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RESEARCH ARTICLE

Gender-specific anterior cruciate ligament – gait forces

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Abstract: The purpose of this study was to investigate gender-based differences in gait biomechanics and to evaluate those effects on forces generated on the ACL during walking. Estimation of gender-specific ACL forces in the frontal plane can provide a better understanding of the biomechanical patterns underlying higher female injury risk. The present study used a sample from the Fels Longitudinal Study to test the hypothesis that there are significant gender-differences in frontal plane ACL loading during walking. A cross-sectional sample of 178 participants, including 79 males and 99 females was used to evaluate differences in gait kinetics. Females walked at higher cadence with narrower steps ($P < 0.05$). No difference was observed in the peak flexion force and knee rotation moment between males and females ($P = 0.51$ and 0.07), respectively. Peak abduction moment was significantly lower among females than in males ($P = 0.05$). A regression equation was developed which considers a person's weight and height in addition to forces which could give better estimate of the forces acting on the ligament. The peak force acting on the ACL during walking reaches as high as 0.44 of BW, regardless of gender.

Keywords: gait, anterior cruciate ligament, kinetics, gender differences

1 Introduction

Anterior cruciate ligament (ACL) provides stability to the knee, and is injured among adolescents and young adults [1] with long term health consequences including premature and disabling osteoarthritis [2]. Rates of injury are particularly high among individuals participating in organized sports, and within that group, females athletes are at significantly greater risk for injury compared to their male counterparts [3, 4]. Extensive literature search which resulted in 33 studies consisted of ACL tear incidence. Meta-analytic principles were used to compare the ACL incidences with respect to gender and sport. Literature also used gender and sex interchangeably, and used in this paper to be consistent with previous citations. Given that there is no difference between incidence of ACL injuries among basketball and soccer players (p =0.80), the ratio of the incidence between females and males can be seen reach as high 5.33 and 9.63 for basketball and soccer, respectively. Risk for ACL injury among females in these sports is statistically significant when compared to males, with p values of 0.03 and 0.02 for basketball and soccer, respectively.

Sex differences in gait patterns are well-known for a given walking speed, females tend to have shorter strides and slower gait speed than their male counterparts [5]. Moreover, females generate higher joint power at the hip and knee joints during the late stance phase of the gait cycle, and have greater knee and hip flexion moments than do young men [6]. Since the ACL is located within the knee joint, its loading can be inferred from these and other in vivo measurements of bony motion at the knee [7]. Only a limited number of studies have calculated knee ligament forces during walking gait, and have focused on forces in the sagittal plane [8, 9]. Other work suggests, however, that ACL injury risk is more closely dependent upon variation in frontal (coronal) plane biomechanics [10, 11]. As such, estimation of sex-specific ACL forces in the frontal plane can provide a better understanding of the biomechanical patterns underlying higher female injury risk. The present study used a sample from the Fels Longitudinal Study to test the hypothesis that there are significant sex-differences in frontal plane ACL loading during walking. A better understanding of sex-specific patterns of ACL loading during lower-intensity habitual movement patterns such as walking can provide insight into sex differences in biomechanical responses to higher-intensity conditions. The focus of this paper is to investigate sex differences in gait biomechanics and to evaluate their effects on forces applied to the ACL during walking.

2 Methods

2.1 Participants

A cross-sectional subset of participants from the Fels Longitudinal Study (FLS) was selected for inclusion in the present study. The FLS is the currently the world's largest and longest-running longitudinal study of human growth, development and aging [12]. Participants are primarily of European descent and live mainly in or near southwest Ohio. The FLS is a study of normal population variation, since participants are/were not recruited on the basis of any health condition or disease.

To be included in the present study, FLS participants had to have at least one gait analysis test in our database, between the ages of 21 and 50 years at the time of testing to exclude gait variation due to growth or aging. To ensure that the present analysis included only people with normal gait, participants were excluded for obesity (BMI $>$ 30.0); chronic musculoskeletal conditions; acute lower limb, pelvic, or vertebral skeletal or soft tissue injury $\leq$ 1 year prior to testing; toe-walking; or prescription shoe inserts. If participants had more than one gait test in the database, only the most recent available and eligible test was included. This resulted in a cross-sectional sample of 178 participants, including 79 males and 99 females.

