Analysis of Gait Parameters and Knee Angles in Ultimate Frisbee Players:

Implications for Balance and Injury

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ABSTRACT

Analysis of Gait Parameters and Knee Angles in Ultimate Frisbee Players:

Implications for Balance and Injury

Ethan E. Nikcevich

Biomechanics research investigating gait and balance of ultimate frisbee players is an unexplored topic. Ultimate requires a wide range of motions that could improve balance and is also a sport prone to frequent injury. This study explores the impact of playing ultimate on gait parameters associated with balance and knee angles associated with joint injury. Gait trials were conducted on 8 ultimate players and 8 control participants between the ages of 18 and 23 to obtain total double support time, stance phase time, single support time, load response time, abduction-adduction (AA) angles, internal-external (IE) rotation angles, and flexion angles of the dominant leg's knee. Knee angles were obtained through the application of a Triangular Cosserat Point Element (TCPE) analysis for Soft-Tissue Artifact (STA) correction of knee kinematics. The gait parameters and knee angles were compared between ultimate players and control group participants using two-sample t tests. The results indicated that (1) playing ultimate may be used to improve balance, and (2) playing ultimate may reduce the range of IE rotation angles.

Keywords: Ultimate, Biomechanics, Triangular Cosserat Point Element, Motion Analysis, Gait

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Chapter 1

INTRODUCTION

Ultimate frisbee (ultimate) is a sport played by an estimated 7 million people across more than 80 countries [1], with more than 18,000 people competing on over 800 teams at the collegiate level in the United States [2]. Despite its popularity, the impact of the sport on the biomechanics of gait and balance has not yet been studied. In a study of college club sports, ultimate accounted for 31% of the 461 injury cases reported across all club sports, including soccer, basketball, baseball, and volleyball. Despite ultimate being a non-contact sport, only rugby caused more injuries [3]. Injuries caused by ultimate are often on the legs, with a study finding that 88% of injured players reported a lower extremity injury during the current season, and all players surveyed reported a prior lower extremity injury in their career [4]. The most frequently injured regions in ultimate players are the thigh and knee, with muscle strains to the thigh and joint sprains to the knee being the most common causes [5]. Despite high injury rates, ultimate has been an underrepresented sport in biomechanics research [6], [7], and the underlying causes of the relatively high injury rates are uncertain.

With knee injuries being common in ultimate, it is possible that competing in the sport impacts how players walk. Certain abnormalities in knee angles can be indicators of an increased risk of injury. Abnormal knee angles may be a direct biomechanical mechanism of knee osteoarthritis (OA) that alter contact load distribution during gait [8]. For example, studies have found that knee abduction angle offset [9], [10], lesser peak knee flexion angles [11], [12], and an external rotation angle offset [10] all increase the risk of incidence and progression of knee OA. By comparing the means and ranges of abduction-adduction (AA), internal-external rotation (IE), and the peaks and minimum flexion angles through a gait study, the knee angles of ultimate players can be assessed for abnormalities that could indicate risk of OA. Whether participating in ultimate would contribute to an increase or a decrease in these measures is uncertain, due to the detrimental frequency of knee injuries in the sport and the benefits of regular exercise.

Postural stability, or balance, is an involuntary mechanism for maintaining an upright posture and reducing fall risk. Nearly 30% of adults over the age of 65 fall each year, with 1 in 5 of these falls being fatal [13]. 48% of young adults reported falling at least once during a 16-week study [14]. Much like other aspects of physical fitness, balance can be improved to reduce fall risk through appropriate exercise. In a study on female college students, soccer players and gymnasts exhibited superior static and dynamic balance skills. It was hypothesized that soccer players and gymnasts have improved dynamic balance because they often use a single legged stance [15]. Another study found that varsity high school soccer players had better dynamic balance than non-athletes [16]. Young male adult soccer players were also found to have superior stability in one-legged stance tests [17].

The impact of ultimate on balance had not been studied before, but the motions used in ultimate suggest that the sport may improve balance. Ultimate requires that players run, jump, dive, make sharp cuts, and balance on their non-dominant leg when lunging and pivoting to throw the disc. Both forward and sideways lunges are used in ultimate, and are exercises proven to activate stabilizing muscles and develop balance [18]. Studies have shown that certain gait parameters can be indicators of improved balance. Double support time is the proportion of time that both feet touch the ground while walking, measured either in absolute time or as a percentage of each gait cycle [19]. An increase in double support time has been related to an increase in fear of falling [20] while a lower double support time is correlated with improved walking stability and a decreased risk of falling [21]. Injuries to the lower extremities increase double support time [22]. Double support time has been found to decrease as body mass decreases [23]–[25]. A study into the effects that unilateral sports have on the gait cycle found that sports such as soccer alter one's gait pattern, including a reduction in double support time [26]. There are other gait parameters (stance phase time, single support time, and load response time) that are also related to balance [27].

The goals of this study were to evaluate balance and knee angles in ultimate players compared to control group participants. Due to regularly playing a sport that requires running, cutting, jumping, and lunging, ultimate players were expected to have better balance compared to control group participants with the same age and BMI (body mass index) ranges. The objective of this study was to investigate if select gait parameters and knee angles would differ between ultimate and control group participants. More specifically, compared to control group participants it was hypothesized that:

1) Total double support time would be lower for ultimate players. Total double support time was defined as the percentage of the gait cycle (heel contact at 0% to the consecutive heel contact of the same foot at 100%) that both feet were contacting the ground as measured by force plate readings.

2) Stance phase time would be lower for ultimate players. Stance phase time measured the percentage of the gait cycle where the leg of interest was contacting the ground, beginning with heel contact and ending with toe off.

3) Single support time would be higher for ultimate players. Single support time measured the percentage of the gait cycle spent with only the leg of interest contacting the ground, beginning with toe off of the opposite leg and ending with heel contact of the same leg.

4) Load response time would be lower for ultimate players. Load response time measured the percentage of the gait cycle beginning with heel contact of the leg of interest and ending with toe off of the opposite leg.

5) AA and IE angle means and ranges would differ between ultimate players and control group participants.

6) Magnitudes of first and second peak flexion angles and magnitude of flexion minimum would differ between ultimate players and control group participants.

These hypotheses were explored for males and females individually and grouped together.

Chapter 2

METHODS

2.1 Participants

All protocols were approved by Cal Poly's Institutional Review Board and were designed to minimize risk to all participants.

2.1.1 Participant Demographics

A total of sixteen 18- to 23-year-olds participated in this study: eight ultimate players (four females [age 21 ± 1 years, BMI 20.5 ± 0.8] and four males [age 19.5 ± 1.5 years, BMI 23.1 ± 2.7]) and eight control group participants (four females [age 20.5 ± 2.5 years, BMI 20.7 ± 1.6] and four males [age 22 ± 1 years, BMI 21.2 ± 2.1]). All participants were right-leg dominant (Table 1). The eight ultimate players were recruited to participate in this study using an advertisement that was approved by the Cal Poly Institutional Review Board (IRB). All ultimate players had competed in a full season of ultimate and had no severe leg injuries in the preceding year. The eight control group participants' data were acquired from previous studies conducted within Cal Poly's Human Motion Biomechanics (HMB) Lab using the same methods as with the ultimate players.

Table 1. Participant data. All participants were right-leg dominant.

Participant ID	Ultimate (U) or Control (C)	Female (F) or Male (M)	Age (years)	BMI (kg/m^2)	Years of Ultimate Played
2023Mar05-01	U	F	20	21.2	2
2023Mar18-02	U	F	20	20.8	3
2023Apr22-01	U	F	22	19.7	7
2023May01-01	U	F	21	20.3	2
2023Feb20-02	U	M	19	20.4	2
2023Feb19-02	U	M	21	21.0	11
2023Apr15-01	U	M	20	25.8	3
2023Feb19-03	U	M	18	23.0	2
2023May04-01	C	F	22	22.3	0
2017Jun19-01	C	F	23	22.3	0
2017Jun27-01	C	F	20	19.1	0
2017Jul24-01	C	F	18	19.5	0
2016Nov05-01	C	M	22	19.1	0
2017Jun27-02	C	M	21	23.2	0
2018Apr13-01	C	M	23	22.9	0
2018Apr24-01	C	M	23	21.3	0

2.1.2 Inclusion and Exclusion Criteria and Informed Consent

All eligible participants were screened through a telephone call or in-person meeting. Inclusion criteria for all participants included adults between 18-23 years of age, and ultimate participants were required to have regularly trained and competed in the most recent season of ultimate. All participants were ensured to be in good health through a Physical Activity Readiness Questionnaire (PAR-Q). Exclusion criteria were any self-reported cardiovascular complications or lower-extremity injuries that had not fully recovered at the time of the experiment. After screening, participants came to Cal Poly's HMB Lab for the experiment, starting with informed consent, which was obtained from each participant. Participants completed standard pre-experiment tests to measure body height and body weight to calculate BMI as body mass divided by height squared (kg/m^2).

2.2 Experiments

2.2.1 Motion Analysis System

Motion capture experiments were conducted at the HMB Lab. This study used 10 optical infrared cameras (Motion Analysis Corp., Santa Rosa, CA, USA) to track retroreflective markers placed on anatomical landmarks and 4 force plates (AccuGait, AMTI, Watertown, MA, USA) embedded in a walking platform to capture the timing of heel contact and toe off during gait. Cortex software (Motion Analysis Corp.) was used to record gait experiments and process the captured static and dynamic trials. Marker trajectories were captured at a rate of 150 Hz, interpolated (third-order spline), and filtered (4^{th} order Butterworth filter, cutoff frequency 6 Hz) in Cortex.

2.2.2 Participant Preparation

Following informed consent, participants changed into skintight compression clothing. Height and mass were measured, and participants self-reported their dominant leg, defined as the preferred leg used to manipulate an object (such as to kick a soccer ball) or to lead out in movement [28]. 32 markers were placed on the participant based on an enhanced Helen Hayes model (Fig 1) to track body segmental motions during gait. An additional 14 markers were placed on the dominant leg for Triangular Cosserat Point Element (TCPE) analysis of knee angles, with 7 placed in a cluster on the thigh and 7 in a cluster on the shank [29].

