Influence of the Lumbopelvic-hip Complex on Female Softball Hitting

by

Jessica Kate Washington

A dissertation submitted to the Graduate Faculty of
Auburn University
in partial fulfillment of the
requirements
for the Degree of
Doctor of Philosophy

Auburn, Alabama
August 4, 2018

Keywords: batting, hand velocity, hips, pelvis, torso, timing

Approved by

Gretchen D. Oliver, Chair, Associate Professor of Kinesiology
Wendi H. Weimar, Professor of Kinesiology
Sheri J. Thornburg-Brock, Associate Professor of Kinesiology
David M. Shannon, Humana Germany Sherman Distinguished Professor of Educational
Research and Evaluation

Abstract

Softball hitting is one of the most difficult skills in sport. Lumbopelvic-hip complex (LPHC) stability, as well as proper proximal to distal segmental sequencing, can directly influence hitting performance, because segment and implement (bat) velocities and positions dictate ball contact, and thus the outcome of the swing. To the author's knowledge, no data exist that examine the influence of the LPHC on hitting performance indicators, such as hand velocity, in softball athletes. Therefore, the purpose of this project was to determine influences of the LPHC on angular hand velocity; specifically examining the relationship of hip internal and external isometric strength; pelvis and torso rotation separation; pelvis and torso rotational jerk; load timing; and temporal components of maximum angular velocity of the hips, pelvis, and torso to hand velocity throughout the swing. Results revealed a significant, negative correlation between pelvis and torso separation and hand angular velocity at ball contact (r = -0.351, p = 0.039), as well as, a significant, negative correlation between timing of peak angular velocity of the pelvis during the acceleration phase and hand angular velocity at ball contact (r = -0.379, p = 0.028). No other statistically significant findings were observed in this study. Temporal components of the kinematics measured in this study may be of most benefit for practical application to performance improvements in hitting; however, future research is needed to reliably support this notion.

Acknowledgements

First, I would like to thank Dr. Gretchen Oliver for her mentorship the past 3 years as a Doctoral student, and Dr. Wendi Weimar for her mentorship as an undergraduate intern and Master's student. Each of you influenced my growth as a student, mentor, researcher, and person more than anything or anyone, and I will be forever grateful.

Next, I would like to thank the members of my dissertation committee, in addition to the outside reader for this dissertation. Thank you, Dr. Gretchen Oliver (chair), Dr. Wendi Weimar, Dr. David Shannon, Dr. Sheri Thornburg-Brock, and Dr. Michael Stallings for your continued support, suggestions, and encouragement throughout this process and my academic career at Auburn University. Also, I would like to thank Jaimie Roper for her advisement and assistance in this dissertation, as well as her guidance in my professional development the past year.

I would like to thank all members of the Sports Medicine and Movement Laboratory: Matthew Hanks, Sarah Gascon, Jeff Barfield, Portia Williams, Kenzie Friesen, Kyle Wasserberger, Jessica Downs, Gabrielle Gilmer, and Abby Brittain for their assistance during data collections and preparations for this study. Lastly, thank you to all members of the Sport Biomechanics Laboratory: Dr. Chris Wilburn, Brandi Decoux, Nick Moore, and Randy Fawcett for your encouragement and assistance, as well.

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List of Abbreviations

LPHC	Lumbopelvic-hip Complex
%BF	Percent Body Fat
Hz	Hertz
ICC	Intra-class Correlation
kgf	Kilogram-force
%BM	Percent Body Mass
IR	Femoral Internal Rotation
ER	Femoral External Rotation
ISO	Isometric Strength
deg	Degrees
deg/s	Degrees per Second – Unit of Angular Velocity
deg/s^3	Degrees per Second Cubed – Unit of Angular Jerk
% Swing	Percent of the Swing
% Acceleration Phase	Percent of the Acceleration Phase of the Swing

List of Symbols

N	Participant Sample
$\mathbf{H}$	Angular Momentum
$\mathbf{I}$	Moment of Inertia
$\boldsymbol{\omega}$	Angular Velocity
$\mathbf{k}$	Radius of Gyration
$^{\circ}$	Degrees
α_3	Skewness
α_4	Kurtosis
r	Correlation Coefficient
R^2	Coefficient of Determination
p	Alpha Level of Significance
SE_{skew}	Standard Error of Skewness
SE_{kurt}	Standard Error of Kurtosis
$\mathbf{b}$	Unstandardized Regression Coefficient
$\boldsymbol{\beta}$	Standardized Regression Coefficient

CHAPTER I

INTRODUCTION

Softball hitting is one of the most difficult skills in sport (Fleisig, Zheng, Stodden, & Andrews, 2002; Welch, Banks, Cook, & Draovitch, 1995; Williams & Underwood, 1986), incorporating precise timing and coordination to successfully execute solid bat impact on a pitched ball. The purpose of this project was to determine kinematic influences of the lumbopelvic-hip complex (LPHC) on angular hand velocity; specifically, examining the relationship of internal and external isometric strength; separation of pelvis and torso rotation; pelvis and torso angular jerk; load timing; and maximum angular velocity timing of the pelvis, and torso to hand velocity throughout the swing. This chapter presents a brief introduction divided into three sections: 1) basic mechanics in hitting, 2) performance improvement in hitting, and 3) purpose, hypotheses, and glossary of terms.

Basic Mechanics in Hitting

In 1995, Welch et al. developed the first full-body quantitative description of the baseball swing (Welch et al., 1995). The swing was divided into three main events: foot off, foot contact, and ball contact (Figure 1). Foot off was defined as the point in which the stride foot broke contact with the ground, and foot contact was the next point in which the stride foot made contact with the ground. Following foot contact, ball contact was defined as the moment in which the bat first made contact with the ball.

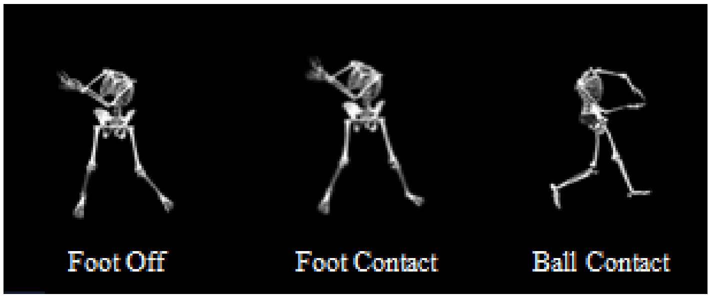

Figure 1. Hitting events defined by Welch et al. (1995).

While this analysis of the swing was crucial in establishing a foundation of hitting research, it failed to account for critical movements prior to foot off and following ball contact. Therefore, the swing has since been further divided into five main events: stance, load, foot contact, ball contact, and follow-through (Figure 2) (Dowling & Fleisig, 2016). Stance was defined as the last frame prior to the hitter moving his pelvis backwards towards the direction of the catcher. Load was the frame in which the hitter's pelvis reached maximum displacement towards the catcher. Foot contact was the frame in which the hitter's heel reached the minimum value of the vertical axes. Ball contact was defined using the reflective markers placed on the bat and on the stationary tee. Once the most minimum distance between the bat marker and tee marker was reached, ball contact was established. Follow-through was defined as the moment the hitter's lead elbow reached full extension. Two phases between load and ball contact were defined as the stride phase between load and foot contact, as well as, the acceleration phase between foot contact and ball contact (Figure 2).

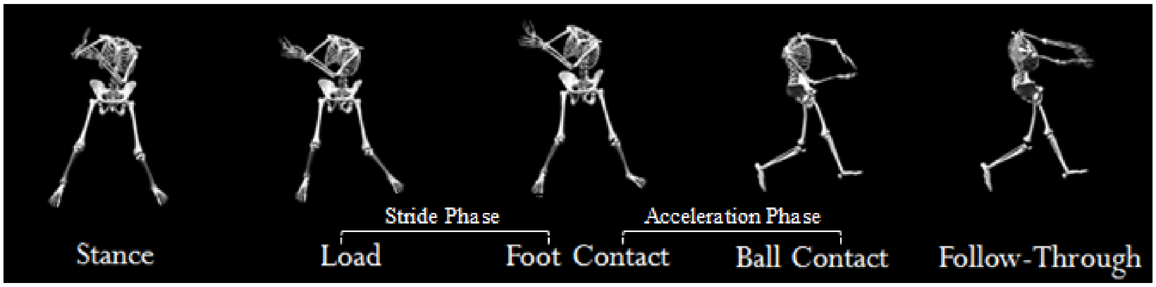

Figure 2. Hitting events defined by Dowling & Fleisig (2016).

In analysis of the swing phases, the first major movement immediately following foot off was stride leg hip abduction directly towards the pitcher, resulting in a linear motion under single-leg support, and initiating the stride phase of the swing. At the instant of foot contact, the linear component began to interact with a rotational component of the proximal segments (hips, pelvis and torso) to initiate the acceleration phase. Hip rotation velocity is reported as the most important of these components, because the swing is a sequential movement (Welch et al., 1995). Segment velocities and accelerations, initiated by the hips and transferred to the hands, follow a proximal to distal sequencing pattern (Elliott, 2000; Escamilla et al., 2009a; Welch et al., 1995). Specifically, maximum rotation velocity of the hips is succeeded by that of the pelvis, torso, shoulders, humerus, forearm, and hand. It has been postulated that if proper proximal to distal sequencing does not occur (i.e. a more distal segment reaches maximum velocity prior to the adjacent proximal segment or at the same moment), performance may decrease, because it can lead to inhibited force production by the trunk and upper extremity musculature (Welch et al., 1995). Applying the kinetic chain theory and the summation of speed principle in sequential motion, a hitter's maximum bat velocity should be a function of the initial rotational velocity established at the most proximal joint, the hips.

Performance Improvement in Hitting

Hand and bat velocity is considered a performance indicator for both baseball and softball athletes, because high swing velocity is a characteristic of a "good" hitter. Increased swing velocity allows a hitter to wait longer to swing at a pitched ball, thereby allowing the hitter to be more selective when choosing which pitches to hit (Adair, 1995; Breen, 1967). Additionally, increased hand and bat velocity prior to ball contact results in a greater batted ball velocity, due to the larger momentum transferred to the ball (Adair, 1995). The increased batted ball velocity then results in a "harder" hit or increased distance the ball travels, both of which are widely accepted as good swing outcomes in baseball and softball.

Existing data have established the potential positive effects of manipulating equipment through various training techniques such as over-weighted bats, under-weighted bats, and alterations in weight distribution within a bat; however, results are contradictory by gender, age, and skill (Dabbs et al., 2010; DeRenne, Buxton, Hetzler, & Ho, 1995; DeRenne & Okasaki, 1983; Escamilla et al., 2009a; Szymanski et al., 2011). Establishing internal techniques, i.e. a hitter altering mechanics within the body to improve swing mechanics via an increase in hand velocity, will benefit baseball and softball athletes alike. To apply this internal approach, investigation of LPHC influence could be crucial, as great importance is placed on LPHC stability within other sequential tasks such as overhead throwing, underhand throwing, the tennis serve, the golf swing, and the cricket bowl (Oliver & Keeley, 2010a, 2010b; Oliver & Plummer, 2011; Oliver, Plummer, & Keeley, 2011; Plummer & Oliver, 2014, 2016, 2017; Portus, Mason, Elliott, Pfitzner, & Done, 2004; Putnam, 1991, 1993). To the author's knowledge, there are no data examining the direct influence of the LPHC on hitting performance indicators, such as hand velocity. Therefore, establishing the role of the LPHC to increase hand velocity during the swing is warranted.

Purpose, Hypotheses and Glossary

Purpose

The purpose of this project was to determine kinematic influences of the lumbopelvic-hip complex (LPHC) on angular hand velocity; specifically, examining the relationship of internal and external isometric strength; separation of pelvis and torso rotation; pelvis and torso angular jerk; load timing; and maximum angular velocity timing of the pelvis, and torso to hand velocity throughout the swing.

Significance

Investigating the role of the LPHC in hitting will improve current sport biomechanics literature in that it will establish pertinent influences of the LPHC that may directly influence performance indicators in hitters, which is not present within the current baseball or softball hitting literature.

Research Questions

RQ1: Is bilateral hip rotational isometric strength correlated to angular velocity of the hands?

RQ2: Is maximum separation of pelvis and torso rotation, during the acceleration phase of the swing, correlated to angular velocity of the hands?

RQ3: Is minimal angular jerk of the pelvis and torso, during the acceleration phase of the swing, correlated to angular velocity of the hands?

RQ4: Is load time correlated to angular velocity of the hands?

RQ5: Is maximum angular velocity timing of the pelvis and torso correlated to angular velocity of the hands?

Hypotheses

H01: Hip rotational isometric strength was positively correlated with angular velocity of the hands at ball contact.

H02: Greater maximum separation of pelvis and torso rotation, during the acceleration phase, was positively correlated with increased angular velocity of the hands at ball contact.

H03: Minimal angular jerk at the pelvis and upper torso, during the acceleration phase, was positively correlated to increased angular velocity of the hands at ball contact.

H04: Greater load time was positively correlated to increased angular velocity of the hands at ball contact.

H05: Maximum angular velocity timing of the pelvis and torso reached later in the acceleration phase was positively correlated to increased angular velocity of the hands at ball contact.

Limitations

Limitations of this study are:

1. Variability in swing mechanics is typically high.
2. A large amount of noise is accumulated in angular jerk data.

Delimitations

Delimitations of this study are below:

1. Hip rotational isometric strength was measured using a hand-held dynamometer.
2. All data collections were executed in a controlled laboratory setting in the Auburn University Sports Medicine and Movement Laboratory.
3. All hitting trials were executed from a stationary hitting tee.

Definitions

Hitting Motion:

Upper Extremity:

Back Side – The back side of the upper extremity is the side furthest from the pitcher.

Lead Side – The lead side of the upper extremity is the side closest to the pitcher.

Lower Extremity:

Load Side – The load side of the lower extremity is the side furthest from the pitcher.

Stride Side – The stride side of the lower extremity is the side closest to the pitcher.

Hitting Strength – The sum of stride leg hip internal rotation isometric strength and load leg hip external rotation isometric strength.

Jerk – Rate of change of acceleration, or the third derivative of displacement, with respect to time.

Kinematics – The spatial and temporal components of motion, i.e. position, velocity, and acceleration (Nordin & Frankel, 2001).

Kinetic Chain – The kinetic chain is a series of linked segments of the body that move together (Blazevich, 2017).

Lumbopelvic-hip Complex (LPHC) – The lumbopelvic-hip complex (LPHC) encompasses the spine, torso, hips, pelvis, proximal lower limbs, and associated musculature of the abdomen and gluteals (W.B. Kibler, Press, & Sciascia, 2006).

Lumbopelvic Stability – The ability of the LPHC to limit excessive displacement of the pelvis and vertebral column, as well as, maintain structural integrity following perturbation, or disruption of the system, during a dynamic task (Pope & Panjabi, 1985; Willson, Dougherty, Ireland, & Davis, 2005).

Proximal to Distal Sequencing – Proximal to distal sequencing describes the temporal order of movements in joints and segments (Herring & Chapman, 1992). In hitting, this order begins at the hips and progresses to the pelvis, torso, humerus, forearm, and hand.

Summation of Speed Principle – The summation of speed principle states that any given segment will reach a maximum velocity greater than that of its most adjacent, proximal segment after the proximal segment has reached its maximum velocity, leading to the highest velocity imparted on the most distal segment of the chain (Bunn, 1972).

CHAPTER II

REVIEW OF LITERATURE

The purpose of this project was to determine kinematic influences of the lumbopelvic-hip complex (LPHC) on angular hand velocity; specifically, examining the relationship of internal and external isometric strength; separation of pelvis and torso rotation; pelvis and torso angular jerk; load timing; and maximum angular velocity timing of the pelvis, and torso to hand velocity throughout the swing. The project objective was to establish pertinent influences of the LPHC that may directly influence performance indicators in hitters, which is not present within the current baseball or softball hitting literature. The following chapter presents relevant literature pertaining to the appropriate facets of this project. It is divided into five sections examining the LPHC in sequential motion, jerk as a measure of movement smoothness, hitting mechanics, hitting performance indicators, and a brief summary of previous literature and the application to this project.

LPHC in Sequential Motion

The LPHC encompasses the spine, hips, pelvis, proximal lower limbs, and abdominal musculature (W.B. Kibler et al., 2006; Plummer & Oliver, 2014), in addition to the torso and gluteal musculature that provide the foundation of the pelvis (Plummer & Oliver, 2014). Muscles of the hips and pelvis are the base of support for the LPHC including large cross-sectional muscles that not only stabilize the lower extremity, but transfer energy through the LPHC to allow the body to generate a significant amount of force and power at the upper extremity in ballistic motion (W.B. Kibler et al., 2006). The latissimus dorsi, pectoralis major, hamstrings, quadriceps, and iliopsoas muscles are all considered prime movers of upper and lower extremity segments (W.B. Kibler, 1998). Major stabilizing muscles of the extremities, such as the upper and lower trapezius, hip rotators, and gluteal musculature also attach within LPHC region (W.B. Kibler, 1998). Each of these muscles are classified by an activation pattern of either length dependent or force dependent (Nichols, 1994). Length dependent muscles are short and small in diameter, typically spanning only one joint, thus providing a small lever for one segment. Force dependent muscles, however, span multiple joints producing large forces for efficient movement and stability of the central body. Length dependent muscles, such as the multifidi, act to stabilize each vertebrae in preparation for extremity movement (Bergmark, 1989; W.B. Kibler, 1998). The rectus abdominus and oblique abdominals are force dependent muscles that activate in direction-specific patterns, contracting prior to segmental movement (Hodges, 2003; Hodges, Butler, McKenzie, & Gandevia, 1997). Therefore, they provide stabilization of the spine before limb movements occur (Aruin & Latash, 1995; Hodges & Richardson, 1997; Zattara & Bouisset, 1988). Coordination of these activation patterns create a ‘neutral zone’ control of the spine, placing minimal tension on ligamentous structures and, ultimately, allowing the LPHC to be an efficient energy transmitter during athletic, kinetic chain tasks (Bergmark, 1989; Panjabi, 1992; Steffen, Nolte, & Pingel, 1994; Wilke, Wolf, Claes, Arand, & Wiesend, 1995; Young, Herring, Press, & Casazza, 1996).

It is well documented that the LPHC functions to provide proximal stability for distal mobility in athletic movement (Baechle, Earle, & Wathen, 2000; W.B. Kibler et al., 2006; Putnam, 1993; Zattara & Bouisset, 1988). The hips, pelvis, and spine provide stabilization to allow for the most distal segments' proper function (W.B. Kibler et al., 2006). Energy and force transfer through the body, during explosive movements, can be explained by the kinetic chain theory and the summation of speed principle (S. K. Chu, Jayabalan, Kibler, & Press, 2016; W.B. Kibler, Wilkes, & Sciascia, 2013; Sciascia, Thigpen, Namdari, & Baldwin, 2012). The kinetic chain has five main functions: (1) utilize muscle activation patterns to temporarily link multiple body segments into one; (2) provide a stable base for distal segment mobility; (3) maximize large force development in the LPHC and energy transfer distally to the hand; (4) produce interactive moments at distal joints greater than the energy and force the joint itself could produce; and (5) produce torques that limit negative acceleration forces linked to injury (W.B. Kibler, Kuhn, et al., 2013). Sequential movement of body segments provides the circumstances capable of placing the most distal segment in the best position, at peak velocity and with optimal timing, to ultimately produce a desired outcome in an athletic task (Putnam, 1993). For example, in the tennis serve, it has been found that hip and torso rotation velocity contribute 50-55% of the kinetic energy and force generated throughout the movement (W.B. Kibler, 1995). Similarly, angular momentum generated during the tennis serve is known to be initiated in the torso and transferred to the shoulder, forearm, hand, and racket (Bahamonde, 2000). However, sequential motion of the kinetic chain is characterized only in the occurrence of proper proximal to distal sequencing.

Proximal to distal sequencing is known to be the most efficient movement pattern in overhead throwing, softball and baseball pitching, kicking, striking, the golf swing, and in the tennis serve (Bahamonde, 2000; Dapena, 1978; W.B. Kibler et al., 2006; Maffet, Jobe, Pink, Brault, & Mathiyakom, 1997; Matsuo, Matsumoto, Mochizuki, Takada, & Saito, 2002; Oliver & Keeley, 2010b; Putnam, 1991, 1993; Willson et al., 2005). When proximal to distal sequencing is executed, potential and kinetic energy is generated in the most proximal segments of the body: the hips, pelvis, and torso of the LPHC. Energy is then transferred to the most distal segments and joints: the humerus, elbow, forearm, wrist, and hand (W.B. Kibler et al., 2006). Proximal to distal sequencing follows two scientific principles: (1) the summation of speed principle and (2) the law of conservation of momentum. The summation of speed principle was originally developed by Bunn in 1972 and states that any given segment will reach a maximum velocity greater than that of the adjacent, proximal segment after the proximal segment has reached its maximum velocity (Bunn, 1972). A subsequent slowing of the proximal segment also occurs as the distal segment's velocity increases. Progression of this sequence results in the maximum velocity imparted on the most distal segment and the most minimal velocity located at the proximal segment.

The summation of speed principle is made possible by the law of conservation of momentum, which states that the momentum of a system remains unchanged unless acted upon by an external force (Blazevich, 2017). Most athletic tasks include rotation; therefore, each segment in the kinetic chain has angular momentum (H). Angular momentum is the product of moment of inertia (I) and angular velocity (ω) (Equation 1) and is maintained unless acted upon by an external force.

$$\text{Equation 1. } H = I\omega$$

When the proximal segment in the kinetic chain is slowed, the angular momentum is transferred to the distal adjacent segment. Moment of inertia (I) is the product of a segment's mass (m) and its radius of gyration (k) squared (Equation 2).

$$\text{Equation 2. } I = mk^2$$

As an athlete gives a segment angular momentum, each segment is given an angular velocity. More distal segments are lighter in mass thereby decreasing the moment of inertia; however, to maintain angular momentum (H), angular velocity (ω) must increase. Therefore, rotation of proximal segments followed by a slowing of these segments transfers the momentum to the distal segments of less mass, which imparts the greatest velocity on the most distal segment. Additionally, distance from the axis of rotation for each segment decreases as momentum progresses through the distal segments, because the axis of rotation becomes the adjacent proximal segment. A decrease in the radius of gyration (k) dramatically decreases moment of inertia (I), which will substantially increase angular velocity (ω).

