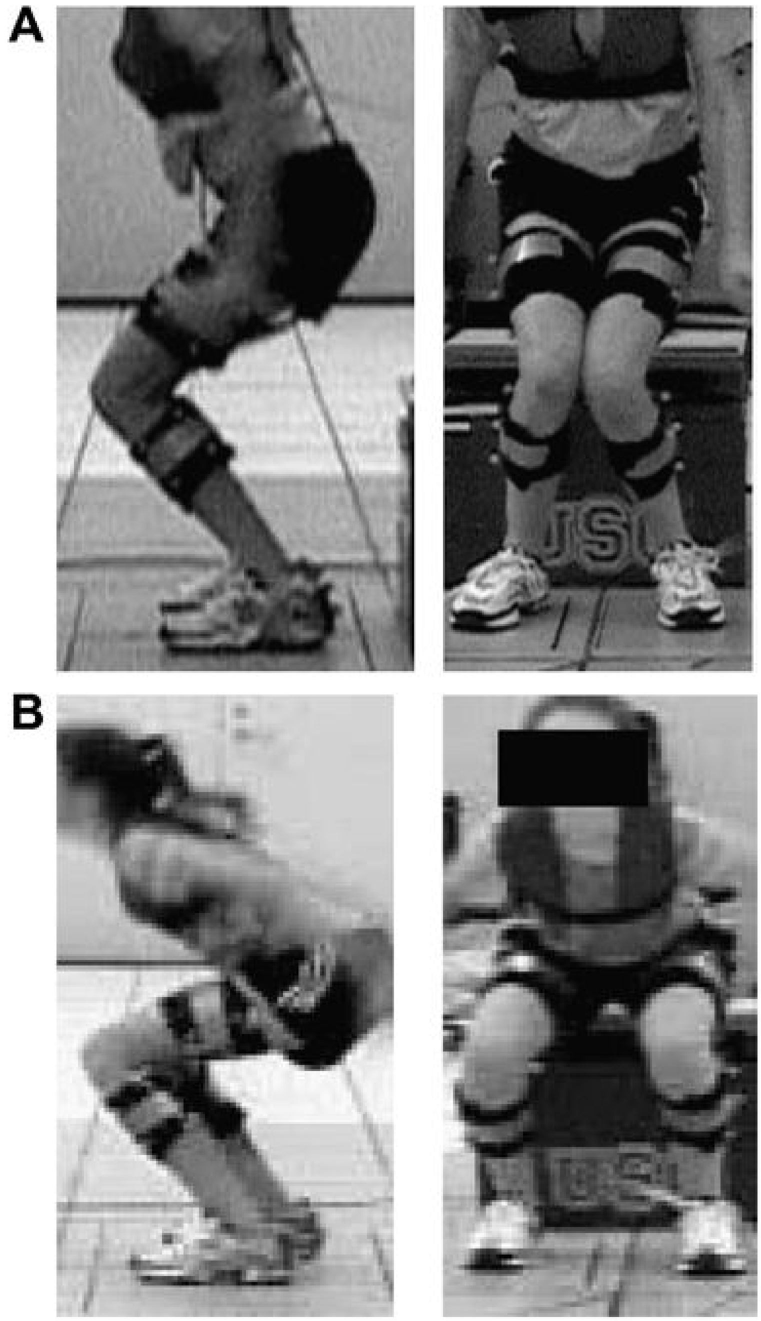

Figure 3.
An example of a “low flexion landing” (A) versus “high flexion landing” (B) strategy.

Table 1

Comparison of kinematic, kinetic and electromyographic data between sagittal plane landing groups

	Landing Groups (Mean (SD))			
	High Flexion	**Low Flexion**	**P value**	**Effect size**
Peak knee flexion (degrees)	100.6 (10.5)	86.5 (8.5)	< 0.001	0.59
Peak hip flexion (degrees)	89.9 (9.2)	67.4 (8.1)	< 0.001	0.79
Peak knee valgus (degrees)	3.9 (3.7)	6.3 (4.4)	0.02	0.27
Average knee extensor moment (Nm/kg)	1.22 (0.22)	1.34 (0.29)	0.05	0.22
Average hip extensor moment (Nm/kg)	0.84 (0.17)	0.65 (0.25)	< 0.001	0.41
Average knee adductor moment (Nm/kg)	0.06 (0.14)	0.13 (0.17)	0.03	0.24
Energy absorption at the knee (Watts/kg)	391.3 (74.1)	354.7 (69.5)	0.02	0.25
Energy absorption at the hip (Watts/kg)	187.2 (62.8)	113.2 (35.1)	< 0.001	0.59
Knee to hip extensor moment ratio	1.5 (0.5)	2.5 (1.4)	< 0.001	0.43
Knee to hip energy absorption ratio	2.3 (0.8)	3.5 (1.5)	< 0.001	0.45
Vastus lateralis muscle activation (% MVIC)	59.2 (23.5)	90.0 (53.1)	0.005	0.35
Medial hamstring muscle activation (% MVIC)	17.4 (9.9)	21.1 (13.5)	0.21	< 0.001
Lateral hamstring muscle activation (% MVIC)	24.5 (14.6)	24.4 (20.5)	0.99	< 0.001