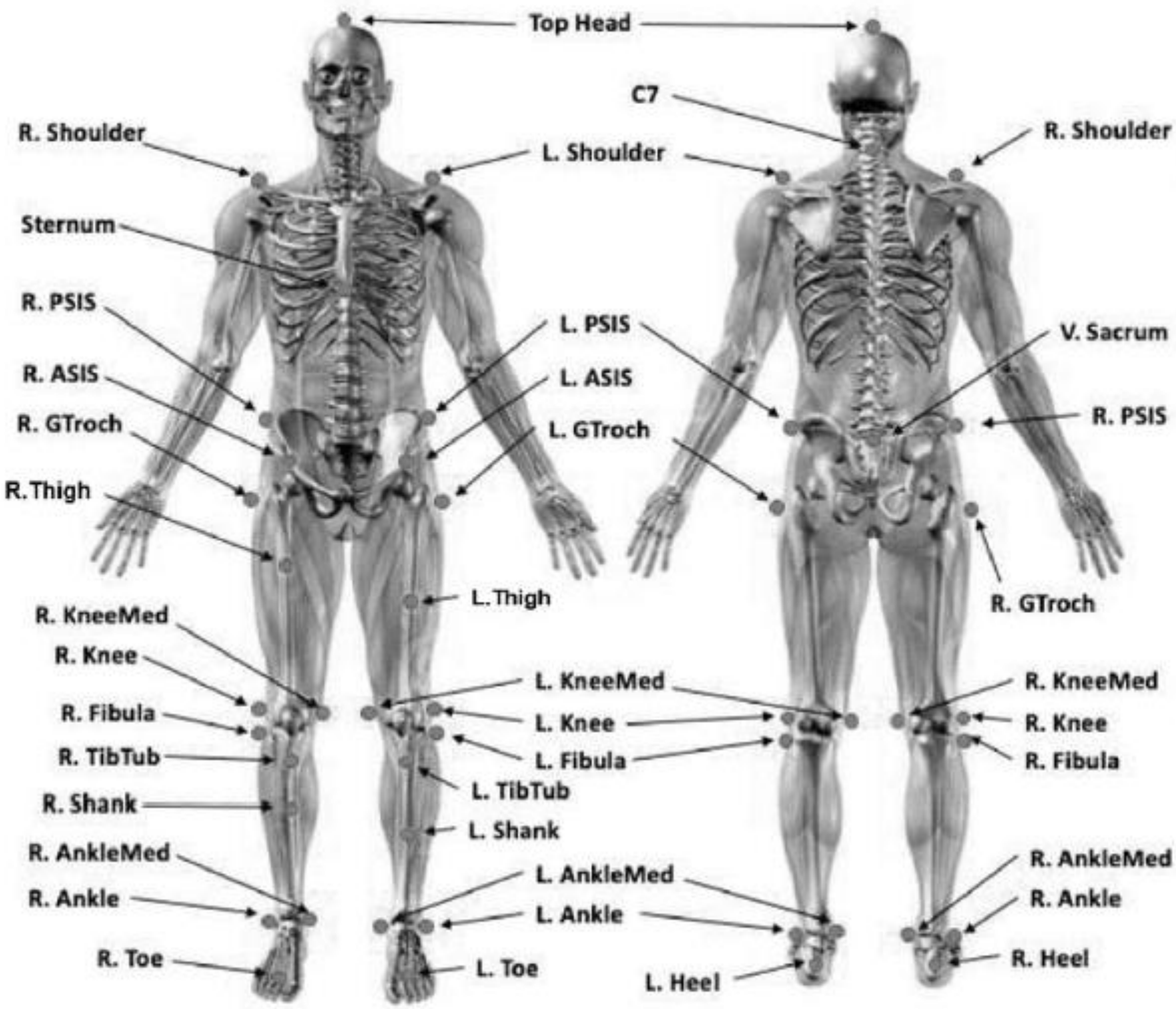

Figure 1: Enhanced Helen Hayes marker set consisting of 32 markers [30]

2.2.3 Static Pose Capture

In order to acquire participant-specific joint centers and initial joint orientations, static pose captures were recorded following marker placement. Participants stood with their feet shoulder-width apart and raised their arms to improve visibility of the markers (Fig. 2). Markers placed on the medial region and top of the head were removed before dynamic motion experiments to avoid markers falling off during gait.

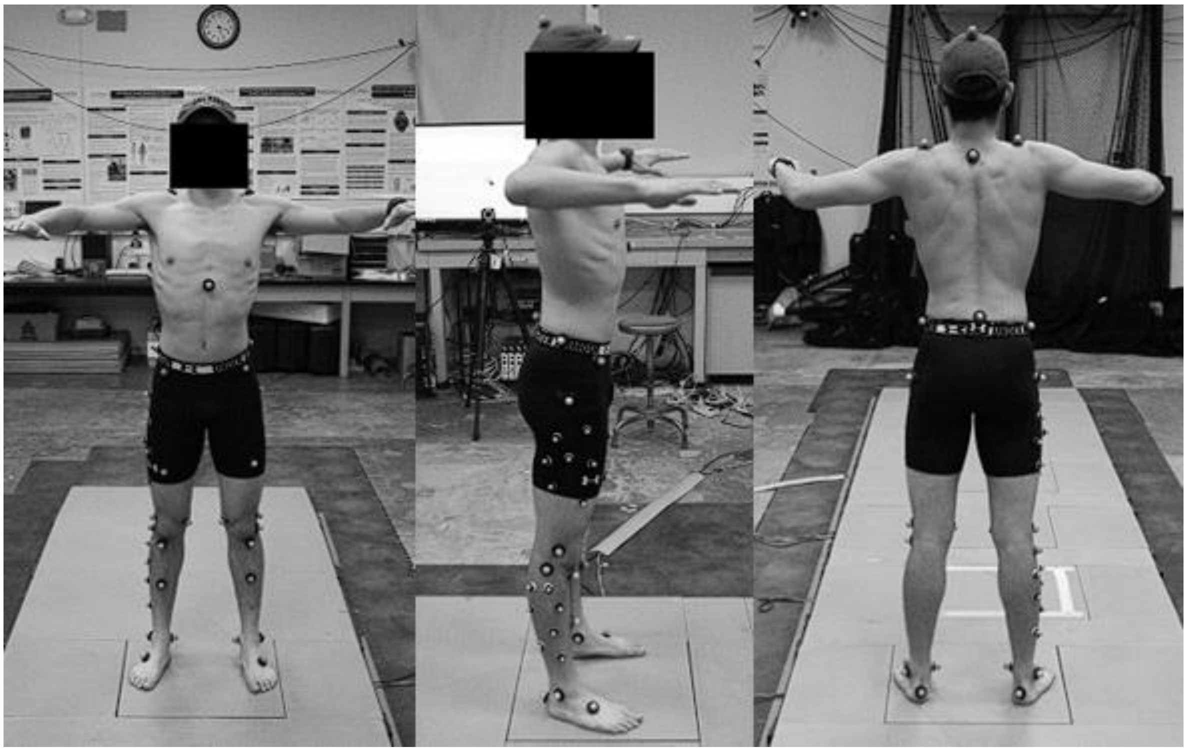

Figure 2: Static pose in the anterior (left), sagittal (middle), and posterior (right) views.

2.2.4 Dynamic Motion Capture

Gait trials consisted of participants walking barefoot at self-selected speeds for four steps across the force plates, along with several steps prior to the force plates for reaching a steady speed. One full gait cycle was defined as the first heel strike of the dominant foot on a force plate (0%) to the second heel strike of the dominant foot on another force plate (100%). A gait trial was permissible when each foot was placed fully atop each force plate during their stride. Participants repeated gait trials until three successful captures were recorded.

2.3 Cortex Processing

2.3.1 Static Trials

One static trial was post-processed in Cortex by identifying markers captured by the cameras according to the enhanced Helen Hayes marker set with thigh and shank marker clusters. For the gait parameters of interest in this study, markers on the anterior superior iliac spine, sacrum, greater trochanter, anterior and lateral midthigh, lateral and medial knee condyles, anterior and lateral shank, second metatarsal, posterior calcaneus, and lateral and medial ankle malleoli were identified and used in post-processing. Other markers associated with the enhanced Helen Hayes marker set were placed on the participant for possible future research but were not used in this study. Virtual markers were created from existing markers to locate joint centers of two consecutive segments.

2.3.2 Dynamic Trials

Three dynamic trials were post-processed in Cortex by identifying markers according to the enhanced Helen Hayes maker set with thigh and shank marker clusters. Cortex was used to obtain marker positions and force plate data.

2.4 Analysis

2.4.1 Knee Angle Parameters

Estimation of the orientation of rigid bones using markers placed on the skin was limited by the relative motion between the bone and the markers, known as the soft tissue artifact (STA) [31]. A method based on continuum mechanics was used to describe the kinematics of a cluster of markers affected by STA to better estimate the orientation of the bone segment. Each cluster was defined by Triangular Cosserat Point Elements (TCPEs) using all possible combinations of three markers in the cluster, containing a minimum of four markers [32], [33]. The TCPE method was used to identify instantaneous subsets of TCPEs with the least physical deformation to predict the best possible bone pose in the presence of STA.

AA, IE rotation, and flexion angles of the knee throughout a gait cycle were calculated using a floating axis coordinate system [34]. The thigh and shank segments had their own respective body-fixed coordinate system axes estimated from the TCPE method using marker clusters (Fig. 3). The thigh and shank coordinate systems were defined by XYZ and xyz axes respectively. The thigh coordinate system was defined with the Z-axis from the knee joint center to the hip joint center, the Y-axis as the cross product of the Z-axis and a temporary vector created from the medial to lateral knee marker, and the X-axis as the cross product of the Y- and Z-axes. The shank coordinate system was defined with the z-axis from the ankle joint center to the knee joint center, the y-axis as the cross product of the z-axis with a vector pointed from the medial to lateral ankle marker, and the x-axis as the cross product of the y- and z-axes. In the floating axis coordinate system, flexion occurs about the X-axis of the thigh, IE rotation occurs about the z-axis of the shank, and AA occurs about a floating axis that is calculated from the normalized cross product of the IE and flexion angle axes (z- and X-axes).

The reference configuration where AA, IE rotation, and flexion angles are zero occurs when the Y-axis is perpendicular to the thigh frontal plane formed by the Z-axis and the lateral knee marker, the y-axis is perpendicular to the shank frontal plane formed by the z-axis and the lateral ankle marker, and the Z- and z-axes are parallel.