The influence of LPHC function in sequential motion is well documented in sport biomechanics literature, specifically in throwing, kicking, and striking tasks (Putnam, 1991, 1993). Studies suggest that the mechanics of proximal segments through a dynamic task, including kinematics and muscle activations, may influence movement of distal segments (Aguinaldo, Buttermore, & Chambers, 2007; W.B. Kibler et al., 2006; Oliver & Keeley, 2010a, 2010b; Plummer & Oliver, 2014; Sabick, Torry, Kim, & Hawkins, 2005). This can be extrapolated to imply that modifications in pelvis position or rotation can ultimately alter joint movement of the torso (Figure 3), and the pattern continues through the shoulder, arm, and hand. For example, it has been found that, in overhead throwing, a 20% decrease in energy transfer from the proximal segments necessitates an 80% increase in mass or a 34% increase in rotational velocity of the shoulder, in order to impart an equivalent resultant force on the hand and ball (W.B. Kibler, 1998). In 2014, upper extremity kinematics were compared among 72 high school baseball pitchers divided into two groups: 1) those who exhibited true proximal to distal sequencing within the LPHC and 2) those who did not. It was found that a true proximal to distal rotation sequence at the torso, i.e. the pelvis reached maximum axial rotation velocity prior to the torso's maximum axial rotation velocity, is correlated with greater maximal external rotation at the humerus as well as greater shoulder proximal force (Oyama et al., 2014).

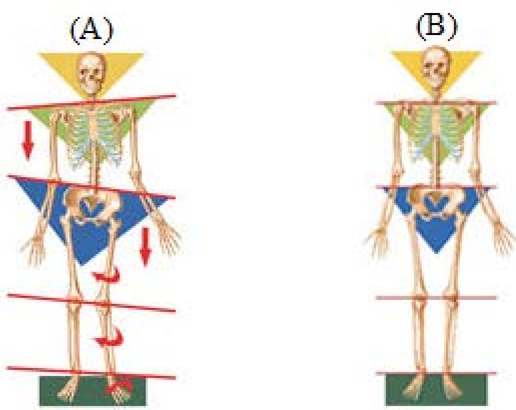

Figure 3. (A) Kinetic chain discrepancies versus (B) kinetic chain balance (ChiroMatrix, 2014).

Additionally, studies confirm that, not only does proximal to distal sequencing affect forces at distal segments; sequential timing may also have an influence on forces of distal segments. Baseball pitchers who exhibit longer time, between the instant of maximal pelvis rotation velocity and the instant of maximal torso rotation velocity, also display a lesser maximum internal rotation torque of the humerus (Aguinaldo et al., 2007). Studies also indicate that hyperangulation of the throwing shoulder, where the humerus moves posterior to the front plane, may result from rapid torso axial rotation too early during the stride phase of the pitch (Aguinaldo et al., 2007; Hedrick, 2000; Oliver & Keeley, 2010b; Sabick et al., 2005). These findings reiterate that the LPHC is extremely influential in sequential motion, as it provides proximal stability of the kinetic chain for distal mobility in athletic tasks.

Jerk

Similar to the wind-up in overhead throwing, hitting employs a loading phase that activates the stretch-shortening cycle in which an eccentric muscle contraction, or 'prestretch', precedes a concentric muscle contraction (Nordin & Frankel, 2001). Upon load initiation, a hitter will shift his or her weight to the back leg of the stance, slightly flexing the knees while shifting (Figure 4). Knee flexion of the stride leg engages an eccentric contraction of the quadriceps and hip extensor muscle groups to resist full knee flexion, building tension and storing elastic potential energy in the muscle groups. The stored potential energy will then assist in concentric muscle contraction of the quadriceps, which allows for a more explosive knee extension in moving towards the ball (Elliott, 2000; Komi & Bosco, 1978).

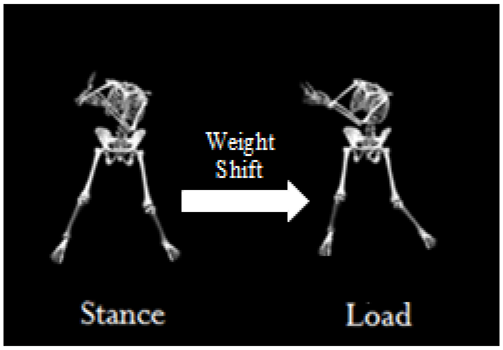

Figure 4. Weight shift toward load leg with slight stride leg knee flexion.

The stretch-shortening cycle is also imposed on the torso during initiation of the acceleration phase. As seen in the baseball pitch, athletes 'wind-up' by rotating the torso away from the pitcher, simultaneously maintaining pelvis positioning. The torso rotation stores elastic potential energy in the lumbopelvic musculature, thereby creating greater rotational velocity of the torso when the pitcher rotates forwards toward the hitter. Based on the summation of speed principle, greater rotation velocity at the torso will ultimately impart greater velocities on the humerus, forearm, and hand. Therefore, efficiency in storing potential energy within the musculature to create kinetic energy in the opposite direction is extremely important in an explosive, sequential task. Preparatory rotation in hitting occurs when the torso is rotated backwards towards the load leg, creating separation of the pelvis and torso (Welch et al., 1995). However, an additional increase in potential energy within the torso musculature occurs as a hitter begins to rotate the pelvis towards the stride leg, while maintaining torso position (Figure 5). As a result, the associated lumbopelvic musculature is stretched in an eccentric manner to produce a more forceful concentric muscle action during the acceleration phase, which will then result in greater velocities imparted on the distal segments.

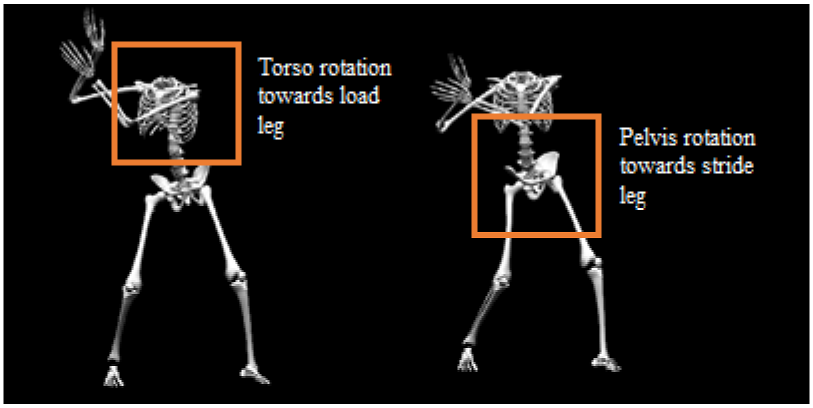

Figure 5. Preparatory torso rotation and pelvis rotation to increase stored potential energy in the lumbopelvic musculature.

Efficiency in engaging the stretch-shortening cycle can be measured using jerk, because jerk is a quantitative representation of movement smoothness (Choi, Joo, Oh, & Mun, 2014). Successful sport performance can be attributed to efficient, accurate, consistent, and smooth movements (Schmidt, 1975). Efficiency and consistency can be jeopardized by rapid changes in acceleration, which is jerk. A rapid change in acceleration could require greater energy expenditure, thereby reducing muscular efficiency. Additionally, force is the product of mass and acceleration, and rapid changes in acceleration will directly result in rapid changes of force produced at any given time. A decrease in consistency of force production throughout a ballistic motion may decrease the amount of muscular force produced from the movement overall.

Analysis of jerk has been executed for both lower and upper extremity movement (Choi et al., 2014; Hreljac, 2000; Yan, Hinrichs, Payne, & Thomas, 2000). In comparison of skilled middle school and high school long-distance runners to soccer and tennis athletes, it was found that skilled runners exhibited less jerk of heel movement in running and fast-walking tasks (Hreljac, 2000). In upper extremity movement, it has been found that jerk is reduced in overhead throwing as one ages, indicating that jerk is minimized as skill improves (Yan et al., 2000). In comparison of skilled versus unskilled golfers, skilled golfers exhibited less jerk in the clubhead during the downswing motion (Choi et al., 2014). While golf and hitting are both bimanual movements, analysis of jerk in hitting has yet to be executed. Therefore, influence of LPHC jerk on a performance indicator, such as hand velocity, is warranted in determining the most influential factors of the LPHC in hitting. These data could ultimately provide a quantitative measure by which athletes, coaches, and sport performance personnel can objectively classify skill level and efficiency in hitting.

Hitting Mechanics

In 1995, the first holistic approach to the biomechanics of the baseball swing was presented to develop understanding through quantitative data (Welch et al., 1995). From this comprehensive analysis, several key factors were derived for producing the most optimal swing. During the stride phase, hitters should emphasize the amount of backwards rotation toward the load leg, followed by the sequential, forward rotational movement towards the stride leg and pitcher. It has been found that baseball hitters rotate their pelvis towards the pitcher at a maximum of 22-28° (Lim, Park, & Kwon, 2016; Welch et al., 1995), while the shoulders continue to rotate directionally opposite the pelvis to a maximum of 52°, increasing tension in the abdominal musculature (Welch et al., 1995). Pelvis and torso separation has been found to increase up to 0.1 seconds prior to ball contact, then is absent at ball contact, indicating that a stretch-shortening cycle occurs in preparation for maximum efficiency at ball contact (Morishita, Yanai, & Hirano, 2010). While evidence is clear that pelvis and torso separation is beneficial for hitters in an attempt to store elastic potential energy, excessive rotation of the pelvis or torso has the potential to reduce muscular efficiency (DeRenne, 2007; Elliott, 2000; Welch et al., 1995). Data indicate that there is a point of diminishing return upon rotation toward the load leg or subsequent rotation towards the stride leg; however, the optimal range of motion to consistently produce the most efficient and highest rotational velocities has yet to be established.

Effects of pitch type and location have been examined to better understand changes in hitting mechanics during a game or competition setting (Katsumata, Himi, Ino, Ogawa, & Matsumoto, 2017; Tago, Ae, Tsuchioka, Ishii, & Wada, 2010; Tago, Kaneko, Tsuchiokat, Ishii, & Wada, 2016). In comparison of varying pitch velocities, it was found that no differences occurred in upper torso rotation between a fast and slow pitch during the initiation of the stride or at foot contact. However, hitters were significantly more rotated towards the pitcher at ball contact when hitting fastballs compared to slowballs (Tago et al., 2016). No significant differences were observed in hip rotation between the two pitch conditions, yet significant differences were observed in center of gravity displacement at ball contact. Hitters displayed greater horizontal displacement in the direction of the pitcher and significantly less displacement in the vertical direction when hitting slowballs compared to fastballs (Tago et al., 2016).

Pitch location was found to have a significant influence on hitting mechanics of both the lower and upper extremities (Katsumata et al., 2017; Tago et al., 2006; Tago et al., 2010; Tago, Ae, Tsuchioka, Ishii, & Wada, 2009). In comparison of high, middle, and low pitch locations, it was found that hitters exhibit greater flexion of both hips from foot contact until ball contact, when hitting a low pitch versus middle or high pitch (Tago et al., 2006). At the torso, axial rotation, flexion, and lateral tilt towards the load side increase when swinging at a low pitch; however, a hitter's center of gravity tends to be more displaced towards the pitcher when swinging at low pitches (Tago et al., 2006).

Table 1. Changes in upper extremity as a function of pitch height.

High	**Middle**	**Low**
(-)	Wrist Extension	(+)
(-)	Elbow Extension	(+)
(+)	Shoulder Rotation	(-)
(-)	Torso Lateral Tilt	(+)
(-)	Torso Rotation	(+)
(-)	Torso Flexion	(+)
(-)	Weight Shift	(+)

Shoulder horizontal adduction of the back side shoulder and bilateral shoulder rotation significantly increase when swinging at inside pitches versus middle or outside (Tago et al., 2009). However, shoulder flexion is least when swinging at a middle pitch compared to an inside or outside pitch (Tago et al., 2009). The torso is more rotated (i.e. more square to the pitcher) on an inside pitch, and a hitter's center of gravity is more displaced towards the pitcher when swinging at an outside pitch versus middle or inside (Table 2) (Tago et al., 2009).

Table 2. Changes in upper extremity as a function of pitch distance.

Outside	**Middle**	**Inside**
(-)	Shoulder Horizontal Adduction	(+)
(-)	Shoulder Rotation	(+)
(+)	Shoulder Flexion *(least in middle)*	(+)
(-)	Torso Rotation	(+)
(+)	Weight Shift	(-)

Just as differences in hitting mechanics have been attributed to variations in pitch locations and velocities (Katsumata et al., 2017; Tago et al., 2006; Tago et al., 2010;

Tago et al., 2009), differences have also been found among groups of age, competition, skill level, and hitting style (Chang et al., 2011; Dowling & Fleisig, 2016; Escamilla et al., 2009a; Inkster, Murphy, Bower, & Watsford, 2010). In comparison of youth and adult baseball hitters, adult hitters displayed a significantly greater amount of time under single-leg support during the stride phase. Adult hitters also exhibited significantly greater lead knee flexion during the stride and acceleration phase compared to youth. Furthermore, adult hitters displayed a more open pelvic position (i.e. the pelvis was square to the pitcher) when the stride foot relinquished contact with the ground, suggesting that adult hitters begin to rotate their pelvis during the stride phase, rather than onset of the acceleration phase. When the hands began to move forward toward the pitch, adult hitters had a more open torso position; however, they were significantly more closed at ball contact compared to the youth. Torso angular velocity was significantly greater in adult hitters, as expected; and peak velocity was executed later in the acceleration phase for adult hitters compared to youth. These data suggest that adult hitters begin to rotate their pelvis earlier in the swing, and with greater velocities, yet do not open their torso as much as youth by ball contact. In these data, adult hitters executed a significantly higher bat velocity compared to youth (Table 3). Contradictory data do exist, however, in comparison of competition levels of youth, high school, collegiate, and professional baseball athletes (Dowling & Fleisig, 2016). No differences in load time were observed, nor were any differences found for timing of peak angular velocities at the torso or pelvis. At ball contact, youth exhibited the greatest amount of back side elbow flexion and the least amount of back shoulder abduction. Youth also displayed significantly greater back knee flexion compared to high school athletes and significantly more pelvic rotation (i.e. square to the pitcher), at ball contact, compared to professional hitters. Professional hitters exhibited the greatest amount of torso extension, leaning away from the pitcher, at ball contact compared to all other competition level groups.

Table 3. Comparison of hitting kinematics as a function of age.

Adult		Youth
Load Phase		
(+)	Load Time	(-)
Stride Phase		
(+)	Lead Knee Flexion	(-)
(+)	Pelvis Rotation	(-)
Acceleration Phase		
(+)	Lead Knee Flexion	(-)
(+)	Torso Rotation	(-)
Ball Contact		
(-)	Torso Rotation	(+)
Full Swing		
(+)	Torso Angular Velocity	(-)
(+)	Bat Velocity	(-)

Similar findings to the age comparisons were observed in the lower extremity when comparing hitters of varying skill, classified as 'high-caliber' and 'low-caliber' athletes (Inkster et al., 2010). At ball contact, 'high-caliber' hitters displayed greater back side knee extension compared to the 'low-caliber' group. Additional findings were observed in the upper extremity, as the high-caliber group executed greater maximum angular velocity of lead elbow extension and greater bat velocity overall (Table 4) (Inkster et al., 2010).

Table 4. Comparison of hitting kinematics as a function of skill.

Skilled		Novice
Acceleration Phase		
(+)	Lead Elbow Extension Velocity	(-)
Ball Contact		
(+)	Load Knee Extension	(-)
Full Swing		
(+)	Bat Velocity	(-)

Variations of the swing have been established through the evolution of softball and baseball to create advantageous effects for hitters. While extreme changes in hitters' approaches do not exist among baseball hitters, two types of swings exist in softball: the traditional swing and the slap hit (Potter & Johnson, 2007). The slap hit consists of left-handed hitters executing a running swing with the ultimate goal of reaching first base faster than with the use of a typical swing. This swing is often employed by hitters whose aim is to simply reach base with their inherent speed rather than hitting the ball for distance. Kinematic comparisons of these two swings has been reported with differences occurring at the torso (Chang et al., 2011). During swing initiation, when the hitter typically loads the back leg, slap hitters exhibit less torso rotation and less pelvis and upper torso separation (Chang et al., 2011). Interestingly, no significant differences were observed in overall torso rotation at ball contact suggesting that the only differences in these two types of swings occur entirely prior to ball contact (Chang et al., 2011).

In baseball, hitters often make adjustments at the hands in an attempt to create an advantage in bat control. In comparison of a normal grip position versus a choke up grip position, in which the hands are moved closer to the barrel of the bat away from the knob, it was found that a choke-up grip resulted in significantly less time during the stride phase and acceleration phases (Escamilla et al., 2009b). Less time accrued during these phases could prove advantageous for a hitter, as he would have more time to prepare for the pitch being thrown. Furthermore, the choke-up grip resulted in significantly greater bilateral elbow flexion at foot contact, a more rotate pelvis position at bat-ball contact (i.e. square to the pitcher), and greater peak back elbow extension angular velocity. The normal grip position resulted in greater rotation overall of the upper torso and pelvis, and perhaps most significantly, greater bat velocity at bat-call contact as compared to the choke up grip position (Escamilla et al., 2009b). These results could be a function of the longer radius of the swing using a normal grip position, rather than an actual increase in velocity of the hands.

Changes in swing kinematics have been observed as a function of equipment, specifically alterations in inertial properties of the bat. In comparison of bats with different lengths and weights, no significant differences were observed in angular displacement (i.e. rotation) of the lower torso, upper torso, or pelvis between conditions (Takahashi et al., 2016). However, a trend was noted as lower and upper torso rotation increased with an increase in moment of inertia (Takahashi et al., 2016). Additionally, in an analysis of peak ground reaction forces of both feet, differences in weight distribution within the bat resulted in significant changes of both the stride and load foot peak force and timing of peak vertical force (Laughlin, Fleisig, Aune, & Diffendaffer, 2016). In observation of anteroposterior, mediolateral, and vertical ground reaction forces, only average peak stride and load foot vertical ground reaction forces and the timings of these peak forces were maintained between bat conditions. These data suggest that changes in bat properties, specifically the location of the center of mass of the implement, have a significant effect on swing mechanics, because temporal differences in peak ground reaction forces of both feet were observed.

Hitting Performance Indicators

In baseball and softball hitting, the most power is produced when the highest possible bat-head velocity is achieved (Miller, Strohmeyer, & Bemben, 2017). Achieving a high velocity bat-head will result in greater projection speed of the batted ball, ultimately yielding a further projection distance of the ball, as well (Sawicki, Hubbard, & Stronge, 2003). Four fundamental elements, based on physiological and mechanical factors, have been suggested for hitters to increase bat velocity (Elliott, 1992). First, hitters are encouraged to flow the backswing phase into the forward swing phase to utilize the most stored elastic potential energy available via the stretch shortening cycle. Second, integrating rotation of the lower limbs into the backswing, rather than a linear backswing, will increase the distance by which the bat travels, thereby increasing the distance over which velocity can be developed prior to ball contact. Third, coordination of body segments, such that angular velocities generate a true proximal to distal sequence from the lower to upper extremity, will integrate the summation of speed principle, culminating with the greatest velocity imparted on the bat. Fourth, increases in overall muscular strength will reduce total percentage of energy expended per swing, thereby increasing endurance and the ability to maintain proper mechanics throughout a competition (Elliott, 1992). These principles were established with a holistic approach; therefore, recent literature has examined each of these individually to determine which anatomical characteristics may have the most influence on bat velocity and, potentially, performance (Dabbs et al., 2010; Horiuchi & Sakurai, 2016; Koike & Mimura, 2016; Smith, Broker, & Nathan, 2003; Takagi, Fujji, Koike, & Ae, 2009; Tsuchikane et al., 2017).

Studies have investigated various training tools, muscular power, and body kinetics that could influence bat velocity in hitting (Dabbs et al., 2010; Horiuchi & Sakurai, 2016; Koike & Mimura, 2016; Miller et al., 2017; Smith et al., 2003; Takagi et al., 2009; Teichler, 2010). Specifically, integration of various warm-up techniques have been extensively examined for acute effects on bat velocity (Dabbs et al., 2010; Reyes & Dolny, 2009; Southard & Groomer, 2003; Szymanski et al., 2011). In comparison of weighted tools, baseball hitters were placed under three practice swing conditions: normal condition, weighted bat condition, and recalibration condition (Figure 6) (Nakamoto, Ishii, Ikudome, & Ohta, 2012). Subjectively, participants felt they swung faster with the standard bat after exposure to either of the weighted bat conditions. These findings were verified in the objective results, as there was a statistically significant increase in bat velocity after exposure to the weighted bat condition or the recalibration condition. Similar results have been seen in a weighted bat warm-up sequence, in which baseball hitters executed swings with a standard bat followed by a lighter weight bat, then by a heavy weighted bat (Reyes & Dolny, 2009). Swing velocity of the standard bat following this warm-up sequence increased, suggesting that alterations in implement weight prior to performance may acutely increase swing velocity (Nakamoto et al., 2012; Reyes & Dolny, 2009).

