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THE EFFECTS OF GOLF STANCE ON THE PEAK KNEE ADDUCTION MOMENT DURING THE GOLF SWING

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Quenten L. Hooker, Student
Dr. Michael Pohl, Major Professor
Dr. Heather Erwin, Director of Graduate Studies
THE EFFECTS OF GOLF STANCE ON THE PEAK KNEE ADDUCTION MOMENT DURING THE GOLF SWING

THESIS

A thesis submitted in partial fulfillment of the requirements for the degree of Master of Science in the College of Education at the University of Kentucky

By

Quenten Lowell Hooker

Lexington, Kentucky

Director: Dr. Michael Pohl, Assistant Professor

Lexington, Kentucky

2017

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INTRODUCTION: The knee joint is one of the most frequently injured structures in the game of golf. The loads experienced by the knee during the golf swing are typically greater than those experienced during walking. In particular, a heightened lead limb peak external knee adduction moment has been linked to the progression of medial compartment knee osteoarthritis (OA). Altering movement patterns is a common strategy that can be used to reduce loading on the knee joint but has received little attention during the golf swing. Also, while such manipulations may be beneficial from an injury prevention perspective, they may have implications on golf performance. The purpose of this study was to analyze the effects altering stance has on the peak knee adduction moment and swing speed during the golf swing.

METHODS: Twenty healthy subjects were recruited for a 3-dimensional biomechanical analysis wherein participants hit three golf shots using different stance positions in which either foot angle or stance width was altered. The following stance conditions were used: self-selected, 0° foot angle (perpendicular to target line), 30° foot angle (externally rotated), wide stance width, and narrow stance width.

RESULTS: Both the 30° foot angle and the wide stance width significantly decreased (p < 0.001) the lead limb peak external knee adduction moment compared to the self-selected golf stance. In contrast, the narrow stance width significantly increased (p = 0.023) the peak knee adduction moment when compared to the self-selected stance. No significant differences were found in the peak knee adduction moment between the 0° foot angle and self-selected stance. Lastly, no significant differences (p = 0.109) were found in swing speed between any of the stance conditions.

CONCLUSION: The externally rotated foot position and wider stance width decreased the lead limb peak knee adduction moment without hindering performance. Considering the prevalence of injury to the lead limb knee joint, modifying a golfer’s stance could potentially be used to increase the longevity of their playing career.

KEYWORDS: golf, biomechanics, knee adduction moment, osteoarthritis

Quenten L Hooker
July 9, 2017
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CHAPTER ONE: INTRODUCTION

The introduction section provides background information on the prevalence of knee injuries in the game of golf and the significance of the peak knee adduction moment. This section justifies the importance of conducting this study.

Introduction

Golf is a popular sport played by roughly 55 million individuals [1]. Considering its perceived low impact nature and aerobic exercise component, golf is widely recommended by medical professionals for patients wishing to remain active in the later stages of life [2, 3]. However, previous research shows that up to 72% of golfers have experienced injury, suggesting there is potential for strain on the musculoskeletal system during the golf swing [2-7]. Specifically, the knee joint is one of the most frequently injured areas, with 6-9.3% (approximately 3 million golfers) of all injuries occurring at the knee [2, 4, 7]. The forces and moments experienced by the body during the golf swing are not believed to be of sufficient magnitude to cause acute injury, which suggests overuse and chronic abnormal loading are the most likely sources of knee injury [2, 4, 5]. Since golf is the main choice of exercise for so many individuals, interventions to reduce detrimental loading of the knee are needed to expand the longevity of one’s playing career.

The external knee adduction moment has traditionally been used as a surrogate measure of the distribution of forces between the medial and lateral compartments of the knee joint [8, 9]. More importantly, previous research has shown a strong association between a high peak external knee adduction moment during gait and the presence, progression, and pain of medial compartment knee osteoarthritis (OA) [8-10]. Furthermore, the peak knee adduction moment may be of interest to a golfing population since it has
been reported to be 29% larger during the golf swing than gait [11]. Therefore, strategies to reduce the peak knee adduction moment during the swing may be helpful in terms of lowering the risk of the development or progression of knee OA in golfers.

Altering movement patterns has been used effectively to reduce loading on the knee joint during gait. Specifically, adaptations such as increasing one’s self-selected foot angle (internal/external rotation of the foot) and stance width have both been shown to decrease the peak knee adduction moment [12-16]. To date, only one study has been published of which included strategies to reduce the knee adduction moment during the golf swing. Lynn et al. [11] reported a reduction in the peak knee adduction moment when both feet were externally rotated 30°, compared to 0° (feet perpendicular to target line). Considering golfers do not typically stand with a 0° foot angle, comparing the peak knee adduction moment during altered stances in relation to a golfer’s self-selected stance may give a better representation of the potential reductions in loading. In addition, to our knowledge no prior study has examined the effects of altering stance width on the peak knee adduction moment during the golf swing. However, since alterations in stance width have been shown to decrease the knee adduction moment during gait, the strategy may also result in beneficial reductions of the moment during the golf swing.