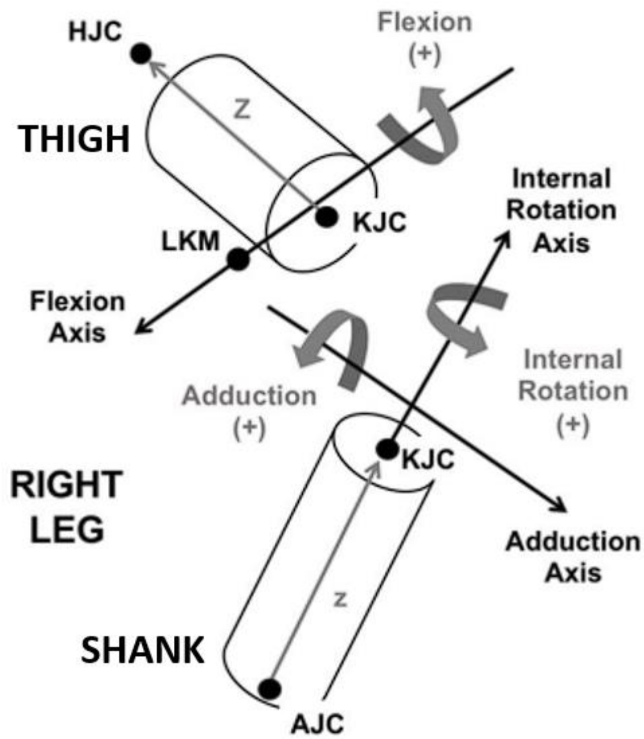

Figure 3: The definition of AA, IE, and flexion with the floating axis coordinate system. The hip joint center (HJC), knee joint center (KJC), ankle joint center (AJC), and lateral knee marker (LKM) are used to define the axes [30].

Knee angle analysis was conducted using equations, procedures, and algorithms implemented in a previous study [35] in MATLAB software (MATLAB R2023a. MathWorks, Inc. Natick, MA, USA). Key equations used for knee angle analysis are included in the appendix (Appendix D).

2.4.2 Gait Parameters

Total double support time, stance phase time, single support time, and load response time were calculated using force plate data to determine the timing of heel strikes and toe offs during gait. These gait phase parameters were represented as a percentage of the gait cycle by dividing the number of frames spent in the relevant gait phase by the total number of frames in the gait cycle. Equations 1-7 were used to calculate each relevant gait phase parameter.

$$Stance\ Phase\ Time\ (R) = \frac{Toe\ off\ (R) - Heel\ strike\ (R, 0\%)}{Heel\ strike\ (R, 100\%) - Heel\ strike\ (R, 0\%)} \tag{1}$$

$$Stance\ Phase\ Time\ (L) = \frac{Toe\ off\ (L) - Heel\ strike\ (R, 0\%) + Heel\ strike\ (R, 100\%) - Heel\ strike\ (L)}{Heel\ strike\ (R, 100\%) - Heel\ strike\ (R, 0\%)} \tag{2}$$

$$Single\ Support\ Time\ (R) = \frac{Heel\ strike\ (L) - Toe\ off\ (L)}{Heel\ strike\ (R, 100\%) - Heel\ strike\ (R, 0\%)} \tag{3}$$

$$Single\ Support\ Time\ (L) = \frac{Heel\ strike\ (R, 100\%) - Toe\ off\ (R)}{Heel\ strike\ (R, 100\%) - Heel\ strike\ (R, 0\%)} \tag{4}$$

$$Load\ Response\ Time\ (R) = \frac{Toe\ off\ (L) - Heel\ strike\ (R, 0\%)}{Heel\ strike\ (R, 100\%) - Heel\ strike\ (R, 0\%)} \tag{5}$$

$$Load\ Response\ Time\ (L) = \frac{Toe\ off\ (R) - Heel\ strike\ (L)}{Heel\ strike\ (R, 100\%) - Heel\ strike\ (R, 0\%)} \tag{6}$$

$$Total\ Double\ Support\ Time = \frac{Toe\ off\ (R) - Heel\ strike\ (L) + Toe\ off\ (L) - Heel\ strike\ (R, 0\%)}{Heel\ strike\ (R, 100\%) - Heel\ strike\ (R, 0\%)} \tag{7}$$

Stride velocity, stride length, and stride width were part of an exploratory hypothesis, with results included in the appendix (Appendix E). Stride velocity was calculated using the time to complete a gait cycle and the 1-dimensional distance travelled along the direction of forward motion by the sacrum marker in that time. Stride length was calculated using the 1-dimensional distance along the direction of forward motion between the heel marker of the dominant foot at the first and second heel strikes of the gait cycle. Velocity and stride length were normalized into non-dimensional quantities to account for variations in participants' leg lengths (distance from greater trochanter to lateral knee to lateral malleolus markers) with equations 8 and 9 [36].

$$ND\ velocity = \frac{velocity}{\sqrt{(leg\ length)g}} \tag{8}$$

$$ND\ stride\ length = \frac{stride\ length}{leg\ length} \tag{9}$$

Stride width was calculated using the distance between the right and left heel markers during double support in the direction orthogonal to the direction of forward motion. Stride width is typically not normalized, so it was reported as an absolute value [37].

2.4.3 Statistical Analysis

To test the hypotheses, two-sample t tests were performed to investigate whether the hypothesized parameters varied between ultimate players (U) and the control group (C). Statistical analysis was performed with females and males both combined and separated, producing results for groups of combined females and males, only females, and only males. A lower-tailed t test was used for stance phase time, load response time, and total double support time where U was expected to be lower than C. An upper-tailed t test was used for single support time where U was expected to be higher than C. Two-tailed t tests were used for knee angle parameters. An alpha level of 0.05 was used to determine statistical significance. Adjustments of p values for multiple comparisons were not made because this was intended as a pilot study to inform a larger follow-up study with more participants, at which point p value adjustment would be more appropriate [38].

Chapter 3

RESULTS

3.1 Female and Male Combined: Ultimate vs. Control

Of the gait parameters studied, there was a significant difference ($p<0.05$) in stance phase time of the dominant leg, single support time of the non-dominant leg, load response time of the dominant leg, and knee internal rotation ranges between ultimate players and control group participants (Table 2).

Table 2. Two-sample t test results of gait and knee angle parameters comparing female and male ultimate playing participants to female and male control group participants. * = significant difference ($p<0.05$).

Female and Male: Ultimate (U) vs Control ©	U: Mean ± 1 SD	C: Mean ± 1 SD	Stat Test Type	p value
Stance Phase, Non-Dominant [% of gait cycle]	60.4 ± 1.0	60.5 ± 1.3	t Test, lower-tailed	0.4504
Stance Phase, Dominant [% of gait cycle]	60.5 ± 1.0	61.5 ± 0.6	t Test, lower-tailed	0.0241*
Single Support, Non-Dominant [% of gait cycle]	39.5 ± 1.0	38.5 ± 0.6	t Test, upper-tailed	0.0241*
Single Support, Dominant [% of gait cycle]	39.6 ± 1.0	39.5 ± 1.3	t Test, upper-tailed	0.4504
Load Response, Non-Dominant [% of gait cycle]	10.7 ± 0.9	10.5 ± 1.1	t Test, lower-tailed	0.5944
Load Response, Dominant [% of gait cycle]	10.3 ± 1.3	11.4 ± 0.4	t Test, lower-tailed	0.0225*
Total Double Support [% of gait cycle]	20.9 ± 1.7	21.9 ± 1.4	t Test, lower-tailed	0.1124
Flexion First Peak [degrees]	20.2 ± 9.4	17.2 ± 4.3	t Test, two-tailed	0.4211
Flexion Second Peak [degrees]	65.7 ± 6.8	67.2 ± 4.8	t Test, two-tailed	0.6147
Flexion Minimum [degrees]	7.4 ± 4.5	3.8 ± 2.3	t Test, two-tailed	0.0666
IE Rotation Range [degrees]	14.4 ± 3.8	19.4 ± 5.0	t Test, two-tailed	0.0457*
IE Rotation Mean [degrees]	-1.3 ± 3.9	3.6 ± 5.5	t Test, two-tailed	0.0599
AA Range [degrees]	10.7 ± 3.4	11.3 ± 2.8	t Test, two-tailed	0.7302
AA Mean [degrees]	1.6 ± 2.1	0.9 ± 2.1	t Test, two-tailed	0.4882

The stance phase times of the dominant leg were 60.5 ± 1.0% and 61.5 ± 0.6% for ultimate players (U) and the control group (C), respectively (p=0.0241). The stance phase times of the non-dominant leg were 60.4 ± 1.0% and 60.5 ± 1.3% for U and C, respectively (p=0.4504) (Fig 4).

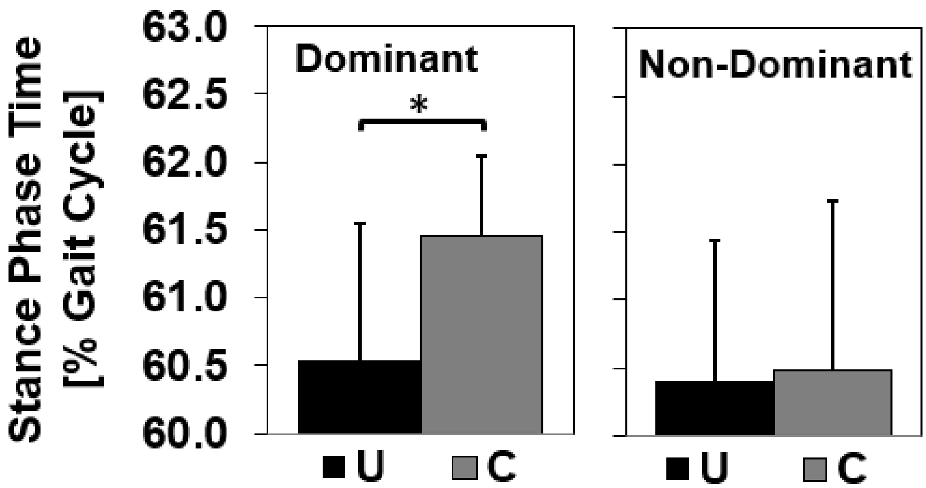

Figure 4: Stance phase time of the dominant (p=0.0241) and non-dominant (p=0.4504) legs of the combined male and female group. * = significant difference (p<0.05).