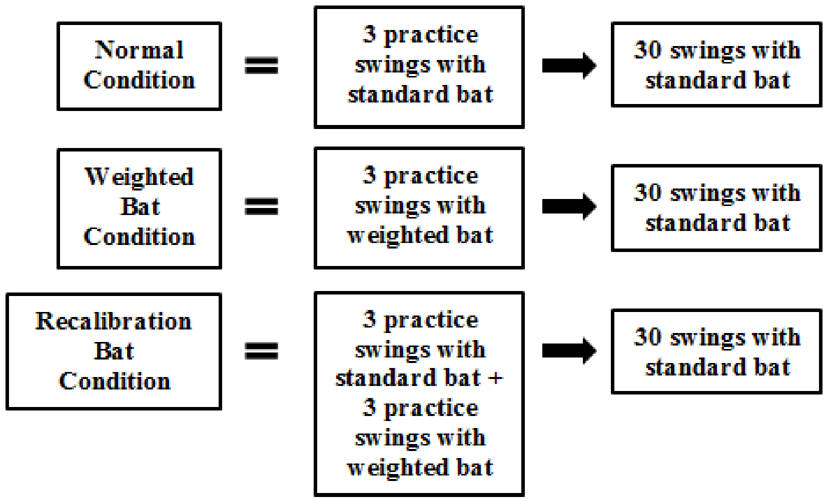

Figure 6. *Weighted bat warm-up techniques investigated by Nakamoto, et al.(Nakamoto et al., 2012)*

However, conflicting results have been observed in softball athletes who tested various warm-up techniques, because no significant increases were observed in bat velocity after implementing eight different warm-up devices (Szymanski et al., 2011). Significant decreases were observed using a traditional, weighted donut ring placed on the barrel of the bat (Szymanski et al., 2011), which indicates the need for softball athletes to refrain from using this particular device in warm-up routines at all (Szymanski et al., 2011). As each of the observed devices integrated resistance via heavier weight, lighter weight, or resistance bands, it was speculated that gender differences exist in effects of warm-up devices. Findings could be a result of too much load added to a bat for female athletes such that they needed to develop a new motor program, or it is possible that female athletes may have a slower feedback time for utilizing post-activation potentiation, the contractile history of skeletal muscle that may facilitate the volitional production of force (Hodgson, Docherty, & Robbins, 2005).

In addition to equipment characteristics, research has shown changes in bat velocity as a function of anatomical characteristics of the hitter (Tsuchikane et al., 2017). In investigation of the relationship between muscular thickness of the torso (upper abdominal rectus, central abdominal rectus, lower abdominal rectus, abdominal wall, and multifidus lumborum), upper limbs (elbow extensors, elbow flexors, and forearm muscles), and lower limbs (knee extensors, knee flexors, ankle dorsiflexors, and ankle plantar flexors) and bat swing velocity, muscle thickness was positively correlated to bat velocity. Specifically, thicker musculature of the abdominal wall and the multifidus lumborum on the back side of the body led to greater bat velocity, overall. All other torso, upper extremity, and lower extremity muscles examined were not significantly correlated with bat velocity, implying that the abdominal wall is critical for increased swing performance (Urquhart & Hodges, 2005). This is most likely due to its increased role in torso rotation during the load phase and the acceleration phase of the swing (Chang et al., 2011; Morishita et al., 2010; Szymanski, McIntyre, et al., 2007; Takagi et al., 2009; Welch et al., 1995), further emphasizing the significant, direct impact of the LPHC in hitting. Furthermore, it has been found that torso rotational strength interventions significantly increase both hand and bat velocity (Szymanski, McIntyre, et al., 2007). Increases in angular hip velocity and dominant torso rotational strength scores were significantly related to increases in bat velocity (Szymanski, McIntyre, et al., 2007), which further indicate the influence the torso and LPHC may have on hitting performance indicators.

Power assessments have been most commonly examined in relation to bat velocity, as it is known that bat swing velocity can be increased via implementation of power or sport-specific resistance training (DeRenne et al., 1995; DeRenne & Okasaki, 1983; Schwendel, 1992; Sergo & Boatwright, 1993). The relationship of percent body fat (%BF), total body mass, lean body mass, dominant and non-dominant grip strength, rotation power, upper body strength, lower body strength, explosive leg power, and peak power to bat velocity have been examined among collegiate baseball athletes (Szymanski et al., 2011). Bat velocity was shown to be significantly correlated to dominant hand grip strength, i.e. increased grip strength of the top hand resulted in increased bat velocity. Additionally, moderate correlations were observed for non-dominant hand grip strength, total body mass, peak power, upper body strength, lean body mass, and lower body strength, suggesting that overall power and body composition can influence hitting performance outcomes as well (Szymanski et al., 2011). Conflicting results have been observed, however, as vertical jump power was found to be moderately correlated with swing velocity in baseball players, yet not in softball players (Miller et al., 2017). Furthermore, it has been determined that upper extremity power is minimally correlated to bat velocity in softball hitters. However, no significant regression was observed in these softball hitters, which indicates that, while upper extremity power is slightly correlated to bat velocity, it may not be a significant predictor of bat velocity (Teichler, 2010). Therefore, these data suggest that specific performance parameters, such as grip strength and maximal power, may have an impact on bat velocity overall, but no cause-effect relationship can be definitively determined across all hitting athletes.

Summary

Previous literature states that the swing is a sequential movement (DeRenne, 2007; Elliott, 2000; Welch et al., 1995). In other sequential tasks, such as kicking, throwing, the tennis tennis serve, and the golf swing, the LPHC was found to be extremely influential in providing a stable base of support to create the best possible circumstances for distal segments to move efficiently (Elliott, 2000; Putnam, 1993). In hitting, a common performance indicator is hand and bat velocity (Adair, 1995; Breen, 1967). An increase in velocity often elicits characteristics of better performance, because it allows a hitter to have more time in preparing for a pitch, and it can assist in increasing the distance the batted ball travels. Previous literature investigated the kinematics of the LPHC in hitting, including range of motion, segmental angular velocities, and the timing of these variables (Escamilla et al., 2009a, 2009b; Inkster et al., 2010; Katsumata et al., 2017; Lim et al., 2016; Welch et al., 1995). However, no study to date has investigated the relationship of LPHC characteristics to hand velocity. Therefore, data from this study established pertinent influences of the LPHC that may directly influence performance indicators in hitting.

CHAPTER III

METHODS

Project objectives were to investigate the correlation of 1) bilateral hip isometric, internal and external rotation strength to angular hand velocity, 2) pelvis and torso separation to angular hand velocity, 3) minimal jerk of the pelvis and torso to angular hand velocity, 4) time of load from stance to the load event to angular hand velocity, and 5) time of maximum lead hip, back hip, pelvis, and torso angular velocity to angular hand velocity at ball contact during the softball swing. The role of this chapter is to outline and describe the methodology that was used for this study as follows: participants, setting, instrumentation, design and procedures, and data analysis.

Participants

Twenty-seven female, collegiate softball athletes volunteered to participate (20.41 ± 1.78 years; 167.47 ± 21.27 cm; 74.97 ± 15.28 kg). Sixteen athletes were right-handed hitters, and eleven were left-handed hitters. Selection criteria included being currently active on a playing roster and medically cleared by all sports medicine staff. Participants with a history of lower extremity, upper extremity, pelvis, low back, or torso injury within the past six months were excluded. All participants were recruited via an approved recruitment letter (Appendix A). Upon arrival to the laboratory, participants completed a health history questionnaire to determine eligibility for participation in the study

(Appendix B). The questionnaire was immediately evaluated by the investigator to eliminate any participants that might be at risk of injury. Prior to participation, all participants read and signed an informed consent document approved by the Auburn University Institutional Review Board (Appendix C). Parental assent was obtained for any participants under the age of 19 (Appendix D). The least number of participants that were chosen to participate was based on an *a-priori* power analysis. The power analysis determined that a minimum of 13 participants were necessary to achieve a power of 0.80 at $\alpha = 0.05$ (Faul, Erdfelder, Buchner, & Lang, 2009).

Setting

All data collections were conducted in a controlled laboratory setting in the Sports Medicine and Movement Laboratory within the School of Kinesiology at Auburn University. This location had the space and necessary equipment to successfully execute and fulfill the objectives of the study.

Instrumentation

Isometric Strength

Strength measurements were assessed using a hand-held dynamometer (Lafayette Instruments, Lafayette, IN) and recorded on a participant data form (Appendix E). The hand-held dynamometer was reported to have a sensitivity of 0.2 Newtons and is capable of measuring up to 500 Newtons of force (Couppe et al., 2014). In addition, overall methodology of a hand-held dynamometer has been deemed reliable and valid for measuring muscular strength (Byl, Richards, & Asturias, 1988; Donatelli et al., 2000;

Magnusson, Glelm, Kolbe, & Nicholas, 1992; Sullivan, Chesley, Hebert, McFaull, & Scullion, 1988; Trakis et al., 2008). Prior to any strength measures, the dynamometer was calibrated using the manufacturer's recommendations.

Kinematics

All kinematic data were collected with The MotionMonitor™ software (Innovative Sports Training, Chicago, IL) synchronized with an electromagnetic tracking system (Track Star, Ascension Technologies Inc., Burlington, VT). Prior to data collection, the system was calibrated using previously established techniques (Day, Murdoch, & Dumas, 2000; Keeley, Oliver, & Dougherty, 2012; Oliver & Keeley, 2010a, 2010b; Oliver & Plummer, 2011; Oliver et al., 2011; Perie, Tate, Cheng, & Dumas, 2002; Plummer & Oliver, 2014, 2016, 2017). Error in determining the position and orientation of the electromagnetic sensors within the world axes system is less than 0.01 meters and 3°, respectively. All kinematic data were sampled at a frequency of 240 Hz (Tago et al., 2016; Takagi et al., 2009; Takahashi et al., 2016). Raw data of sensor orientation and position were transformed to a locally-based coordinate system for each respective body segment and independently filtered using a 4^{th} order Butterworth filter with a cutoff frequency of 13.3 Hz (Fleisig, Hsu, Fortenbaugh, Cordover, & Press, 2013; Welch et al., 1995). However, raw jerk data has been reported as sensitive to smoothing methods due to the large amount of noise accumulated in the data (Hreljac, 2000). Therefore, potential error was minimized by visually inspecting the frequency spectrum of the raw jerk data and selecting a cut-off frequency of 14.5 Hz for pelvis angular jerk and 16 Hz for trunk angular jerk (Choi et al., 2014).

Design and Procedures

Athletic shorts and a loose-fitting t-shirt were worn by all participants to allow for unobstructed access to necessary anatomical landmarks for sensor placement. Isometric strength measures were collected immediately after informed consent was obtained. Measurements were performed for bilateral hip internal and external rotation. High intra-class correlation coefficients (ICC) were determined during pilot testing and reported for all isometric strength measurement tests (ICC = 0.836 - 0.986). Internal and external rotation of the hips was conducted in a seated position, legs hanging off the side of a table and knees in 90° of flexion (Figure 7) (Dorf, Chhabra, Golish, McGinty, & Pannunzio, 2007; Ellenbecker et al., 2007; Laudner, Moore, Sipes, & Meister, 2010; Robb et al., 2010; Sauers, Huxel Bliven, Johnson, Falsone, & Walters, 2014; Scher et al., 2010). A towel was placed under the distal femur to maintain the femur in a horizontal plane, and participants were allowed to place their hands on the edge of the table to assist in torso stability (Norkin & White, 2016). The dynamometer was placed approximately three inches proximal to the medial malleolus for femoral external rotation and three inches proximal to the lateral malleolus for femoral internal rotation (Corben et al., 2015; Mora-Custodio, Rodríguez-Rosell, Pareja-Blanco, Yañez-García, & González-Badillo, 2016; Pua, Wrigley, Cowan, & Bennell, 2008). On a verbal cue by the investigator, the participant internally or externally rotated while being resisted by the investigator for three seconds. All strength tests were normalized and reported as a percentage of the participant's body mass.

Figure 7. Isometric strength measurement protocol.

Following collection of strength measures, 14 electromagnetic sensors, approximately the size of a pencil eraser, were attached to the skin using double-sided tape and PowerFlex cohesive tape (Andover Healthcare, Inc., Salisbury, MA) to ensure the sensors remain in place throughout testing (Figure 8). The sensors were placed on the bilateral dorsal aspect of the foot [1-2], bilateral lateral aspect of the shank [3-4], bilateral lateral aspect of the femur [5-6], sacrum at S2 vertebrae [7], thorax at the junction of cervical vertebrae 7 (C7) and thoracic vertebrae 1 (T1) [8], bilateral lateral aspect of the humerus [9-10], bilateral lateral aspect of the forearm [11-12], and bilateral dorsal aspect of the hand [13-14] (Figure 9).

Figure 8. Electromagnetic sensor attachment.

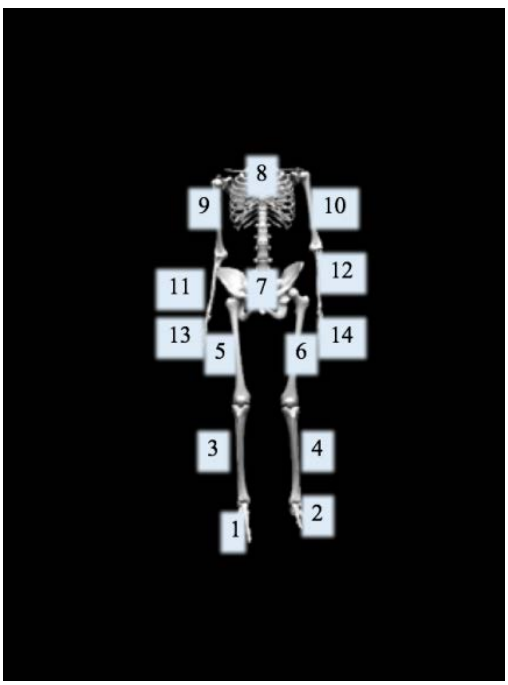

Figure 9. Electromagnetic sensor placement.

Once all sensors were securely attached to the body, a fifteenth sensor attached to a stylus was used to digitize various bony landmarks on the thorax (middle of torso), humerus (upper arm), radius and ulna (lower arm), pelvis, femur (thigh), tibia and fibula (lower leg), and foot (second toe) (Table 5) (Myers, Laudner, Pasquale, Bradley, & Lephart, 2005; Myers, Oyama, & Hibberd, 2013; Oliver & Keeley, 2010b; Oliver & Plummer, 2011; Plummer & Oliver, 2014, 2016). Medial and lateral aspects of each joint were identified and digitized, and joint centers were calculated by the midpoint of the two points digitized. A link segment model was developed through digitization of bony landmarks used to estimate the joint centers for the ankle, knee, shoulder, hips, thoracic vertebrae 12 (T12) to lumbar vertebrae 1 (L1), and cervical vertebrae 7 (C7) to thoracic vertebrae 1 (T1). The spinal column was defined as the digitized space between the associated spinous processes, whereas the ankle and knee was defined as the midpoints of the digitized medial and lateral malleoli in the frontal plane, and the medial and lateral femoral condyles in the frontal plane, respectively (Oliver & Keeley, 2010b; Oliver & Plummer, 2011; Plummer & Oliver, 2014; Wu et al., 2002).

Table 5. Description of the bony landmarks to be palpated and digitized to create the skeletal model of each participant in anatomical neutral.

Bony Landmark	**Digitized Bony Process**
Lower Extremity	
Foot	Second phalange metacarpal joint
Medial Ankle	Medial malleolus
Lateral Ankle	Lateral malleolus
Medial Knee	Distal aspect of medial femoral condyle
Lateral Knee	Distal aspect of lateral femoral condyle
Pelvis	Bilateral anterior superior iliac crest
	Bilateral posterior superior iliac crest
Torso	
Seventh Cervical Vertebra [C7]	C7 spinous process
Thoracic Vertebra 12 [T12]	T12 spinous process
Eighth Thoracic Vertebra [T8]	T8 spinous process
Suprasternal Notch	Most cranial aspect of sternum
Xiphoid Process	Most distal aspect of sternum
Upper Extremity	
Anterior Shoulder	Anterior aspect of humeral head
Posterior Shoulder	Posterior aspect of humeral head
Medial Elbow	Medial epicondyle
Lateral Elbow	Lateral epicondyle
Medial Wrist	Most distal aspect of ulna
Lateral Wrist	Most distal aspect of radius
Hand	Tip of third phalange and third metacarpal

Two points described the longitudinal axis of each segment and the third point defined the plane of the segment. A second axis was defined perpendicular to the plane and the third axis was defined as perpendicular to the first and second axes. Neutral stance was defined as the y-axis in the vertical direction, anterior/posterior of y in the direction of movement was the positive x-axis, and orthogonal to x-y-axis and to the right was the positive z-axis. Pelvis, torso, and upper extremity kinematics was defined by the standards and conventions of The International Shoulder Group and International Society of Biomechanics (Wu et al., 2002; Wu et al., 2005). Specifically, ZX'Y" was used for the hip, pelvis, torso, and hand (Table 6).

Table 6. Angle orientation decomposition sequences.(Wu et al., 2002; Wu et al., 2005)

Segment	Axis of Rotation	Angle
Hip		
Flexion/Extension	Z	Flexion [-]/Extension [+]
Abduction/Adduction	X’	Abduction [-]/Adduction [+]
Axial Rotation	Y”	Right Rotation [-]/Left Rotation [+]
Pelvis		
Anterior/Posterior Tilt	Z	Anterior [+]/Posterior [-]
Lateral Tilt	X’	Right Tilt [-]/Left Tilt [+]
Axial Rotation	Y”	Right Rotation [-]/Left Rotation [+]
Torso		
Flexion/Extension	Z	Flexion [+]/Extension [-]
Lateral Tilt	X’	Right Tilt [-]/Left Tilt [+]
Axial Rotation	Y”	Right Rotation [-]/Left Rotation [+]
Hand		
Flexion/Extension	Z	Flexion [+]/Extension [-]
Radial/Ulnar Deviation	X’	Radial Deviation [-]/Ulnar Deviation [+]
Pronation/Supination	Y”	Pronation [-]/Supination [+]

* Prime [‘] and double prime [“] notations represent previously rotated axes due to the rotation of the local coordinate system resulting in all axes within that system being rotated. [Rotation about X axis also results in rotation of both Y and Z axes resulting in a new system of X’, Y’, Z’. Subsequent rotations are then about those axes.]

After digitization was complete, participants were given verbal instruction to perform their own specified warm-up. The average number of warm-up swings was 4.48 ± 2.47. Warm-up was not standardized, because some hitters needed more time than others to feel sufficiently warm and capable of executing maximum effort swings without risk of injury. Each participant used their personal softball bat to reduce variability otherwise accrued by adaption to unfamiliar equipment (Table 7) (Szymanski, McIntyre, et al., 2007).

Table 7. Bat properties of all participants.

N	Length (L)	Mass (M)	Drop (L - M)	Brand
14	33	23	-10	Demarini
5	34	24	-10	Demarini
4	34	25	-9	Rawlings
2	33	23	-10	Rawlings
1	34	24	-10	Rawlings
1	33	23	-10	Axe

Participants were instructed to execute three maximum effort swings off a stationary tee at nine different ‘strike-zone’ locations (Figure 10) (Tago et al., 2010). Tee location was randomized for each participant using a random number generator, such that every participant did not execute the same order of swing locations during each collection (Tago et al., 2010). Sequence of tee location by trial was recorded on each participant’s data form (Appendix E). Saved trial criteria included 1) result of a line drive and 2) verbal approval by the participant as a “good” swing (Horiuchi & Sakurai, 2016; Inkster et al., 2010; Welch et al., 1995). Approval from the participant was desired because the softball swing varies from athlete to athlete, and because the temporal and “feel” components of a hitter’s swing is essential to successful performance (Williams & Underwood, 1986). Stationary tee height was placed approximately midway between the knee and hip for the middle strike zone locations 4-6, at the hips for high strike zone locations 1-3, and at the knees for low strike zone locations 7-9 (Dowling & Fleisig, 2016). In this study, the back leg when facing perpendicular to the pitcher or the stationary tee was defined as the load leg while the leg closest to the pitcher or tee when facing perpendicular was defined as the stride leg. The swing was divided into the events of stance, load, stride foot contact, ball contact, and follow through (Figure 11) (Dowling

& Fleisig, 2016). Additionally, data were analyzed during the acceleration phase of the swing, defined as the time between foot contact and ball contact (Figure 12) (Escamilla et al., 2009b).

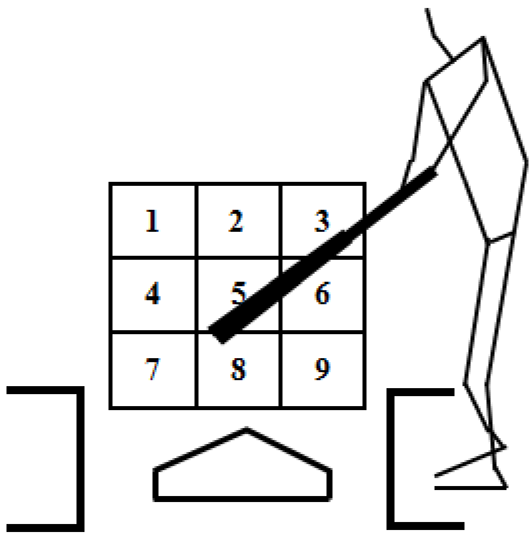

Figure 10. 'Strike zone' stationary tee locations.

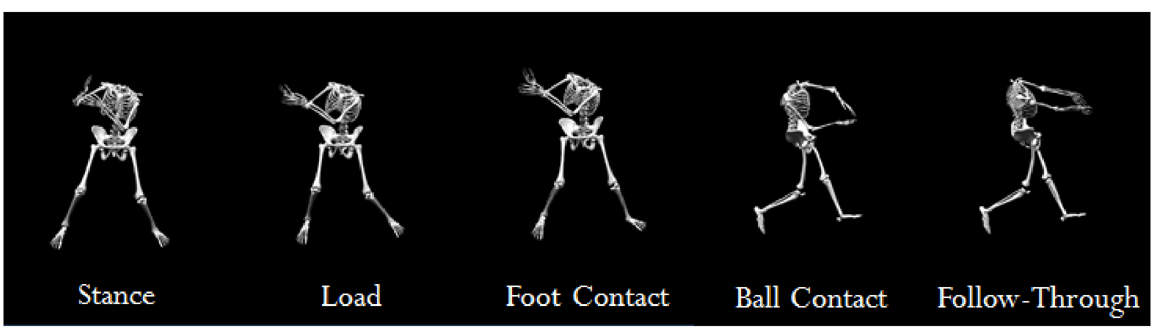

Figure 11. Hitting motion events.

