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Differences in normal and perturbed walking kinematics between male and female athletes

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Abstract

Objective. To identify differences in lower extremity kinematic movement patterns between genders during walking through the application of an expected perturbation.

Design. Randomized limb kinematics were compared between healthy active males and females.

Background. Lower extremity kinematics during jump landing and cutting have been implicated as a potential source of the discrepancy in anterior cruciate ligament injury rates between genders. Kinematic differences between genders have been identified during tasks that are not provocative of anterior cruciate ligament injury but do result in increased ligament strain. Repetition of movement patterns that increase anterior cruciate ligament strain may increase the likelihood they will be reproduced during athletic tasks that produce force loads that exceed anterior cruciate ligament tensile strength.

Methods. Twenty subjects (10 women, 10 men) classified as a level I or II athlete underwent motion analyses while performing self-paced walking trials. Five trials were undisturbed, and five each with a platform translating either laterally or anteriorly at heel contact. Sagittal, frontal, and transverse hip angles as well as sagittal and frontal knee angles were collected during stance.

Results. Excursions in the frontal and transverse planes were greater at the hip and knee for females compared to males in each walking condition. The rate of these excursions also occurred more rapidly for females than males. There was no difference for joint angles at initial contact between genders, and there was no difference in the amount of sagittal plane excursion for the hip and knee when comparing genders.

Conclusions. Females demonstrate characteristics during both normal and perturbed gait that may potentially contribute to increased anterior cruciate ligament strain. Repetition of these potentially harmful movement patterns during provocative athletic maneuvers may lead to anterior cruciate ligament injury.

Relevance

Females exhibit lower extremity kinematic patterns that differ from males. Female kinematic patterns may contribute to an increased risk for anterior cruciate ligament injury.

Keywords: Anterior cruciate ligament; Gender; Gait

1. Introduction

The female anterior cruciate ligament (ACL) injury epidemic has been well documented. Women who are active in jumping and cutting sports injure their ACL's 2–9 times more frequently than their male counterparts (DeHaven and Lintner, 1986; Arendt and Dick, 1995; Myklebust et al., 1998). The overwhelming majority of these injuries occur through non-contact mechanisms (Noyes et al., 1983; Myklebust et al., 1998; Myklebust et al., 1997). The most common athletic maneuvers resulting in non-contact ACL injuries for both genders include planting and cutting, straight knee landings, and rapid, 1-step stops (McNair et al., 1990). Increased ACL

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strain can be triggered during these athletic maneuvers as a consequence of the collective positioning of the hip, knee and ankle. Lower extremity positioning at the time of injury often includes relatively low hip and knee flexion angles accompanied by high femoral internal rotation, adduction, and knee valgus angles, and external rotation of the foot relative to the knee (Boden et al., 2000; McNair et al., 1990; Noyes et al., 1983; Feagin and Lambert, 1985; Dufek and Bates, 1991). Previous studies have found female athletes are more likely to reproduce lower extremity movement patterns that contribute to increased ACL strain when compared to male athletes (Chappell et al., 2002; McLean et al., 1999). Thus, kinematic differences may explain, in part, the gender bias in ACL injuries.

In vitro and in vivo investigations have directly measured ACL strain and load during isolated and combined movements of the knee. Isolated anterior tibial translation strains the ACL with the joint near full extension. Most studies show the ACL is also strained in response to internal tibial rotation torques, (Markolf et al., 1990, 1995; Fleming et al., 2001) but not in response to external tibial rotation torques (Berns et al., 1992; Markolf et al., 1995; Fleming et al., 2001). Few studies have measured isolated and combined effects of varus and valgus torques on the strain behavior of the ACL. Markolf and colleagues in 1990 (Markolf et al., 1990) and 1995 (Markolf et al., 1995) demonstrated an increase in ACL strain with varus and valgus torques, though he concluded, "The overall risk of injury to the ligament from varus or valgus moment applied in combination with internal tibial torque is similar to the risk from internal tibial torque alone." Fleming and colleagues (Fleming et al., 2001) and Berns and colleagues (Berns et al., 1992) found no increase in ACL load or strain with the application of isolated varus or valgus torques. Fleming and colleagues evaluated the effect of weight bearing on these isolated motions and found at 20° of knee flexion with a load of 40% body weight, the strains in the ACL were significantly higher for both varus and valgus.