Although altering stance has shown to decrease loading at the knee [11], manipulations to a golfer’s stance may have implications on performance [17-19]. Swing speed is a relatively simple marker of performance since it is strongly correlated to total driving distance [20]. Therefore, when considering alterations to swing technique, it is pertinent to examine whether they have negative implications for swing speed.
In summary, there is limited research exploring potential strategies to reduce the external knee adduction moment during the golf swing. Reducing knee loading during the golf swing may be beneficial for individuals with or at risk of developing medial compartment knee OA. Therefore, the primary aim of this study was to examine the effect altering golf stance has on the lead limb peak knee adduction moment. It was hypothesized that increasing foot angle and/or increasing stance width in a golfer’s setup would significantly decrease the peak external knee adduction moment. The secondary aim of this study was to examine the effect that the previously mentioned stance alterations have on swing speed.
CHAPTER TWO: METHODOLOGY

The methodology section details the specific steps that were performed during this study. This section contains information regarding the research design, study sample, instrumentation, data collection procedures, and data analysis.

Experimental Design

This was a laboratory based, quasi-experimental study design. The independent variables include foot angle (self-selected, 0°, and 30°) and stance width (self-selected, narrow, and wide). The dependent variables were the golfer’s lead limb peak external knee adduction moment and swing speed at impact.

Participants

Participants were recruited using a convenience sample from the University of Kentucky campus and local Lexington area. The following methods for recruitment were used to fulfill the final sample: word of mouth, flyers, UK PR and HealthCare social media pages: Facebook, UKTwitter, and other social media outlets. Participants were between the ages of 18-55 with a USGA golf handicap of 20 or below. Participants were excluded if they: were unable to perform multiple golf swings without pain or injury; had undergone major orthopedic surgery; or had current or previous injuries that limited golf activity in the past 3 months. A total of 22 participants were recruited, but two participants were excluded from this study. The first exclusion was due to a previous lower extremity injury. The second exclusion was due to the manner in which the golfer addressed the golf ball. Specifically, the participant’s self-selected stance used a substantially “closed” position, meaning that the golfer’s lead limb was roughly .5 meters closer to the ball than the trail
Researchers believed this posture could create an additional confounding variable. Therefore, the final sample included 20 right handed participants (Table 2.1).

**Table 2.1: Participant demographics (Mean ± SD)**

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>26.05 ± 7.13</td>
</tr>
<tr>
<td>Height (meters)</td>
<td>1.76 ± 0.09</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>78.86 ± 13.62</td>
</tr>
<tr>
<td>USGA Handicap</td>
<td>10.2 ± 6.9</td>
</tr>
<tr>
<td>Male/Female (n)</td>
<td>16/4</td>
</tr>
</tbody>
</table>

**Procedures**

Participants were asked to complete one data collection session of roughly 90 minutes. All data collections occurred in room 161 (Biodynamics Laboratory) of the Multi-Disciplinary Sciences Building, University of Kentucky. All participants provided informed consent using a form approved by the Institutional Review Board. Participants were asked to complete the Par-Q activity form to determine their physical activity readiness. Lastly, participants were asked to complete an injury history questionnaire to collect information regarding previous injuries and basic demographics.

**Data Collection Preparation**

Participants were asked to change into sports clothing, and a standardized neutral shoe (Nike Xccelerator TR, Beaverton, OR) was provided for the entire data collection. Participants were asked to address the golf ball with their normal (self-selected) golf stance (Figure 2.1A). The investigators used a pen to mark the ground next to the heel and toe of each foot and then drew a line representing the longitudinal axis of the foot in the transverse plane (Figure 2.1B). A goniometer was used to draw additional longitudinal foot axes that were standardized at 0° (i.e. feet pointing straight ahead) and 30° external rotation for both
feet, while keeping stance width constant. These lines were subsequently used to standardize the alignment of the subjects’ feet for the different stance conditions. The investigators also used the distance between the heels during the self-selected stance to calculate wide (20% wider than self-selected) and narrow (20% narrower than self-selected) stance conditions. Appropriate lines and markings were again made on the floor to assist in altering the subjects’ stance width while keeping the foot angle constant (self-selected).

Figure 2.1: Experimental setup: A) example of a participant taking a self-selected stance; B) example illustration for foot angle (larger angle represents more external rotation) and stance width.
Fifty-seven reflective markers were placed on the participant's skin over the following anatomical landmarks: bilateral acromion process, sternal notch, spinous process of the seventh cervical vertebrae (C7), spinous process of the twelfth thoracic vertebrae (T12), bilateral iliac crest, bilateral ASIS & PSIS, bilateral greater trochanter, bilateral medial & lateral knee, bilateral medial & lateral malleoli, bilateral lateral heel, bilateral proximal & distal heel, bilateral 1st & 5th metatarsal head, bilateral toe, and an offset marker on the right foot (Appendix A). Lastly, rigid body clusters of 4 markers were placed on the anterior/lateral aspect of the subject’s right shank and left thigh/shank, while 5 markers were used on the right thigh (Appendix A).

Data Collection

Three-dimensional marker co-ordinate data were collected for both a static standing trial and the dynamic golf swing trials using ten high speed Motion Analysis Cameras (Motion Analysis Corp, Santa Rosa, CA) at a sampling rate of 200 Hz. A SC100 radar device (Voice Caddie Corp, La Mirada, CA) was placed 2 meters behind the golf ball to measure swing speed. Furthermore, kinetic data were collected at 1000 Hz using two Bertec force plates (Bertec Corporation, Columbus, OH), synchronized with Cortex 5.5 (Motion Analysis Corporation, Santa Rosa, CA, USA) motion software. Initially, participants were asked to stand in the calibrated volume, while a static calibration was taken. Next, participants were given a warm-up period (approximately 5 minutes) until comfortable taking full length golf swings. Participants were then asked to hit 3 golf drives into the net using the following stance conditions: self-selected, 0º foot angle, 30º foot angle, narrow stance width, wide stance width. The order of the stance conditions was block randomized. Participants were allowed rest as needed between each of the various
trials, and an acclimation period was given during the transition of each stance position.