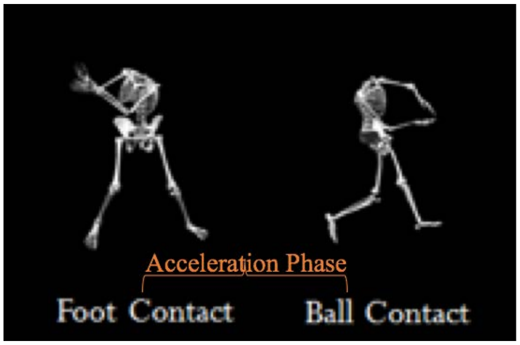

Figure 12. Hitting acceleration phase.

Data Analysis

Data were analyzed using IBM SPSS Statistics 23 (IBM corp., Armonk, NY). A total of 27 trials were collected: 3 trials at 9 stationary tee locations (Figure 10). For the purpose of this study alone, only the trials collected in the center strike zone tee location (location 5) were used for analysis. Prior to statistical analysis, all variables were checked for normality of distribution using Shapiro-Wilk Tests of Normality. Correlation and regression analyses were conducted for Research Questions 1, 3 and 5; however, only correlation analyses were conducted for Research Questions 2 and 4, as each question had one independent variable (Table 8). The alpha level was set *a priori* at $p \leq 0.05$.

Table 8. Correlation and Regression Analyses.

Research Question (RQ)		Independent Variable (IV)	Dependent Variable (DV)	Swing Event/Phase
RQ1	Is hitting strength correlated to angular velocity of the hand?	a. Stride hip internal rotation strength b. Load hip external rotation strength	Hand angular velocity	DV at ball contact
RQ2	Does pelvis and torso separation, during the acceleration phase of the swing, correlate to angular velocity of the hands?	Degrees of separation between pelvis and torso	Hand angular velocity	a. IV during acceleration phase b. DV at ball contact
RQ3	Does jerk of the pelvis and torso correlate to angular velocity of the hands?	a. Minimum pelvis jerk b. Minimum torso jerk	Hand angular velocity	a. IV during acceleration phase b. DV at ball contact
RQ4	Do load time correlate to angular velocity of the hands?	Load time	Hand angular velocity	a. IV during load phase b. DV at ball contact
RQ5	Does maximum angular velocity timing of the pelvis and torso correlate to angular velocity of the hands?	Maximum angular velocity timing of: a. Pelvis b. Torso	Hand angular velocity	a. IV during acceleration phase b. DV at ball contact

CHAPTER IV

RESULTS

The purpose of this project was to determine kinematic influences of the lumbopelvic-hip complex (LPHC) on angular hand velocity; specifically, examining the relationship of internal and external isometric strength; separation of pelvis and torso rotation; pelvis and torso angular jerk; load timing; and maximum angular velocity timing of the pelvis, and torso to hand velocity throughout the swing. This chapter describes and outlines the results from each research question and is sectioned accordingly:

RQ1: Is bilateral hip isometric strength correlated to angular velocity of the hand?

RQ2: Is maximum pelvis and torso separation, during the acceleration phase of the swing, correlated to angular velocity of the hand?

RQ3: Is minimal angular jerk of the pelvis and torso, during the acceleration phase of the swing, correlated to angular velocity of the hand?

RQ4: Is load time correlated to angular velocity of the hand?

RQ5: Is maximum angular velocity timing of the pelvis and torso correlated to angular velocity of the hand?

Research Question 1: Relationship of Hip Isometric Strength to Hand Velocity

Twenty-seven female softball athletes volunteered to participate; however, due to non-normal distribution of hand velocity (Appendix F), one outlier was removed from the dataset rendering a final participant pool of 26 participants (20.5 ± 1.7 years; 167.1 ± 21.2 cm; 74.8 ± 15.3 kg), as hand velocity was the dependent variable for all statistical tests. Correlation and multiple regression analyses were conducted to examine the relationship of stride hip internal rotation isometric strength and load hip external rotation isometric strength to hand velocity at ball contact. Descriptive statistics and statistical results are presented below (Tables 9-10). No significant relationship was found between stride hip internal rotation isometric strength and hand velocity at ball contact (r = -0.129, p = 0.264) (Figure 13), nor between load hip external rotation isometric strength and hand velocity at ball contact (r = -0.120, p = 0.279) (Figure 14). The multiple regression model with two predictors produced: R^2 = 0.017; $F(2, 25)$ = 0.196; p = 0.823.

Model: Hand Velocity = (-4.903*stride hip IR ISO) + (0.273*load hip ER ISO) +1343.928

Therefore, stride hip internal rotation isometric strength and load hip external rotation isometric strength were not significant predictors of hand velocity in the multiple regression model.

Table 9. Means and standard errors for Research Question 1.

	Mean ± Standard Error	**N**
Hand Velocity (deg/s)	1236.994 ± 47.968	26
Stride Hip IR[a] ISO[c](%BM)	19.933 ± 0.973	26
Load Hip ER[b] ISO[c](%BM)	22.922 ± 1.317	26

[a]IR – Femoral Internal Rotation
[b]ER – Femoral External Rotation
[c]ISO – Isometric Strength

Table 10. Multiple Regression model for Research Question 1.

	b	β	Sig.
Hand Velocity (deg/s)	1343.928		
Stride Hip IR[a] ISO[c] (%BM)	-4.903	-0.135	0.818
Load Hip ER[b] ISO[c] (%BM)	-0.273	0.006	0.992

[a]IR – Femoral Internal Rotation
[b]ER – Femoral External Rotation
[c]ISO – Isometric Strength

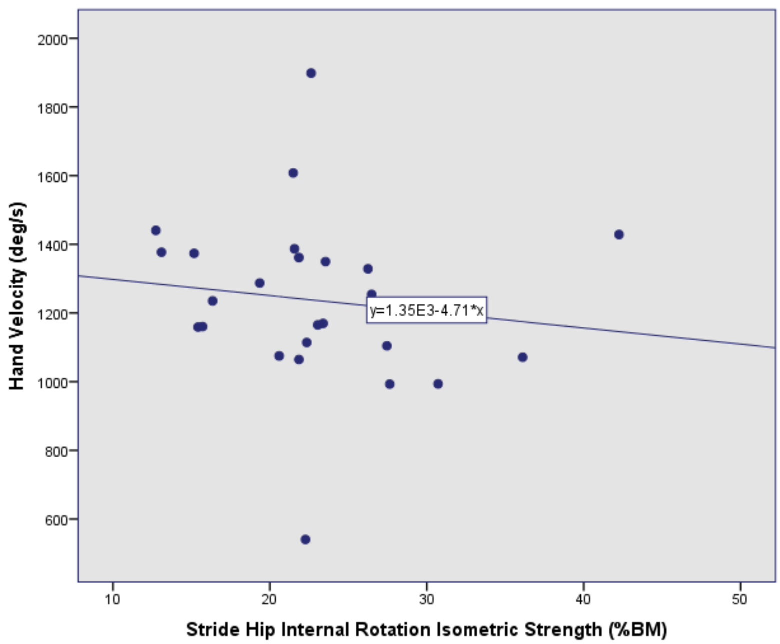

Figure 13. Correlation of average stride hip internal rotation isometric strength to hand velocity.

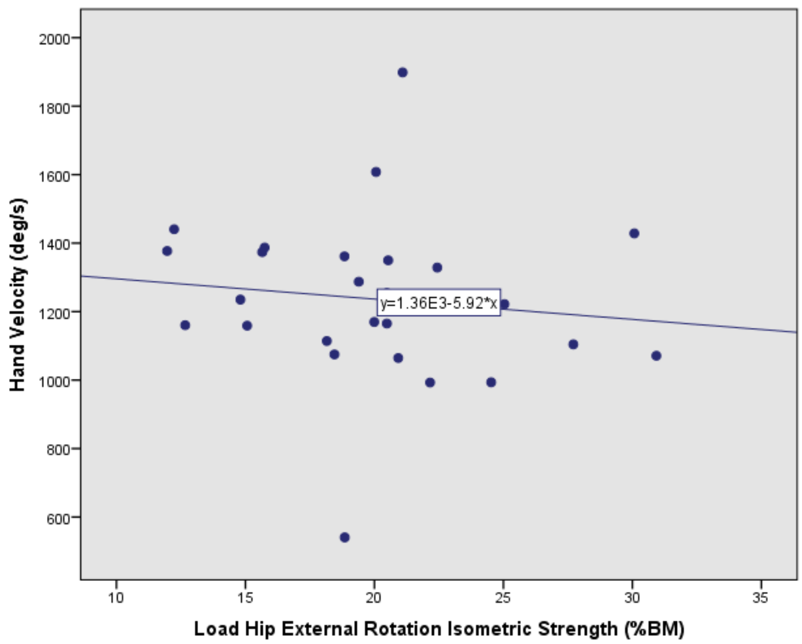

Figure 14. Correlation of average load hip external rotation isometric strength to hand velocity.

Research Question 2: Relationship of Separation to Hand Velocity

Correlation analyses were conducted to examine the relationship between maximum pelvis and torso separation, during the acceleration phase, and hand velocity at ball contact. Descriptive statistics and statistical results are presented below (Table 11). A significant, negative correlation was found between pelvis and torso separation and hand velocity at ball contact (r = -0.351, p = 0.039) (Figure 15).

Table 11. Means and standard errors for Research Question 2.

	Mean ± Standard Error	**N**
Hand Velocity (deg/s)	1236.994 ± 47.968	26
Separation (deg)	29.157 ± 3.422	26

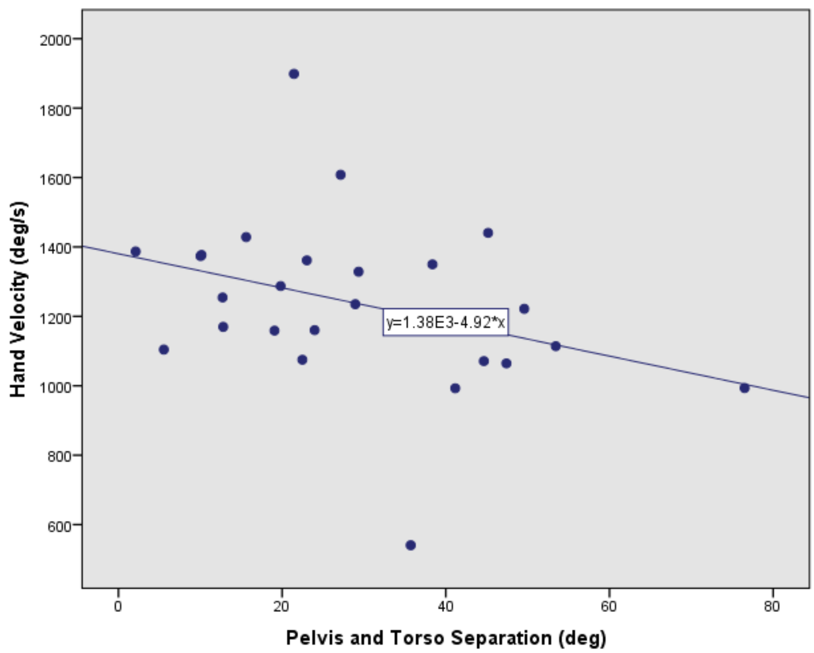

Figure 15. Correlation of average pelvis and torso separation during the acceleration phase to hand velocity.

Research Question 3: Relationship of Jerk to Hand Velocity

Due to non-normal distribution of pelvis and upper torso angular jerk (Appendix F), eight additional outliers were removed from the dataset, rendering a final participant pool of 18 participants (20.3 ± 1.6 years; 164.9 ± 24.9 cm; 74.4 ± 14.0 kg) for this research question. Curvilinear, quadratic regression analyses were conducted to examine the relationship of minimal pelvis angular jerk and minimal torso angular jerk, during the acceleration phase, to hand velocity at ball contact. Quadratic analyses were conducted, because the positive and negative values of jerk indicate a directional change, rather than an increase in the positive or negative x-direction. Therefore, minimal jerk is quantitatively closest to zero, regardless of direction. Descriptive statistics and statistical results are presented below (Tables 12-14). No significant relationship was found between pelvis angular jerk and hand velocity at ball contact (r = 0.192, p = 0.754) (Figure 16). The curvilinear regression model for pelvis angular jerk produced: R^2 = 0.037; F (2, 17) = 0.288; p = 0.754.

Model: Hand Velocity = (-0.004*pelvis jerk) + (-3.058x10^{-6}*pelvis jerk^2) + 1264.354

Additionally, no significant relationship was found between torso angular jerk and hand velocity at ball contact (r = 0.370, p = 0.331) (Figure 17). The curvilinear regression model for torso angular jerk produced: R^2 = 0.137; F (2, 17) = 1.190; p = 0.331.

Model: Hand Velocity = (0.014*torso jerk) + (9.915x10^{-7} *torso jerk^2) + 1243.558

Therefore, pelvis and torso angular jerk were not significant predictors of hand velocity in the regression models.

Table 12. Means and standard errors for Research Question 3.

	Mean ± Standard Error	**N**
Hand Velocity (deg/s)	1251.879 ± 35.564	18
Pelvis Jerk (deg/s^3)	-979.583 ± 507.427	18
Torso Jerk (deg/s^3)	-536.453 ± 956.837	18

Table 13. Curvilinear regression model of pelvis angular jerk.

	b	β	Sig.
Hand Velocity (deg/s)	1264.354		
Pelvis Jerk (deg/s^3)	-0.004	-0.056	0.851
Pelvis Jerk^2 (deg/s^3)^2	-3.058x10^{-6}	-0.214	0.476

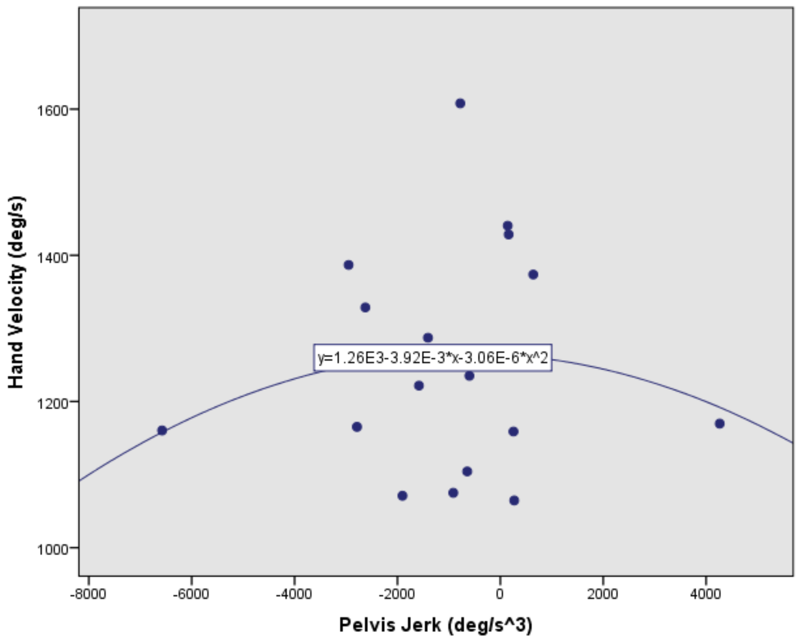

Figure 16. Correlation of average pelvis jerk during the acceleration phase to hand velocity.

Table 14. Curvilinear regression model of torso angular jerk.

	b	β	Sig.
Hand Velocity (deg/s)	1243.558		
Torso Jerk (deg/s^3)	0.014	0.371	0.157
Torso Jerk^2(deg/s^3)^2	9.915x10^{-7}	0.198	0.439

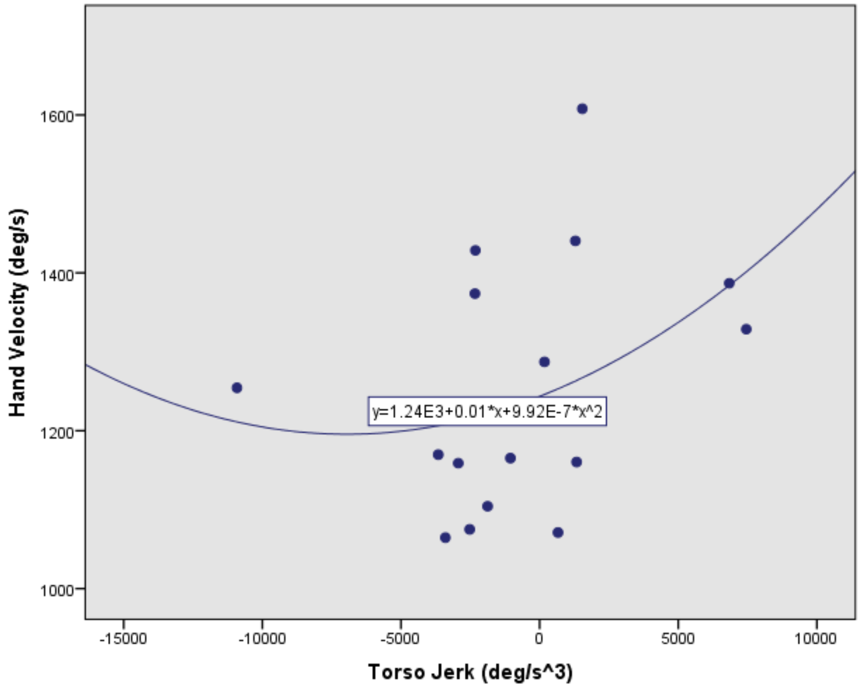

Figure 17. Correlation of average torso jerk during the acceleration phase to hand velocity.

Research Question 4: Relationship of Load Time to Hand Velocity

Correlation analyses were conducted to examine the relationship of load time to hand velocity at ball contact. Descriptive statistics and statistical results are presented below (Tables 15). No significant relationship was found between load time and hand velocity at ball contact (r = -0.208, p = 0.155) (Figure 18).

Table 15. Means and standard errors for Research Question 4.

	Mean ± Standard Error	**N**
Hand Velocity (deg/s)	1236.994 ± 47.968	26
Load Time (% swing)	44.017 ± 1.559	26

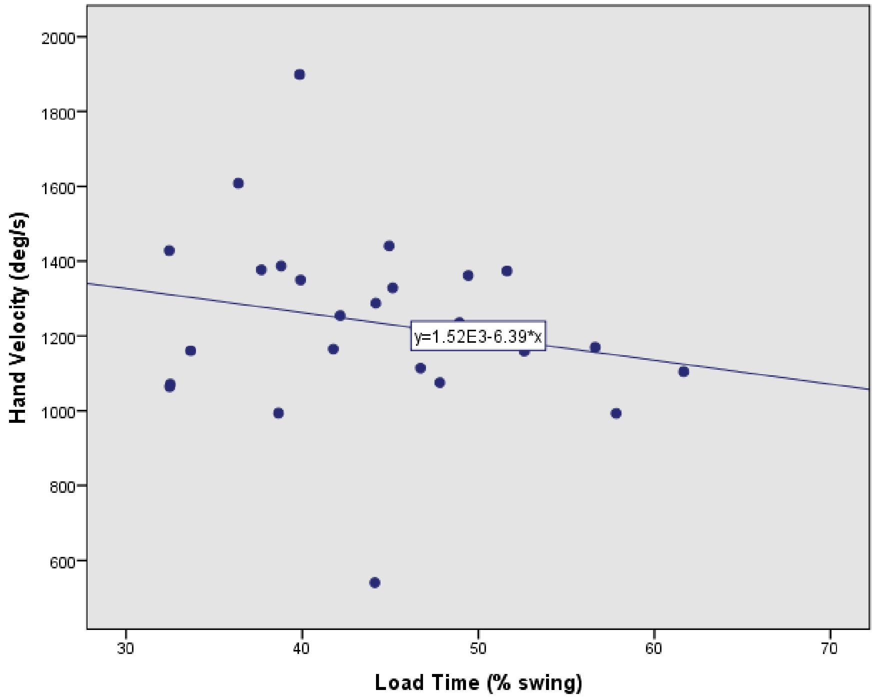

Figure 18. Correlation of average load time to hand velocity.

Research Question 5: Relationship of Rotation Timing to Hand Velocity

Correlation and multiple regression analyses were conducted to examine the relationship of maximum pelvis angular velocity timing and maximum torso angular velocity timing to hand velocity at ball contact. Descriptive statistics and statistical results are presented below (Tables 16-17). A significant, inverse relationship was found between timing of peak angular velocity of the pelvis, during the acceleration phase, and hand velocity at ball contact (r = -0.379, p = 0.028) (Figure 19). However, no significant relationship was found between timing of peak angular velocity of the torso, during the acceleration phase, and hand velocity at ball contact (r = -0.316, p = 0.058) (Figure 20). The multiple regression model with two predictors produced: R^2 = 0.161; F (2, 25) = 2.208; p = 0.133.

Model: Hand Velocity = (-5.044*pelvis rotation time) + (-3.155*torso rotation time) + 1790.231

Therefore, timing of peak angular velocity of the pelvis and torso, during the acceleration phase, were not significant predictors of hand velocity in the multiple regression model.

Table 16. Means and standard errors for Research Question 5.

	Mean ± Standard Error	**N**
Hand Velocity (deg/s)	1236.994 ± 47.968	26
Pelvis Rotation Time (% acc. phase [a])	58.399 ± 2.794	26
Torso Rotation Time (% acc. phase [a])	81.975 ± 2.397	26

[a]Acceleration phase of the swing.

Table 17. Multiple Regression model for Research Question 5.

	b	**β**	**Sig.**
Hand Velocity (deg/s)	1790.231		
Pelvis Rotation Time (% acc. phase[a])	-5.044	-0.294	0.208
Torso Rotation Time (% acc. phase[a])	-3.155	-0.158	0.494

[a]Acceleration phase of the swing.

