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MEDIAL FOOT LOADING ON ANKLE AND KNEE BIOMECHANICS

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Abstract

Background—The incidence of anterior cruciate ligament (ACL) injuries among females continues at disproportionate rates compared to males, with research indicating inconclusive multifactorial causality. Data from previous retrospective studies suggest an effect of abnormal foot and ankle biomechanics on pathology at the knee, including the ACL.

Objective—To determine if a relationship exists between plantar foot loading patterns during normal gait and high risk biomechanics purported to increase risk of ACL injury.

Methods—Dynamic barefoot plantar pressure distribution was measured on 33 female collegiate soccer players. Groups were divided according to their predominant gait loading pattern (medial or lateral). Three dimensional (3-D) motion analysis was conducted during drop vertical jumps to assess vertical ground reaction force and discrete angle and joint moment variables of the lower extremities.

Results—No significant differences occured in sagittal or coronal plane knee joint kinematics and kinetics between the medial and lateral loading groups.

Discussion—Dynamic foot and ankle biomechanics during gait do not appear to be related to lower extremity kinematics or kinetics during landing in collegiate female soccer players.

Conclusion—The exact cause of the abnormal differences in female landing biomechanics has not been irrefutably defined. This study suggests no effect of foot and ankle biomechanics exists on the landing mechanics of female soccer players.

Keywords

anterior cruciate ligament; foot pressures; valgus; pronation

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INTRODUCTION

The incidence of anterior cruciate ligament (ACL) injuries among females is disproportionately greater in females compared to male athletes. Escalating numbers of ACL injuries in the last three decades has led to many investigations to determine the underlying factors for this important health disparity. The current evidence indicates that multiple factors exist regarding the causality of this disproportionate rate of ACL injuries.[1,2] Though prior studies have isolated solitary significant factors related to the increased rates in females, attempts continued to more clearly define a comprehensive understanding of the pathogenesis for these injuries, as well as to facilitate the development of preventative strategies to decrease the incidence of these injuries.[1,3-12] Most ACL injuries in females are non-contact in nature and occur during deceleration, cutting, pivoting, and jumping; either independently or in combination.[9,13] Prior work indicates that females sustaining ACL injuries demonstrate the previously mentioned actions with the lower extremity in a position of dynamic knee valgus at ground contact.[9,14] In this position of dynamic knee valgus, the lower extremity is described to demonstrate excessive coronal and transverse plane motion, with hip adduction and internal rotation, knee abduction with subsequent tibial rotation, and ankle eversion.[4,5,9,10,14-18] In addition, many studies have implicated deficient biomechanics and decreased neuromuscular control during activities which simulate the mode in which females sustain non-contact ACL injuries.[1,4,7-10,12,14,16,17,19-26]

Although the evidence is controversial, improper foot and ankle kinematics may influence more proximal joints and may also be a factor which underlies increased susceptibility to ACL tears.[27,28] The lower extremity functions in a closed kinetic chain during certain phases of activities such as ambulation, running, jumping, and cutting.[27,28] It is commonly accepted in biomechanics that structure dictates function; however, a lack of consistent support exists for structural variability as a measure of function during dynamic activities.[29] Pathological biomechanics of the foot have been implicated in the etiology of ankle, knee, hip, and even low back, pathology.[27,30] Some authors have suggested excessive pronation is an underlying factor contributing to the dynamic knee valgus position associated with the increased rates of ACL injuries.[30-32] Retrospective analyses by Becket et al,[30] Woodford-Rogers et al,[32] and Loudon et al[31] have demonstrated a significant correlation between excessive pronation and ACL injuries. The authors suggested excessive pronation with subsequent internal tibial rotation during the stance phase of gait limits the ability of the ACL to restrain the natural increase in anterior tibial translation and internal rotation torque increasing stress to the ACL, ligamentous laxity, and susceptibility to injury.