Data Processing

Marker trajectory data were tracked using Cortex software (Motion Analysis Corp., Santa Rosa, CA), while further data processing was conducted using Visual 3D software (C-Motion Inc., Germantown, MD). Raw marker trajectory data were filtered using a fourth order low-pass Butterworth filter with a cut-off frequency of 12 Hz [11]. An X-Y-Z cardan sequence (sagittal-frontal-transverse) was used to quantify joint angles, in which the distal segment is expressed relative to the proximal segment [21, 22]. In addition, the pelvis was expressed relative to a virtual lab coordinate system. An adapted version of the marker set and model from Joyce et al. [21] and Nesbit et al. [22] were utilized for the lower extremity kinematic and kinetic data (Appendix B). Discrete variables of interest included the lead limb peak external knee adduction moment. The peak knee adduction moment was determined by the greatest value observed between the “top of the backswing” and the “finish” of the golf swing. Lastly, “top of backswing” and “finish” events were used to time normalize data, and an ensemble mean value for three consecutive golf swings was calculated for each subject for each of the five stance conditions.

Statistical Analysis

Repeated measures ANOVA analyses were used to determine if there were differences in the peak knee adduction moment and swing speed between the stance conditions. A planned contrasts analysis was used to determine which (if any) stance conditions were significantly different from the self-selected condition at an alpha level of $p < 0.05$. Furthermore, Pearson product-moment correlations were performed between: i) the change in foot angle vs. the change in the peak knee adduction moment between the
self-selected and 30° foot angle conditions, ii) the change in stance width vs. the change in the peak knee adduction moment between the self-selected and wide stance width conditions. Interpretation of Pearson product-moment correlation coefficient (r) followed guidelines by Hinkle et al. [23] for very strong (r>0.90), strong (r = 0.70-0.90), moderate (r = 0.50-0.69), low (r = 0.30 – 0.49), and negligible (r = 0.00 – 0.29). All statistical analyses were performed using SPSS 24 statistical software (SPSS Inc., Chicago, IL).
CHAPTER THREE: RESULTS

The results section presents the findings of this study, including the ANOVA, post-hoc, and correlation analyses that were conducted.

Results

Descriptive statistics for the peak external knee adduction moment and swing speed for all stance conditions are presented in Table 3.1, while ensemble mean curves of the peak knee adduction moment are shown in Figure 3.1. On average, the participants addressed the ball with a foot angle of 11.3 ± 5.3º external rotation and stance width of 0.49 ± 0.07 meters. The peak knee adduction moment was significantly different (p < .001) between the five stance conditions (Table 3.1). The planned contrasts analyses revealed both the 30º foot angle and wide stance width conditions significantly decreased the peak knee adduction moment (p < .001) when compared to self-selected. In contrast, the narrow stance width condition significantly increased (p = .023) the peak knee adduction moment when compared to self-selected. No significant differences (p = 0.605) were found in the peak knee adduction moment between the 0º and self-selected foot angle conditions. Furthermore, a negligible correlation was found between the change in foot angle vs. the change in the knee adduction moment (r = -.228, p = 0.333) and the change in stance width vs. the change in knee adduction moment (r = .040, p = 0.866) (Figure 3.2). In terms of the secondary aim there were no significant differences (p = .109) in swing speed between any of the stance conditions (Table 3.1).

A similar trend in terms of the change in peak knee adduction moment between stance conditions was observed for males and females. Specifically, males experienced 29% and 23% reductions in the knee adduction moment during the 30º and wide stance
conditions respectively, while females experienced a 35% and 23% reductions in loading for the equivalent conditions. Also, a similar increase in the peak knee adduction moment was observed in males (5%) and females (16%) for the narrow stance width, when compared to a self-selected stance width.

Table 3.1: Mean ± SD for peak knee adduction moment and swing speed for golf stance conditions

<table>
<thead>
<tr>
<th>Stance Condition</th>
<th>Peak Knee Adduction Moment (Nm/kg)</th>
<th>Swing Speed (mph)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Self-selected</td>
<td>1.15 ± 0.58</td>
<td>98.87 ± 10.53</td>
</tr>
<tr>
<td>0 degree</td>
<td>1.12 ± 0.54</td>
<td>97.90 ± 10.43</td>
</tr>
<tr>
<td>30 degree</td>
<td>*0.80 ± 0.51</td>
<td>98.30 ± 9.85</td>
</tr>
<tr>
<td>Narrow</td>
<td>*1.23 ± 0.57</td>
<td>97.30 ± 9.41</td>
</tr>
<tr>
<td>Wide</td>
<td>*0.89 ± 0.49</td>
<td>98.53 ± 10.64</td>
</tr>
</tbody>
</table>

* Significantly different from Self-selected (p<.05)

Figure 3.1: Ensemble mean of frontal plane moments for lead limb knee joint. Percent swing is normalized from “top of backswing” to “follow through”. Vertical dashed line represents “impact”.
Figure 3.2: Relationship between A) change in foot angle vs. the change in the peak knee adduction moment (self-selected vs. 30° foot angle); B) change in stance width vs. the change in the peak knee adduction moment (self-selected vs. wide stance width)
The discussion and conclusion sections interpret the results that were reported in chapter three. It also includes a listing of potential limitations of the study and recommendations for future research.