The single support times of the non-dominant leg were 39.5 ± 1.0% and 38.5 ± 0.6% for U and C, respectively (p=0.0241). The single support times of the dominant leg were 39.6 ± 1.0% and 39.5 ± 1.3% for U and C, respectively (p=0.4504) (Fig 5).

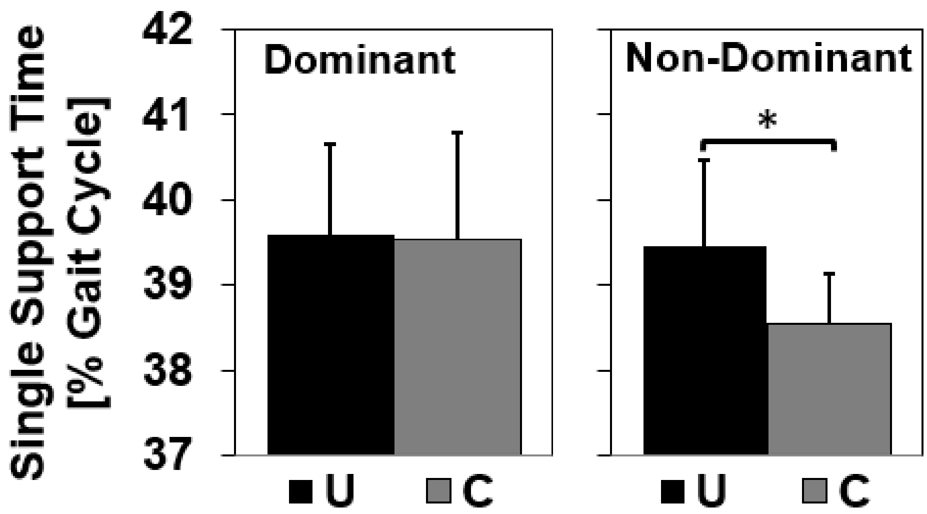

Figure 5: Single support time of the dominant (p=0.4504) and non-dominant (p=0.0241) legs of the combined male and female group. * = significant difference (p<0.05).

The load response times of the dominant leg were 10.3 ± 1.3% and 11.4 ± 0.4% for U and C, respectively (p=0.0225). The load response times of the non-dominant leg were 10.7 ± 0.9% and 10.5 ± 1.1% for U and C, respectively (p=0.5944) (Fig 6).

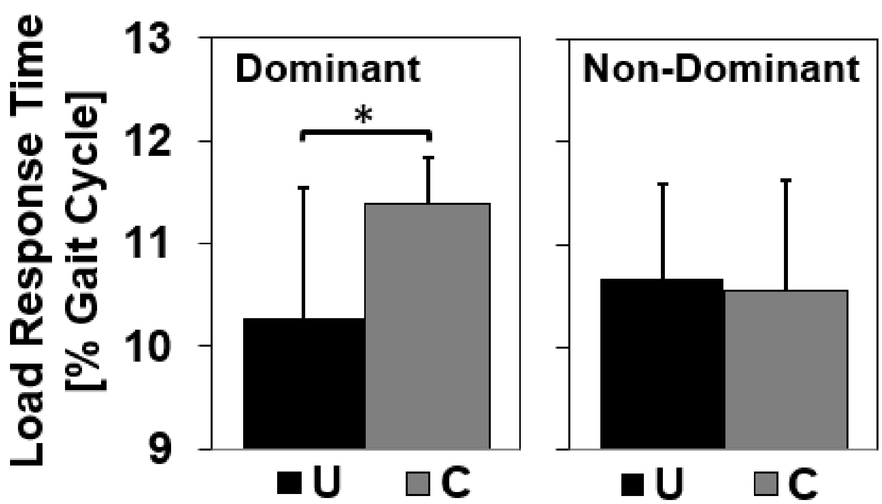

Figure 6: Load response time of the dominant (p=0.0225) and non-dominant (p=0.5944) legs of the combined male and female group. * = significant difference (p<0.05).

Knee internal rotation angle ranges were 14.4 ± 3.8° and 19.4 ± 5.0° for U and C, respectively (p=0.0457) (Fig 7, Fig 8).

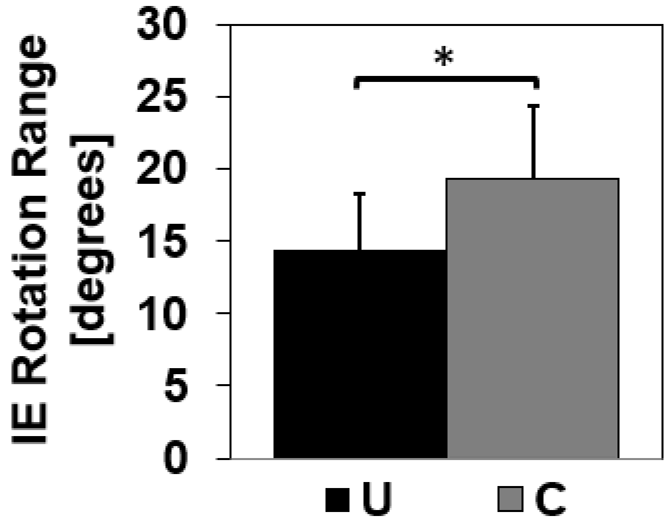

Figure 7: IE rotation range (p=0.0457) of the combined male and female group. * = significant difference (p<0.05).

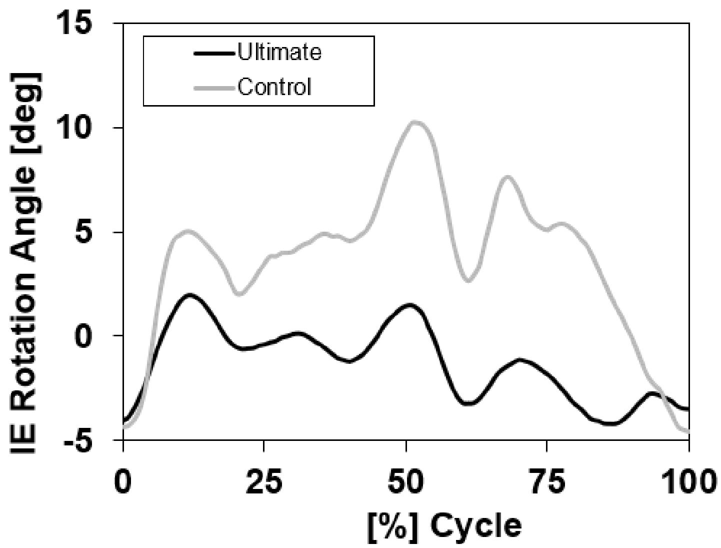

Figure 8: IE rotation through a gait cycle of the combined male and female group, averaged across all participants.

3.2 Female: Ultimate vs. Control

Of the gait parameters studied, there was a significant difference in stance phase time of the dominant leg, single support time of the non-dominant leg, load response time of the dominant leg, and total double support time between female ultimate players and female control group participants (Table 3).

Table 3. Two-sample t test results of gait and knee angle parameters comparing female ultimate playing participants to female control group participants. * = significant difference ($p<0.05$).

Female: Ultimate vs Control	U: Mean ± 1 SD	C: Mean ± 1 SD	Stat Test Type	p value
Female Stance Phase, Non-Dominant [% of gait cycle]	59.6 ± 0.6	60.5 ± 1.8	t Test, lower-tailed	0.2138
Female Stance Phase, Dominant [% of gait cycle]	59.8 ± 0.8	61.5 ± 0.3	t Test, lower-tailed	0.0089*
Female Single Support, Non-Dominant [% of gait cycle]	40.2 ± 0.8	38.5 ± 0.3	t Test, upper-tailed	0.0089*
Female Single Support, Dominant [% of gait cycle]	40.4 ± 0.6	39.5 ± 1.8	t Test, upper-tailed	0.2138
Female Load Response, Non-Dominant [% of gait cycle]	10.1 ± 1.0	10.5 ± 1.4	t Test, lower-tailed	0.3490
Female Load Response, Dominant [% of gait cycle]	9.3 ± 0.8	11.5 ± 0.6	t Test, lower-tailed	0.0034*
Female Total Double Support [% of gait cycle]	19.4 ± 0.5	21.9 ± 1.9	t Test, lower-tailed	0.0371*
Female Flexion First Peak [degrees]	23.9 ± 5.0	18.0 ± 5.7	t Test, two-tailed	0.1689
Female Flexion Second Peak [degrees]	66.6 ± 5.3	69.8 ± 4.1	t Test, two-tailed	0.3739
Female Flexion Minimum [degrees]	8.4 ± 3.1	5.1 ± 1.7	t Test, two-tailed	0.1314
Female IE Rotation Range [degrees]	16 ± 4.4	19.5 ± 6.6	t Test, two-tailed	0.4245
Female IE Rotation Mean [degrees]	0.0 ± 4.1	1.7 ± 2.9	t Test, two-tailed	0.5185
Female AA Range [degrees]	10.1 ± 2.6	9.8 ± 1.4	t Test, two-tailed	0.8648
Female AA Mean [degrees]	0.6 ± 1.0	1.0 ± 1.2	t Test, two-tailed	0.6835

The stance phase times of the dominant leg were 59.8 ± 0.8% and 61.5 ± 0.3% for female U and C, respectively (p=0.0089). The stance phase times of the non-dominant leg were 59.6 ± 0.6% and 60.5 ± 1.8% for female U and C, respectively (p = 0.2138) (Fig 9).

