Published in final edited form as:
Clin Biomech (Bristol, Avon). 2011 June ; 26(5): 504–508. doi:10.1016/j.clinbiomech.2011.01.004.

Patterns of Hip Flexion Motion Predict Frontal and Transverse Plane Knee Torques during a Single-leg Land-and-Cut Maneuver

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Abstract

Background—Impulsive frontal plane knee joint torques directly strain the anterior cruciate ligament and therefore may contribute to injury risk. Because deleterious torques may in part be related to aberrant kinematic movement patterns the primary purpose of this study was to establish a prediction model for frontal plane knee joint torques based on motion characteristics derived from Principal Component Analysis.

Methods—Eighteen healthy NCAA Division I female athletes performed a single-leg land-and-cut maneuver (n = 5 trials) with their dominant limb. Ensemble average lower extremity joint angles for the hip, knee, and ankle along with normalized external knee abduction torques were calculated for the entire stance phase. The ensemble kinematic data were individually submitted to a Principal Component Analysis. Principal component scores were used in a forward step-wise regression model to establish a prediction equation for peak ensemble-averaged knee abduction torque.

Findings—Approximately 31% of the variance in knee abduction torque was explained by a principal component that captured relative magnitudes of hip flexion motion during early stance. Likewise, approximately 32% of the variance in knee internal rotation torque was explained by a principal component that captured overall hip flexion during stance.

Interpretation—Rapid hip flexion motion during the first half of the stance phase of a single-leg land-and-cut maneuver is associated with greater knee abduction joint torques, whereas greater overall flexion during the entire stance phase is associated with smaller internal rotation torques.

Keywords

principal components analysis; landing biomechanics; regression analysis

Introduction

It is reported that between 80,000 – 250,000 anterior cruciate ligament (ACL) injuries occur each year in the United States alone (Griffin et al., 2006). It is interesting to note that

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roughly 70% of these injuries occur in the absence of physical contact and that when matched by sport, females are two to eight times more likely to sustain a non-contact ACL injury than their male counterparts (Boden et al., 2010, Hootman et al., 2007). While the sex bias in ACL injury rates appears to result from a combination of factors, differences in neuromechanical control are considered the greatest extrinsic contributors to this discrepancy (Griffin et al., 2006). Moreover, the ability to influence neuromechanical control with targeted interventions has increased focus on research in this area.

With respect to neuromechanical control, it is known that females display different movement patterns during dynamic landing, cutting and change of direction tasks than males (Chappell et al., 2002, Decker et al., 2003, Ford et al., 2005, Huston et al., 2001, Landry et al., 2007, Landry et al., 2007). Collectively, sex-differences in joint postures and movement patterns are implicated to result in a more deleterious joint loading environment that ultimately exposes females to a higher risk of ACL injury (Hewett et al., 2005, McLean et al., 2005). Regarding the loading environment of the knee joint, results from in-vitro analyses show that frontal and transverse plane knee joint torques directly strain the ACL (Markolf et al., 1995). Furthermore, the most compelling prospective evidence available suggests that neuromuscular control of frontal plane mechanics, especially frontal plane knee abduction torque, during jump-landing tasks can prospectively predict ACL injury risk in young female athletes (Hewett et al., 2005).

Modeling studies have demonstrated knee joint torques are sensitive to realistic perturbations in explicit lower limb kinematics (McLean et al., 2004, McLean et al., 2003). More specifically, initial contact lower limb postures during dynamic landings can predict peak frontal plane knee joint torques (McLean et al., 2005). Unfortunately, no such data is available for transverse plane knee joint torques. Moreover, though previous studies provide important information on the relationship between discrete variables (i.e. initial contact postures) and peak frontal plane knee torques, the analysis of solely peak kinematic variables provides limited information with respect to the structure of motion patterns across the time course of an entire movement. For example, Wrigley (Wrigley et al., 2005) demonstrated that traditional empirical analyses of peak variables were unable to prospectively discriminate between industrial workers that developed low back pain and those that did not, whereas variables derived from a principal components analysis (PCA), which capture spatial and temporal movement characteristics across time-series data, of the kinematic and kinetic waveforms were able to characterize biomechanical aspects of lifting technique prior to the development of low back pain in industrial workers. Other authors have recently used PCA to examine sex differences in neuromechanical control variables purported to influence ACL injury risk across a variety of movement tasks (Landry et al., 2007, Landry et al., 2007, O'Connor and Bottum, 2009). Results from these studies support the notion that PCA-derived variables may provide more sensitive measures that better describe the structure and variability of the time-varying movement characteristics and allow for a better understanding of injury risk factors.