Timing of non-contact ACL injuries usually occurs during single limb support ranging from initial contact through loading duration (e.g., the point of heel strike to peak knee flexion). ACL injury occurring at foot contact is strongly related to positioning of the lower extremity (i.e., hip and knee extension, knee abduction, tibia internal rotation, etc.). However, the ability to control knee excursion throughout the loading duration can also determine whether ligamentous loading occurs. Joint excursion is also influenced by the passive and active stiffness properties of the musculo-tendinous components. Knee joint excursion and positioning are also directly related to hip and ankle kinematics. Consequently, undesirable hip and ankle positioning and excursions may result in *relative* knee positioning that leads to injury. Excessive lower extremity excursions in the frontal and transverse planes result in knee positioning that can load the ACL. Conversely, limited sagittal plane excursions can leave the knee in a more extended position and thus lead to increased ACL strain. Maneuvers with a combination of decreased hip and knee extension, increased knee valgus and hip internal rotation bring athletes close to body positions in which ACL injuries may occur (Malinzak et al., 2001).

Griffin and colleagues (Griffin et al., 2000) and Davis and Ireland (2003) summarized the results from consensus conferences held to investigate the female athlete's ACL injury problem. Biomechanical, anatomic, environmental, and hormonal factors were identified as factors that may contribute to gender differences in injury rates. Lower extremity kinematics and neuromuscular characteristics are two components of biomechanics that differ between males and females. Gender differences in lower extremity kinematics during provocative (i.e., jump landing and cutting) athletic maneuvers have been identified by multiple investigators (Huston and Wojtys, 1996; Hewett et al., 1996; Malinzak et al., 2001; McLean et al., 1999; Lephart et al., 2002; Besier et al., 2003). In each of these studies, female athletes were more likely than males to reproduce lower extremity movement pattern(s) that are known to increase ACL strain. Neuromuscular characteristics exhibited by females also contribute to deleterious movement patterns. Females have earlier muscle activation in anticipation of landing (Schmitz and Thompson, 2002; Cowling and Steele, 2001), rely heavily on their quadriceps to stabilize the knee (Huston and Wojtys, 1996; Malinzak et al., 2001), and take longer to produce peak muscular tension (Bell and Jacobs, 1986; Komi and Karlsson, 1979). Specialized training programs (Hewett et al., 1996; Chmielewski et al., 2002; Beard et al., 1994; Ihara and Nakayama, 1986) have resulted in changes in ground reaction forces, strength, and muscle firing patterns. The use of these specialized training programs has resulted in variable effectiveness in reducing the female injury rate.

Investigators have identified gender differences in movement patterns during walking and running (Ferber et al., 2003; Malinzak et al., 2001). Kerrigan and colleagues (Kerrigan et al., 1998) reported sagittal plane hip flexion was greater and knee flexion was lower in females in preparation for weight bearing. Kerrigan and colleagues also described changes in knee flexion moment and power absorption during pre-swing (a time in the gait cycle not usually associated with injury). Two studies have reported on differences between genders during running. Malinzak and colleagues (Malinzak et al., 2001) found women to have significantly lower peak knee flexion and knee flexion excursion compared to men. Additional findings from this study include similar frontal plane knee excursions between genders, but with females remaining in valgus throughout the stance phase. Ferber and colleagues (Ferber et al., 2003) reported kinematic differences between genders during running for both the hip and knee. Ferber and colleagues found no kinematic differences between genders in the sagittal plane. Ferber did find significant differences in peak knee abduction, hip internal rotation, and hip adduction in females when compared to males. Subjects in the study by Ferber and colleagues also demonstrated higher hip angular velocities and power absorption during the loading phase.

While ACL injury does not typically happen during running and walking, individuals who demonstrate potentially harmful movement patterns during these activities may also reproduce them during athletic tasks that can injure the ACL. Little attention has been devoted to gender differences in global movement patterns and how these patterns may contribute to the greater ACL injury risk for female athletes. Therefore, the purpose of this study was to identify differences in lower extremity kinematic strategies between genders during walking through the application of an expected perturbation. The perturbed walking paradigm was chosen as a testing model based on similar reports of its usefulness in revealing dynamic stabilization strategies (Ferber et al., 2002; Nashner, 1980). The variables of interest were those that have been implicated during ACL injury and are known to increase ACL strain and/or impingement. It was hypothesized that males and females would demonstrate different lower extremity kinematic patterns. Specifically, we hypothesized that (1) females would have less hip and knee extension at initial contact, and greater hip adduction, internal rotation, and knee valgus at initial contact; (2) females would have less sagittal plane hip and knee excursion and greater transverse and frontal plane hip and knee excursions measured during the loading duration of stance when compared to males and (3) that sagittal, frontal, and transverse plane hip and sagittal and frontal plane knee excursion rates measured during the loading duration of stance would occur more quickly in females than in males.