Discussion

The primary aim of this study was to examine the effect of altering golf stance on the peak knee adduction moment. Our hypothesis was confirmed in that both the 30° foot angle and wide stance width conditions significantly decreased (p < .001) the peak knee adduction moment when compared to the self-selected stance. Although previous literature has also reported a reduction in the peak knee adduction moment when the feet are placed in greater external rotation, the magnitude of the change differed slightly from our findings. Specifically, Lynn et al. [11] found a 14.3% reduction in the peak knee adduction moment when the feet were externally rotated 30°, while we found a 30.4% reduction between the 30° and self-selected conditions. Considering participants in the present study used a driver for all golf shots whereas Lynn et al. [11] utilized a 5 iron, we attribute our greater reduction of the knee adduction moment to the various kinematic differences in technique and typically greater exertion when using a driver [24-26].

To date, our results are the first to examine the effect of stance width on the peak knee adduction moment during the golf swing. However, our findings mirror those reported during gait modification studies. Specifically, Favre et al. [14] and Fregley et al. [15] found 17.1% and 9% reductions in the peak knee adduction moment respectively when participants widened their stance during walking, while we found a 22.6% reduction between the wide and self-selected stance width conditions. In addition, Favre et al. [14]
found a 13.7% increase in the peak knee adduction moment in gait when participants utilized a narrow stance width, while we found a 6.9% increase between narrow and self-selected during the golf swing. Given that the knee adduction moment is larger during the golf swing in comparison to walking, a 22.6% reduction in the peak knee adduction moment during the golf swing would correspond to an even larger reduction in the absolute moment than those reported in walking.

Although the 30º foot angle and wide stance width conditions successfully reduced loading at the knee, a negligible correlation was found between the magnitude of the change in the stance parameter (foot angle or stance width) and the change in knee adduction moment (Figure 3.2). Participants in this study addressed the golf ball with a range of self-selected foot angles (1.1-23.5º) and stance widths (0.36-0.64m), thus requiring individuals to change their foot angles and stance widths varying extents to match the appropriate modification. Collectively, 19 and 18 out of the total 20 golfers reduced their peak knee adduction moment when altering stance to 30º foot angle and wide stance width conditions respectively. However, considering the negligible correlation, a greater change in foot angle or stance width does not necessarily result in a greater reduction in loading at the knee. It is possible that the negligible correlation may be partially explained by individual differences in anatomy such as frontal plane knee alignment, which may cause golfers to respond differently to the stance modifications.

The findings of the present study may also have clinical implications for populations who have or at risk of developing medial compartment knee OA. It has been widely proposed that reducing the knee adduction moment may in turn reduce loads placed on the medial compartment knee joint [8-12]. Therefore, reducing the knee adduction
moment has become a common strategy to not only slow the development/progression of medial compartment knee OA, but also alleviate symptoms from the disease [8-12]. The findings of the present study suggest that adopting an externally rotated foot position or wider stance width may potentially be beneficial to golfers with medial compartment knee OA or those at risk of developing the disease.

The second purpose of this study was to analyze the effect altering a golfer’s stance has on performance. Golf performance has been previously broken down into two components; distance and accuracy. For this study, we analyzed swing speed since it has been strongly associated to total distance [20]. Our results found no significant differences in swing speed between the five stance conditions; indicating the alterations in stance did not hinder the ability for the golfer to generate maximum swing speed. Furthermore, researchers believe that the previously mentioned stance conditions will not prevent a golfer from hitting the golf ball his or her maximum distance potential. This suggests that the externally rotated foot position and the wider stance were both successful in terms of decreasing the peak knee adduction moment without hindering performance.

**Limitations**

As with all studies, this investigation was not without limitations. Errors due to marker placement error are always possibilities in 3-dimensional motion capture, although marker placement was performed by one trained investigator to avoid inter-tester variability. In addition, the biomechanical model used in this study cannot account for individual differences in anatomy such as frontal plane knee alignment. Another limitation in this study is researchers were only able to measure swing speed as a performance variable. Measuring accuracy in addition to swing speed may provide a more complete
picture as to how altering a golfer’s stance effects performance. Therefore, future researchers should assess both swing speed and accuracy as performance variables.

Next, only healthy participants were recruited for data collection. Authors have reported that symptomatic OA patients walk differently when compared to asymptomatic controls [8, 27, 28]. In addition, patients tend to vary their walking style depending on the severity of knee OA; possibly in attempt to further decrease the adduction moment [27, 28]. Therefore, current findings should be interpreted with caution when relating to a population with medial compartment knee osteoarthritis. Furthermore, there is a possibility that heightened frontal plane knee moments impact not only the development and progression of osteoarthritis, but also increase the potential for ACL damage [29]. Therefore, future research should investigate the relationship between frontal plane knee moments and ACL injury during the golf swing.