Based on the ability of PCA to extract features of the underlying structure of lower extremity motions across entire time-series waveforms, this approach may offer unique insight beyond that available from discrete peak variables. The use of PCA-derived variables to establish whether certain sets of spatial and temporal movement characteristics are associated with deleterious joint torques would represent a novel extension of PCA and provide valuable information. From an applied clinical standpoint, this information could be used to complement and refine currently used injury screening and prevention protocols in that lower extremity movement characteristics may be targeted in an attempt to decrease deleterious joint torques. Therefore, the primary purpose of this study was to establish prediction models for frontal and transverse plane knee joint torques based on movement characteristics derived from PCA.

Methods

Eighteen healthy female athletes (mean(SD) age: 19.7(1.7) yrs; height: 1.67(0.06) m; mass: 61.5(6.0) kg) were recruited from NCAA Division I basketball, soccer, and volleyball teams to participate in this study. Exclusion criteria for participation in the study were as follows: 1) prior history of knee injury and/or surgery, 2) lower extremity pain before testing, 3) any recent injury to the lower extremity (previous 6 months), 4) participation in any exercise within 24 h of testing, and 5) current pregnancy. Before conducting the study, IRB approval was gained from the local institution. Written informed consent was obtained from all participants. All subjects wore spandex bike shorts, sports shoes, and a sport brassier during testing.

Biomechanical data were collected during the execution of a single-leg land-and-cut maneuver that required participants to initiate a forward two-legged jump, land on the dominant leg, and execute a side cut away from the landing leg (McLean and Samorezov, 2009). Participants performed this maneuver with their dominant limb and were required to perform five successful trials. Trials were deemed successful if the respective foot of the dominant limb made complete contact with an AMTI force plate (Advanced Mechanical Technology, Inc., Watertown, MA, USA) and within the field of view of an eight-camera high-speed motion analysis system (Motion Analysis Corp, Santa Rosa, CA, USA). Reflective markers were attached to anatomical landmarks on the pelvis, thigh, shank, and foot (McLean and Samorezov, 2009). In order to minimize marker movement subjects wore ankle length spandex tights over the markers, small incisions into the tights were made to pull the markers through so that they were visible to the motion capture cameras. Three-dimensional body segment orientations and joint center locations during each land-and-cut maneuver were reconstructed from the trajectories of the reflective markers that were attached to participants. These data were then used as input to a kinematic model that consisted of nine skeletal segments (foot, talus, shank, and thigh, and pelvis) with 24 degrees of freedom. A Mocap Solver (Motion Analysis Corp, Santa Rosa, CA, USA), software 6.17, was used to calculate the 3D lower extremity rotations at each time frame. Joint angles data were expressed relative to each subject's neutral standing position.

Marker trajectories and force plate data were low-pass filtered with a cubic smoothing spline at 12 Hz (Woltring et al., 1985). Joint angles of the hip, knee, and ankle were then computed in three dimensions based on a Cardan rotation sequence of flexion/extension, abduction/adduction, and internal/external rotation of the distal segment. Body segment masses, center of mass locations, and mass moments of inertia were calculated from measured anthropometrics and published relationships (de Leva, 1996). The resultant external joint torque acting at the knee in the frontal and transverse plane were then calculated with a conventional inverse dynamics approach and normalized to participant mass and height (McLean and Samorezov, 2009).

Kinematic and kinetic time-series waveform data were time-normalized to 100% of stance phase and ensemble-averaged. The stance phase was identified as the instant when the vertical component of the ground reaction force exceeded and fell below 10 N (Borotikar et al., 2008). The mean ensemble average of all participants was then subtracted from each individual participant's kinematic waveform. The individual mean-removed ensemble averages were then combined into 8 data matrices (one for each joint rotation) that consisted of 18 rows (one for each participant's mean-removed ensemble average) by 101 columns (time normalized stance phase). These data matrices were then used as input to the PCA

(Landry et al., 2007, Landry et al., 2007, O'Connor and Bottum, 2009). An eigenvector decomposition method was used to extract principal components from the covariance matrices of the input data. A scree plot was used to eliminate principal components that explained trivial proportions of the waveform variance. Principal components that explained a significant amount of the waveform variance were retained for further analysis. The retained principal components were each normalized in a manner similar to a unit vector in order to account for magnitude differences between principal components. These magnitude-normalized principal components were then projected onto the respective waveforms from which the principal components were originally extracted. The summation of these projections across the stance phase gave a set of PC scores that expressed the extent to which each principal component contributed to the original waveforms for each participant. The PC scores for each of the retained principal components were then subjected to statistical analysis.