2.2 Data collection

All data collection occurred at the Lifespan Health Research Center (LHRC), Wright State University . All study procedures were approved by the Wright State University Institutional Review Board, and each participant gave his or her informed consent prior to any testing. Anthropometric measurements were taken using standard methods [13] on barefoot participants wearing light clothes, and included stature, body weight, sitting height, and bicristal breadth; BMI was determined from weight and height; subischial leg length was calculated as: height – sitting height.

2.3 Statistical analysis

Before the inter-subject comparison of gait analysis data, a data normalization to remove the effect of body size was essential [14]. In this study, ad hoc method was used; speed (cm/min) was divided by $\sqrt{L_o}$ (L_o: leg length), and kinetic data were divided by body weight and height. The ANCOVA (Table 1) with height as the covariate, and sex as the fixed factor were practiced for leg length, speed, stride length, step width, and ratio of step width to pelvic width.

Table 1 FANOVA analysis estimating the risk of ACL injury

Source	Nparm	DF	Sum of Squares	F Ratio	Prob > F
Weight	1	1	4562811	168.14	< 0.0001*
Flexion force	1	1	105329	3.88	0.0490*
Rotation moment	1	1	228709	8.43	0.0037*
Abduction moment	1	1	45275772	1668.43	< 0.0001*

Analyses were performed using JMP version 11.0 (SAS Inc., Cary, NC) and were two sided with α=0.05 as the significance level. The two genders were analyzed separately and were compared.

2.4 ACL force calculation

Monte Carlo Simulations (N = 2000) were performed to predict the effect of subject's variability in the kinetic patterns during the gait and a hypothetical 50% variability in the quadriceps and hamstrings activation on peak ACL force (Table 2). The force on the ACL can be found by simultaneously solving three equations of dynamic equilibrium, which includes forces in equilibrium on the tibia in both inferior and superior directions, and equilibrium of moments about the joint center [15, 16]. Therefore, the force on the ACL can be estimated based on the following linear Equation (1):

$$\text{ACL Force} = 1.66 \times \text{weight} + 103.56 \times \text{flexion force} + 356.76 \times \text{rotation moment} + 52.86 \times \text{varus moment} \quad (1)$$

Table 2 Regression parameter to determine the ACL force using Monte Carlo Simulations

Term	Estimate	Std Error	t Ratio	p-value
Intercept	39.66753	29.69584	1.34	0.1818
Weight	1.668482	0.090787	18.38	< 0.0001
Flexion Force	103.5677	25.24477	4.10	< 0.0001
Rotation Moment	356.7613	21.1592	16.86	< 0.0001
Varus Moment	52.86783	1.295256	40.82	< 0.0001

Where F_{AP} is the anterior-posterior joint load calculated from sagittal plane joint reaction and muscle flexion-extension (quadriceps and hamstring) forces, and M_{VV} and M_{IE} are varus-valgus (abduction-adduction) and internal-external joint rotation moments, respectively. The linear regression equation (Equation (1)) had considered not only the forces during the gait cycles but also included the person's weight.

3 Results

Participant demographics are listed in Table 3. Average ages were 35.3 and 35.9 for males and females, respectively. There were no statistically significant sex differences in age, weight, or forward velocity. Females were significantly shorter than males ($P < 0.05$), and walked at higher cadence with narrower steps ($P < 0.05$).