Conclusion

The results of the present study indicate that an externally rotated foot position or a wider stance width decreased the peak knee adduction moment when compared to a self-selected golf stance. The non-significant changes in swing speed between stance conditions suggest the previously mentioned alterations in stance may be used to decrease joint loading without hindering the golfer’s performance. Therefore, adopting a 30º foot angle or a wider stance width may be viable options for golfers who wish to continue playing the sport at a high level, while reducing potential detrimental loads at the knee joint. In particular, the findings may have clinical implications for those individuals who are at risk of the development or progression of medial compartment knee osteoarthritis.
APPENDIX A: MARKER SET

Anatomical Marker

Tracking Marker

17
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<th>Description</th>
<th>Abbreviation</th>
<th>Description</th>
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<tr>
<td>RTOE</td>
<td>Right Toe</td>
<td>LTOE</td>
<td>Left Toe</td>
</tr>
<tr>
<td>ROFF</td>
<td>Right Offset</td>
<td>L1MH</td>
<td>Left 1&lt;sup&gt;st&lt;/sup&gt; Metatarsal head.</td>
</tr>
<tr>
<td>R1MH</td>
<td>Right 1&lt;sup&gt;st&lt;/sup&gt; Metatarsal head.</td>
<td>L5MH</td>
<td>Left 5&lt;sup&gt;th&lt;/sup&gt; Metatarsal head.</td>
</tr>
<tr>
<td>R5MH</td>
<td>Right 5&lt;sup&gt;th&lt;/sup&gt; Metatarsal head.</td>
<td>LLHE</td>
<td>Left Lateral Heel</td>
</tr>
<tr>
<td>RLHE</td>
<td>Right Lateral Heel</td>
<td>LDHE</td>
<td>Left Distal Heel</td>
</tr>
<tr>
<td>RDHE</td>
<td>Right Distal Heel</td>
<td>LPHE</td>
<td>Left Proximal Heel</td>
</tr>
<tr>
<td>RPHE</td>
<td>Right Proximal Heel</td>
<td>LLMA</td>
<td>Left Lateral Malleolus</td>
</tr>
<tr>
<td>RLMA</td>
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<td>LMMA</td>
<td>Left Medial Malleolus</td>
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<tr>
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<td>LTAS</td>
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<tr>
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<td>LTPS</td>
<td>Left Top Posterior Shank</td>
</tr>
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<td>LBPS</td>
<td>Left Bottom Posterior Shank</td>
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<td>Right Top Posterior Thigh</td>
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<td>Left Greater Trochanter</td>
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<td>LASI</td>
<td>Left Anterior Superior Iliac Spine</td>
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<tr>
<td>RGTR</td>
<td>Right Greater Trochanter</td>
<td>LPSI</td>
<td>Left Posterior Superior Iliac Spine</td>
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<td>LICR</td>
<td>Left Iliac Crest</td>
</tr>
<tr>
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<td>LSHL</td>
<td>Left Shoulder (Acromion Process)</td>
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<tr>
<td>RICR</td>
<td>Right Iliac Crest</td>
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<td>Proximal Shaft</td>
</tr>
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<td>DSHA</td>
<td>Distal Shaft</td>
</tr>
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<td>Cervical Vertebrae #7</td>
<td>PFCL</td>
<td>Proximal Anterior Club</td>
</tr>
<tr>
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<td>Sternal Notch</td>
<td>DFCL</td>
<td>Distal Anterior Club</td>
</tr>
<tr>
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<td>Right Shoulder (Acromion Process)</td>
<td>MBCL</td>
<td>Posterior Club</td>
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<td>Ball</td>
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APPENDIX B: 3-D MODEL DEFINITION

Virtual lab

Due to the orientation of the data collections, a virtual laboratory coordinate system was defined. The virtual lab coordinate system was identical to the original lab coordinate system, with a 90° clockwise rotation about the superior-inferior (Z) axis. The superior-inferior axis was defined as Z (superior +), the medio-lateral axis was defined as X (right +), and the antero-posterior axis was defined as Y (anterior +).

Pelvis

A CODA pelvis was used to define a pelvic coordinate system relative to the virtual laboratory [21, 30, 31]. Initially, the Y axis (anterior +) was defined from the midpoint of the PSIS markers to the midpoint of the ASIS markers. Next, a temporary axis was defined from the midpoint of the PSIS markers to the left ASIS marker. The Z axis (superior +) was defined by the cross products between the Y and temporary axis. Finally, the X axis (right +) was defined by the cross product between the Y and Z axis.

Thigh

A temporary axis was formed between the LMKN and LLKN markers and the LGTR marker. Next the Z axis (superior +) was defined from the midpoint between LMKN and LLKN markers to the left hip joint center. The Y axis (anterior +) was defined by the cross product between the temporary axis and the Z axis. Lastly, the X axis (right +) was defined as the cross product between the Y axis in the Z axis. The tracking markers were a rigid cluster of four (LTAT, LTPT, LBAT, and LBPT) or five markers (RTAT, RTPT, RBAT, RBPT, and RMID) placed on the antero-lateral portion of the distal thigh.
Shank

The Z axis (superior +) was defined from the midpoint between the LLMA and LMMA markers to the LLKN and LMKN markers. A temporary axis was defined by the LLMA and LMMA markers to the LLKN marker. The Y axis (anterior +) defined by the cross product between the temporary axis and the Z axis. Lastly the X axis (right +) was defined as the cross product between the Y and Z axis. Tracking markers were defined by the left and right shank clusters. The tracking markers were a rigid cluster of four markers (LTAS, LTPS, LBAS, and LBPS) or (RTAS, RTPS, RBAS, and RBPS) placed on the antero-lateral portion of the distal shank.