Two forward step-wise regression models, with PC scores as dependent variables, were used to establish prediction equations for peak ensemble-averaged knee abduction and internal rotation torque. Normality and residual plots were checked to ensure that assumptions of the test statistic were met. Entry and removal conditions for the regression model were set at 0.05 and 0.10, respectively. All statistical analyses were performed with SPSS version 17 (SPSS, Chicago, IL, USA).

Results

The principal-component analysis extracted between 3–4 principal components for each of the joint angle waveform. For each joint angle rotation, the sum of all principal components explained more than 95% of the variance for the respective kinematic waveform. The combined total of 25 principal components was thus sufficient to describe the time-varying movement characteristics of the lower extremity during the single-leg land-and-cut maneuver.

PCA-derived kinematic principal components were able to account for approximately 31% and 32% of the variance in knee abduction and internal rotation torques, respectively (Table 1). The variance in knee abduction torque was explained by only one principal component; the second principal component extracted from the hip joint flexion waveform ($PC2_{HipFlexion}$). This principal component captured a relative magnitude in hip flexion motion during early landing around 25% of stance (Figure 1). Figure 2 illustrates the difference in knee abduction torques between subjects with high and low $PC2_{HipFlexion}$ scores. Similarly, the variance in knee internal rotation torque was also explained by a single principal component; the first principal component extracted from the hip joint flexion waveform ($PC1_{HipFlexion}$). This principal component captured overall magnitude of hip flexion motion across the entire stance phase (Figure 3). Figure 4 illustrates the difference in knee internal rotation torques between subjects with high and low $PC1_{HipFlexion}$ scores.

Discussion

In an effort to prevent ACL injuries, training programs have focused on the modification of aberrant joint motion under the premise that abnormal neuromuscular control is associated with deleterious knee joint loading in the frontal and transverse plane (Myklebust et al., 2003, Olsen et al., 2005). Though previous studies demonstrate that initial contact lower extremity postures are predictive of frontal plane knee joint torques, no such information is available for transverse plane knee joint torques. Furthermore, a major drawback associated with the use of discrete variables like initial contact postures is that they provide limited information about the underlying time-varying structure of lower extremity motions. The purpose of this study was to establish whether PCA-derived movement characteristics are predictive of deleterious frontal and transverse joint torques. It was anticipated that the results would reveal novel insight between kinematic movement characteristics and kinetic joint torques, which would ultimately help refine injury screening and prevention protocols.

The PCA extracted between 3–4 principal components for each of the joint angle waveforms. In total there were 25 principal components that described the time-varying kinematic movement characteristics of the lower extremity during the single-leg land-and-cut maneuver. Surprisingly, however, each regression model included only a single significant predictor; the second principal component for hip joint flexion waveform ($PC2_{HipFlexion}$) in the case of knee abduction torque and the first principal component for hip joint flexion waveform ($PC1_{HipFlexion}$) in the case of knee internal rotation torque. Our analysis indicated that $PC2_{HipFlexion}$ captured a relative peak in hip flexion motion immediately after landing, whereas $PC1_{HipFlexion}$ captured the overall magnitude of hip flexion motion across the entire landing phase (Figures 1 and 3). It therefore appears that rapid hip flexion upon impact is associated with a greater abduction torque while a more erect posture across the entire movement is associated with a greater internal rotation torque. We should emphasize that since PCA uses an orthogonal decomposition method, the components that are extracted from each waveform are independent and not correlated to each other. Thus, even though both regression models consist of predictors that capture characteristics of hip joint flexion, these variables represent unique features within the underlying time-varying structure of that motion.