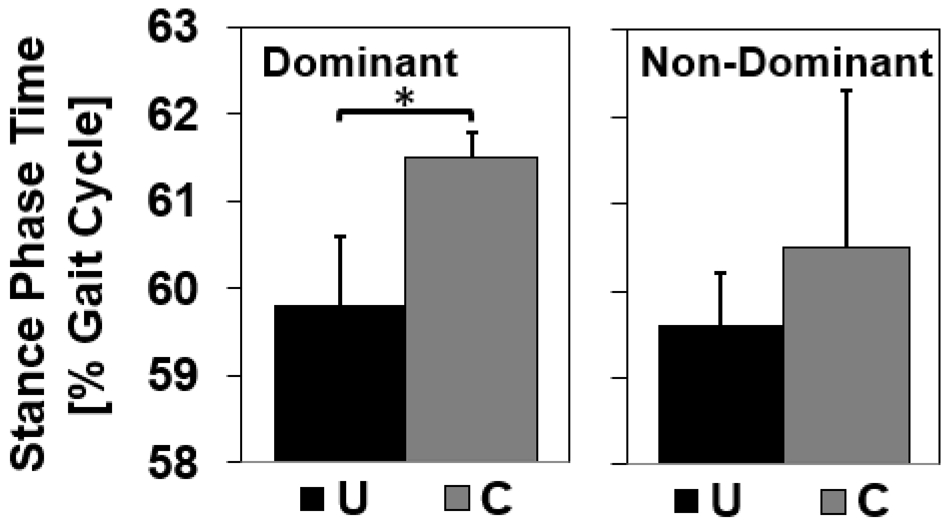

Figure 9: Stance phase time of the dominant (p=0.0089) and non-dominant (p=0.2138) legs of the female group. * = significant difference (p<0.05).

The single support times of the non-dominant leg were 40.2 ± 0.8% and 38.5 ± 0.3% for female U and C, respectively (p=0.0089). The single support times of the dominant leg were 40.4 ± 0.6% and 39.5 ± 1.8% for female U and C, respectively (p=0.2138) (Fig 10).

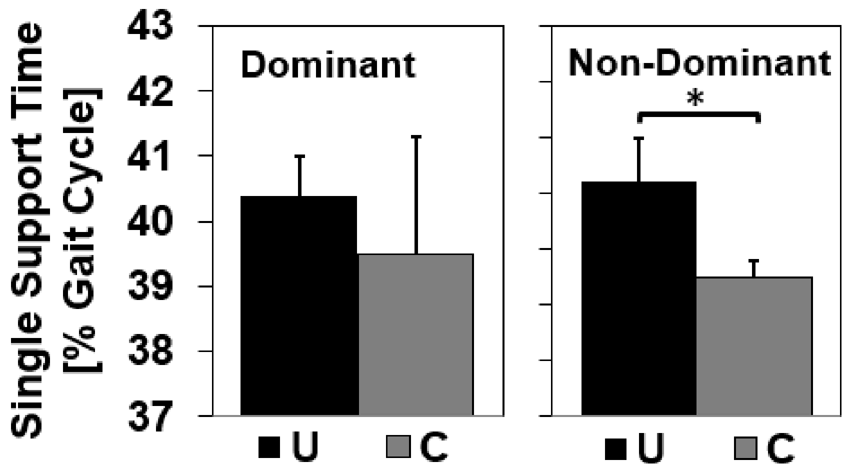

Figure 10: Single support time of the dominant (p=0.2138) and non-dominant (p=0.0089) legs of the female group. * = significant difference (p<0.05).

The load response times of the dominant leg were 9.3 ± 0.8% and 11.5 ± 0.6% for female U and C, respectively (p=0.0034). The load response times of the non-dominant leg were 10.1 ± 1.0% and 10.5 ± 1.4% for female U and C, respectively (p=0.3490) (Fig 11).

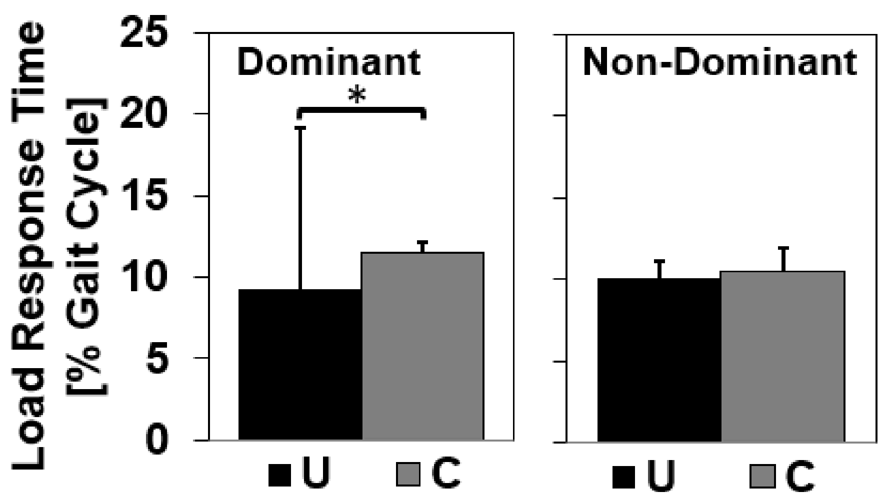

Figure 11: Load response time of the dominant (p=0.0034) and non-dominant (p=0.3490) legs of the female group. * = significant difference (p<0.05).

Double support time was 19.4 ± 0.5% and 21.9 ± 1.9% for female U and C, respectively (p=0.0371) (Fig 12).

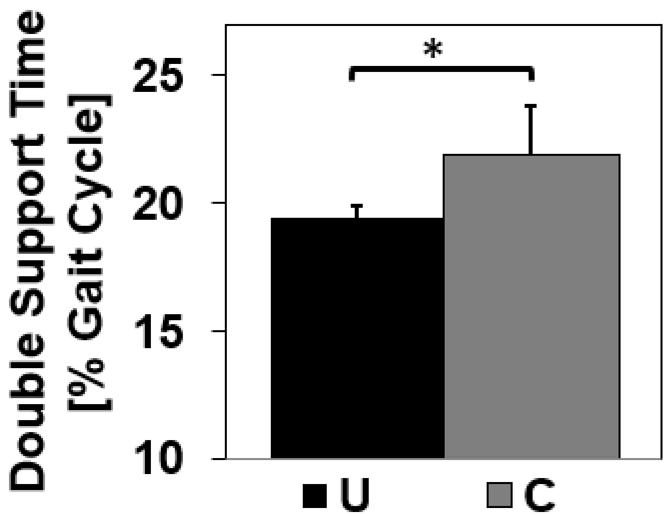

Figure 12: Double support time (p=0.0371) of the female group. * = significant difference (p<0.05).

3.3 Male: Ultimate vs. Control

Of the gait parameters studied, there was a significant difference in knee internal rotation ranges between male ultimate players and male control group participants (Table 4).

Table 4. Two-sample t test results of gait and knee angle parameters comparing male ultimate playing participants to male control group participants. * = significant difference ($p<0.05$).

Male: Ultimate vs Control	U: Mean ± 1 SD	C: Mean ± 1 SD	Stat Test Type	p value
Male Stance Phase, Non-Dominant [% of gait cycle]	61.2 ± 0.7	60.5 ± 0.5	t Test, lower-tailed	0.9220
Male Stance Phase, Dominant [% of gait cycle]	61.3 ± 0.4	61.4 ± 0.8	t Test, lower-tailed	0.3895
Male Single Support, Non-Dominant [% of gait cycle]	38.7 ± 0.4	38.6 ± 0.8	t Test, upper-tailed	0.3895
Male Single Support, Dominant [% of gait cycle]	38.8 ± 0.7	39.5 ± 0.5	t Test, upper-tailed	0.9220
Male Load Response, Non-Dominant [% of gait cycle]	11.2 ± 0.4	10.6 ± 0.9	t Test, lower-tailed	0.8576
Male Load Response, Dominant [% of gait cycle]	11.3 ± 0.6	11.3 ± 0.2	t Test, lower-tailed	0.4823
Male Total Double Support [% of gait cycle]	22.5 ± 0.4	21.9 ± 0.7	t Test, lower-tailed	0.8843
Male Flexion First Peak [degrees]	16.5 ± 11.9	16.3 ± 2.9	t Test, two-tailed	0.9825
Male Flexion Second Peak [degrees]	64.9 ± 8.8	64.7 ± 4.5	t Test, two-tailed	0.9737
Male Flexion Minimum [degrees]	6.5 ± 5.9	2.4 ± 2.1	t Test, two-tailed	0.2715
Male IE Rotation Range [degrees]	12.8 ± 2.8	19.2 ± 3.8	t Test, two-tailed	0.0383*
Male IE Rotation Mean [degrees]	-2.6 ± 3.8	5.5 ± 7.3	t Test, two-tailed	0.1096
Male AA Range [degrees]	11.4 ± 4.3	12.7 ± 3.3	t Test, two-tailed	0.6353
Male AA Mean [degrees]	2.6 ± 2.5	0.8 ± 2.9	t Test, two-tailed	0.3852

Knee internal rotation angle ranges were 12.8 ± 2.8° and 19.2 ± 3.8° for male U and C, respectively (p=0.0383) (Fig 13, Fig 14).

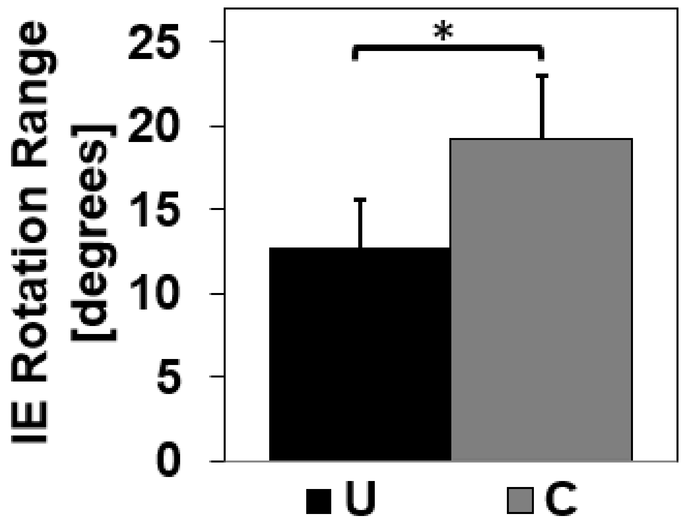

Figure 13: IE rotation range (p=0.0383) of the male group. * = significant difference (p<0.05).