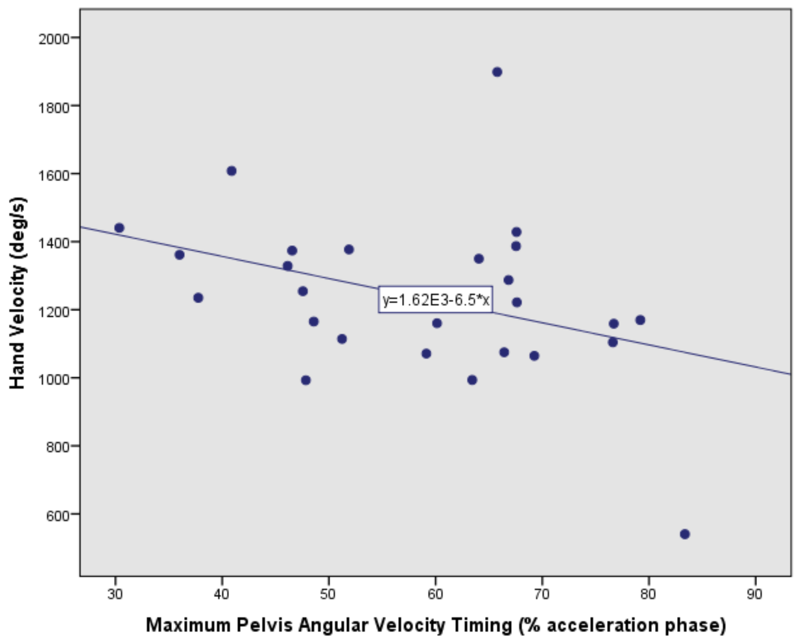

Figure 19. Correlation of average maximum pelvis rotation angular velocity timing to hand velocity.

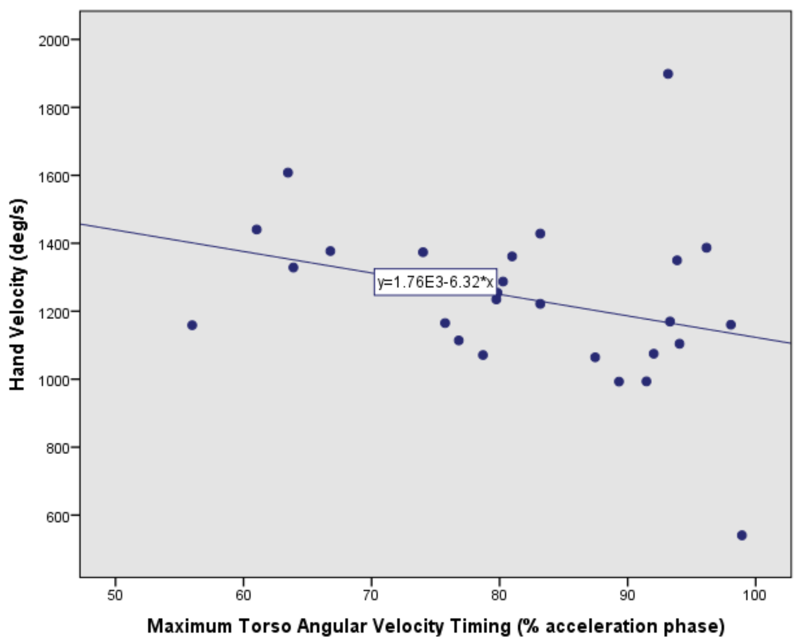

Figure 20. Correlation of average maximum torso rotation angular velocity timing to hand velocity.

CHAPTER V

DISCUSSION

The purpose of this project was to determine kinematic influences of the lumbopelvic-hip complex (LPHC) on angular hand velocity; specifically, examining the relationship of internal and external isometric strength; separation of pelvis and torso rotation; pelvis and torso angular jerk; load timing; and maximum angular velocity timing of the pelvis, and torso to hand velocity throughout the swing. This chapter discusses findings in accordance with each research question and addresses the applications of these findings to softball athletes.

Relationship of Hip Isometric Strength to Hand Velocity

The author hypothesized that external rotation strength of the load hip and internal rotation strength of the stride hip would be positively correlated with angular velocity of the hands at ball contact, however no significant relationship was found. A significant regression was also not found, indicating that load hip external rotation strength and stride hip internal rotation strength were not significant predictors of hand velocity. Though no statistical significance was observed, an inverse trend was exposed signifying that, as external rotation strength of the load hip and internal rotation strength of the stride hip increases, hand velocity at ball contact decreases.

In a similar study, Bailey et al. investigated multi-joint isometric force production using ground reaction force data during an isometric mid-thigh pull (Bailey, Sato, & Hornsby, 2013). Peak force, weight-normalized peak force, and instantaneous forces at 50, 90, and 250 milliseconds were measured and used in bivariate correlation analyses to investigate the relationship with batting average, doubles hit, home runs hit, and slugging percentage across four NCAA Division I baseball seasons. Strong, positive correlations were found between the rate of force development (calculated using the instantaneous forces) to doubles, home runs, and slugging percentage. Additionally, peak force and instantaneous forces at each time increment were strongly correlated to home runs and slugging percentage. The authors suggested that, based on their data, several strong relationships exist between isometric force production characteristics and baseball hitting performance. Most notably, rate of force development yielded the strongest relationships to hitting performance (Bailey et al., 2013).

Relative to bat and hand velocity, Szymanski et al. examined contribution of dynamic LPHC musculature strength to linear hand and bat velocity among high school baseball players (Szymanski, McIntyre, et al., 2007; Szymanski, Szymanski, Bradford, Schade, & Pascoe, 2007). A twelve-week intervention, in which a series of rotational medicine ball throws were added to a stepwise periodized strength training regimen, resulted in increased linear bat and hand velocity (Szymanski, McIntyre, et al., 2007). The training protocol significantly increased torso rotational strength, which was believed to be the central explanation for increases in bat and hand velocity (Szymanski, McIntyre, et al., 2007).

Results from the current study indicate a potential inverse relationship between isometric strength of the hips and hand velocity. This could be a consequence of differences in hip position during isometric testing of the hip during the swing. During isometric testing of the hip internal and external rotators, the hips and knees were in a position of 90º of flexion in open chain motion. Bailey et al. preformed isometric strength tests with the participant in a standing position of 175º hip flexion and 125º knee flexion (Bailey et al., 2013), which better represents the hitting stance position. Investigation of isometric strength of the internal and external hip rotators with the participant standing, or in a hitting stance, could significantly change the results found in this study.

Movement characteristics may have also influenced results, as isometric testing only examines one position of the limb, while dynamic movement requires the limb to go through a range of motion and is not positional dependent. During the isometric measurement, the lower extremity was in open chain, such that movement at the hip is isolated. During the swing, the body is in closed chain, causing the primary rotation to occur at the pelvis, which induces internal and external hip rotation. Therefore, musculature that internally or externally rotates the femur must assist in transverse pelvis rotation, as well. For example, the iliacus and psoas major and minor act to only externally rotate the femur during isometric testing; however, these muscles assist in transverse pelvis rotation as well as, externally rotate the load side femur during the swing. Similarly, the sartorius, gluteus medius, and gluteus minimus assist in hip abduction, as well as, internal and external rotation of the femur (Table 18). Therefore, the inverse relationship could simply be a result of differences in the musculature most employed during the isometric test versus the swing.

Table 18. Muscles most involved within the LPHC during the isometric strength test and swing.

Hip Motion - Isometric Strength Test	Hip Motion - Swing
External Rotation	***External Rotation***
Iliacus	Iliacus
Psoas Major and Minor	Psoas Major and Minor
Sartorius	Sartorius
Pectineus	Pectineus
Biceps Femoris	Biceps Femoris
Gluteus Maximus	Gluteus Maximus
	Adductor Magnus
Internal Rotation	***Internal Rotation***
Gracilis	Gracilis
Semitendinosus	Semitendinosus
Semimembranosus	Semimembranosus
Gluteus Minimus	Gluteus Minimus
Tensor Fasciae Latae	Tensor Fasciae Latae
	Hip Adduction
	Pectineus
	Adductor Brevis
	Adductor Longus
	Adductor Magnus
	Gracilis
	Hip Abduction
	Sartorius
	Gluteus Medius
	Gluteus Minimus
	Pelvis Rotation
	Iliacus
	Psoas Major and Minor

The lack of a significant relationship between hip isometric strength and hand velocity could also indicate that greater hand velocity is not a direct result of hip isometric strength, but rather a function of how a hitter utilizes hip leverage. The primary lever in the swing involves the torso, arms, and the bat with the fulcrum located at the load hip joint, the force located at a point on the pelvis where the pelvic stabilizers insert, and the resistance located at the center of gravity of the torso, arms, and bat lever similar to that of a tennis swing (Hamilton, Weimar, & Luttgens, 2008). Additional levers exist as a result of torso axial rotation, shoulder flexion, shoulder horizontal abduction and adduction, elbow extension, and wrist extension. Adequate strength of the stabilizing musculature surrounding the pelvis, torso, and shoulder to collectively maintain proper pelvis, torso, and shoulder position, may influence the maximum value of hand velocity at ball contact in the greatest manner.

Future research should examine hip rotational strength using different methodology to better analyze this question of hip strength influence. Use of Biodex or Cybex equipment, with a participant standing in the hitting stance position and executing transverse pelvis rotation in closed chain, would be more applicable and reliable in determining whether there is a relationship between hip rotation strength and hand or bat velocity (Szymanski, McIntyre, et al., 2007; Szymanski, Szymanski, et al., 2007). Because positive correlations have been established between hand velocity and torso rotation strength (Szymanski, McIntyre, et al., 2007), it is postulated that there is also a relationship between hand velocity and hip rotational strength as well, given different data collection methods.

Investigation into use of the hips could also be of value in identifying the best predictors of greater hand and bat velocity. Two theories of thought may be derived from these data: use of the stride hip as an anchor for transverse pelvis rotation, or use of both hips as a force couple from simultaneous clockwise and counter-clockwise rotation of the pelvis caused by the femurs. As the stride knee extends, transverse pelvis rotation occurs (Welch et al., 1995). A longer stride may cause a hitter to use the stride hip as an anchor, because the weight may be distributed more on the load side of the body. In this manner, the lead hip becomes an axis of rotation for the pelvis and trunk while the load side is pulled forward, similar to that of a golf swing (Y. Chu, Sell, & Lephart, 2010). However, some hitters may execute a shorter stride and a simultaneous clockwise and counter-clockwise rotation of the pelvis. Using this technique, the hips are used as means of initiating rotation and transferring force up the kinetic chain (Welch et al., 1995). Data identifying hand and bat velocity differences between these techniques could provide insight into whether the hip is more efficiently used as a force transmitter that initiates the rotation component of the swing, or as the axis of rotation to increase leverage of the load side.

Relationship of Pelvis and Torso Separation to Hand Velocity

It was hypothesized that participants who exhibited greater maximum separation of pelvis and torso rotation would have increased angular velocity of the hands at ball contact. Results showed a significant, negative correlation between degrees of pelvis and torso separation during the acceleration phase and angular hand velocity at ball contact, which does not support the original hypothesis. The inverse relationship signified that, during the acceleration phase, as degrees of maximum separation between the pelvis and torso decreased, hand velocity at ball contact increased.

Pelvis and torso separation has not been investigated relative to hand or bat velocity; however, studies have compared pelvis and torso separation between age, competition level, and alternative grip types in hitting (Dowling & Fleisig, 2016; Escamilla et al., 2009a, 2009b). In previous literature, pelvis and torso separation was calculated when the stride foot broke contact with the ground, foot contact, initiation of the hands moving towards the ball, and at ball contact (Escamilla et al., 2009a). In order to apply these data to the current study, which examined separation during the acceleration phase, the average degrees of separation across foot contact, hands moving towards the ball, and ball contact was calculated. Hitters in the current study displayed greater pelvis and torso separation at 29°, while previous literature found youth baseball hitters displayed approximately 15.3º of pelvis and torso separation and adult baseball hitters displayed approximately 14.3º. Using the same method in comparing normal grip versus choke-up grip, the normal grip yielded an average of approximately 15º of pelvis and torso separation, but only 12.3º while using a choke-up grip (Escamilla et al., 2009b). However, average pelvis and torso separation at ball contact was 22º for youth hitters and 23º for adult hitters, suggesting that maximum pelvis and torso separation occurred at ball contact, rather than prior to ball contact (Escamilla et al., 2009a). Dowling and Fleisig reported degrees of pelvis and upper torso rotation at the stance, load, foot contact, ball contact, and follow-through for the various competition levels of youth, high school, college, and professional (Dowling & Fleisig, 2016). Using the difference of upper torso rotation and pelvis rotation, and averaging these values across the events of foot contact and ball contact, the college group displayed approximately 7º of rotation during the acceleration phase of the swing. In each of the three studies, significant differences in bat or hand velocity were found between groups, yet no significant differences in pelvis and torso separation were observed, calling into question whether a relationship between hand or bat velocity and pelvis and torso separation exists.

However, the evidence of an inverse relationship in the current study may indicate that the value of pelvis and torso separation itself is not indicative of hand or bat velocity, but the timing at which maximum pelvis and torso separation occurs may be of interest.

The stretch-shortening cycle employs energy storage capabilities of the series elastic components and stimulation of the stretch reflex, which is the involuntary response of the body, to an external stimulus that stretches the muscle to facilitate a maximal increase in muscle recruitment over a minimal amount of time (Haff & Triplett, 2015). The stretch-shortening cycle consists of three phases: 1) eccentric muscle contraction, 2) amortization, and 3) concentric muscle contraction. The eccentric phase is a preload of the agonist muscle where the muscle stretches, increasing tension, and elastic energy is stored in the series elastic components (Komi, 1986). Amortization is the time between the eccentric and concentric phases, and potentially, is the most important component of the stretch-shortening cycle. If the amortization phase is long in duration, the energy stored is dissipated as heat, and the stretch reflex will not be able to increase muscle activity during the concentric phase (Nordin & Frankel, 2001). Thus, the amortization phase must be kept short in duration in attempt to increase force production. Using visual observation of data from the current study, it is evident that maximum separation occurs just prior to ball contact, rather than at ball contact (Figure 21). Preloading the LPHC musculature late in the phase could prove to be beneficial to those who exhibit less separation, yet a hindrance to those with greater separation, because those with less pelvis and torso separation would display a greater hand velocity since the distance over which the recoil into the concentric muscle action must occur is decreased (Baechle et al., 2000; Nordin & Frankel, 2001). Therefore, the elastic potential energy stored will be sufficiently expended and create a greater force production during the concentric phase, allowing for greater angular velocity of the segments. Displaying peak separation this late in the phase, with a greater amount of separation, could hinder angular velocity for the same reasons. Greater separation at this point will create an increased distance over which the musculature must recoil, which may result in much of the stored elastic energy being dissipated as heat (Nordin & Frankel, 2001). As such, this energy will not be rendered in the concentric phase, leading to a decrease in force production and a potential subsequent decrease in angular velocity of the segments (Szymanski, McIntyre, et al., 2007; Szymanski, Szymanski, et al., 2007).

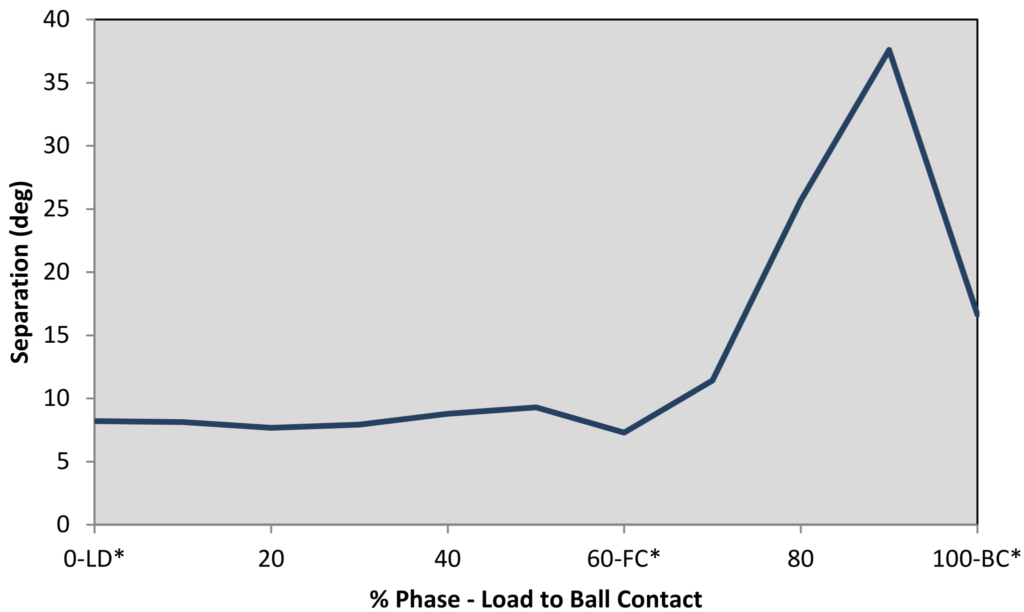

Figure 21. Timing of maximum separation.
**LD- Load*
**FC- Foot Contact*
**BC- Ball Contact*

Furthermore, it is postulated that the hitters who displayed less separation during the acceleration phase could be utilizing the amortization timing in a more beneficial manner, eliciting a greater return during the concentric phase from the stored potential energy. A desirable amortization time is 0.9 seconds or less (Hamill & Knutzen, 2015). Because the average swing time is approximately 0.2 seconds, it can be assumed that the hitters in this study executed a desirable amortization time of less than 0.9 seconds. However, use of the amortization timing is crucial and may have been different according to degrees of separation exhibited. The stretch prior to the concentric muscle action initiates stimulation through reflex potentiation, which accounts for approximately 30% of increase in concentric muscle action (Komi, 1984). The remaining increase is attributed to stored energy. A short-range or low-amplitude pre-stretch, occurring over a short time, is known to be the best technique to significantly improve the output of concentric action through return of elastic energy and increased activation of the muscle (Asmussen & Bonde-Petersen, 1974; Komi, 1984). Therefore, it has been suggested that an athlete should execute a pre-stretch, and then move immediately into the concentric phase. It is possible that hitters in the current study who executed greater maximum pelvis and torso separation rotated too far for the muscles to store the elastic energy properly, because the filaments were stretched beyond optimal length. It is also possible that these hitters paused after the pre-stretch, allowing the stored energy to be dissipated as heat, rather than expressed during the concentric muscle action.

Further research is necessary to determine the accuracy of either of these assumptions. Timing of maximum pelvis and torso separation and duration of the amortization phase in swinging athletes could be of most benefit to determine the overall influence of pelvis and torso separation on performance indicators in hitting. Results of this research could significantly influence training methods in hitters, once the true role of pelvis and torso separation is defined.

Relationship of Angular Jerk to Hand Velocity

In the current study, it was hypothesized that minimal angular jerk in axial rotation of the pelvis and torso, during the acceleration phase, would be positively correlated with increased angular velocity of the hands at ball contact. Specifically, it was anticipated that an angular jerk quantity closest to zero during the acceleration phase will result in greater hand velocity at ball contact. Results from this study did not reveal a significant relationship between pelvis or torso angular jerk with hand velocity. The trend of pelvis angular jerk was as expected, though, as seen in the concave parabolic curve (Figure 14). The pelvis angular jerk trend indicated that minimal pelvis angular jerk, quantitatively closest to zero, resulted in increased hand velocity. However, the trend of torso angular jerk was not expected, as it was anticipated that the torso trend would follow that of the pelvis.

Results from this study are somewhat similar to those found by Choi et al. in 2014, who analyzed normalized jerk of the lower extremity, upper extremity, and club head during the golf swing. Additionally, it has been found that smoothness of a movement is associated with greater skill and better performance (Schmidt, 1975). This claim is supported by Choi et al. in the examination of skilled and unskilled golfers in which they found that skilled golfers executed less normalized jerk of the club head during the swing. It was also found that skilled golfers had significantly lower normalized jerk of the lower extremity compare to unskilled golfers; however, jerk in the upper extremity was not significantly different. Additionally, a significant, positive correlation between normalized jerk of the lower extremity and normalized jerk of the club head was found, indicating that smoothness of movement in the lower extremity was directly related to smoothness of the club used to strike the ball (Choi et al., 2014). Choi et al.'s findings are similar to this study, because minimal jerk of the lower extremity was related to more efficient movement of the distal segments, i.e. increased hand velocity.

Due to the large amount of error present in the jerk measure, several ideas should be explored to determine why trends in this study occurred. Timing of minimal jerk through the acceleration phase could be a large predictor of hand velocity, rather than the value of jerk itself. Visual analysis of timing of minimum jerk data indicates that the timing of minimum and maximum pelvis and torso angular jerk throughout the phase are very similar (Figures 20-21). The least amount of jerk was displayed at the start of the acceleration phase, just after foot contact. However, there was a significant increase in jerk approaching ball contact, implying that there was a large amount of change in acceleration in the transverse plane as the hitter prepared to make contact with the ball. This trend could be a direct effect of speed-accuracy tradeoff (Fitts, 1954). Increase in jerk innately represents sudden changes in movement, which can be a representation of negative acceleration in order to be more accurate with the head of the bat. This occurs at both the pelvis and torso, and evidence of this can be seen in the value of torso angular jerk initially analyzed as a function of hand velocity (Figures 22-23). Subsequently, a decrease in hand velocity and jerk suggests that the hitters in this study made sudden changes in acceleration when approaching ball contact, potentially in order to make solid contact off of the tee, rather than strike the ball with the highest bat velocity. However, this was not seen at the pelvis, which may indicate that these hitters were more focused on speed of the swing while rotating the pelvis but became increasingly focused on accuracy of hitting the ball off the tee in approaching ball contact.