2. Methods

2.1. Subjects

Preliminary data indicated 10 subjects would be needed in each group to achieve adequate power (alpha $\leqslant 0.1$, power $\geqslant 80$). A total of 20 subjects (10 males and 10 activity-level and age matched females) were recruited for the study (Table 1). Participation criteria included being regular participants in sports that include jumping, cutting, and/or pivoting >50 h/year.

Table 1
Subject characteristics reported as mean (SD)

	Males	Females	P-value
Age	21.3 (8.8)	20.9 (8.4)	0.853
Mass[a]	82.9 (14.9)	71.3 (8.7)	0.089*
Pelvic width[b]	38.2 (2.3)	38.6 (2.7)	0.859
Femur length[c]	41.9 (2.0)	41.3 (1.6)	0.631
Limb length[d]	87.3 (6.2)	82.5 (1.7)	0.063*

[a] Reported in kg.
[b] Measured in cm from greater trochanter to greater trochanter.
[c] Measured in cm from greater trochanter to lateral femoral condyle.
[d] Measured in cm from greater trochanter to lateral malleolus.
* Significant at $P < 0.1$.

Exclusion criteria were a history of, or current injury to either lower extremity; no history of vestibular dysfunction; and no recent history (last 6 months) of back pain which necessitated physician attention. All participants signed an informed consent form approved by the University of Delaware Institutional Review Board.

2.2. Motion analysis

Kinematic data were collected with a six camera passive, three-dimensional motion analysis system (VICON, Oxford Metrics, London, England). The cameras were calibrated to a volume of 1.67 m^3. Calibration errors were held below 3 mm. Kinematic data were sampled at 120 Hz and low-pass filtered 6 Hz with a fourth-order zero lag Butterworth filter.

Retro-reflective anatomical markers defining the joint centers were placed bilaterally over the greater trochanter, lateral femoral condyle and lateral malleolus. Rigid thermoplastic shells with markers firmly secured for tracking three-dimensional movement were attached bilaterally to the postero-lateral aspect of the pelvis, thigh and shank. Two additional markers were placed on the posterior heel counter of the shoe and one on the fifth metatarsal head to track foot motion. After a static standing calibration was collected, the retro-reflective anatomical markers were removed and dynamic trials were collected. Approximately six 30 mm foot switches (Motion Lab Systems, Baton Rouge, LA, USA) were affixed to the bottom of the subject's test side shoe to assist in determination of initial contact and toe-off. The exact number of foot switches necessary to determine stance varied based on subject shoe size. The testing limb was randomized and determined prior to data collection.

Subjects walked along a 13 m walkway at their preferred speed. A perturbation platform embedded in the floor was located at its center. Five trials each were collected during three separate walking conditions. The first condition was with the perturbation platform locked (not moving). There were also two separate moving conditions during which the perturbation platform would translate either anteriorly or laterally. The sequence of executing the anterior and lateral perturbation trials was randomized prior to the data collection. Subjects first performed 3–5 practice trials before each condition until speed was consistent and platform contact could be achieved without targeting. Walking speed was monitored using photoelectric cells placed 2.86 m apart along the walkway. All subjects were tested in the locked condition first. The perturbation platform was then programmed to translate in the pre-determined direction after initial contact of the test leg. The platform translation rate was 40 cm/s over a distance of 5.8 cm. Time required for the platform to reach translation speed was 5 m s. Then, the platform was repositioned to collect perturbed trials in the untested movement direction. Data collection for all trials was initiated at toe-off of the test leg prior to contact with the perturbation platform. Only trials in which walking speed was within 5% of the mean speed were accepted.

MoveSD software (MOVE3D, NIH Biomotion Laboratory, Bethesda, MD, USA), was used to calculate kinematic variables. Kinematic variables of interest included hip flexion, adduction, and internal rotation at initial contact; knee adduction and flexion at initial contact; and loading duration (LRD), which was defined as the time from initial foot contact to peak knee flexion.