Foot

The Y axis (anterior +) was defined from the midpoint between the LLMA and LMMA markers to the L5MH and L1MH markers. A temporary axis was defined from the midpoint between LLMA AND LLMMA markers to the L5MH marker. Next the Z axis (superior +) was defined as the cross product between the Y axis and the temporary axis. Lastly the X axis (right +) was defined as the cross product between the Y axis in the Z axis. Tracking markers for the foot include LDHE, LPHE, and LLHE. The “Foot” model was used primarily for kinetic calculations.

Virtual foot

Initially LDHE marker was projected onto the frontal plane made up by LPHE markers and LPHE landmarks. Next, both L5MH and L1MH were projected upward on to the transverse plane of LPHE markers and LPHE landmarks. The Z axis (superior +) was defined from the projected LDHE marker to LPHE. The Y axis (anterior +) was defined from the LPHE marker to the midpoint between the projected L5MH and L1MH markers.
Lastly, the X axis (right +) was defined by the cross product between the Y and Z axis. The virtual foot model was only used for kinematic variables such as foot angle and stance with.
APPENDIX C: ANKLE, KNEE, AND HIP MOMENTS

Ankle Moment (X)

Ankle Moment (Y)

Ankle Moment (Z)

Knee Moment (X)

Knee Moment (Y)

Knee Moment (Z)

Hip Moment (X)

Hip Moment (Y)

Hip Moment (Z)
APPENDIX D: EXPANDED LITERATURE REVIEW

Golf Background

The popularity of golf has continually increased since the game’s origin around the 13th century [32]. As of 2003, it was estimated that there were 30,000 golf courses and 55 million people playing golf around the world [1]. One appealing factor of the game of golf is that there are no age, gender, or skill limitations to participate [1]. Although golfers can be of all ages, participation is shown to be higher in older age brackets [4]. This may result from golf being perceived to be a low impact activity with general aerobic benefits, which makes the sport a popular recommendation for health care providers that promote an active lifestyle in older individuals [4]. Not only is golf a popular option to remain active, but a golfer reaches his or her peak performance at an older age than most other sports [33].

In golf, skill level is primarily determined by the number of shots one completes during an 18 hole round of golf. The USGA Handicap system is a quantitative measure of the average strokes over par and is established by taking the lowest 10 scores out of the last 20 golf rounds [34]. For example, if an individual takes 82 shots to complete a par 72 golf course, they would have an approximate handicap of 10.0 (this varies depending on course characteristics and best 10 out of previous 20 rounds). The lower the handicap, the more skillful the golfer. Currently, the average USGA handicap is roughly 15.0 [34]. Considering that golfer’s wish to better their score through learning new information and practice, many researchers have aimed to examine various factors relating to golf performance.

Golf Performance
Golf performance has been a popular research topic for centuries. Specifically, the golf swing has attracted considerable research from a biomechanical analysis perspective, since a better performance could potentially lead to significant financial compensation at the professional level. Although numerous researchers have studied how to improve one’s golf performance, there has yet to be a conclusive biomechanical explanation for how to properly perform the golf swing [35]. Some researchers believe the variability in anatomy, mobility, strength, and motor control between golfers make it impossible to determine a perfect golf swing [35]. Thus, a large portion of golf performance literature suggest there are certain components of a golf swing one might improve, but the overarching kinematic and kinetic sequence will vary by the individual.

Although it is difficult to analyze biomechanical factors that can increase every golfer’s skill, measuring golf performance is relatively simple. The most common markers of golf performance are swing speed and accuracy. Swing speed refers to the speed the club is traveling at the instant the golf club meets the ball [18, 36]. Furthermore, swing speed and resultant force is shown to be the largest when utilizing a driver, compared to other golf clubs (i.e. irons or wedges) [24, 26]. This measure is closely related to the total distance the golf ball travels and can be easily measured with various radar devices [18, 36]. Accuracy refers to the distance from the intended target the golf ball travels. Considering the goal of golf is to hit the ball at a very specific target, accuracy is widely considered the most important performance variable in the game of golf [17]. Measuring accuracy in golf is slightly more complex, and requires the use of devices that can measure various characteristics of impact (i.e. impact speed, face angle, spine rate). When altering swing biomechanics, it is critical that performance be assessed because it is highly unlikely
that a golfer would use an intervention that is detrimental to their ability to swing the golf club effectively.

**Golf Swing Phases**

Due to the complex nature of the golf swing, it is common for researchers to evaluate the golf swing in phases [37, 38]. The three phases of the golf swing are the backswing, downswing, and follow-through.

![Golf Swing Phases Diagram](image)

**A) Setup**  **B) Top of backswing**  **C) Impact**  **D) Finish**

**Backswing**

The purpose of the backswing is to align the golfer’s body and club so that the golfer can execute a powerful and accurate downswing [38]. For the right handed golfer, the backswing is initiated at the setup (A) with a clockwise movement of the golf club in the frontal plane and ends when the golf club reaches maximum height (B). In addition, the end of the backswing phase can be determined by a club velocity of 0 mph (change from clockwise to counter-clockwise speed in the frontal plane).

**Downswing**

The purpose of the downswing is to return the club head to the ball with maximum velocity [38]. The downswing phase starts at the end of the backswing phase, when the
club is at maximum height (B). The golf club initiates a counter-clockwise velocity, which continues to accelerate until the end of the downswing phase at ball contact (C).