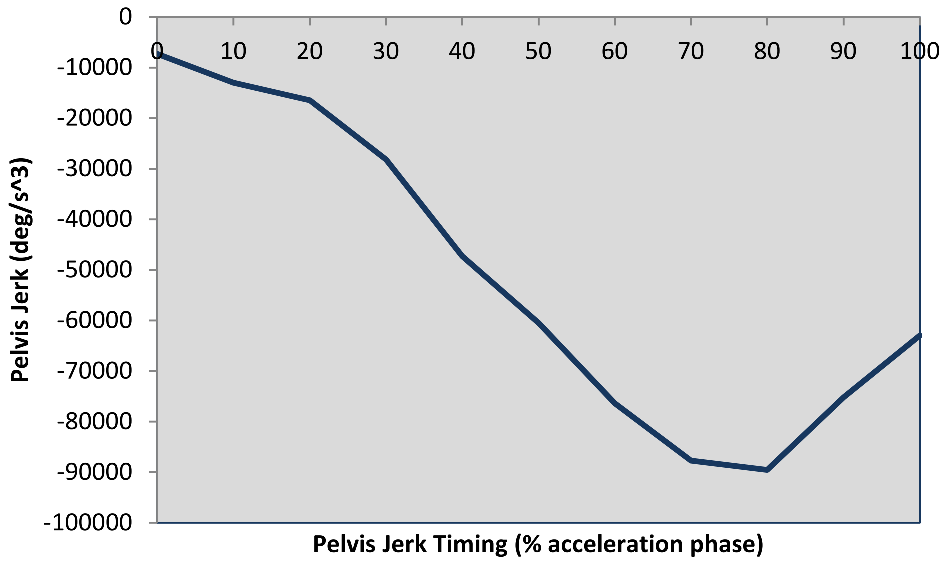

Figure 22. Pelvis angular jerk timing during the acceleration phase.

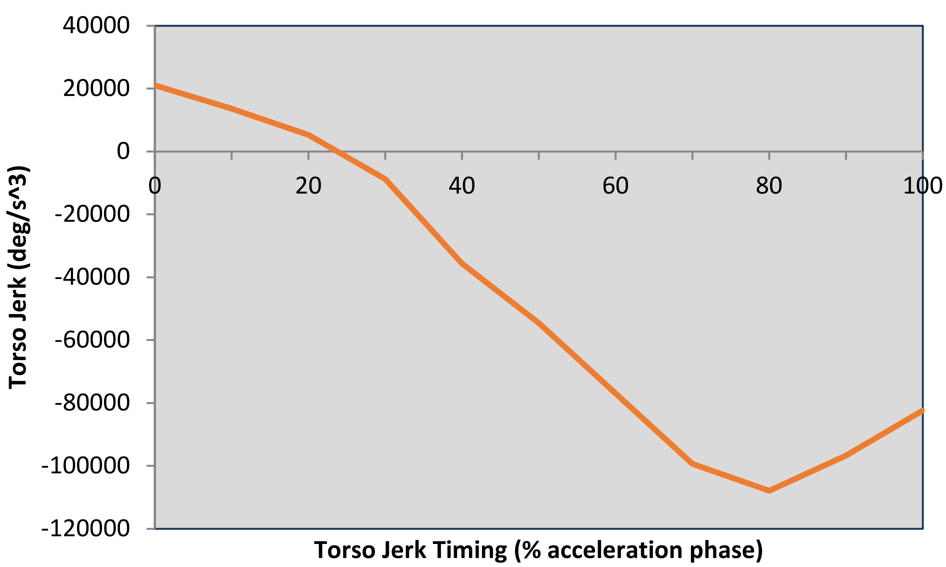

Figure 23. Torso angular jerk timing during the acceleration phase.

Future research should investigate angular jerk in hitters who are preparing to hit a ball thrown to them from a pitcher, rather than off a stationary tee. While the stationary tee is mostly viewed as a method for establishing fundamental hitting mechanics, hitting from a pitcher is most similar to a competition. Hitting a thrown ball requires the hitter to not only focus on their swing mechanics, but also on the location and velocity of the ball thrown (Winkin, Leggett, Johnson, & McMahon, 2001). The change in condition could significantly alter the trend of torso angular jerk, as the hitters will make a greater assumption of pitch location and bat placement through the hitting zone. With only enough time to assume pitch location, influence of the speed-accuracy tradeoff may be reduced and may create a torso angular jerk trend more similar to that of pelvis angular jerk through the acceleration phase of the swing.

Relationship of Load Time to Hand Velocity

In the current study, it was hypothesized that load time would be positively correlated with angular velocity of the hands at ball contact. No significant relationship was observed, however, an inverse trend between load time and hand velocity was exposed, suggesting that a longer load time is related to decreased hand velocity.

Previous literature has stated that hitters with faster hands or bat velocity exhibit a longer load time, because the increased velocity allows for more time to decide whether he or she will swing at the pitched ball (Adair, 1995; Breen, 1967). Dowling and Fleisig analyzed timing of the load event across competition levels of youth, high school, college, and professional baseball hitters (Dowling & Fleisig, 2016). Although there were no significant differences found of load time between the competition level groups, there was a progressive increase in the amount of time spent during the load from youth to professional hitters. It was concluded that older and more skilled hitters spent more time preparing for the upcoming swing compared to younger, less skilled hitters (Dowling & Fleisig, 2016). Furthermore, hand velocity progressively increased across competition level, indicating a potential positive relationship between load time and hand velocity. However, a ceiling effect may be present in these data as load time is very short in duration, eliciting small variability and minimizing the potential for a statistically significant relationship.

The inverse trend between load time and hand velocity found in this study could simply be based on the definition of load time utilized in the current study. Dowling and Fleisig's study, as well as studies from Escamilla et al. and Tago et al., defined load timing by milliseconds (Dowling & Fleisig, 2016; Escamilla et al., 2009b; Tago et al., 2006). However, this study defined load as a percentage of the swing in order to normalize timing across all participants, as correlation and regression analyses can be considered a between subjects statistical test. Future research should investigate load timing in milliseconds to more accurately compare results to previous hitting literature. Findings from this study could also suggest that a hitter's load time is an indicator of hand velocity, rather than hand velocity an indicator of load time. It is possible that hitters with greater hand velocity have the ability to wait longer, thus allowing the execution of a longer load time. To support this notion, future research should investigate the correlation and regression using load time as the dependent variable to determine whether hand velocity is an accurate predictor of load time in hitting athletes.

Additionally, use of the stationary tee condition could have influenced results of load time. Previous literature within the sport of cricket found that cricket batters had more consistent load timing when facing a live cricket bowler (Gibson & Adams, 1989).

Many hitters in baseball and cricket use the load as a timing mechanism to ensure the absence of swinging too late or too early at a given pitch (Dowling & Fleisig, 2016; Escamilla et al., 2009a; Gibson & Adams, 1989; Williams & Underwood, 1986). The tee condition utilized in the current study could have elicited a disregard of timing, because the ball is stationary, which may reduce focus on the load and increase focus on accuracy of making solid contact off the tee. Therefore, the consistency of load timing may be decreased and could influence results of the potential relationship between load timing and hand velocity during the swing. Future studies should investigate the relationship of load timing to hand velocity under a live pitcher condition in order to further distinguish the relationship of load timing to hand velocity and to identify whether load timing is a result of a hitter's swing velocity abilities, rather than hand velocity a resultant factor of load timing.

Relationship of Maximum Angular Velocity Timing to Hand Velocity

It was hypothesized that timing of maximum angular velocity of the pelvis and torso, during the acceleration phase, would be positively correlated with increased angular velocity of the hands at ball contact. Specifically, occurrence of maximum velocity of both segments later in the acceleration phase would result in increased hand velocity at ball contact. Results revealed a statistically significant, negative correlation between the timing of maximum pelvis angular velocity and hand angular velocity at ball contact, indicating that reaching maximum pelvis angular velocity earlier in the acceleration phase of the swing resulted in greater hand velocity at ball contact. A negative correlation was also identified between the timing of maximum torso and angular velocity; however, this relationship was not statistically significant. Multiple regression analyses were also not significant, indicating that timing of maximum pelvis and torso angular velocities, during the acceleration phase of the swing, are not significant predictors of hand velocity at ball contact.

Timing of the swing has been noted as the most important component for improvements in hitting performance (Dowling & Fleisig, 2016). Because the swing is a sequential motion, not only is execution of a true proximal to distal sequence important, but the timing of maximum angular velocities of each segment are paramount in effectively utilizing the momentum accumulated (Dowling & Fleisig, 2016). If maximum segment rotation velocity commences too early or too late, the potential maximum velocity of the implement on the ball will not be fulfilled. All participants in this study exhibited a true proximal to distal sequencing pattern of the LPHC, with maximum pelvis angular velocity occurring approximately 20% earlier than maximum torso angular velocity (Figure 24).

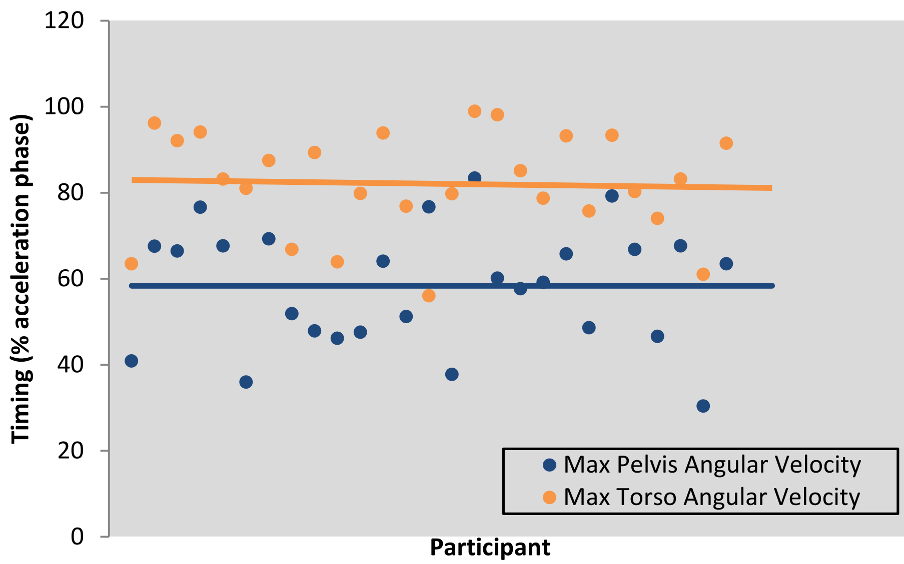

Figure 24. Maximum angular velocity timing of the pelvis and torso during the acceleration phase.

Escamilla et al. analyzed maximum pelvis and upper torso angular velocity timing as a percentage of the swing from initiation of the stride to ball contact, and found that maximum pelvis angular velocity occurred at 82% of the swing and upper torso angular velocity at 88% of the swing (Escamilla et al., 2009a). Dowling and Fleisig also investigated timing of peak angular velocity of the pelvis and upper torso and found that peak pelvis velocity occurred at approximately 70% of the acceleration phase of the swing; however, peak upper trunk velocity was reported after ball contact because analysis consisted of the entire swing, which included follow-through (Dowling & Fleisig, 2016). However, neither study investigated the relationship of maximum angular velocity timing to hand or bat velocity, thus it is difficult to make inferences of the current study's findings and previous literature.

The negative relationship between maximum pelvis angular velocity and hand velocity at ball contact indicated that reaching maximum pelvis angular velocity approximately 60% of the acceleration phase, or earlier, may elicit a greater hand velocity at ball contact. However, it should be noted that reaching this maximum velocity too early may be detrimental to the hitters, just as reaching this velocity too late (Welch et al., 1995). Future research should examine these data as a function of the full swing in order to more accurately compare data from previous studies, as well as compare findings from softball to baseball athletes. Additional studies are needed to identify an optimal threshold for both pelvis and torso maximum velocities to better understand the relationship of proximal to distal sequencing timing, rather than the progression of segment rotations alone.

Summary

Previous literature states that the swing is a sequential movement (DeRenne, 2007; Elliott, 2000; Welch et al., 1995). In other sequential tasks, such as kicking, throwing, the tennis serve, and the golf swing, the LPHC was found to be extremely influential in providing a stable base of support to create the best possible circumstances for distal segments to move efficiently (Elliott, 2000; Putnam, 1993). In hitting, a common performance indicator is hand and bat velocity (Adair, 1995; Breen, 1967). An increase in velocity often elicits characteristics of better performance, because it allows a hitter to have more time in preparing for a pitch, and it can assist in increasing the distance the batted ball travels. Previous literature investigated the kinematics of the LPHC in hitting, including range of motion, segmental angular velocities, and the timing of these variables (Escamilla et al., 2009a, 2009b; Inkster et al., 2010; Katsumata et al., 2017; Lim et al., 2016; Welch et al., 1995). However, no study to date has investigated the relationship of LPHC characteristics to hand velocity.

It was found that timing of pelvis and torso separation, rather than the value of separation itself, may be of most importance in predicting hand velocity in hitting. Furthermore, the timing of segmental sequencing is also crucial in efficiently transferring energy through the kinetic chain to allow for increased hand velocity. While few statistically significant findings were established, these data may point to the notion that the variables investigated in this study may not be the best indicators of improvements in performance. Therefore, future research should investigate the timing of appropriate variables, as well as, other LPHC characteristics to more definitively determine particular predictors within the LPHC that allow for increased hand velocity and potential performance improvements in hitting.

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Appendix A

Participant Recruitment Letter

Hello All,

The Sports Medicine & Movement Lab is currently conducting a research study to examine the influence of the lumbopelvic-hip complex on softball hitting mechanics. You must be a hitting athlete between the ages of 17 – 23 and you must be injury, surgery, and pain free for the last 6 months. You must also not have an allergy to adhesive tape.

We will measure hitting mechanics, muscle activation, and hip and shoulder range of motion. Range of motion and strength of the shoulder and hip will be first measured and recorded. Next, 14 sensors will be affixed to the skin using non-adhesive tape. Eight EMG sensors will be placed on the following muscles: bilateral external oblique [lateral abdomen], bilateral latissimus dorsi [mid lateral back], bilateral gluteus medius [lateral hip], and bilateral gluteus maximus [mid buttocks]. You will be allotted an unlimited time to warm-up after all sensors are affixed. You will perform a total of 45 maximal effort swings off a stationary hitting tee, 5 swings from 9 tee locations. You will be given a 15-second rest period between each swing to mitigate any potential fatigue that could influence testing or increase your susceptibility for injury. Testing should take approximately 1 hour and 15 minutes to complete. Please wear loose fitting athletic attire. If you, or anyone you know, are interested in participating in this study please let me know. The trials are non-invasive and possible risks and discomforts associated with this study are no greater than those involved in competitive hitting. Included are death, muscle strain, muscle soreness, and ligament and tendon damage.

If you have any questions about the study, don't hesitate to contact me or Dr. Gretchen Oliver (goliver@auburn.edu).

Thank you,
Jessica Washington
jkw0011@auburn.edu

Appendix B

HEALTH and SPORT HISTORY QUESTIONNAIRE

Part 1. Participant Information
[Please print]

ID Number ________________

Age: _______ State: ___________________ **Phone: ___________________Email:**
____

Height: _____ft _____in   Weight: ________lbs

Part 2. Athletic Participation
(Circle or fill in your responses)
1. Are you currently cleared to participate in hitting activities? **YES** **NO**

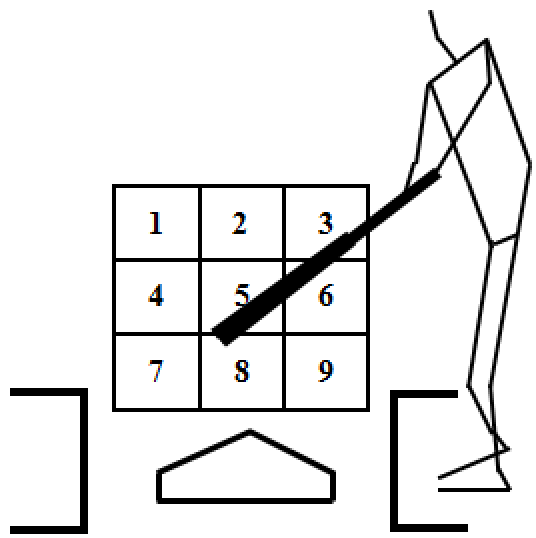
2. From which side do you hit? **RIGHT** **LEFT**

3. What position is your primary position?

4. Approximately, how many swings do you take in practice? _______ In a game? _______

5. What is "your" pitch? _________ (Please indicate by number from the grid below.)

Grid Key: **Inside pitch** for right-handed (3,6,9) and left-handed (1,4,7).
Outside pitch for right-handed (1,4,7) and left-handed (3,6,9).



6. At what competition level are you currently playing? [Please circle]

NCAA Div. I **NCAA Div. II** **NCAA Div. III** **Junior College** **High School**
Junior High **Youth League** **Other** ______________________

7. For how many years have you been participating at this level? ________

8. List all leagues you played in within the past year
____
____

9. Do you play in every game? **YES** **NO**
10. Is softball your primary sport? **YES** **NO**
List all sports you play competitively
____

11. At what age did you begin to play competitive softball? __________

12. At what age did you begin hitting? ___________

Part 3. Medical History
13. Are you allergic to adhesive tape or other adhesive products? **YES** **NO**
If **YES, explain**:
____

14. Have you ever had surgery before? **YES** **NO**
If **YES, explain**:
____
____

If YES, how long ago? ______ **Years**

15. In the past year, have you had any injury to your upper-extremity that has caused you to miss a practice or game? **YES** **NO**
If **YES, explain**:
____
____

If YES, on what part(s)? **SHOULDER** **ELBOW** **WRIST** **HAND/FINGER**

16. Do you currently experience pain/stiffness before, during or after hitting?
YES **NO**
If **YES, please explain** and continue onto question 17:
____
____

If **NO, please sign on page 3**.

IF you answered YES to question 16:

17. For how long have you been experiencing pain? (Indicate a number next to 1 category)

_____Years _____Months _____ Days

18. When you do experience pain, how would you describe the onset of pain? (Circle one)

SUDDEN GRADUAL

19. When you do experience pain, how is it related to activity? (Circle one)

ASSOCIATED WITH USE INTERMITTENT

ALL THE TIME

20. Have you changed your training/competition habits because of upper extremity pain?

YES NO

If **YES, explain**:

____

____

____

21. Have your activities of daily living been effected by your pain? **YES NO**

If **YES, explain**:

____

____

____

22. Has your pain disrupted your sleep? **YES NO**

If **YES, explain**: _____________-

____

____

____

23. Have you sought medical consultation because of your pain? **YES NO**

If **YES, explain**:

____

____

____

24. Have you been given treatment for your pain? **YES NO**

If **YES, explain**:

____

____

______

25. When you do experience pain, what is the intensity of the pain (1= NO pain; 10= unbearable pain)?

NO PAIN **UNBEARABLE PAIN**

1 2 3 4 5 6 7 8 9 10

I hereby state, to the best of my knowledge, my answers to the above questions are complete and correct.

Signature of Participant (or parent/guardian):

____

Appendix C

Informed Consent

SCHOOL OF KINESIOLOGY
301 Wire Road
Auburn, AL 36849
(334) 884-4483

AUBURN UNIVERSITY

COLLEGE OF EDUCATION

(NOTE: DO NOT SIGN THIS DOCUMENT UNLESS AN IRB APPROVAL STAMP WITH CURRENT DATES HAS BEEN APPLIED TO THIS DOCUMENT.)

Influence of Lumbopelvic-hip Complex on Hitting kinematics
INFORMED CONSENT TO PARTICIPATE IN RESEARCH

Explanation and Purpose of the Research

You are being asked to participate in a research study for the Sports Medicine & Movement Group in the Department of Kinesiology by Dr. Gretchen Oliver. Before agreeing to participate in this study, it is vital that you understand certain aspects of what might occur. This statement describes the purpose, methodology, benefits, risks, discomforts, and precautions of this research. This statement describes your right to confidentiality and to discontinue your participation at any time during the course of this research without penalty or prejudice. No assurances or guarantees can be made concerning the results of this study.

This study is designed to examine the influence of the lumbopelvic-hip complex on hitting mechanics of softball players. To investigate this, joint kinematic (where your body segments are in space), kinetic (estimates of how much force is produced), temporal (timing of movement), isometric strength (ability to contract your muscle against resistance), and range of motion data will be collected during hitting.

Research Procedures

To be considered for this study, you must be a hitting athlete between the ages of 19 – 23 and you must be injury, surgery, and pain free for the last 6 months. You must also not have an allergy to adhesive tape. Hitting dominance will not be a selection factor for this study.

Testing in this research will require the evaluation of height, body mass, age, and range of motion. Body mass and height will be measured with the Motion Monitor motion capture system and will be recorded to the nearest hundredth of a kilogram and centimeter. Age will be determined from this consent form and will be recorded to the nearest month. Range of motion will be measured with a goniometer and will be recorded to the nearest degree.

Once all preliminary paperwork has been completed, you will need to be dressed in loose fitting athletic attire for testing. Range of motion of the shoulder and hip will be first be measured and recorded. To measure shoulder range of motion, you will lay supine on the table with your arm hanging off the side at the shoulder. An investigator will hold the your arm parallel to the ground with your elbow bent to ninety degrees. The investigator will then passively rotate your arm until maximal internal rotation is reached. This will then be repeated for maximal external rotation. For hip range of motion, you will sit on a table with your lower leg hanging off the side. An investigator will passively rotate your lower leg until maximal internal rotation is reached. This will then be repeated for maximal external rotation.

Similar procedures will be performed to measure isometric strength. To measure shoulder strength you will be positioned lying on a table with your upper arm parallel to the floor and elbow bent. You will then externally/internally rotate against the investigator for 3 seconds. The investigator will hold a hand held dynamometer against your forearm to record the force. In order to reduce the effects of fatigue, a rest period of 20-30 seconds will be allotted between trials.

Next, electromagnetic sensors will be placed on your legs, arms, torso, and neck. Placement of the markers at these locations will allow the movement of the joint centers to be properly monitored during testing. Eight surface electrodes will be placed on the following muscles: bilateral external oblique [lateral abdomen], bilateral latissimus dorsi [mid lateral back], bilateral gluteus medius [lateral hip], and bilateral gluteus maximus [mid buttocks]. Manual muscle testing will be performed to establish baseline muscle activity in which all data will be compared.

Once these measurements have been collected and following the placement of the markers, you will perform your own specified pre-competition warm-up routine. During the warm-up period, we ask that you contribute five minutes to maximal effort swings. After completing the warm-up, a total of 45 maximal effort swings will be made while hitting off a tee randomly positioned in 9 different locations. A 15 second rest period will be allotted between each swing to mitigate potential effects of fatigue.