Our finding that a hip flexion magnitude during early landing is associated with peak frontal plane knee joint loading is similar to that of a previous study that found larger hip flexion at initial contact was related to frontal plane knee joint loading (McLean et al., 2005). In a recent video analysis of non-contact ACL injuries that occurred during athletic competitions, Boden (Boden et al., 2009) found that hip flexion motion immediately after landing was statistically greater in the group who sustained an injury than in the control group. The observed early hip flexion movement during the initial landing phase in this study is therefore consistent with that reported in the literature as occurring during ACL injury. A potential explanation for the association between this characteristic of rapid hip flexion and peak frontal plane knee joint loading during a single-leg land-and-cut maneuver is that forward rotation of the trunk away from the landing leg and in the cutting direction may cause the center of pressure underneath the landing foot to translate in a more anterior and lateral direction. Such a change in the center of pressure would increase the moment arm and alter the vector direction of the ground reaction force in a way that could increase the resultant external torque about the knee joint. Because the hip extensor musculature actively controls forward rotation of the torso during landing, it could be argued that insufficient muscle activation around the time of initial contact leads to quick and rapid trunk flexion about the hip joint immediately after landing. Since the hamstring musculature plays an active role in the support of external knee abduction torques (Lloyd et al., 2005), aberrant hamstring muscle activation may not only lead to rapid initial hip flexion upon landing, but could also contribute to frontal plane knee joint torques. Seeing as attenuated eccentric activation of the hamstring muscles may also increase the relative strain within the ACL during simulated landings (Withrow et al., 2008), proper neuromuscular control therefore appears crucial in assuaging deleterious knee joint forces and torques.

Although a more erect posture is frequently cited as a risk factor for ACL injury (Griffin et al., 2006) and previous research shows that active hip and knee flexion are associated with smaller ground reaction forces (Yu et al., 2006) and simulated ACL injury via a valgus load mechanism (McLean et al., 2008), the exact effects of maintaining an erect posture throughout the stance phase of single-leg landings on knee joint torques are not known. Our results indicate that the overall magnitude of hip flexion over the entire landing phase is associated with peak transverse plane knee joint loading. More specifically, less overall hip flexion (i.e. a more erect posture) throughout landing is associated with greater peak internal rotation torque at the knee during a single-leg land-and-cut maneuver. This finding is significant because an internal rotation torque applied at the knee is considered to be an important dynamic loading mechanism of the ACL (Griffin et al., 2006). Markolf and colleagues (Markolf et al., 1995) showed that the overall risk of injury to the ACL from internal rotation torque alone is similar to the risk from a combination of an abduction and internal rotation torque. Although it is not immediately intuitive why greater overall hip flexion should serve to decrease knee rotational torques, it may be that active flexion in the sagittal plane enables the major muscle groups that cross the knee to better absorb energy during landing and more effectively neutralize torques acting in other planes of rotation.

The regression analyses in this study considered only PCA-derived scores, which capture kinematic movement characteristics, as potential contributors to frontal and transverse plane knee joint loading. While each of the regression models explained slightly less than approximately one third of the variance in the respective knee joint loadings, several other sources likely account for the remaining contributors to the loading environment. For example, the biomechanical profile during the landing phase of dynamic tasks may also be influenced by joint laxity or lower extremity strength (Shultz and Schmitz, 2009, Shultz et al., 2009). In turn, in-vitro models show that the kinematic profile during simulated landing has the potential to increase strain of the ACL (Withrow et al., 2006). Furthermore, internal forces from lower extremity muscles that cross the knee can also significantly alter the overall loading environment in the joint during simulated dynamic activities (Withrow et al., 2006). Future studies should consider investigating these factors in combination in order to increase the ability to predict which variables are most deleterious to knee joint loading and risk of injury.

Conclusions

This study examined the relationship between lower extremity kinematic movement characteristics derived from PCA and frontal and transverse plane knee joint loading in female athletes. The major findings were that patterns of hip flexion motion during a single-leg land-and-cut maneuver were associated with external knee abduction and internal rotation torques. Effective control of hip flexion movements play an important role with respect to frontal and transverse plane knee joint loading and therefore underscore the importance of proximal neuromechanical control during single-leg land-and-cutting maneuvers. From a clinical standpoint the findings indicate that gradual and controlled hip flexion during early landing and greater overall hip flexion motion during the entire landing should minimize knee abduction and internal rotation torques, respectively. Injury screening and prevention protocols should consider these aspects of controlled hip flexion during a single-leg land-and-cut maneuver as a trainable characteristic that could decrease the magnitude of deleterious knee joint torques and potentially decrease the likelihood of ACL injury in females.