Table 3 Demographics for healthy volunteers

	Males (N = 79)	Females (N = 99)	*P* -value (two sample t-test)
Age (Years)	35.2 (SD 8.5)	35.9 (SD 8.7)	0.5
Height (cm)	179.7 (SD 7.3)	166 (SD 6.8)	< 0.05
Weight (kg)	78.6 (SD 12.2)	65.1 (SD 10.13)	< 0.05
BMI (kg/m^2)	24.2 (SD 3.1)	23.5 (SD 2.9)	0.2
Gait speed (cm/second)	126.4 (SD 13.7)	128 (SD 13.1)	0.4
Cadence (/second)	108.7 (SD 6.5)	117.9 (SD 7.0)	< 0.05
Step width (cm)	11.4 (SD 2.9)	10.1 (SD 1.8)	< 0.01

We focus on gender differences in maximal forces and moments here, since these in theory place the greatest strain on the ACL. Peak flexion force was observed during mid-stance in both sexes, reaching as high as 0.110 ± 0.04 N/kg and 0.106 ± 0.04 N/kg in males and females, respectively (P = 0.51). The highest magnitude knee rotation moment (internal rotation) also occurred during mid-stance in both sexes. The value was smaller in females than in males, but the difference was not significant (-0.231 ± 0.10 Nm/kg vs. -0.258 ± 0.10 Nm/kg, respectively; P = 0.07). Finally, peak abduction moment occurred during mid-stance in both sexes, and was significantly lower in females than in males (0.408 ± 0.268 Nm/kg vs. 0.508 ± 0.381, respectively; P = 0.05).

3.1 ACL forces

The ACL is loaded during the entire stance phase of the gait cycle. The higher forces in the leg are dependent upon which leg is dominant, regardless of the gender. For example, in subjects with their right leg in the first quarter of the gait cycle had observed a peak ACL force of 300 N (Figure 1), in the swing phase of the cycle force will likely decrease. Peak ACL force in the left leg would reach as high as 150 N, during the swing phase of the cycle. Similarly, the opposite forces were observed in a patient who have started off with their left leg.

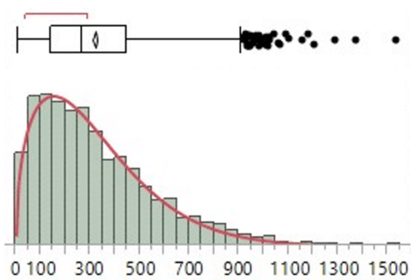

Figure 1 Distribution of ACL forces

ACL forces among females is lower when compared their male counterparts. The forces experience in both legs are quite similar (Figure 1). The peak ACL force was 221 N which had occurred in the stance phase in the gait cycle. The forces in both legs were similar. Regardless of the gender, the peak ACL force observed during normal walking had reached as high as 0.44 times BW (Figure 2).

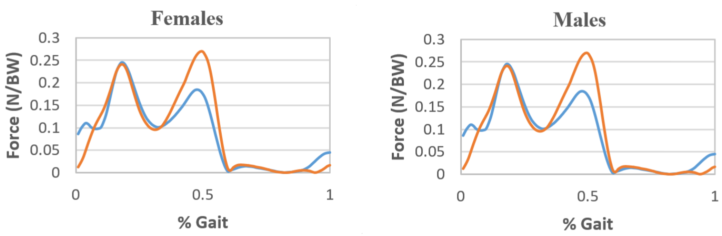

Figure 2 ACL force per kg body weight in each gender. BW = body weight. The red line represents the dominant, and the blue line the non-dominant leg.

4 Discussion and conclusion

An attempt was made to discuss and compare the pattern of forces transmitted to the ACL in normal gait seen by healthy volunteers. The analysis focused on body motions, and leg muscle forces which were obtained from gait analysis among healthy individuals.

The major limitation for this analysis was the static equilibrium assumed in the calculation of ligament forces, similarly the effects of centrifugal and inertial forces neglected. During the stance phase of the gait, the muscle and gravitational forces play a key role in the forces transmitted to the lower limb joints [17].

In our study we have noticed differences in the anthropometrics, stride characteristics and walking biomechanics between males and females, that were quite consistent with literature [5, 6, 18]. Significantly higher stride length observed among males can be due to differences in height. This difference in stride length disappears when stride length normalized with height.

In our study, female physique was significantly smaller ($P<0.05$), but ANCOVA showed that their shorter leg length could not be one of the gender features of female gait. The unnormalized female pelvic width, which was as wide as of males, could be regarded as greater than that of males with the same height. Thus, females' wide pelvis can influence the large joints below pelvis, especially in coronal plane.