2.3. Data management and analysis

Kinematic data from the five trials for each subject were averaged. Excursions for kinematic variables were measured as joint movement during LRD. Excursions were divided by LRD to calculate excursion rates. A repeated measures MANOVA was used to assess joint angles at initial contact, excursion and excursion rates for each combination of motions (sagittal plane hip and knee motion; transverse and frontal plane hip and frontal plane knee motions). Post hoc testing was performed with independent *t*-tests. Differences in movement patterns were also analyzed using frequency counts for movement direction during loading response. A chi-square statistic was used to analyze these data. Given the highly variable nature of healthy, adult gait patterns (Perry, 1992), statistical significance was set at $P < 0.1$.

3. Results

MANOVA testing (Table 2) revealed no significant differences between groups for joint angle at initial contact across conditions. There were also no differences between groups for sagittal plane excursion or excursion rates across conditions. Females did demonstrate significantly greater excursions for each testing condition when assessing transverse/frontal plane movements at the hip and knee. Excursion rates in transverse/frontal planes were also significant for each condition. Post hoc testing (Tables 3–5) indicates group differences in frontal plane knee and transverse hip excursion and excursion rates for all conditions. Additionally, excursions for females in the frontal plane of the hip were significantly greater in the locked condition. Figs. 1 and 2 present the 3-D angular motions of the hip and knee during the lateral condition. Similar motion patterns were observed in the locked and anterior conditions.

Both genders across conditions demonstrated unidirectional excursions for knee adduction (into adduction), knee flexion (into flexion), hip adduction (into adduction), and hip flexion (into extension). Excursions for hip rotation were inconsistent for each condition. In the locked condition 3/10 males and 8/10 females

Table 2
MANOVA results (*P*-values) for differences between groups

	Locked	Anterior	Lateral
Initial contact			
Sagittal	0.233	0.417	0.364
Transverse/frontal	0.892	0.887	0.932
Joint excursion			
Sagittal	0.820	0.129	0.449
Transverse/frontal	0.001*	0.021*	0.002*
Excursion rate			
Sagittal	0.197	0.903	0.232
Transverse/Frontal	0.000*	0.018*	0.000*

* Denotes significance at $P < 0.1$.

Table 3
Locked condition: comparisons (mean (SD)) of angles, excursions, excursion rate, and gait descriptives

Variable of interest	Male	Female	*P*-value
Angle at initial contact (degrees)			
Knee adduction	1.17 (2.99)	−0.09 (1.48)	0.247
Knee flexion	8.39 (3.51)	8.23 (3.47)	0.929
Hip adduction	−1.37 (3.79)	−0.119 (3.21)	0.434
Hip flexion	30.87 (3.51)	33.94 (7.54)	0.259
Hip internal rotation	3.41 (6.97)	3.03 (4.52)	0.885
Joint excursion (absolute value in degrees)			
Frontal knee	2.17 (1.62)	4.21 (2.83)	0.063*
Sagittal knee	16.7 (4.48)	14.25 (3.92)	0.210
Frontal hip	4.91 (1.98)	6.46 (1.89)	0.091*
Sagittal hip	7.66 (4.06)	9.67 (3.58)	0.255
Transverse hip	2.14 (1.13)	6.18 (4.1)	0.008*
Excursion rate			
Frontal knee	0.0149 (0.01)	0.0325 (0.02)	0.024*
Sagittal knee	0.1165 (0.04)	0.1115 (0.03)	0.732
Frontal hip	0.0338 (0.01)	0.0506 (0.01)	0.014*
Sagittal hip	0.0523 (0.03)	0.0771 (0.03)	0.071*
Transverse hip	0.0149 (0.01)	0.0489 (0.03)	0.006*
Gait characteristics			
Velocity (m/s)	1.54 (0.16)	1.63 (0.13)	0.229
LRD (m s)	144.5 (8.86)	127.4 (11.3)	0.001*

* Denotes significant difference between groups ($P < 0.1$).