Acknowledgments

This research was funded in part through the National Football League Charities. The authors would also like to acknowledge assistance from the National Institutes of Health (NIH T-32 HD007422-15).

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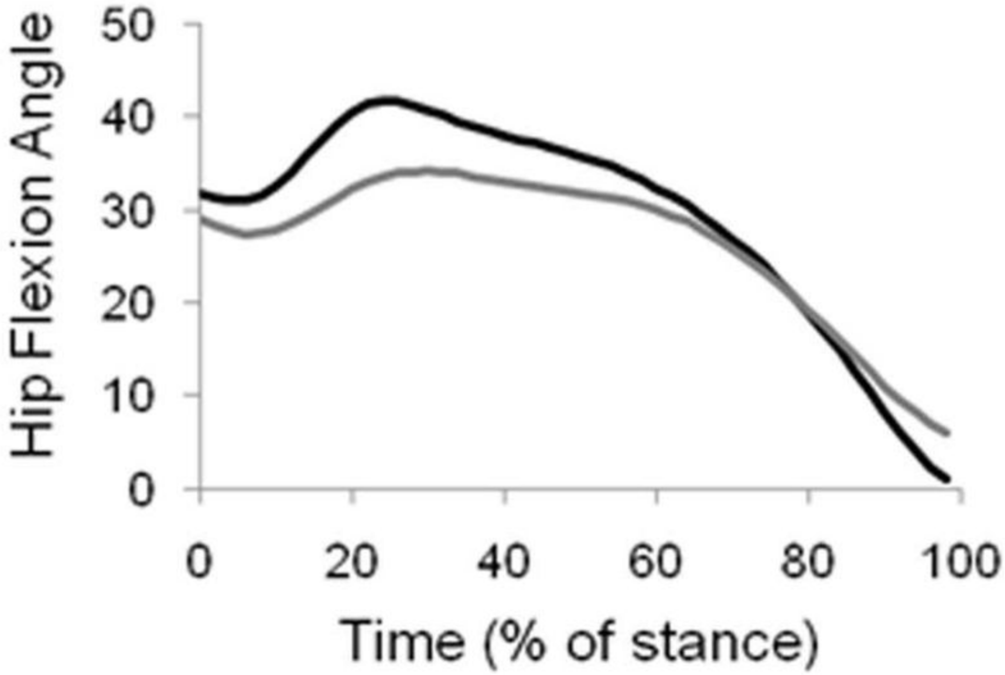

Figure 1.
Ensemble average hip flexion angles (degrees) across the stance phase of the single-leg land-and-cut maneuver for subjects with high (black line) and low (grey line) $PC2_{HipFlexion}$ scores.

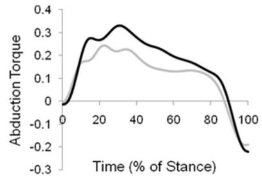

Figure 2.

Normalized ensemble average external knee abduction torques (N m/k gm) across the stance phase of the single-leg land-and-cut maneuver for subjects with high (black line) and low (grey line) $PC2_{HipFlexion}$ scores.