The female cadence, which was as great as males' in spite of their shorter leg length, did not seem to be influenced by the size of physical size. Also, ANCOVA showed that slower walking speed in female was due to their smaller physique, which could not be a gender feature of female gait. Consequently, the spatiotemporal data did not show any additional effort for females to walk as fast as males. The female stride length, both the female step width, and its ratio to pelvic width were smaller than the males' even with ANCOVA. This could imply the females protrude their legs less vigorously with greater hip joint adduction. Since the female pelvis is wider than the males, it can be assumed that the anatomical or habitual differences can explain their step width to be narrower than the male counterparts.

From an anatomical perspective, greater knee joint valgus seen in females in this study might contribute to the narrower step width. Because most of our data were derived from gait analysis, we can show that females would have significantly greater abduction angles ($p < 0.005$). Similarly, males exhibit significantly larger adduction angles than the females ($p < 0.05$). Another reason for the narrower step width can be explained by the differences in the narrower base of support.

The loads experienced on the ACL can be explained by the shear forces that are acting at the knee joint. The shear force acting on the knee depends on a balance of muscle forces, ground reaction forces, and joint contact forces applied to the leg. Forces are experienced on the ACL when the shear forces are applied in an anterior direction. In the early stance phase, the shear forces are often caused by the patellar tendon, which is then later transmitted to the ACL. The shear forces caused by the patellar tendon are often large in the stance phase due to forces generated by quadriceps since the patellar tendon is located more anteriorly to the long axis of the tibia.

In this study, there were two peaks of ACL loading in stance with the maximum occurring in early stance right around the time of contralateral toe-off, consistent with findings from literature [19]. Even though the pattern of ACL loading was quite consistent with previous literature, there are significant differences in the predicted values of peak ACL forces. In this study, the peak ACL force observed during normal walking was 0.44 BW and is quite similar to loads of ACL seen in various other activities such as isokinetic knee extension exercise.

The regression equation, shown in Equation (1), and linear equation derived from [11] had utilized the three joint load components on the ACL which are independent and additive. However, Equation (1) had considered a person's weight as an additional factor that can play a role in the forces experienced on the ligament. There is no statistical difference in forces resulted from each of these equations (p = 0.56, and 0.41) for males and females, respectively. In addition, the correlation factors for both genders can be seen in Figure 3, with R^2 of 0.99, regardless of gender. However, the regression equation considers weight as additional factor providing more accurate forces acting on the ligament.

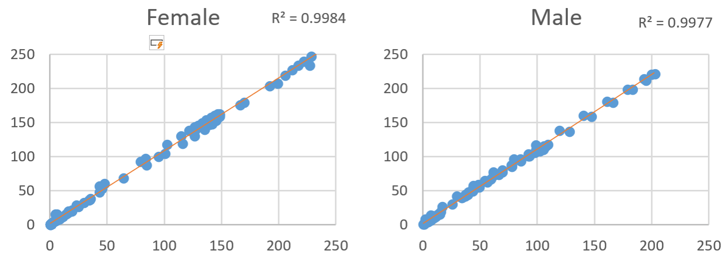

Figure 3 Predicted versus observed ACL forces using Equation (1)

A study in 1970 had used inverse dynamics approach to estimate muscle, ligament and joint forces at the knee during normal level walking [8]. The ACL has loaded during the stance phase of gait cycle, and the peak ACL force was 156N (˜ 0.25 body weight). More recent study which have used a three dimensional model of the whole body with dynamic optimization theory to estimate body segmental motions, ground reaction forces and leg muscle forces for one cycle of gait [9]. The peak ACL force had reached as high as 303 N and have occurred at the beginning of single leg stance.

Previous studies have utilized 3 dimensional models of the whole body in conjunction with dynamic optimization to calculate forces during the gait cycle. To date, this study is one of the first which used healthy volunteers to determine forces experienced during gait cycle. Results obtained from this research will be valuable in simulating 3D ACL stress analysis using various constitutive models determining critical conditions for tear.

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