Positive correlations between abnormal foot and ankle biomechanics with tibiofemoral joint pathomechanics have also been previously reported. Trimble et al[17] demonstrated a significant correlation between sex and navicular drop as indicators of excessive anterior tibial translation measured by a KT 1000 athrometer©. Similarly, Coplan and colleagues[33] demonstrated that females with excessive pronation exhibited excessive passive transverse plane rotation at the tibiofemoral joint. Female athletes are also reported to exhibit greater hip adduction, knee valgus, and foot pronation during landing. Results from one study suggested knee valgus and foot pronation to be the most significant factors affecting the increased motion in the coronal plane.[24]

Physical therapists and other health professionals often visually assess barefoot gait and implement their interventions, such as orthotic prescription, based upon their observations of an abnormal foot loading pattern. Technology continually evolves and enhances the ability to more accurately and objectively assess biomechanics. These associations of the effects of abnormal foot and ankle kinematics with pathomechanics and injuries in more proximal kinetic chain structures warrants further investigation by an accurate and objective measurement system to more precisely define the relationships between the biomechanics of the foot/ankle complex and the tibiofemoral joint. Therefore, the primary purpose of this study was to determine if subjects with more medial foot loading patterns compared to lateral foot loading patterns, as measured by the emed-x system (Novel GMBH, Munich), would correlate with differences in ankle and knee coronal plane motion and torque measures during a drop vertical jump task. The hypothesis to be tested was if greater ankle eversion and knee abduction motion and torque, measured during a drop vertical jump, would occur in female collegiate soccer players which have more medial as compared to lateral foot loading patterns.

METHODS

Subjects

A power analysis was performed a priori with sagittal plane kinematic measures in the sample population.[1] Based on the group differences measured during the drop vertical jump performance, it was determined that in order to achieve 80% power (alpha level 0.05) a minimum of 31 measures were required in each group. Based on these analyses, 33 subjects (66 legs) were recruited to participate in the current investigation. Thirty-three female collegiate soccer players (Division I and III) volunteered to participate in the study. The mean age of the subjects was 19.8 + 1.2 years, height was 165.1 + 5.9 cm, and mass was 62.6 + 8.3 kg. All participants read and signed informed consent approved by the Institutional Review Board of the Cincinnati Children's Hospital Medical Center.

Barefoot Plantar Pressure Measures

Dynamic barefoot plantar pressure distribution was obtained on each subject. The subjects were positioned on a 6 m walkway. An emed-x system was mounted in the middle of the walkway and level to the surface. The platform consisted of a 48×32cm matrix of capacitive sensors (4 sensors/cm2) collected at 100Hz. Subjects were instructed to walk normally at a self-selected speed[34] until five successful trials on each side were collected. A trial was accepted if the entire foot was within the sensor area. Morag and Cavanaugh35 and Cavanaugh et al36 reported on the reliability of foot pressures to represent differences in foot structure during dynamic activity. Most importantly, they reported the arch structure reliably dictated the function of the foot and pressures under the midfoot during gait.35,[36] Hughes et al[37] demonstrated excellent reliability of the emed-f system, which has the same technology as the emed-x system, as high as .91, when the mean result of three trials was utilized. The authors recommended utilizing three trials or more to obtain reliable data as reliability increased reciprocally with the number of trials analyzed.[37] Gurney et al[38] demonstrated the between day reliability of the emed technology to be of high reliability with an average ICC value of 0.85.

Motion Analysis

Each subject was instrumented with 37 retroreflective markers placed on the sacrum, left posterior superior iliac spine (PSIS), sternum, and bilaterally on the shoulder, elbow, wrist, anterior superior iliac spine (ASIS), greater trochanter, mid thigh, medial and lateral knee, tibial tubercle, mid shank, distal shank, medial and lateral ankle, heel, dorsal surface of the midfoot, lateral foot (5th metatarsal), and toe (between 2nd and 3rd metatarsals). A static trial was first collected in which the subject was instructed to stand still and to allow for alignment with the laboratory coordinate system. This static measurement was used as each subject's neutral (zero) alignment; subsequent kinematic measures were referenced in relation to this position. Each subject performed a drop vertical jump (DVJ) which consisted of starting on top of a 31 cm box with their feet positioned 35 cm apart (distance measured between toe markers).[10] They were instructed to drop directly down off the box and immediately perform a maximum vertical jump. Three successful trials were recorded for each subject.

Trials were collected with EVaRT (Version 4, Motion Analysis Corporation, Santa Rosa, CA) using a motion analysis system consisting of eight digital cameras (Eagle cameras, Motion Analysis Corporation, Santa Rosa, CA) positioned in the laboratory and sampled at 240 Hz. Prior to data collection the motion analysis system was calibrated as previously described Cowley et al.[39] Two force platforms were sampled at 1200 Hz and time synchronized with the motion analysis system. The force platforms were embedded into the floor and positioned 8 cm apart so that each foot would contact a different platform during stance phase of the drop vertical jump.