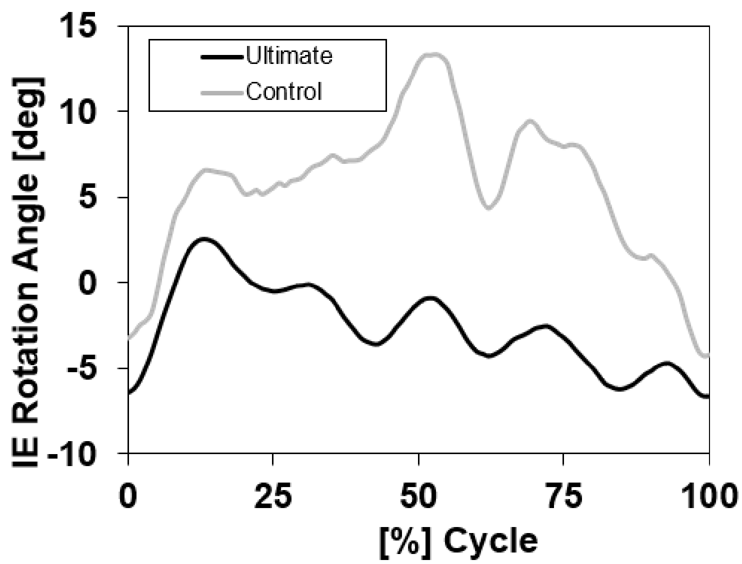

Figure 14: IE rotation through a gait cycle of the male group, averaged across all male participants.

Chapter 4

DISCUSSION

4.1 Summary of Key Findings

This was the first study in our knowledge to investigate the biomechanics of gait and balance in ultimate players. The purpose of this study was to investigate if select gait parameters and knee angles would differ between ultimate players and control group participants with the same age and BMI ranges to provide evidence of ultimate affecting balance and knee angles during gait.

Total double support time, stance phase time, and load response time were hypothesized to be lower for ultimate players, while single support time was hypothesized to be higher for ultimate players. The results partially support these hypotheses. The combined group consisting of males and females found that ultimate players had a significant decrease in stance phase time and load response time of the dominant leg, and a significant increase in single support time of the non-dominant leg. Female ultimate players had a significant decrease in stance phase time and load response time of the dominant leg, and a significant increase in single support time of the non-dominant leg. Female ultimate players also had a significant decrease in double support time. Male ultimate players had no statistically significant results for the hypothesized gait phase parameters. These results suggest that the female ultimate players studied did have improved balance over the control group female participants.

The hypothesized changes in stance phase time, load response time, and single support time were each observed in one leg and not both. A significant difference in load response time and stance phase time were observed in the dominant leg, while a significant difference in single support time was observed in the non-dominant leg. It is possible that this asymmetry is caused by ultimate players repeatedly lunging with their dominant leg. Ultimate players are required to always lunge with their dominant leg leading to throw the disc, regularly pivoting and balancing on their non-dominant leg for support. Lunges as used in ultimate are an asymmetrical lower body exercise that strengthens the quadriceps of the leading leg and the hamstrings of the supporting leg [39]. Repeated loading of the dominant leg and balancing on the non-dominant leg could cause an asymmetry in muscle growth that directly impacts postural stability, resulting in improved balance in both the dominant and non-dominant legs, but being observable through different gait parameters. For instance, single support time of the non-dominant leg was significantly higher in ultimate players, while the load response of the dominant leg was significantly lower in ultimate players. Both results indicate improved balance but were observed in different legs. A study on unilateral sports such as soccer also found asymmetry in gait phase parameters due to the players' tendencies to kick and dribble with their dominant leg [26], suggesting that asymmetries in gait could arise from ultimate as well.

Of the hypotheses that AA and IE rotation means and ranges would differ between ultimate players and control group participants, only IE rotation range was observed to be significantly different for males and the combined group of males and females. A larger IE rotation range has been linked to OA [10], so due to the observed lower IE rotation range in ultimate players, it is possible that playing ultimate might be beneficial for those who have abnormal knee angles during gait. The hypothesis that the magnitude and timing of first and second peak flexion angles and magnitude of flexion minimum would differ between ultimate players and control group participants was not supported by the results, as none of the tested flexion angle parameters were significantly different.

4.2 Limitations

There were several limitations in the present study. The first limitation is the low number of participants. Four female ultimate players, four male ultimate players, four female control participants, and four male control participants were used in this study. A sample size this small is not ideal and would need to be increased to have confidence in the results. With the low number of participants used in this study, several of the investigated parameters did not pass the Shapiro-Wilk test for normality, which is an issue that could potentially be resolved through the inclusion of more participants and non-parametric statistics.

A second limitation is that knee angle data were only collected from the dominant leg of the participant. With the evident asymmetry in gait phase parameters, it is possible that there was asymmetry in knee angles between the dominant and non-dominant legs of ultimate players as well. Since TCPE marker clusters were only placed on the dominant leg of participants, only knee angles for the dominant leg using the TCPE method to account for STA could be calculated. Although marker clusters were not placed on the non-dominant leg, markers were placed according to the enhanced Helen Hayes marker protocol, so knee angles could be calculated for the non-dominant leg, although the effects of STA would be more prevalent.

A third limitation is that there was no principal component analysis (PCA) correction for crosstalk errors when determining knee angles. When using 3-dimensional motion capture, misplacement in markers on the knee causes a phenomenon known as crosstalk, which affects the kinematics of joints undergoing relatively large rotations about a major axis, resulting in incorrect results for knee angles. The knee angle most susceptible to crosstalk in gait is flexion, which can become incorrectly correlated with AA. PCA is a post hoc method that can be used to reduce crosstalk errors and minimize the correlation between the knee angles. Although crosstalk was not addressed through PCA in this study, a previous study conducted within the HMB lab at Cal Poly found that the protocols as used in this study produce results that are not affected by crosstalk, so using PCA to correct the angles would not yield a difference in the results. [40].

4.3 Future Research

Future research on the effects of ultimate on balance should increase the number of participants used in the study. The results from the present study indicate that ultimate may have a larger impact on the gait phase parameters of females, so future studies should continue to separate females and males in statistical analysis testing. The age range could be increased to allow recruitment from elite clubs or professional teams to reach players at the highest level of play. Future studies should also consider measuring the knee angles of both the dominant and the non-dominant leg, to check for any asymmetrical results, like what was found in gait phase parameters. Although gait phase parameters have been found to be good metrics for measuring balance [21], static balance tests in the single-leg stance on each leg to test for balance differences using force plates [41] or a smartwatch [42] could provide more reliable data on postural stability and balance between dominant and non-dominant legs.

4.4 Conclusion

Although ultimate has a high risk of injury, most injuries are minor and should not discourage children and young adults from trying the sport. While there is a high risk of getting a sprain or pulling a muscle while playing ultimate, the results of this study indicate that playing ultimate does not adversely impact knee angles that increase the risk of a more serious condition such as OA. If the risk of injury that comes with playing ultimate is especially concerning, there are safer alternatives including yoga, tai chi, and simple exercises such as one-legged stands that improve balance and have a low risk of injury at any age when done correctly. Balance and exercise are both invaluable to living a long and healthy life, and playing a sport like ultimate can be a rewarding way to obtain both.

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APPENDIX A: NOMENCLATURE

AA – Abduction-Adduction

IE – Internal-External

TCPE – Triangular Cosserat Point Element

STA – Soft-Tissue Artifact

OA – Osteoarthritis

BMI – Body Mass Index

PCA – Principal Component Analysis

APPENDIX B: ORIGINAL DATA

Original participant gait data were collected for this research. The data include raw marker position and force plate values for each trial and processed data in the form of gait phase parameters and knee angles. The data do not contain identifiable participant information. The data are available upon request by contacting Dr. Stephen Klisch at sklisch@calpoly.edu.

APPENDIX C: MATLAB SOFTWARE

MATLAB software was developed for this research. This MATLAB software is available upon request by contacting Dr. Stephen Klisch at sklisch@calpoly.edu

APPENDIX D: TCPE EQUATIONS

Direction vectors are created from each TCPE for the static pose, used as the reference configuration, and for the dynamic trials at each frame. The present configuration direction vector, $\boldsymbol{d_i}$, and the reciprocal of the reference configuration direction vector, $\boldsymbol{D^i}$, are used to calculate the deformation gradient tensor, $\boldsymbol{F}$, for each TCPE at each frame in the dynamic trial using equation (1).

$$\boldsymbol{F} = \boldsymbol{d_i} \otimes \boldsymbol{D^i} \tag{1}$$

Through polar decomposition, the deformation gradient tensor is multiplied by the inverse of the stretch tensor, **M**, to solve for the rotation tensor, **R** using equation (2).

$$\boldsymbol{R} = \boldsymbol{F}\boldsymbol{M}^{-1} \tag{2}$$

To determine the strain in each TCPE, the Lagrangian strain tensor, **E**, is calculated using **I,** the identity tensor and **C,** the right Cauchy-Green deformation tensor in equation (3).

$$\boldsymbol{E} = \frac{1}{2}\boldsymbol{C} - \boldsymbol{I} = \frac{1}{2}\boldsymbol{F}^T\boldsymbol{F} - \boldsymbol{I} \tag{3}$$

The translation of each TCPE is calculated between the centroid of a TCPE and the knee joint center. The translation of this reference point, $\boldsymbol{t}^{(B)}$, is calculated as the difference in the present and reference configuration vectors, $\boldsymbol{x}^{(B)}$ and $\boldsymbol{X}^{(B)}$, respectively. The present configuration vector, $\boldsymbol{x}_{(B)}$, is calculated from the present configuration TCPE centroid, $\bar{\boldsymbol{x}}$, minus the deformed difference between the static configuration centroid, $\bar{\boldsymbol{X}}$, and reference configuration vector, $\boldsymbol{X}^{(B)}$ (Eqn. 4).

$$\boldsymbol{t}^{(B)} = \boldsymbol{x}^{(B)} - \boldsymbol{X}^{(B)} = \overline{\boldsymbol{x}} - \boldsymbol{F}\left(\overline{\boldsymbol{X}} - \boldsymbol{X}^{(B)}\right) - \boldsymbol{X}^{(B)}$$

(4)

The relative rotation of a TCPE is found by first calculating the relative angle between each combination of two TCPEs where the relative angle, $\phi_{j/k}$, between the j and k^{th} TCPE is:

$$\phi_{j/k} = \cos^{-1}[\frac{1}{2}(\boldsymbol{R}_j \cdot \boldsymbol{R}_k - 1)]$$

(5)

where $\mathbf{R}_j$ and $\mathbf{R}_k$ are the rotation tensors of the j and k^{th} TCPE, respectively. The relative angle of the j^{th} TCPE relative to all TCPEs at one time point is shown in equation (6), where N is the total number of TCPEs.

$$\phi_j = \frac{1}{N-1}\sum_{k=1}^{N} \phi_{j/k}$$

(6)

The translation of the j^{th} TCPE relative to the translation of all the TCPEs can be calculated from the translation of the j and k^{th} translation vectors, $\boldsymbol{t}_j^{(B)}$ and $\boldsymbol{t}_k^{(B)}$ (Eq. 7).

$$T_j = \frac{1}{N-1}\sum_{k=1}^{N} \left|\boldsymbol{t}_j^{(B)} - \boldsymbol{t}_k^{(B)}\right|$$

(7)

The magnitude of the strain tensor of the j^{th} TCPE is calculated from equation (8).

$$E_j = \sqrt{\boldsymbol{E}_j \cdot \boldsymbol{E}_j}$$

(8)

The relative rotation, relative translation and strain were normalized by their respective ranges of magnitude to yield the parameters N_ϕ, N_T, and N_E, representing normalized rotation, normalized translation and normalized strain, respectively. The average of the three parameters yields the combined normalized filtering parameter, $N_{combined}$, calculated in equation (9).

$$N_{combined} = \left(N_{\phi} + N_T + N_E\right)/3 \qquad (9)$$

A canonical extension of Solav's filtering algorithm is used. If the normalized filtering parameter is less than or equal to 0.1, a TCPE is considered a good estimate of the motion of the underlying bone segment. The TCPEs which meet this criterion are selected and their rotation tensors are averaged in order to get the best predicted rotation tensor for each time point. If no TCPEs meet the criterion, the three TCPEs with the lowest normalized filtering parameters are selected.