Follow-through

The purpose of the follow-through phase is to decelerate the body and golf club [38]. Although the follow-through does not influence the distance or accuracy of the golf shot, this phase is believed to decrease the likelihood of injury by slowing rotational forces over a large time window [37]. The follow-through phase starts at ball impact (C), and the golf club continues a counter-clockwise velocity in the frontal plane until the club reaches a velocity of 0 mph at the finish position (D).

Golf Injuries

Golf is often considered to be a low impact and injury free sport, yet previous literature suggests up to 72% of golfers have experienced an injury during their golfing career [7]. Numerous authors have examined the prevalence of specific injury locations in the game of golf. McHardy et al. [3] investigated injury occurrence by surveying over 1600 professional and amateur golfers. Results indicated the low back (23.7-55%) was the most common injury location, followed by the wrist (18-35%), elbow (8-30%), shoulder (8-12%), and knee (6-9.3%), respectively [2, 3]. A similar trend in injury location is shown across gender and performance level, but injury prevalence in the game of golf is shown to be higher in professional golfers (low handicap) compared to amateurs (high handicap) [2]. Considering the rather high prevalence for injury in the game of golf, research is needed to further understand underlying mechanisms that can contribute to injury during the golf swing.
To date, previous research shows overuse and chronic loading are the most likely mechanisms for injury during the golf swing [2, 3, 39, 40]. Although not as common as the low back, knee injuries occur during the golf swing [2, 3]. Specifically at the golfer’s lead limb knee joint, forces and moments are not believed to be of sufficient magnitude to cause acute injury [2, 3, 39, 40]. Thus, previous researchers suggest the cumulative abnormal loading of knee joint is most likely responsible for the prevalence of knee injuries during the golf swing [2]. Furthermore, Bechler et al. [41] aimed to further investigate injuries in golf by measuring EMG activity of a wide range of muscles used in the golf swing. Results showed that activity of muscles in the golfer’s lead limb reach up to 83% of peak contraction [41]. Bechler et al. [41] also suggested that the increased muscle activity in the lead limb could be present to attenuate the substantial loads placed on the ankle, knee, and hip joints during the downswing and follow-through. Therefore, a chronic abnormal loading coupled with intense muscular contractions could put the golfer’s knee joint at a heightened chance of injury.

**Knee Joint Anatomy**

The knee is located at the junction of the thigh and shank and is made up of three joints: tibiofemoral, patellofemoral, and superior tibiofibular [42]. These joints, along with the supporting structures, combine to create a very stable yet mobile structure that is critical for human locomotion.

**Bones and Ligamentous Structures**

The knee is composed of four bones: femur, patella, tibia, and fibula. Specifically, the medial and lateral condyles of the femur articulate with the condyles of the tibia. This allows for a large contact area at the tibiofemoral joint, with the medial and lateral menisci
acting to absorb shock and enhance joint congruity [42]. The knee also contains four ligaments which help to guide, control, and limit the various actions at the knee. The anterior cruciate ligament (ACL) resists anterior tibial movement (posterior femoral movement) and medial tibial rotation, while the posterior cruciate ligament (PCL) restricts posterior tibial movement (anterior femoral movement) and lateral rotation of the tibia. The medial collateral ligament (MCL) restricts tibial abduction (knee valgus) and lateral rotation of the tibia. Lastly, the lateral collateral ligament (LCL) limits tibial adduction (knee varus) and medial rotation of the tibia. All of the previously mentioned bony structures and ligaments help to promote dynamic and static stability [42].

Musculature

Muscles functioning to control the knee joint act to predominately move the thigh and shank in the sagittal plane. The anterior thigh muscles (rectus femoris, vastus lateralis, vastus medialis, and vastus intermedius) act to extend the knee while the posterior thigh muscles (semitendinosus, semimembranosus, and biceps femoris) act to flex the knee. Although motion primarily occurs in the sagittal plane, the knee is capable of small joint excursions in the frontal and transverse planes. Muscles in the medial thigh (adductor longus, adductor brevis, and adductor magnus) facilitate hip adduction, which alternatively places the knee in an abducted (valgus) position. The medial thigh muscles also work to internally rotate the femur on the tibia. Lastly, muscles on the posterior-lateral portion of the proximal thigh (gluteus maximus, gluteus minimus, tensor fascia latae) facilitate hip abduction, thus a relative knee adduction (varus) position, and laterally rotate the thigh on the tibia. This is a non-exhaustive list and there are many additional connective tissues within the knee joint complex [42].
Osteoarthritis

Osteoarthritis (OA) is a degenerative joint disease which affects over 40 million people in the United States [43-45]. Although the description varies, OA can be defined pathologically, radiographically, or clinically [43-45]. Radiographic OA has previously been considered the gold standard, with the presence of osteophytes, joint space narrowing, sclerosis, cysts, and deformity ultimately determining the severity of OA [43, 44]. Although the exact etiology is unknown, OA can be the product of many factors. Old age, female gender, obesity, injury, repetitive use of joints, bone density, muscle weakness, and joint laxity have a role in the development of joint osteoarthritis [43, 44]. In addition to the high prevalence of OA, approximately 10 to 30% of those affected by OA have significant pain leading to disability [44]. The estimated cost of this disability is nearly 65 billion dollars annually [44].