Data Analysis

Motion analysis data were imported into Visual3D (Version 3.65, C-Motion, Inc. Germantown, MD) and MATLAB (Version 7.0, The Mathworks, Natick, MA) for data reduction and analysis. Three-dimensional Cartesian marker trajectories from each trial were filtered through a low-pass fourth order Butterworth filter at a cutoff frequency of 12 Hz. Three dimensional (3D) joint angles were calculated for both the left and right side according to the cardan/euler rotation sequence.[40] To minimize possible peak impact errors in joint moment calculations, the force plate data were filtered through a low-pass fourth order Butterworth filter at a cutoff frequency of 12 Hz.41 These data were used with the kinematic data to calculate joint moments using inverse dynamics.42 Net external moments are described in this paper and represent the external load on the joint.

The vertical ground reaction force (VGRF) data were utilized to calculate initial contact with the ground immediately after the subject dropped from the box. Initial contact was defined when VGRF first exceeded 10 N. Toe off was subsequently calculated after initial contact when the VGRF fell below 10 N. Kinematic and kinetic data were normalized to 100% of stance phase (between initial contact and toe off). The following discrete angle and joint moment variables were calculated during stance phase for each lower extremity: maximum ankle dorsiflexion, maximum knee flexion, maximum and minimum ankle inversion/ eversion, and knee abduction/adduction.

Pressure distribution trials were analyzed within a commercial software package (Projects, Novel GMBH, Munich). From each walking trial the midfoot and forefoot were combined and subdivided into separate medial and lateral regions based on an algorithm dividing the entire foot previously described by Cavanagh et al.43 The force time integral within the two regions were calculated and then divided by the total foot force time integral in order to determine the relative load in each region.44 The region (medial or lateral) which underwent greater loading during the walking trials was used to stratify the foot as a medial (MLP) or lateral (LLP) load pattern. Each leg from the box drop trials was analyzed separately based on the load pattern from the initial plantar pressure analysis.

Statistical Analysis

Statistical means and standard deviations were calculated for each subject. A one-way analysis of variance (ANOVA) was utilized to determine the effect of plantar load (MDL and LDL) of the foot on each dependent variable. An alpha level of .05 was selected to identify statistical significance. Statistical analyses were conducted in SPSS (Version 15.0, Chicago, IL).

RESULTS

Analysis of the 33 subjects (66 extremities) plantar pressure distributions revealed 34 lateral load patterns and 32 medial load patterns. The relative load for the medial plantar region (MLP = 33.8 ± 4.5%; LLP = 26.8 ± 3.6%) and lateral plantar region (MLP = 26.4 ± 5.5%; LLP = 37.6 ± 5.5%) were s i g n i f i c a n t l y different between groups (p<0.001). Ensemble average knee joint angles (abduction/ adduction) during landing are shown in Table 1. No significant differences existed in coronal plane knee joint kinematics during landing between the medial (MLP) and lateral (LLP) load pattern groups. The LLP mean knee abduction angles were 8.8 ± 7.4 o compared to MLP mean abduction angles of 6.3 ± 5.4o (p= 0.125). The LLP mean knee adduction angles were −0.1 ± 6.6 o compared to MLP mean adduction angles were 1.3 ± 4.6 o (p= 0.325).

Ensemble average ankle joint angles (eversion/inversion) during landing are also shown in Table 1. There were also no significant differences in coronal plane ankle joint kinematics during landing between the MLP and LLP groups. The LLP mean ankle eversion angles were 7.1 ± 4.2 o as compared to the MLP mean eversion angles of 6.3 ± 5.8 (p= 0.540). The LLP mean ankle inversion angles were 8.6 ± 4.4 o compared to mean MLP inversion angles, which were 9.4 ± 4.3o (p = 0.465).

Ensemble average knee joint angles (flexion) during landing are shown in Table 2. There were no significant differences in sagittal plane knee joint kinematics during landing between the LLP and medial foot load patterns. The LLP mean knee flexion angles were 80.1 ± 10.3 o as compared to MLP mean flexion angles of 82.1 ± 10.8 o (p= 0.440). Ensemble average ankle joint angles (dorsiflexion) during landing are shown in Table 2. No significant differences were found in sagittal plane ankle joint kinematics during landing between the MLP and LLP. The LLP mean ankle dorsiflexion angles were 25.9 ± 5.59 o compared to MLP mean dorsiflexion angles of 25.7 ± 5.48 o (p=0.919).