Table 4
Anterior condition: comparisons (mean (SD)) of angles, excursions, excursion rate, and gait descriptives

Variable of interest	Male	Female	*P*-value
Angle at initial contact (degrees)			
Knee adduction	1.00 (3.62)	−0.19 (1.82)	0.363
Knee flexion	7.16 (3.61)	9.36 (4.08)	0.218
Hip adduction	−0.77 (3.91)	0.28 (3.03)	0.507
Hip flexion	30.21 (3.22)	34.98 (7.34)	0.076*
Hip internal rotation	4.32 (4.79)	4.81 (5.19)	0.832
Joint excursion (absolute value in degrees)			
Frontal knee	2.00 (1.28)	3.51 (2.4)	0.097*
Sagittal knee	12.0 (5.17)	8.58 (4.5)	0.131
Frontal hip	4.22 (1.98)	4.61 (2.39)	0.697
Sagittal hip	8.61 (5.86)	8.29 (3.03)	0.881
Transverse hip	1.39 (0.95)	4.36 (2.95)	0.007*
Excursion rate			
Frontal knee	0.0132 (0.01)	0.0272 (0.02)	0.051*
Sagittal knee	0.0788 (0.04)	0.0642 (0.03)	0.312
Frontal hip	0.0272 (0.01)	0.0294 (0.03)	0.823
Sagittal hip	0.0541 (0.03)	0.0670 (0.03)	0.353
Transverse hip	0.0093 (0.01)	0.0329 (0.02)	0.002*
Gait characteristics			
Velocity (m/s)	1.60 (0.16)	1.66 (0.12)	0.385
LRD (ms)	155.6 (28.2)	133.9 (37.8)	0.163

* Denotes significant difference between groups ($P < 0.1$).

Table 5
Lateral condition: comparisons (mean (SD)) of angles, excursions, excursion rate, and gait descriptives

Variable of interest	Male	Female	*P*-value
Angle at initial contact (degrees)			
Knee adduction	1.04 (3.24)	−0.06 (1.65)	0.348
Knee flexion	8.62 (3.59)	9.53 (4.48)	0.620
Hip adduction	−0.67 (3.36)	0.61 (3.21)	0.392
Hip flexion	30.77 (3.14)	34.54 (8.00)	0.183
Hip internal rotation	4.07 (4.62)	4.06 (5.51)	1.000
Joint excursion (absolute value in degrees)			
Frontal knee	1.73 (1.29)	3.42 (2.41)	0.068*
Sagittal knee	14.56 (4.09)	12.12 (4.19)	0.205
Frontal hip	2.54 (1.48)	3.58 (2.3)	0.246
Sagittal hip	8.93 (3.71)	9.98 (3.42)	0.519
Transverse hip	2.03 (0.90)	6.8 (4.65)	0.005*
Excursion rate			
Frontal knee	0.0125 (0.01)	0.0267 (0.02)	0.045*
Sagittal knee	0.1059 (0.04)	0.0951 (0.03)	0.422
Frontal hip	0.0185 (0.02)	0.0272 (0.03)	0.246
Sagittal hip	0.0648 (0.03)	0.0786 (0.02)	0.049*
Transverse hip	0.0146 (0.01)	0.0502 (0.03)	0.003*
Gait characteristics			
Velocity (m/s)	1.60 (0.16)	1.66 (0.14)	0.436
LRD (ms)	137.8 (7.62)	129.3 (25.48)	0.326

* Denotes significant difference between groups ($P < 0.1$).

demonstrated excursion into internal rotation (chi square $P = 0.021$). In the anterior condition 5/5 males and 10/10 females were measured to have hip excursion into internal rotation (chi square $P = 0.003$). Finally, in the lateral condition only one male had hip excursion into external rotation; all remaining subjects were measured to have greater hip internal rotation excursion in the lateral condition. There was no consistent pattern across the different conditions for magnifying group differences in specific planes of movement.