Potential Risks

As with any movement research, certain risks and discomforts may arise. The possible risks and discomforts associated with this study are no greater than those involved in hitting and may include: death, muscle strain, muscle soreness, ligament and tendon damage, and general overuse injury to the hitting athlete. Every effort will be made to minimize these risks and discomforts. It is your responsibility, as a participant, to inform the investigators if you notice any indications of injury or fatigue, or feel symptoms of any other possible complications that might occur during testing.

To reduce the risk of injury, certain precautions will be taken. During the hitting protocol, two board certified athletic trainers will be present to monitor you as you hit. Ample warm-up and cool-down periods will be required of you, water will be provided to you as needed, and ice will be made available after testing.

The researcher will try to prevent any problem that could happen because of this research. If at any time there is a problem, you should let the researcher know and she will help you. Should an emergency arise, we will call 911 and follow our Emergency Action Plan. In the unlikely event that you sustain an injury from participation in this study, the investigators have no current plans to provide funds for any medical expenses or other costs you may incur.

Confidentiality

All information gathered in completing this study will remain confidential. Your individual performance will not be made available for public use, and will not be disclosed to any person(s) outside of the research team. The results of this study may be published as scientific research. Your name or identity shall not be revealed should such publication occur.

Participation and Benefits

Participation in this research is strictly voluntary and refusal to participate will result in no penalty. If you change your mind about participating, you can withdraw at any time during the study. Your participation is completely voluntary. If you choose to withdraw, your data can be withdrawn as long as it is identifiable. Your decision about whether or not to participate or to stop participating will not jeopardize your future relations with Auburn University or the School of Kinesiology.

By participating in this study, you will receive information regarding hitting mechanics that may help prevent injury. This will allow you the opportunity to alter your training programs in an effort to minimize injury resulting from fatigue, etc.

Questions Regarding the Study

If you have questions about this study, please ask them now. If you have questions later you may contact Dr. Gretchen Oliver, 844-1497 or goliver@auburn.edu.

If you have any questions about your rights as a research participant, you may contact the Auburn University Office of Research Compliance or the Institutional Review Board by phone (334)-844-5966 or email at irbadmin@auburn.edu or IRBChair@auburn.edu.

HAVING READ THE INFORMATION PROVIDED, YOU MUST DECIDE WHETHER OR NOT YOU WISH TO PARTICIPATE IN THIS RESEARCH STUDY. YOUR SIGNATURE INDICATES YOUR WILLINGNESS TO PARTICIPATE.

______________________________	______ yr. ______ mo.
Printed Name of Participant	Age of Participant
______________________________	______________
Signature of Participant	Date

The above consent form was read, discussed, and signed in my presence. In my opinion, the person signing said consent form did so freely and with full knowledge of its contents.

____ ___________________
Signature of Investigator, Jessica Washington Date

Appendix D
Parental Assent

SCHOOL OF KINESIOLOGY
301 Wire Road
Auburn, AL 36849
(334) 884-4483

AUBURN UNIVERSITY
COLLEGE OF EDUCATION

(NOTE: DO NOT SIGN THIS DOCUMENT UNLESS AN IRB APPROVAL STAMP WITH CURRENT DATES HAS BEEN APPLIED TO THIS DOCUMENT.)

Parental Permission/Minor Assent

Influence of Lumbopelvic-hip Complex on Hitting kinematics
CONSENT TO PARTICIPATE IN RESEARCH

Explanation and Purpose of the Research

Your child is being asked to participate in a research study for the Sports Medicine & Movement Group in the Department of Kinesiology by Dr. Gretchen Oliver. Before agreeing to participate in this study, it is vital that you and your child understand certain aspects of what might occur. This statement describes the purpose, methodology, benefits, risks, discomforts, and precautions of this research. This statement describes your right to confidentiality and your child's right to discontinue their participation at any time during the course of this research without penalty or prejudice. No assurances or guarantees can be made concerning the results of this study.

This study is designed to examine the influence of the lumbopelvic-hip complex on hitting mechanics of softball players. To investigate this, joint kinematic (where your body segments are in space), kinetic (estimates of how much force is produced), temporal (timing of movement), isometric strength (ability to contract your muscle against resistance), and range of motion data will be collected during hitting.

Research Procedures

To be considered for this study, your child must be a hitting athlete between the ages of 17 – 18 and you must be injury, surgery, and pain free for the last 6 months. Your child must also not have an allergy to adhesive tape. Hitting dominance will not be a selection factor for this study.

Testing in this research will require the evaluation of height, body mass, age, and range of motion. Body mass and height will be measured with the Motion Monitor motion capture system and will be recorded to the nearest hundredth of a kilogram and centimeter. Age will be determined from this consent form and will be recorded to the nearest month. Range of motion will be measured with a goniometer and will be recorded to the nearest degree.

Once all preliminary paperwork has been completed, your child will need to be dressed in loose fitting athletic attire for testing. Range of motion of the shoulder and hip will be first be measured and recorded. To measure shoulder range of motion, your child will lay supine on the table with arm hanging off the side at the shoulder. An investigator will hold the their arm parallel to the ground with their elbow bent to ninety degrees. The investigator will then passively rotate their arm until maximal internal rotation is reached. This will then be repeated for maximal external rotation. For hip range of motion, your child will sit on a table with their lower leg hanging off the side. An investigator will passively rotate their lower leg until maximal internal rotation is reached. This will then be repeated for maximal external rotation.

Similar procedures will be performed to measure isometric strength. To measure shoulder strength your child will be positioned lying on a table with their upper arm parallel to the floor and elbow bent. They will then externally/internally rotate against the investigator for 3 seconds. The investigator will hold a hand held dynamometer against their forearm to record the force. In order to reduce the effects of fatigue, a rest period of 20-30 seconds will be allotted between trials.

Next, electromagnetic sensors will be placed on your child's legs, arms, torso, and neck. Placement of the markers at these locations will allow the movement of the joint centers to be properly monitored during testing. Eight surface electrodes will be placed on the following muscles: bilateral external oblique [lateral abdomen], bilateral latissimus dorsi [mid lateral back], bilateral gluteus medius [lateral hip], and bilateral gluteus maximus [mid buttocks]. Manual muscle testing will be performed to establish baseline muscle activity in which all data will be compared.

Once these measurements have been collected and following the placement of the markers, your child will perform their own specified pre-competition warm-up routine. During the warm-up period, we ask that they contribute five minutes to maximal effort hitting. After completing the warm-up, a total of 45 maximal effort swings will be made while hitting off a tee randomly positioned in 9 different locations. A 15 second rest period will be allotted between each swing to mitigate potential effects of fatigue.

Potential Risks

As with any movement research, certain risks and discomforts may arise. The possible risks and discomforts associated with this study are no greater than those involved in hitting and may include: death, muscle strain, muscle soreness, ligament and tendon damage, and general overuse injury to the athlete. Every effort will be made to minimize these risks and discomforts. It is your child's responsibility, as a participant, to inform the investigators if they notice any indications of injury or fatigue, or feel symptoms of any other possible complications that might occur during testing.

To reduce the risk of injury, certain precautions will be taken. During the hitting protocol, two board certified athletic trainers will be present to monitor your child as they hit. Ample warm-up and cool-down periods will be required of your child, water will be provided to your child as needed, and ice will be made available after testing.

The researcher will try to prevent any problem that could happen because of this research. If at any time there is a problem, you should let the researcher know and she will help you. Should an emergency arise, we will call 911 and follow our Emergency Action Plan. In the unlikely event that you sustain an injury from participation in this study, the investigators have no current plans to provide funds for any medical expenses or other costs you may incur.

Confidentiality
All information gathered in completing this study will remain confidential. Your child's individual performance will not be made available for public use, and will not be disclosed to any person(s)
outside of the research team. The results of this study may be published as scientific research. Your child's name or identity shall not be revealed should such publication occur.

Participation and Benefits
Participation in this research is strictly voluntary and refusal to participate will result in no penalty. If you or your child changes your mind about participating, they can withdraw at any time during the study. Your child's participation is completely voluntary. If you or your child chooses to withdraw, their data can be withdrawn as long as it is identifiable. Your decision about whether or not to participate or to stop participating will not jeopardize you or your child's future relations with Auburn University or the School of Kinesiology.

By participating in this study, your child will receive information regarding hitting mechanics that may help prevent injury. This will allow your child the opportunity to alter their training programs in an effort to minimize injury resulting from fatigue, etc.

Questions Regarding the Study

If you have questions about this study, please ask them now. If you have questions later you may contact Dr. Gretchen Oliver, 844-1497 or goliver@auburn.edu.

If you have any questions about your child's rights as a research participant, you may contact the Auburn University Office of Research Compliance or the Institutional Review Board by phone (334)-844-5966 or email at irbadmin@auburn.edu or IRBChair@auburn.edu.

HAVING READ THE INFORMATION PROVIDED, YOU MUST DECIDE WHETHER OR NOT YOU WISH FOR YOUR CHILD TO PARTICIPATE IN THIS RESEARCH STUDY. YOUR SIGNATURE INDICATES YOUR WILLINGNESS TO ALLOW YOUR CHILD'S PARTICIPATION.

______________________________	______ yr. ______ mo.
Printed Name of Parent	Age of Participant
______________________________	______________
Signature of Parent	Date
______________________________	
Printed Name of Participant	

Signature of Participant Date

The above consent form was read, discussed, and signed in my presence. In my opinion, the person signing said consent form did so freely and with full knowledge of its contents.

Signature of Investigator, Jessica Washington Date

Appendix E

Participant Data Form

Participant ID: ______________ Age: _____ Date: ___________

Height: _________ Weight: _________

	IR	ER
	ISO	ISO
Back Hip	1)	1)
	2)	2)
	3)	3)
Lead Hip	1)	1)
	2)	2)
	3)	3)

Total # Warm-up Swings: ________

Total # Swings: _________

Bat Length: _________ in.

Bat Mass: _________ oz.

Bat Brand: ___________________

Trial	Tee Location
1	
2	
3	
4	
5	
6	
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11	
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Appendix F

Hand Velocity Normality

Average hand velocity (1279.790 ± 62.944, N = 27) was non-normally distributed with a Skewness of $\alpha_3 = 1.384$ ($SE_{skew} = 0.448$) and a Kurtosis of $\alpha_4 = 5.146$ ($SE_{kurt} = 0.872$) (Figure F1, Table F1).

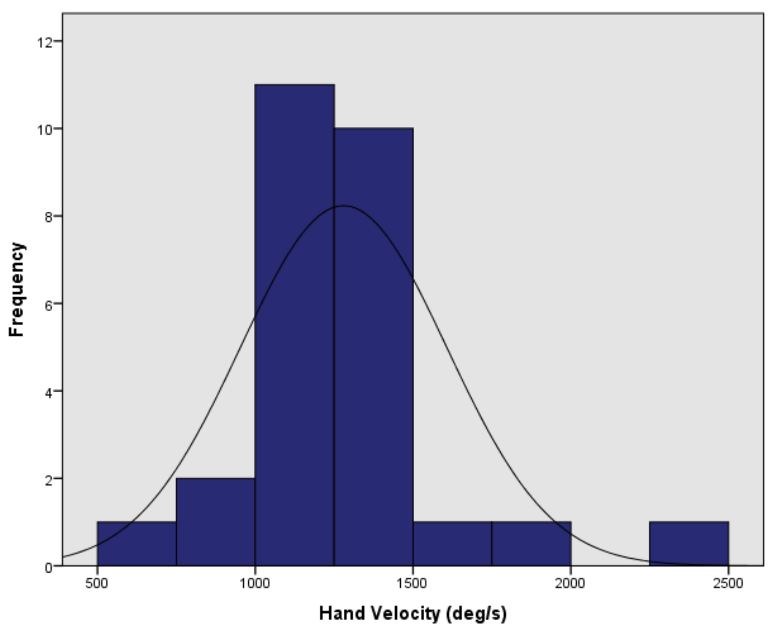

Figure F1. Distribution of hand velocity, N = 27.

Table F1. Tests of normality for hand velocity, N = 27.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Hand Velocity	0.200	27	0.007	0.854	27	0.001

a. Lilliefors Significance Correction

Logarithmic transformation of hand velocity also resulted in a non-normal distribution with a Skewness of $\alpha_3 = -0.571$ ($SE_{skew} = 0.448$) and a Kurtosis of $\alpha_4 = 4.907$ ($SE_{kurt} = 0.872$) (Figure F2, Table F2).

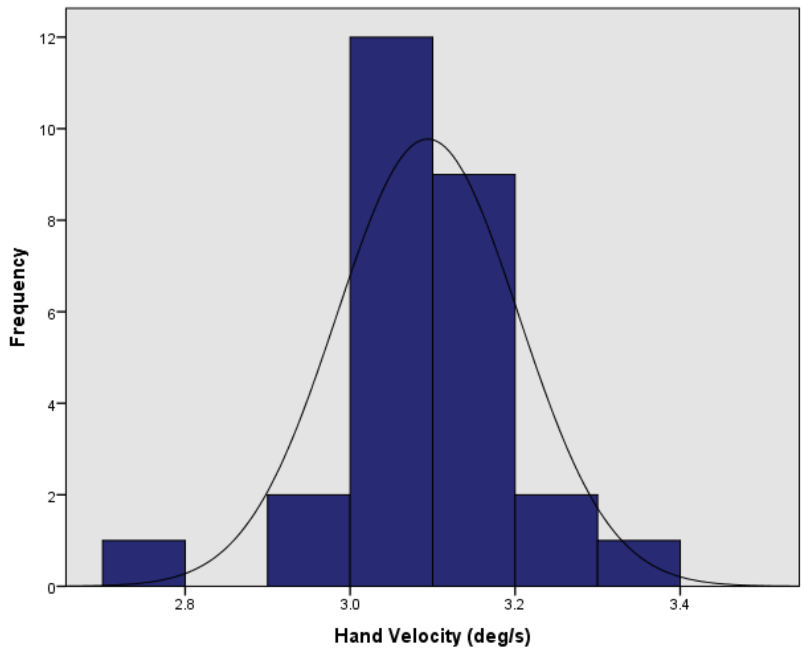

Figure F2. Distribution of log transformed hand velocity, N = 27.

Table F2. Tests of normality for log transformed hand velocity, N = 27.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Hand Velocity Lg10	0.168	27	0.049	0.877	27	0.004

a. Lilliefors Significance Correction

One significant outlier was found within the dataset and removed for all regression analyses, as hand velocity was the dependent variable for each regression. Therefore, average hand velocity (1236.994 ± 47.968, $N = 26$) was then normally distributed with a Skewness of α_3 = -0.052 (SE_{skew} = 0.456) and a Kurtosis of α_4 = 3.152 (SE_{kurt} = 0.887) (Figure F3, Table F3).

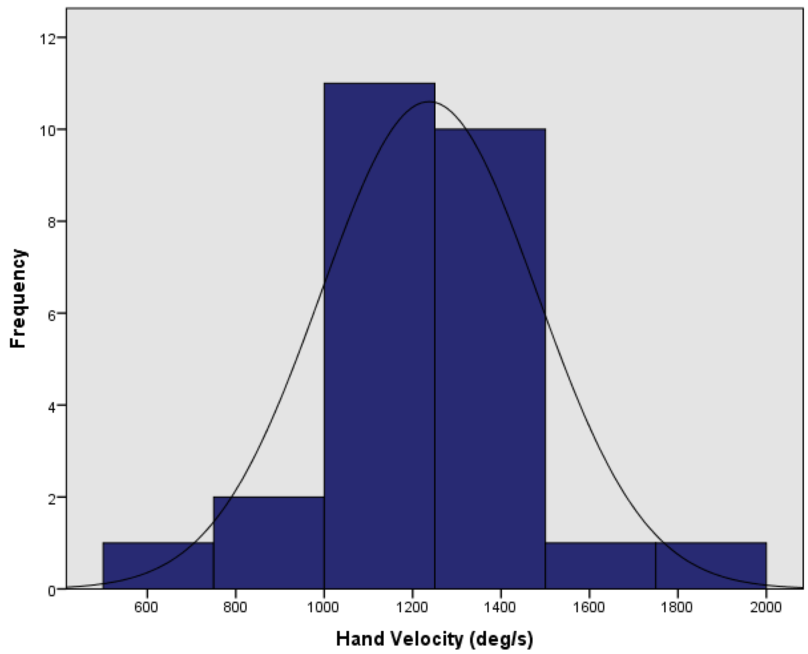

Figure F3. Distribution of hand velocity with outliers removed, N = 26.

Table F3. Tests of normality for hand velocity with outliers removed, N = 26.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Hand Velocity	0.126	26	0.200*	0.929	26	0.075

*. This is a lower bound of the true significance.
a. Lilliefors Significance Correction

Research Question 1 Normality

Average stride hip internal rotation isometric strength (22.922 ± 1.317, *N* = 26), expressed as a percentage of body weight, was normally distributed with a Skewness of α_3 = 0.959 (SE_{skew} = 0.456) and a Kurtosis of α_4 = 1.776 (SE_{kurt} = 0.887) (Figure F4, Table F4).

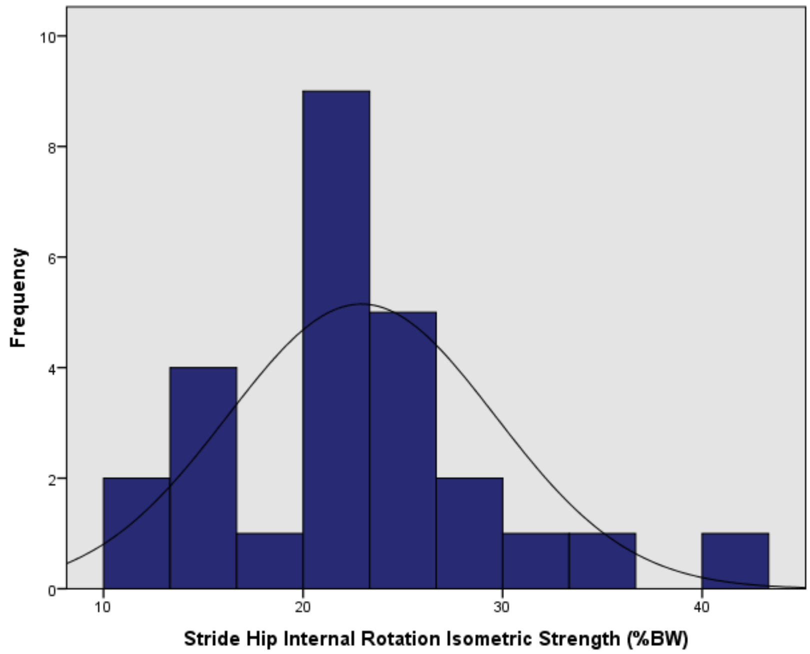

Figure F4. Distribution of average stride hip internal rotation isometric strength, N = 26.

Table F4. Tests of Normality for average load hip external rotation isometric strength, N = 26.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Stride Hip IR[b] ISO[c]	0.155	26	0.108	0.926	26	0.061

[a]Lilliefors Significance Correction
[b]IR – Femoral Internal Rotation
[c]ISO – Isometric Strength

Average load hip external rotation isometric strength (19.933 ± 0.973, N = 26), expressed as a percentage of body weight, was normally distributed with a Skewness of $\alpha_3 = 0.476$ ($SE_{skew} = 0.456$) and a Kurtosis of $\alpha_4 = 0.154$ ($SE_{kurt} = 0.887$) (Figure F5, Table F5).

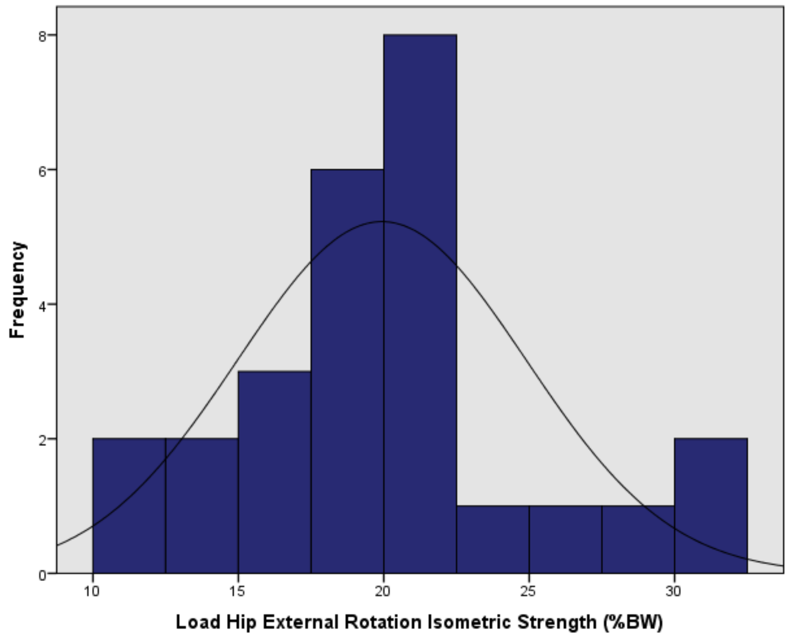

Figure F5. Distribution of average load hip external rotation isometric strength, N = 26.

Table F5. Tests of Normality for average load hip external rotation isometric strength, N = 26.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Load Hip ER[b] ISO[c]	0.138	26	0.200*	0.954	26	0.280

*. This is a lower bound of the true significance.
[a]Lilliefors Significance Correction
[b]ER – Femoral External Rotation
[c]ISO – Isometric Strength

Research Question 2 Normality

Average degrees of pelvis and torso separation (29.157 ± 3.422, N = 26) were normally distributed with a Skewness of $\alpha_3 = 0.707$ ($SE_{skew} = 0.456$) and a Kurtosis of $\alpha_4 = 0.528$ ($SE_{kurt} = 0.887$) (Figure F6, Table F6).