Ensemble average knee joint moments (abduction/adduction) during landing are shown in Table 3. There were no significant differences in coronal plane knee joint kinetics during landing between the MLP and LLP. The LLP mean knee abduction moments were −0.504 ± 0.288 N.m/kg) as compared to MLP mean abduction moments of −0.388 ± 0.230 N.m/kg (p= 0.077). The LLP mean knee adduction moments were 0.069 ± 0.162 N.m/kg compared to MLP mean adduction moments of 0.100 ± 0.110 N.m/kg (p= 0.362). Ensemble average ankle joint moments (eversion/inversion) during landing are shown in Table 3. There were no significant differences in coronal plane ankle joint kinetics during landing between the MLP and LLP. The LLP mean ankle eversion moments were −0.455 ± 0.202 N.m/kg compared to MLP mean eversion moments of −0.385 ± 0.197 N.m/kg (p=0.161). The LLP mean ankle inversion moments were .077 ± 0.098 N.m/kg as compared MLP mean inversion moments of 0.115 ± 0.120 N.m/kg (p=0.161).

Ensemble average knee joint moments (flexion) during landing are shown in Table 4. No significant differences existed in sagittal plane knee joint kinetics during landing between the MLP and LLP. The LLP mean knee flexion moments were −1.92 ± 0.578 N.m/kg as compared to MLP mean flexion moments of −1.70 ± 0.454 N.m/kg (p=0.080). Ensemble average ankle joint moments (dorsiflexion) during landing are shown in Table 4. No significant differences occurred in sagittal plane ankle joint kinetics during landing between the MLP and LLP. The LLP mean ankle dorsiflexion moments were 1.56 ± 0.465 N.m/kg as compared to MLP mean dorsiflexion moments of 1.42 ± 0.349 N.m/kg (p=0.159).

DISCUSSION

The primary purpose of this study was to determine if subjects with primarily medial foot loading patterns compared to lateral foot loading patterns would exhibit differences in ankle and knee coronal plane motion and torque measures. Our original hypothesis was females that demonstrated increased medial foot loading during gait as measured by the emed-x pressure system would also demonstrate greater deviations in ankle and knee angles and moments in the coronal plane during a dynamic landing task. However, no significant differences were found with ankle and knee measures between lateral and medial foot loading patterns identified from foot pressure and motion analysis. Contrary to our original hypothesis, the results from this study suggest additional mechanisms are likely responsible for contributing to dynamic knee valgus apart from foot pressure during gait.

Previous studies have examined the effects of the structure and function of the foot and ankle complex on tibiofemoral biomechanics and ACL injury.[30-32,45] Studies by Becket et al,[30] Loudon et al,[31] and Woodford-Rogers et al[32] reported that subjects with ACL injuries demonstrated a significant association with excessive foot pronation. These retrospective studies presented useful data but the retrospective design of the studies caused speculation and limited their power. These studies lacked the ability to determine if the condition of the foot was causative or resultant of the injury. Similar to the current study, Smith et al[46] found no significant difference in navicular drop height between groups of ACL injured subjects and uninjured subjects. Contrary to the current study and Smith et al,[46] biomechanical studies by Kernozek et al[24] and Ford et al[47] demonstrated that females landed with increased coronal plane ankle kinematics and kinetics during drops jumps as compared to their male counter parts. These authors concluded that the ankle joint may demonstrate this excessive motion to compensate for increased force absorption. Ankle motion may have an influence on the less advantageous landing biomechanics in females and more in depth exploration should be considered.[24,47] In a recent study, McLean et al[48] examined the effects of fatigue on the differences in landing mechanics between males and females. Females demonstrated greater peak stance ankle supination in conjunction with greater dynamic knee valgus measures than their male counterparts. The authors attributed the difference to possible skill level differences or error in obtaining position of the talocrural joint axis. The lack of definitive certainty derived from the studies previously mentioned highlights the necessity for the more detailed analysis conducted during the present study.