APPENDIX E: VELOCITY, STRIDE LENGTH, AND STRIDE WIDTH

Female and Male: Ultimate vs Control	U: Mean ± 1 SD	C: Mean ± 1 SD	Stat Test Type	p value
ND Velocity: U vs C	0.43 ± 0.04	0.44 ± 0.03	t Test, two-tailed	0.5683
ND Stride Length: U vs C	1.56 ± 0.13	1.59 ± 0.08	t Test, two-tailed	0.5804
Stride Width [mm]: U vs C	77.8 ± 19.4	75.3 ± 28.4	t Test, lower-tailed	0.5777

Female: Ultimate vs Control	U: Mean ± 1 SD	C: Mean ± 1 SD	Stat Test Type	p value
Female ND Velocity: U vs C	0.45 ± 0.04	0.45 ± 0.04	t Test, two-tailed	0.9071
Female ND Stride Length: U vs C	1.64 ± 0.04	1.61 ± 0.10	t Test, two-tailed	0.5889
Female Stride Width [mm]: U vs C	91.1 ± 6.5	67.5 ± 23.1	t Test, lower-tailed	0.9343

Male: Ultimate vs Control	U: Mean ± 1 SD	C: Mean ± 1 SD	Stat Test Type	p value
Male ND Velocity: U vs C	0.42 ± 0.04	0.43 ± 0.01	t Test, two-tailed	0.4547
Male ND Stride Length: U vs C	1.47 ± 0.14	1.56 ± 0.04	t Test, two-tailed	0.2642
Male Stride Width [mm]: U vs C	64.4 ± 18.9	83.2 ± 34.3	t Test, lower-tailed	0.1926

Ultimate: Female vs Male	F: Mean ± 1 SD	M: Mean ± 1 SD	Stat Test Type	p value
Ultimate ND Velocity: F vs M	0.45 ± 0.04	0.42 ± 0.04	t Test, two-tailed	0.2818
Ultimate ND Stride Length: F vs M	1.64 ± 0.04	1.47 ± 0.14	t Test, two-tailed	0.0742
Ultimate Stride Width [mm]: F vs M	113.5 ± 20.1	88.3 ± 17.9	t Test, two-tailed	0.1113

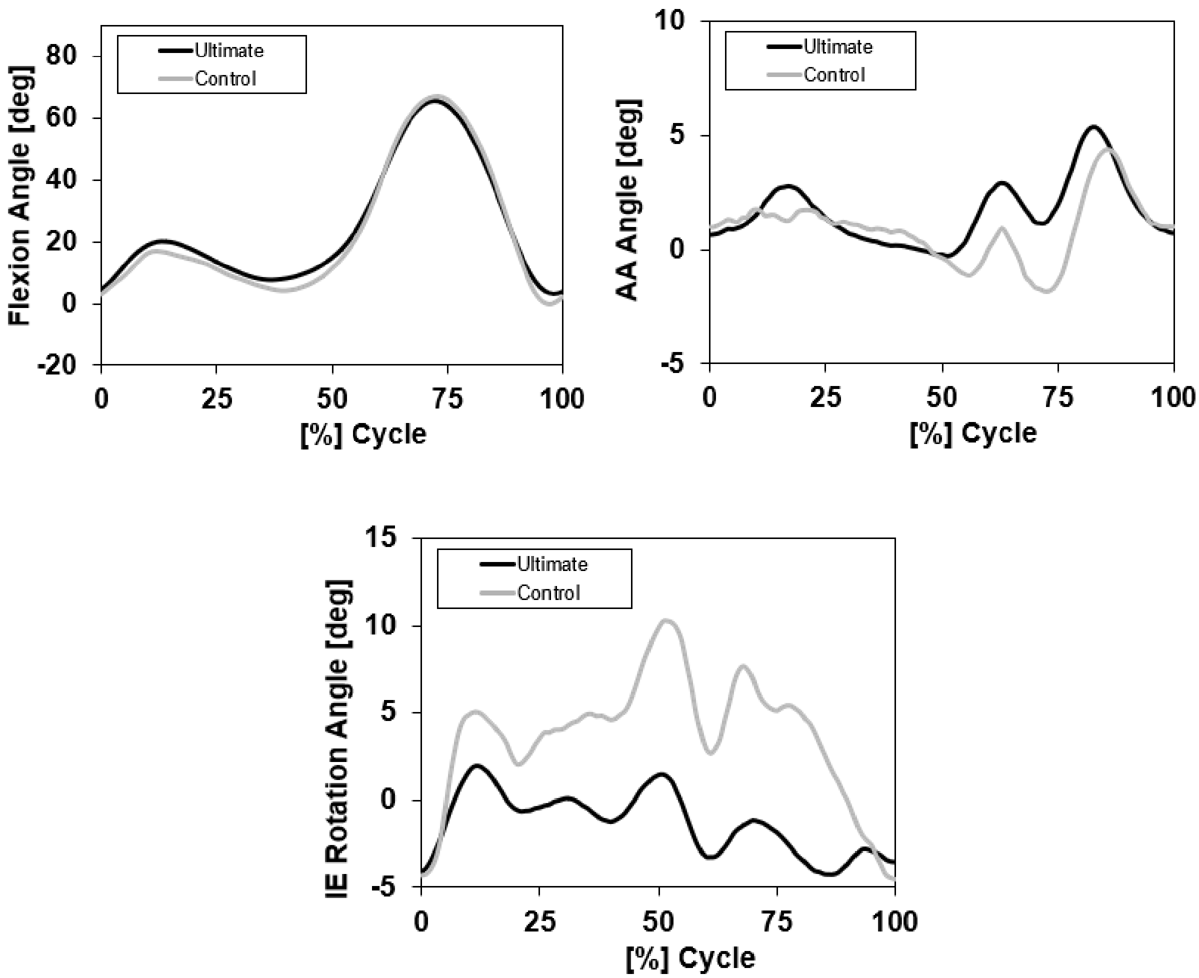
Control: Female vs Male	F: Mean ± 1 SD	M: Mean ± 1 SD	Stat Test Type	p value
Control ND Velocity: F vs M	0.45 ± 0.04	0.43 ± 0.01	t Test, two-tailed	0.4399
Control ND Stride Length: F vs M	1.61 ± 0.10	1.56 ± 0.04	t Test, two-tailed	0.4005
Control Stride Width [mm]: F vs M	67.5 ± 23.1	83.2 ± 34.3	t Test, two-tailed	0.4819