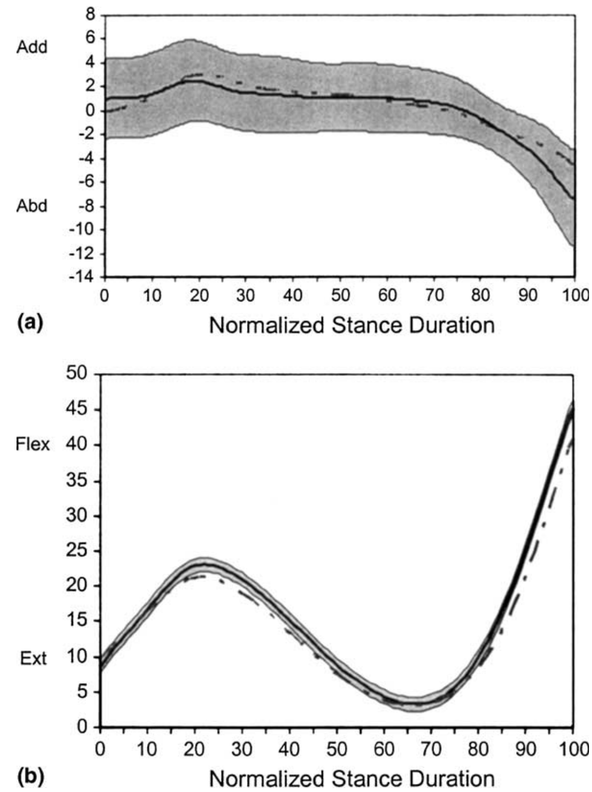

Fig. 1. Knee angular motion in the (a) frontal and (b) sagittal planes. The solid black line represents the average for males, the dashed black line the average for females. The gray area indicates the standard deviation for males ±1.

Loading response duration was significantly shorter for females in the locked condition. Though not reaching the significance level, loading response duration for females was shorter in the anterior and lateral conditions. As a consequence, differences between genders were more pronounced when examining excursion as a function of time.

4. Discussion

The purpose of this study was to compare lower extremity kinematics between genders during perturbed walking. We hypothesized females would demonstrate distinct lower extremity movement patterns which have been associated with ACL injury. The rationale for this hypothesis was that movement patterns are not isolated to specific athletic maneuvers, but instead represent global strategies that are repeated across a variety of activities. Our hypotheses that significant differences

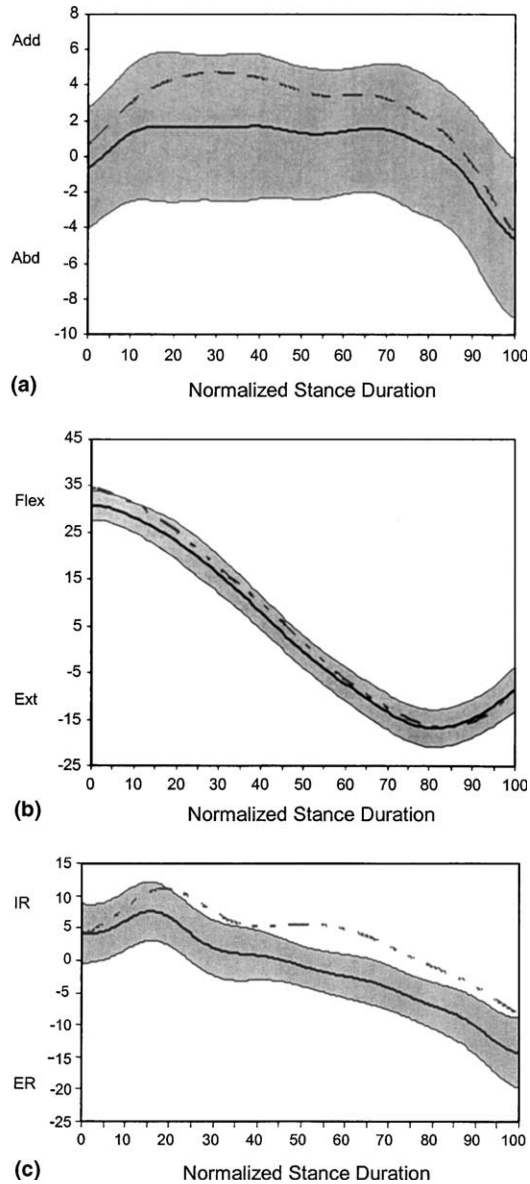

Fig. 2. Hip angular motion in the (a) frontal, (b) sagittal and (c) transverse planes. The solid black line represents the average for males, the dashed black line the average for females. The gray area indicates the standard deviation for males ±1.

would exist between genders for hip and knee kinematics at initial contact were not collectively supported with the results of this study. During the anterior walking condition, females demonstrated greater hip flexion at initial contact. Our hypothesis was females would make initial contact with less hip flexion compared to males, which is consistent with the results obtained by Kerrigan and colleagues (Kerrigan et al., 1998). The pattern of increased hip internal rotation, adduction, and knee abduction excursion and excursion rates in females are consistent with the findings of Ferber and colleagues (Ferber et al., 2003). Ferber and colleagues reported differences in peak values of hip internal rotation, adduction, and knee abduction, angular velocity and power absorption. We demonstrated similar findings through the use of kinematics by measuring differences in excursions and excursion rates during loading response.