Clinical studies, patient evaluations, and orthotic assessment, clinicians assess the function of the medial longitudinal arch utilizing static and dynamic clinical measurements.[49,50] The dynamic function of the medial longitudinal arch is commonly assessed in clinical situations, with visual observation and less frequently, video analysis. This study was the first to utilize both of these highly technological dynamic measurements of foot pressures and motion analysis of lower extremity biomechanics to determine the association of the foot and ankle complex biomechanics with the tibiofemoral joint during dynamic activities. In the present study, foot characteristics were measured with the emed-x system to determine the differences in foot pressures during the dynamic activity of gait and then correlated with biomechanical data obtained during landing. The present findings differed from those of the clinical studies by Woodford-Rogers et al,[32] Beckett et al,[30] and Loudon et al.[31]

The exact cause of the abnormal differences in female landing biomechanics has yet to be irrefutably defined and is likely multifactorial. While the necessity for additional research on the etiology and prevention of ACL injuries continues to perpetuate, current evidence supports that identifying and subsequently correcting disadvantageous biomechanics and neuromuscular control will have a significant effect on decreasing the risk for ACL injuries.

CONCLUSION

The results of the present study did not demonstrate any statistical significance regarding the difference in foot loading pressure patterns during gait and dynamic biomechanics during landing in female collegiate soccer players. Data collection with an in-shoe pressure distribution system during drop landings may have proved useful in determining what pressures occurred under the foot during these landings. With this information a correlation may have been established between lower extremity biomechanics and foot pressures during drop jumps. However, clinically, dynamic assessment of an individual's foot mechanics is most often through visual observation of gait and not drop jumps. Future studies may concentrate on utilizing the emed-x pressure system or an in-shoe system during the drop jumps to evaluate the differences in foot loading pressures that occur during landing.

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Table 1

Coronal Plane Knee and Ankle Kinematics

	Descriptive Statistics			**Univariate Statistical Significance**
	Lateral Load Pattern	**Medial Load Pattern**	**Total**	**Task**
Ankle Eversion/Inversion (*)				
Max Eversion	− 7.05 (4.17)	− 6.28 (5.82)	− 6.67 (5.01)	$F_{1,65}$ = .379, p= 0.540
Max Inversion	8.64 (4.35)	9.43 (4.33)	9.03 (4.32)	$F_{1,65}$ = .539, p = 0.465
Knee Abduction/Adduction (*)				
Max Abduction	− 8.81 (7.43)	− 6.31 (5.41)	− 7.60 (6.60)	$F_{1,65}$ = 2.415, p= 0.125
Max Adduction	− 0.131 (6.61)	1.27 (4.58)	0.546 (5.72)	$F_{1,65}$ = .985, p= 0.325

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Table 2

Sagittal Plane Knee and Ankle Kinematics

	Descriptive Statistics			Univariate Statistical Significance
	Lateral Load Pattern	**Medial Load Pattern**	**Total**	**Task**
Ankle Dorsiflexion (*)				
Max Dorsiflexion	25.9 (5.59)	25.7 (5.48)	25.8 (5.49)	$F_{1,65} = 0.011$, p= 0.919
Knee Flexion (*)				
Max Flexion	− 80.1 (10.3)	− 82.1 (10.8)	− 81.1 (10.5)	$F_{1,65} = 0.603$, p= 0.440

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Table 3

Coronal Plane Knee and Ankle Kinetics

	Descriptive Statistics			Univariate Statistical Significance
	Lateral Load Pattern	**Medial Load Pattern**	**Total**	**Task**
Ankle Eversion/Inversion (N'm/kg)				
Max Eversion	− .455 (.202)	− .385 (.197)	− .421 (.201)	$F_{1,65} = 2.007$, p= 0.161
Max Inversion	.077 (.098)	.115 (.120)	.095 (.110)	$F_{1,65} = 2.011$, p= 0.161
Knee Abduction/Adduction (N'm/kg)				
Max Abduction	− .504 (0.288)	− .388 (0.230)	− .448 (2.66)	$F_{1,65} = 3.225$, p= 0.077
Max Adduction	.069 (0.162)	.100 (0.110)	.084 (0.139)	$F_{1,65} = 0.844$, p= 0.362

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Table 4

Sagittal Plane Knee and Ankle Kinetics

	Descriptive Statistics			Univariate Statistical Significance
	Lateral Load Pattern	**Medial Load Pattern**	**Total**	**Task**
Ankle Dorsiflexion (N'm/kg)				
Max Dorsiflexion	1.56 (0.465)	1.42 (0.349)	1.49 (0.416)	$F_{1,65}$ = 2.033, p= 0.159
Knee Flexion (N'm/kg)				
Max Flexion	−1.92 (0.578)	−1.70 (0.454)	−1.82 (0.530)	$F_{1,65}$ = 3,171, p=0.080