Although most joints in the lower extremity, including the ankle and hip, can develop OA, the knee is the most commonly affected joint [46]. Current literature shows 33-68% of persons over 55 years have radiographic evidence of knee OA [46]. In addition, knee OA can be present in the medial, lateral, and/or patellofemoral joints [46-48]. The prevalence of medial compartment knee OA (13.8-28.6%) is much higher than patellofemoral OA (5.4-16.7%) and lateral compartment OA (2.0-6.0%), which may be due to the heightened loading transferred to the medial compartment during normal gait [46-48].

Both abnormal knee joint anatomy and mechanical loading of the joint have been researched as mechanisms contributing to changes in OA. Specifically, those with varus alignment of the lower limb or heightened loading of the medial compartment are more
likely to develop medial knee joint OA, whereas individuals with valgus alignment or heightened loading of the lateral compartment have an increased risk of lateral joint disease development/progression [44, 49]. Therefore, characteristics such as joint anatomy and loading are additional precursors to changes in knee osteoarthritis.

**External Knee Adduction Moment**

The external knee adduction moment has traditionally been used as a surrogate measure of the distribution of forces between the medial and lateral compartments of the knee joint [8, 9]. More importantly, a high external knee adduction moment has been strongly associated with the presence, progression, and pain of medial compartment knee osteoarthritis [8, 9]. Thus, interventions to reduce the external knee adduction moment during movement has received attention in the literature.

**Orthotic Devices**

The use of various orthotic devices (i.e. lateral shoe wedges, valgus knee braces, etc) have been explored as a strategy to reduce the knee adduction moment [50]. First, lateral wedge insoles force a lateral shift in the center of pressure, ultimately decreasing the peak knee adduction moment [51, 52]. By decreasing loads on the medial compartment, patients using lateral wedge insoles have experienced improvements in pain and function [51, 52]. Similarly, valgus knee braces can also be used to reduce the knee adduction moment [53-55]. The valgus knee brace increases the extent of valgus knee alignment, which may decrease excessive medial compartment load responsible for tibiofemoral OA [53-55]. Although both previously mentioned orthotic devices have been proven to decrease the peak knee adduction moment, the success of such interventions (i.e. decreased pain and slowed progression of OA) rely on continual use of the device. Considering the
valgus knee braces may be bulky and lateral wedges may be uncomfortable, some
investigators question the efficacy of such orthotic devices for long-term use [50].

Gait Modifications

Altering movement patterns during gait has also been explored as a strategy to
reduce resultant loading on the joints of the lower extremity, and thus reduce the risk of
developing/progressing knee OA. Specifically, gait modifications such as decreasing gait
velocity, increasing step width, and increasing foot progression angle have significant
effects on frontal plane loading of the knee joint. Robbins et al. [56] compared peak knee
adduction moments between slow, fast, and self-selected walking speeds. Results showed
significant differences in peak knee adduction moment between the three conditions, with
a greater peak knee adduction moment during the fast condition when compared to the
slow and self-selected conditions [56]. Both Fregly et al. [15] and Street et al. [16]
investigated step width’s role in the knee adduction moment, finding that narrowing and
widening both have the potential to lower the frontal plane loading when compared to a
self-selected amount of step width. Lastly foot progression, foot angle in the transverse
plane (i.e. toe out vs toe in) during stance, has also been investigated in attempt to lower
the knee adduction moment [12, 13]. Gebrands et al. [12] found an increase in foot
progression (toe out) resulted in a decrease of the second peak knee adduction moment.
Similarly, Lynn et al. [13] demonstrated that when the foot was externally or internally
rotated, the knee adduction moment significantly decreased (.25 Nm/kg to .02 Nm/kg) and
increased (.25 Nm/kg to .41 Nm/kg) respectively late in stance. These results regarding
walking speed, step width, and foot progression were confirmed by Favre et al [14].

Knee Adduction Moment in Golf
The external knee adduction moment during the golf swing can be up to 1.5 times greater than the values reported during gait [6, 11]. Although only one group of researchers have explored strategies to reduce the knee adduction moment during the swing, the interventions previously discussed for walking might also be helpful in golf. Lynn et al. [11] examined how altering foot angle (angle that the feet point out at) could potentially affect the knee adduction moment. Golfers were asked to slightly alter their stance between conditions of neutral (both feet pointing straight ahead) and externally rotated (30º toe out angle) [11]. A reduction in the peak knee adduction moment was found when the target limb was in the externally rotated position [11]. Considering the paucity of studies analyzing the peak knee adduction moment and various interventions to reduce loading, future research is needed to see if modifications in a golfer’s stance can be used to alter the peak knee adduction moment during the golf swing.

**Summary**

In summary, golf is a popular game with numerous health benefits. Although golf is believed to be a low impact activity, previous research suggests players could be at risk of injury from both overuse and chronic abnormal loading. Specifically, the peak external knee adduction moment, which is directly correlated to the presence, progression, and pain of medial compartment knee osteoarthritis, is higher during the golf swing than normal gait [8, 10, 27]. Few researchers have analyzed the peak knee adduction moment in golf and potential modifications in a golfer’s stance to decrease the loading. Thus, future research is needed to investigate how altering stance affects the peak knee adduction moment during the golf swing, since reducing loading may have clinical implications in terms of helping lower an individual’s risk of developing medial compartment knee OA.
REFERENCES


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