Excursion patterns observed in our study were in partial support of our hypotheses. Females consistently demonstrated significantly greater excursion for frontal plane knee and transverse plane hip motions. Hip and knee movement collectively result in a *relative* positioning of the lower extremity that may potentially increase ACL strain and lead to injury. Females made foot contact with the knee near zero degrees in the frontal plane and the excursion was into varus; knee excursion into varus is normal in walking (Perry, 1992). Excessive excursion in either direction of the knee in the frontal plane, however, produces undesirable ACL strain. Directional excursion for hip rotation among our subjects was variable. The females tended to move into internal rotation in the locked and anterior conditions while the males did not. In the lateral condition, however, all subjects but one moved into internal rotation during loading response, presumably to counter the lateral movement of the platform.

Our finding of earlier peak knee flexion (as indicated by the shorter loading response duration) among females is consistent with the work from previous studies. McLean and colleagues (McLean et al., 1999) observed earlier peak knee flexion among females during cutting; McLean and colleagues hypothesized males exhibited later knee flexion during cutting secondary to greater eccentric quadriceps activation during LRD. Schmitz and Thompson (2002) studied kinematics during single-leg jump landings and measured earlier muscle activation among females. Schmitz and Thompson proposed it was the earlier onset of muscle activation that resulted in earlier peak knee flexion for females. We propose the rate (and also the amount) of joint excursion exhibited by females is influenced by multiple variables that are both related and independent of gender.

The first two factors contributing to more rapid excursion rates for female subjects include temporal and structural features. There was no significant difference between groups for gait velocity. Total lower extremity limb length was lower for our female subjects. While not measured in the current study, it is a reasonable assumption that, at a fixed speed, subjects with a shorter limb length would have to ambulate at a higher cadence and would experience a shorter stance duration compared to taller subjects. Individuals with equal or greater joint excursion must go through their range of motion more rapidly as a consequence of the shorter stance duration. When applied to athletic maneuvers, the same logic would apply. When performing cutting maneuvers at comparable speeds, individuals with shorter limbs will most likely demonstrate an increased cadence and shorter stance duration, and therefore undergo a (relative) rapid joint excursion.

Another factor potentially contributing to joint excursion is the ability to generate muscular stiffness. Wojtys and colleagues measured sagittal (Wojtys et al., 2002) and torsional (Wojtys et al., 2003) tibial translation in size and activity matched males and females in relaxed and contracted states. Both genders were able to significantly decrease tibial translation in relaxed and contracted states. The percentage increase in knee stiffness was significantly greater in men compared to women, indicating women have decreased muscular protection capabilities than men. Both groups in the study by Wojtys and colleagues had comparable physiologic tibial translation, and regression analysis indicated strength was not the primary contributor to increased stiffness. Wojtys and colleagues concluded the decreased ability to generate muscular stiffness and correspondingly decrease tibial translation was gender specific. Ferber and colleagues (Ferber et al., 2003) suggested that greater excursions were secondary to increased eccentric demands placed on the hip abductors in women compared to men as a result of their greater hip adduction angle and velocity. Our finding of a larger excursion rate during loading in women supports a concept of increased eccentric demand on the hip and knee musculature. While this is a logical inference made regarding muscle activity based on kinematics, kinetic and electromyography data would strengthen our conclusions.

If, as Wojtys and colleagues surmised, women have a diminished ability to generate muscular stiffness, the higher demand and lower capacity mismatch presents a potential injury scenario: the muscular stiffness is insufficient to control joint excursion and ACL strain is developed. Results from the perturbed walking model suggest there are multiple gender independent and dependent factors contributing to lower extremity kinematic differences between genders. These findings, in conjunction with previous running and walking studies, also support the notion that gender differences in movement patterns are not isolated to specific athletic maneuvers, but rather manifest themselves in lower level activities like walking. Application of these results to ACL injury prevention programs may enhance training by influencing global movement patterns rather than attempting to improve jumping and cutting mechanics.

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