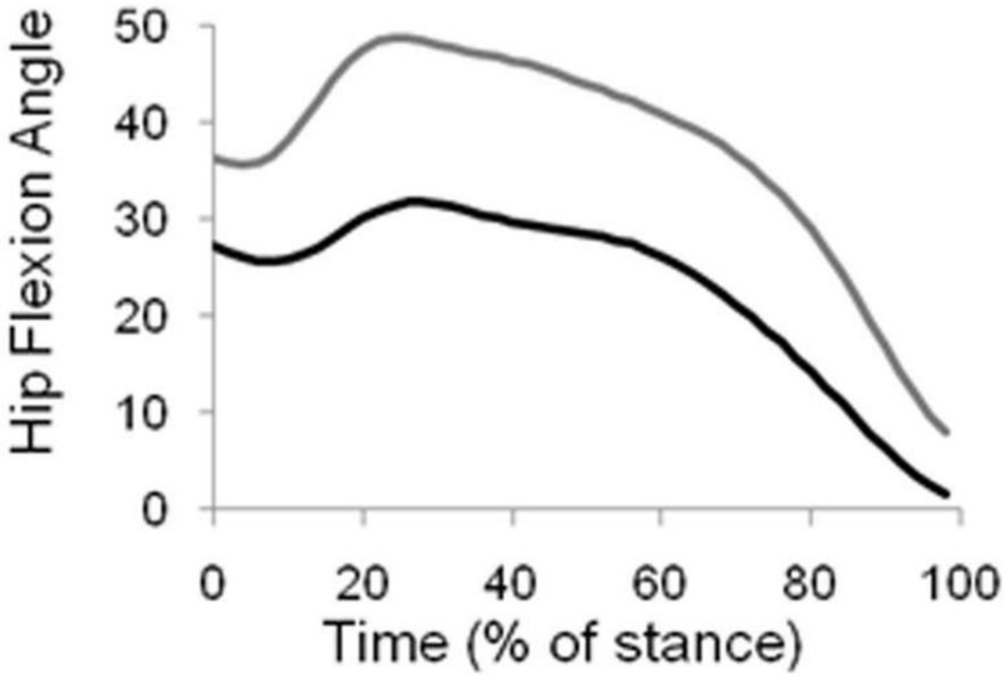

Figure 3.
Ensemble average hip flexion angles (degrees) across the stance phase of the single-leg land-and-cut maneuver for subjects with high (black line) and low (grey line) $PC1_{HipFlexion}$ scores.

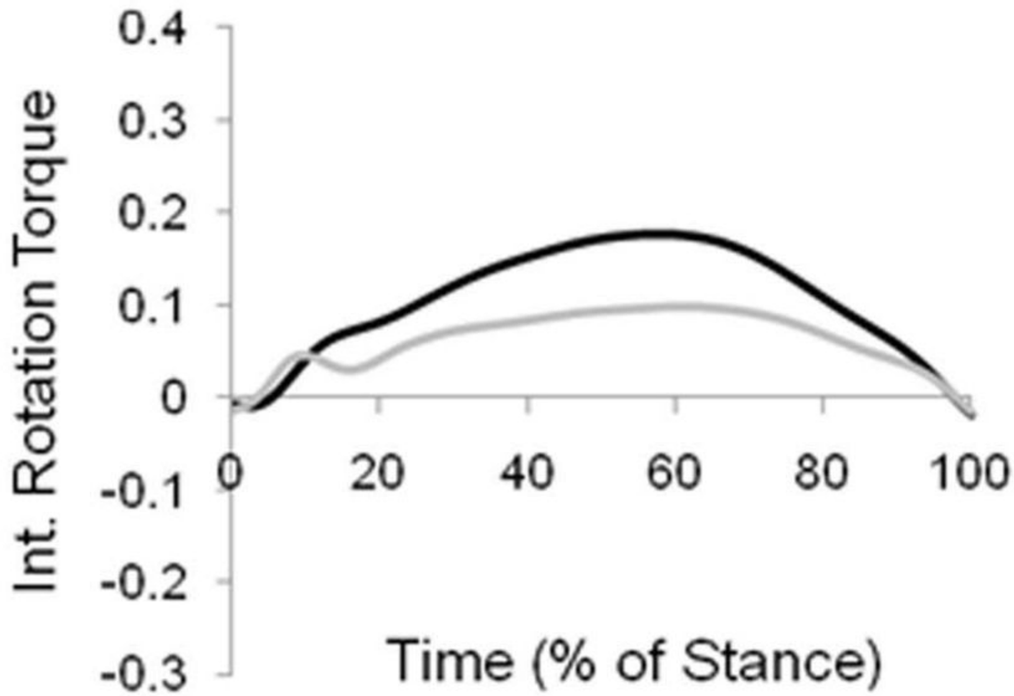

Figure 4.

Normalized ensemble average external knee internal rotation torques (N m/k gm) across the stance phase of the single-leg land-and-cut maneuver for subjects with high (black line) and low (grey line) $PC1_{HipFlexion}$ scores.

Table 1

Regression models with standardized regression coefficient for normalized knee abduction and internal rotation torque (N · m/k · gm).

Torque	Regression Model	R^2_{Adj}
Abduction	$.556\ PC2_{HipFlexion}$*	.309
Internal Rotation	$-.598\ PC1_{HipFlexion}$*	.317

*$p < .05$