APPENDIX F: FEMALE VS MALE GAIT PHASE PARAMETERS AND KNEE ANGLES

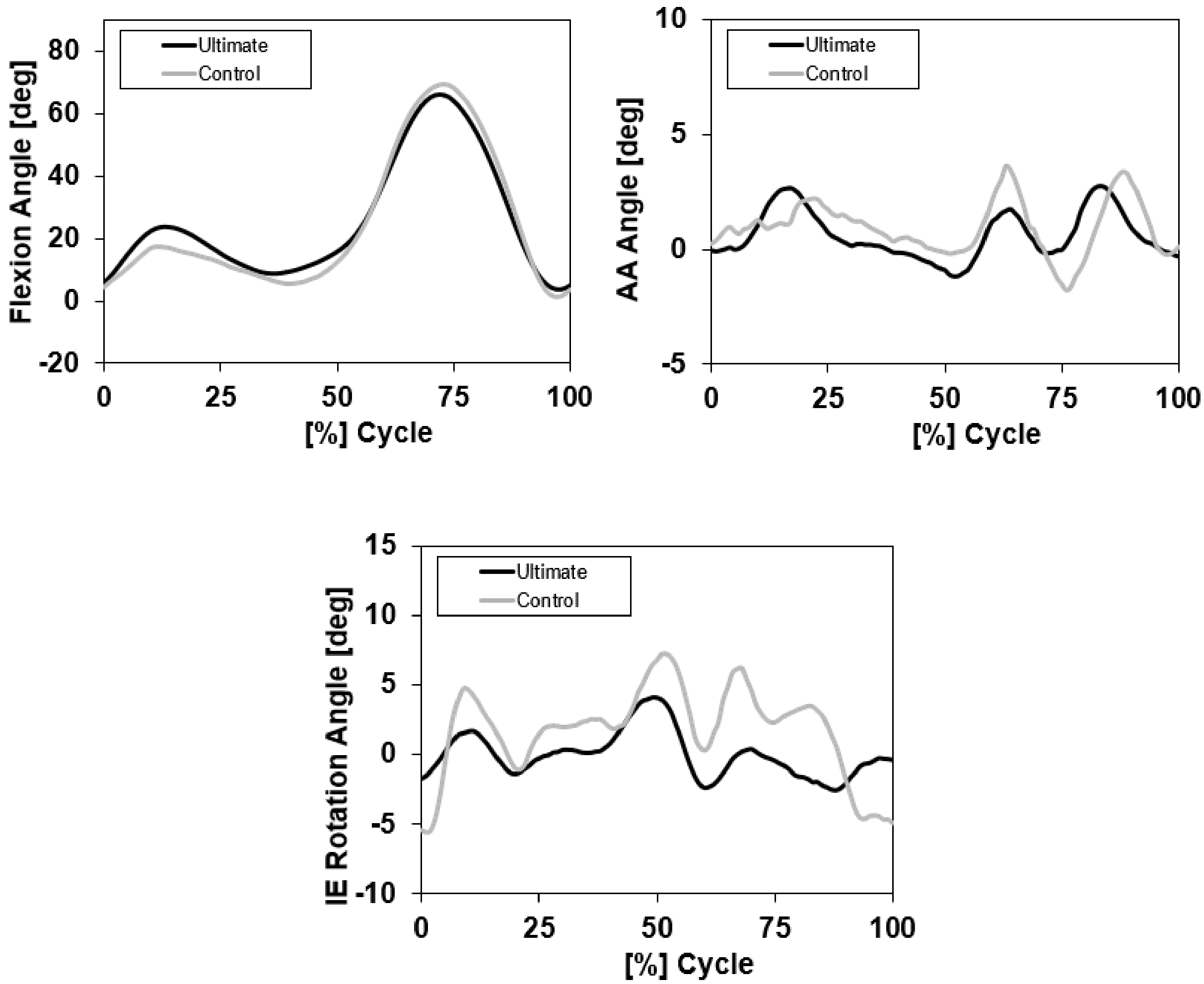
Ultimate: Female vs Male	F: Mean ± 1 SD	M: Mean ± 1 SD	Stat Test Type	p value
Ultimate Flexion First Peak [degrees]: F vs M	23.9 ± 5.0	16.5 ± 11.9	t Test, two-tailed	0.3130
Ultimate Flexion Second Peak [degrees]: F vs M	66.6 ± 5.3	64.9 ± 8.8	t Test, two-tailed	0.7557
Ultimate Flexion Minimum [degrees]: F vs M	8.4 ± 3.1	6.5 ± 5.9	t Test, two-tailed	0.5938
Ultimate Internal Rotation Range [degrees]: F vs M	16.0 ± 4.4	12.8 ± 2.8	t Test, two-tailed	0.2751
Ultimate Internal Rotation Mean [degrees]: F vs M	0.0 ± 4.1	-2.6 ± 3.8	t Test, two-tailed	0.3967
Ultimate Abduction Range [degrees]: F vs M	10.1 ± 2.6	11.4 ± 4.3	t Test, two-tailed	0.6276
Ultimate Abduction Mean [degrees]: F vs M	0.6 ± 1.0	2.6 ± 2.5	t Test, two-tailed	0.2130
Ultimate Stance Phase, Non-Dominant [% of gait cycle]: F vs M	59.6 ± 0.6	61.2 ± 0.7	t Test, two-tailed	0.0139*
Ultimate Stance Phase, Dominant [% of gait cycle]: F vs M	59.8 ± 0.8	61.3 ± 0.4	t Test, two-tailed	0.0241*
Ultimate Single Support, Non-Dominant [% of gait cycle]: F vs M	40.2 ± 0.8	38.7 ± 0.4	t Test, two-tailed	0.0241*
Ultimate Single Support, Dominant [% of gait cycle]: F vs M	40.4 ± 0.6	38.8 ± 0.7	t Test, two-tailed	0.0139*
Ultimate Load Response, Non-Dominant [% of gait cycle]: F vs M	10.1 ± 1.0	11.2 ± 0.4	t Test, two-tailed	0.1128
Ultimate Load Response, Dominant [% of gait cycle]: F vs M	9.3 ± 0.8	11.3 ± 0.6	t Test, two-tailed	0.0096*
Ultimate Total Double Support [% of gait cycle]: F vs M	19.4 ± 0.5	22.5 ± 0.4	t Test, two-tailed	0.0001*

Control: Female vs Male	F: Mean ± 1 SD	M: Mean ± 1 SD	Stat Test Type	p value
Control Flexion First Peak [degrees]: F vs M	18.0 ± 5.7	16.3 ± 2.9	t Test, two-tailed	0.6348
Control Flexion Second Peak [degrees]: F vs M	69.8 ± 4.1	64.7 ± 4.5	t Test, two-tailed	0.1453
Control Flexion Minimum [degrees]: F vs M	5.1 ± 1.7	2.4 ± 2.1	t Test, two-tailed	0.0918
Control Internal Rotation Range [degrees]: F vs M	19.5 ± 6.6	19.2 ± 3.8	t Test, two-tailed	0.9472
Control Internal Rotation Mean [degrees]: F vs M	1.7 ± 2.9	5.5 ± 7.3	t Test, two-tailed	0.3808
Control Abduction Range [degrees]: F vs M	9.8 ± 1.4	12.7 ± 3.3	t Test, two-tailed	0.1787
Control Abduction Mean [degrees]: F vs M	1.0 ± 1.2	0.8 ± 2.9	t Test, two-tailed	0.9334
Control Stance Phase, Non-Dominant [% of gait cycle]: F vs M	60.5 ± 1.8	60.5 ± 0.5	t Test, two-tailed	0.9944
Control Stance Phase, Dominant [% of gait cycle]: F vs M	61.5 ± 0.3	61.4 ± 0.8	t Test, two-tailed	0.9589
Control Single Support, Non-Dominant [% of gait cycle]: F vs M	38.5 ± 0.3	38.6 ± 0.8	t Test, two-tailed	0.9589
Control Single Support, Dominant [% of gait cycle]: F vs M	39.5 ± 1.8	39.5 ± 0.5	t Test, two-tailed	0.9944
Control Load Response, Non-Dominant [% of gait cycle]: F vs M	10.5 ± 1.4	10.6 ± 0.9	t Test, two-tailed	0.8540
Control Load Response, Dominant [% of gait cycle]: F vs M	11.5 ± 0.6	11.3 ± 0.2	t Test, two-tailed	0.6299
Control Total Double Support [% of gait cycle]: F vs M	21.9 ± 1.9	21.9 ± 0.7	t Test, two-tailed	0.9838

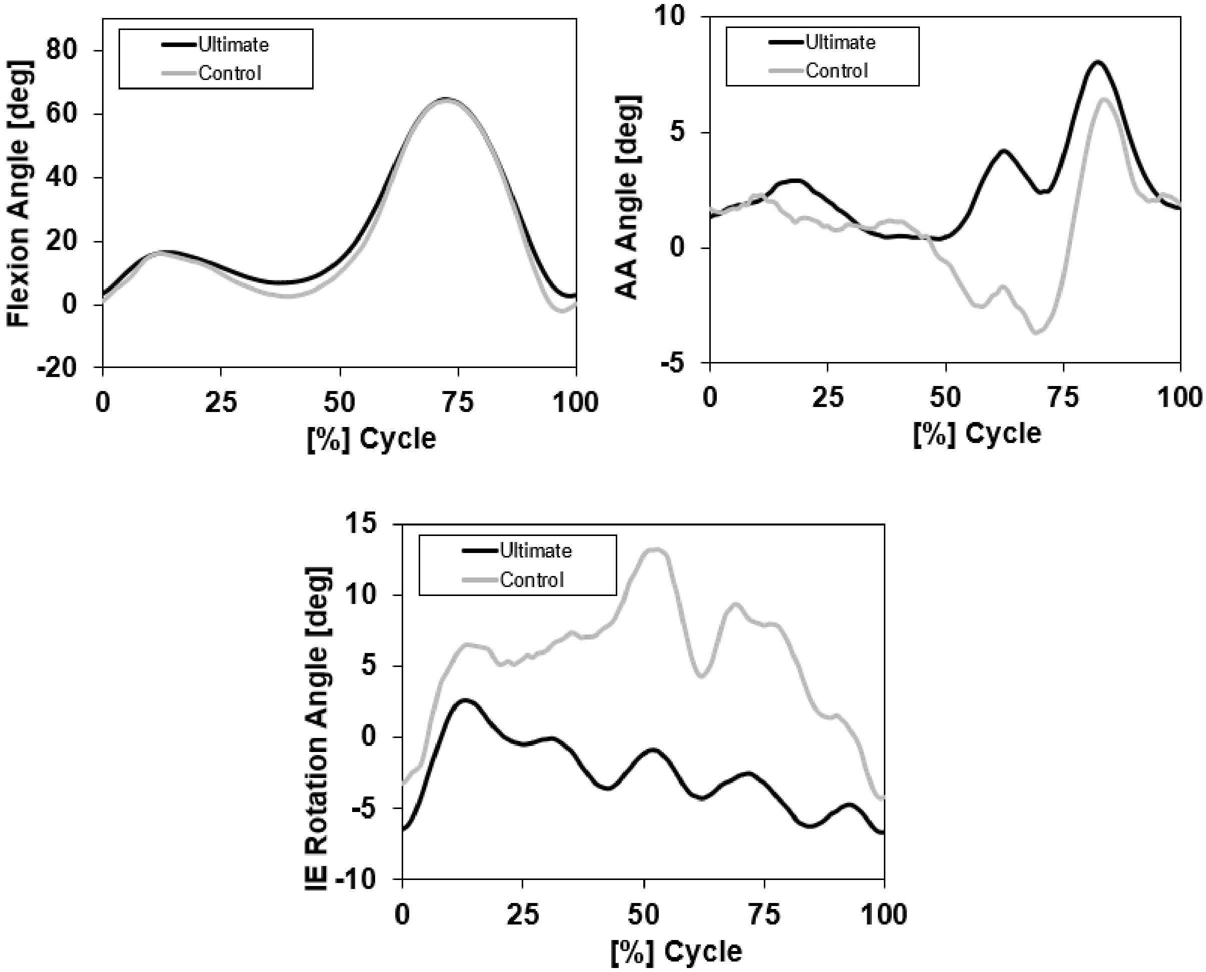
Appendix G: KNEE ANGLES



Flexion angle (Left), AA angle (right), and IE rotation angle (bottom) through a gait cycle of the combined female and male group.



Flexion angle (Left), AA angle (right), and IE rotation angle (bottom) through a gait cycle of the female group.



Flexion angle (Left), AA angle (right), and IE rotation angle (bottom) through a gait cycle of the male group.