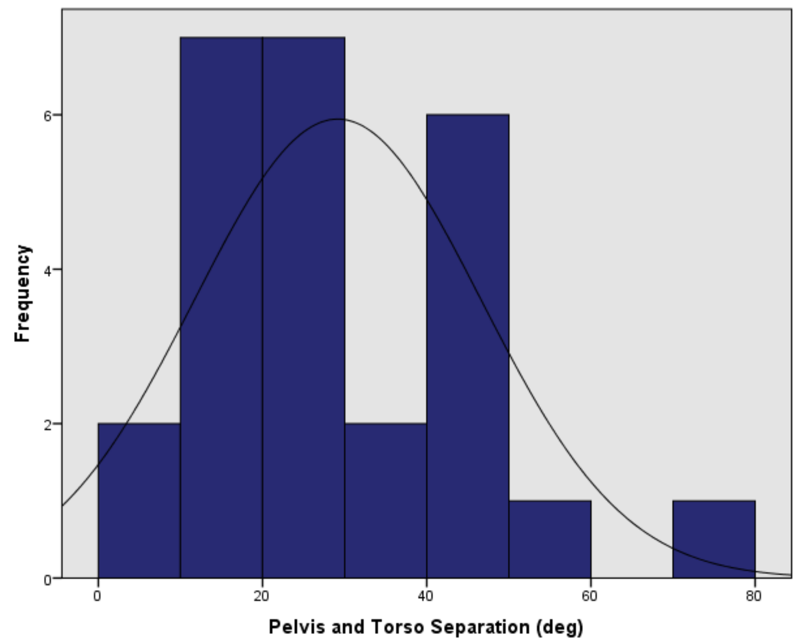

Figure F6. Distribution of average pelvis and torso separation during the acceleration phase, N = 26.

Table F6. Tests of normality for pelvis and torso separation during the acceleration phase, N = 26.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Separation	0.117	26	0.200*	0.956	26	0.318

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Question 3 Normality

Average rotational pelvis jerk (-226.675 ± 3109.719, N = 26) was non-normally distributed with a Skewness of $\alpha_3 = 0.845$ ($SE_{skew} = 0.456$) and a Kurtosis of $\alpha_4 = 4.298$ ($SE_{kurt} = 0.887$) (Figure F7, Table F7).

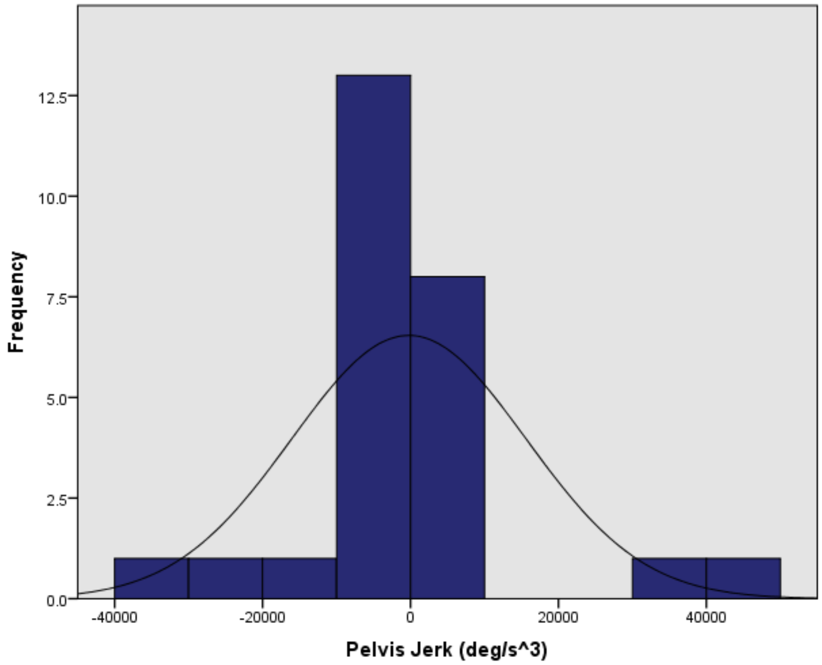

Figure F7. Distribution of average pelvis jerk during the acceleration phase, N = 26.

Table F7. Tests of normality for average pelvis jerk during the acceleration phase, N = 26.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Pelvis Jerk	0.286	26	0.000	0.786	26	0.000

a. Lilliefors Significance Correction

Eight significant outliers were found within the dataset and removed from the multiple regression analyses. Therefore, average rotational pelvis jerk (-979.583 ± 507.427, N = 18) was then normally distributed with a Skewness of $\alpha_3 = -0.259$ ($SE_{skew} = 0.536$) and a Kurtosis of $\alpha_4 = 3.358$ ($SE_{kurt} = 1.038$) (Figure F8, Table F8).

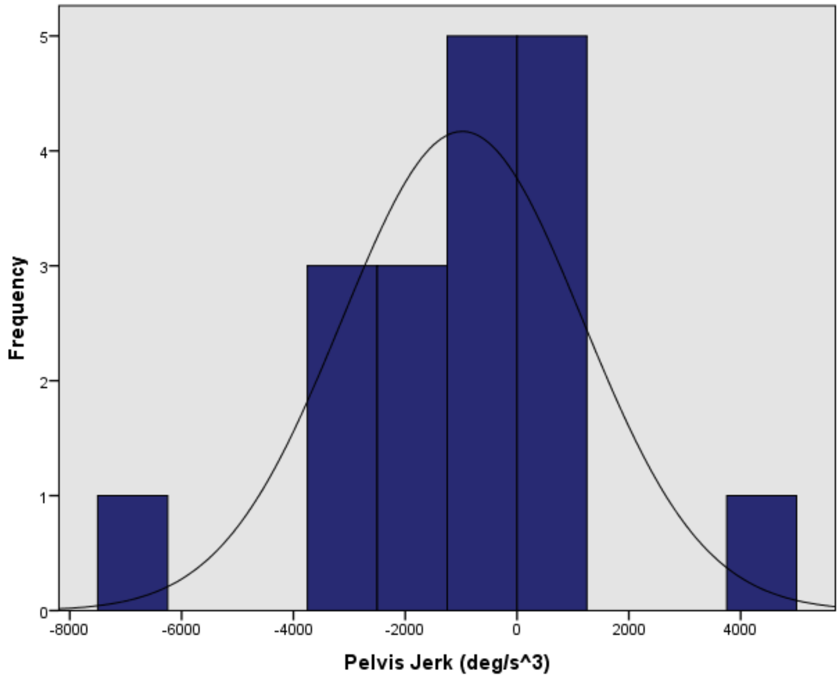

Figure F8. Distribution of average pelvis jerk during the acceleration phase with additional outliers removed, N = 18.

Table F8. Tests of normality for average pelvis jerk during the acceleration phase with additional outliers removed, N = 18.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Pelvis Jerk	0.171	18	0.178	0.906	18	0.072

a. Lilliefors Significance Correction

Average rotational torso jerk (-536.453 ± 956.837 , N = 18) was normally distributed with a Skewness of α_3 = -0.262 (SE_{skew} = 0.536) and a Kurtosis of α_4 = 2.231 (SE_{kurt} = 1.038) (Figure F9, Table F9).

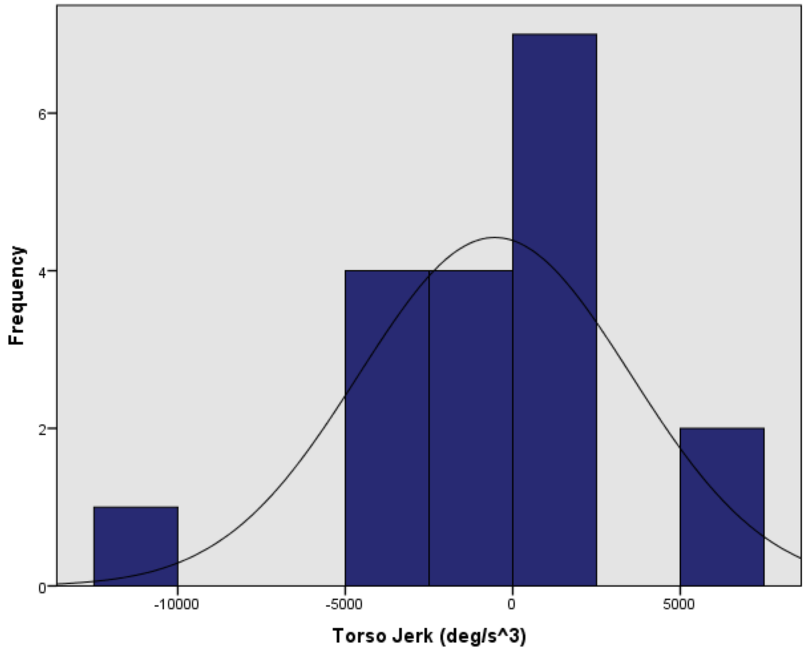

Figure F9. Distribution of average torso jerk during the acceleration phase with additional outliers removed, N = 18.

Table F9. Tests of normality for average torso jerk during the acceleration phase with additional outliers removed, N = 18.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Torso Jerk	0.193	18	0.073	0.909	18	0.083

a. Lilliefors Significance Correction

Question 4 Normality

Average load time, represented as a percentage of the swing from stance to ball contact (44.017 ± 1.559, N = 26), was normally distributed with a Skewness of $\alpha_3 = 0.407$ ($SE_{skew} = 0.456$) and a Kurtosis of $\alpha_4 = -0.326$ ($SE_{kurt} = 0.887$) (Figure F10, Table F10).

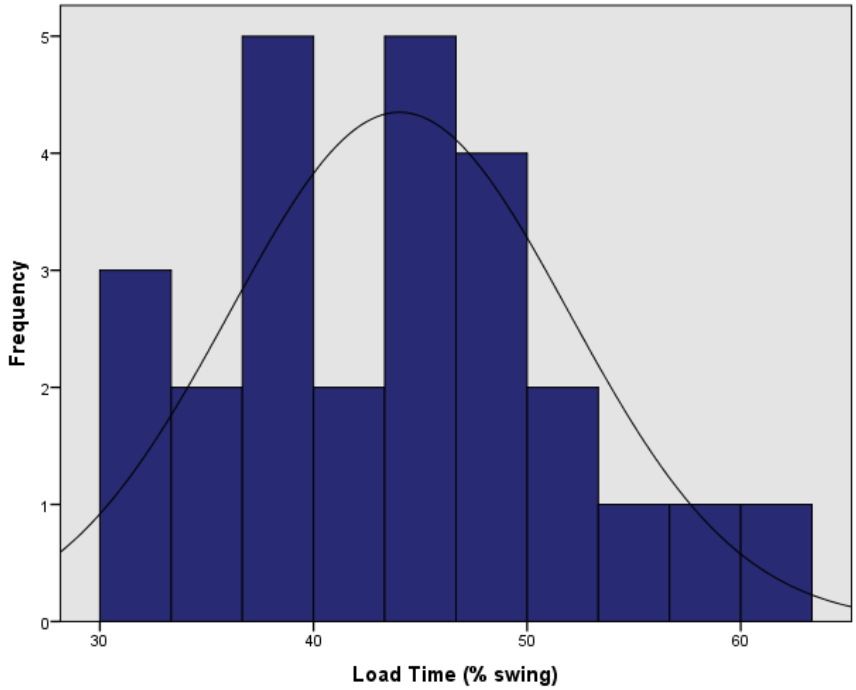

Figure F10. Distribution of load time, N = 26.

Table F10. Tests of normality for load time, N = 26.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Load Time	0.082	26	0.200*	0.966	26	0.516

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Question 5 Normality

Average timing of maximum pelvis rotation angular velocity, represented as the percent at which the maximum value occurred during the acceleration phase (58.399 ± 2.794, N = 26), was normally distributed with a Skewness of α_3 = -0.193 (SE_{skew} = 0.456) and a Kurtosis of α_4 = -0.835 (SE_{kurt} = 0.887) (Figure F11, Table F11).

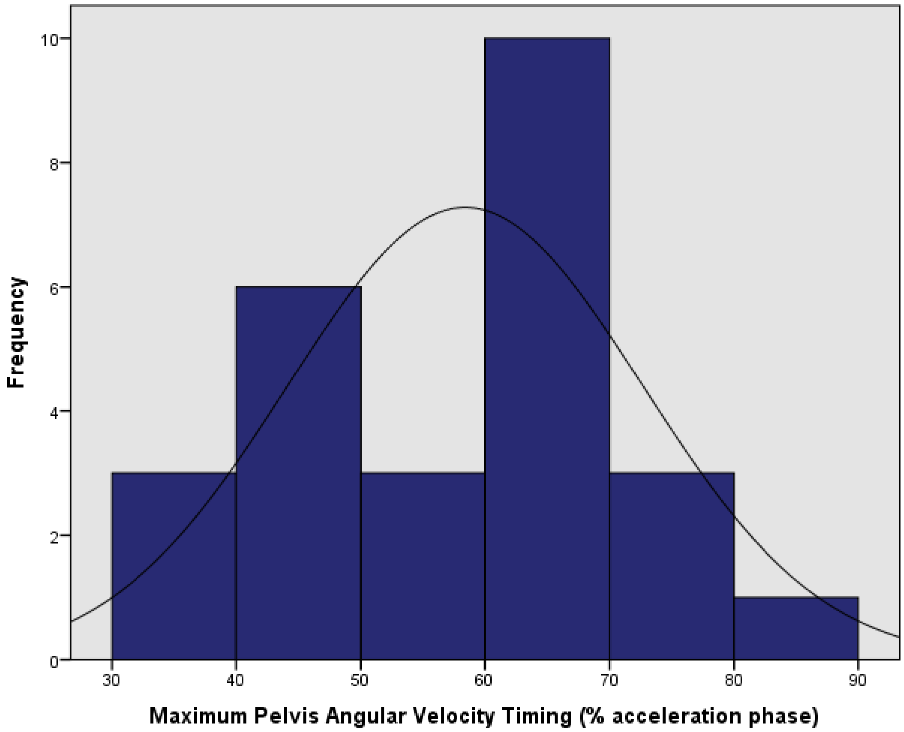

Figure F11. Distribution of maximum pelvis rotation angular velocity timing, N = 26.

Table F11. Tests of normality for maximum pelvis rotation angular velocity timing, N = 26.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Pelvis Time	0.138	26	0.200*	0.962	26	0.436

*. This is a lower bound of the true significance.
a. Lilliefors Significance Correction

Average timing of maximum torso rotation angular velocity, represented as the percent at which the maximum value occurred during the acceleration phase (81.975 ± 2.397, N = 26), was normally distributed with a Skewness of α_3 = -0.543 (SE_{skew} = 0.456) and a Kurtosis of α_4 = -0.618 (SE_{kurt} = 0.887) (Figure F12, Table F12).

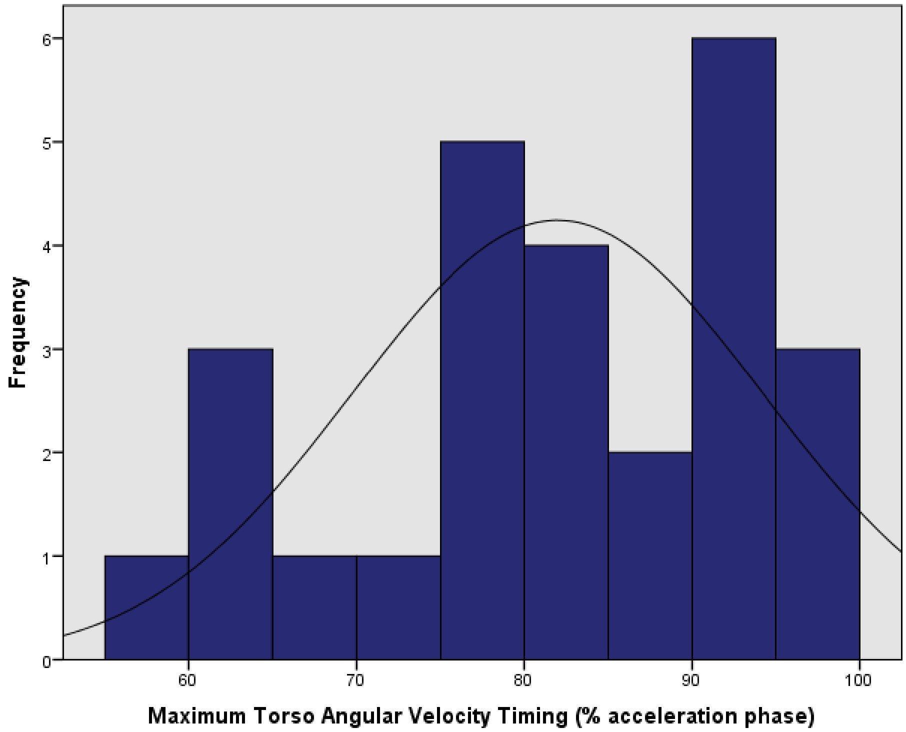

Figure F12. Distribution of maximum torso rotation angular velocity timing, N = 26.

Table F12. Tests of normality for maximum torso rotation angular velocity timing, N = 24.

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Trunk Time	0.127	26	0.200*	0.938	26	0.121

*. This is a lower bound of the true significance.

a. Lilliefors Significance Correction

Appendix G

Statistical Outputs

Research Question 1

Descriptives

			Statistic	Std. Error
HandVel_BC	Mean		1236.994153	47.96761627
	95% Confidence Interval for Mean	Lower Bound	1138.202998	
		Upper Bound	1335.785308	
	5% Trimmed Mean		1236.860021	
	Median		1228.427821	
	Variance		59823.197	
	Std. Deviation		244.5878114	
	Minimum		540.4760410	
	Maximum		1898.591945	
	Range		1358.115904	
	Interquartile Range		277.5421625	
	Skewness		-.052	.456
	Kurtosis		3.152	.887

Descriptives

			Statistic	Std. Error
Stride Hip IR %BW	Mean		22.92205411	1.316611810
	95% Confidence Interval for Mean	Lower Bound	20.21044133	
		Upper Bound	25.63366689	
	5% Trimmed Mean		22.48854211	
	Median		22.30578284	
	Variance		45.070	
	Std. Deviation		6.713429311	
	Minimum		12.73212844	
	Maximum		42.24883566	
	Range		29.51670722	
	Interquartile Range		7.925157105	
	Skewness		.959	.456
	Kurtosis		1.776	.887

Descriptives

			Statistic	Std. Error
Load Hip ER %BW	Mean		19.93276164	.9730002306
	95% Confidence Interval for Mean	Lower Bound	17.92883015	
		Upper Bound	21.93669313	
	5% Trimmed Mean		19.77206590	
	Median		20.02790645	
	Variance		24.615	
	Std. Deviation		4.961347162	
	Minimum		11.96509009	
	Maximum		30.92783505	
	Range		18.96274496	
	Interquartile Range		6.505291787	
	Skewness		.476	.456
	Kurtosis		.154	.887

Correlations

		HandVel_BC	Stride Hip IR %BW	Load Hip ER %BW
Pearson Correlation	HandVel_BC	1.000	-.129	-.120
	Stride Hip IR %BW	-.129	1.000	.934
	Load Hip ER %BW	-.120	.934	1.000
Sig. (1-tailed)	HandVel_BC	.	.264	.279
	Stride Hip IR %BW	.264	.	.000
	Load Hip ER %BW	.279	.000	.
N	HandVel_BC	26	26	26
	Stride Hip IR %BW	26	26	26
	Load Hip ER %BW	26	26	26

Variables Entered/Removed[a]

Model	Variables Entered	Variables Removed	Method
1	Load Hip ER %BW, Stride Hip IR %BW[b]	.	Enter

a. Dependent Variable: HandVel_BC

b. All requested variables entered.

Model Summary[b]

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics					Durbin-Watson
					R Square Change	F Change	df1	df2	Sig. F Change	
1	.129[a]	.017	-.069	252.8561893	.017	.196	2	23	.823	1.858

a. Predictors: (Constant), Load Hip ER %BW, Stride Hip IR %BW

b. Dependent Variable: HandVel_BC

ANOVA[a]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	25046.130	2	12523.065	.196	.823[b]
	Residual	1470533.807	23	63936.252		
	Total	1495579.937	25			

a. Dependent Variable: HandVel_BC

b. Predictors: (Constant), Load Hip ER %BW, Stride Hip IR %BW

Coefficients[a]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95.0% Confidence Interval for B		Correlations			Collinearity Statistics	
		B	Std. Error	Beta			Lower Bound	Upper Bound	Zero-order	Partial	Part	Tolerance	VIF
1	(Constant)	1343.928	214.473		6.266	.000	900.258	1787.599					
	Stride Hip IR %BW	-4.903	21.030	-.135	-.233	.818	-48.406	38.601	-.129	-.049	-.048	.128	7.794
	Load Hip ER %BW	.273	28.456	.006	.010	.992	-58.593	59.140	-.120	.002	.002	.128	7.794

a. Dependent Variable: HandVel_BC

Research Question 2

Descriptives

			Statistic	Std. Error
HandVel_BC	Mean		1236.994153	47.96761627
	95% Confidence Interval for Mean	Lower Bound	1138.202998	
		Upper Bound	1335.785308	
	5% Trimmed Mean		1236.860021	
	Median		1228.427821	
	Variance		59823.197	
	Std. Deviation		244.5878114	
	Minimum		540.4760410	
	Maximum		1898.591945	
	Range		1358.115904	
	Interquartile Range		277.5421625	
	Skewness		-.052	.456
	Kurtosis		3.152	.887

Descriptives

			Statistic	Std. Error
Q2_Separation_Acc phase	Mean		29.15697233	3.422050735
	95% Confidence Interval for Mean	Lower Bound	22.10912692	
		Upper Bound	36.20481775	
	5% Trimmed Mean		28.28123278	
	Median		25.55903133	
	Variance		304.471	
	Std. Deviation		17.44910347	
	Minimum		2.100685333	
	Maximum		76.50586667	
	Range		74.40518133	
	Interquartile Range		27.70106517	
	Skewness		.707	.456
	Kurtosis		.528	.887

Correlations

		HandVel_BC	Q2_Separation_Acc phase
Pearson Correlation	HandVel_BC	1.000	-.351
	Q2_Separation_Acc phase	-.351	1.000
Sig. (1-tailed)	HandVel_BC	.	.039
	Q2_Separation_Acc phase	.039	.
N	HandVel_BC	26	26
	Q2_Separation_Acc phase	26	26

Research Question 3

Descriptives

			Statistic	Std. Error
HandVel_BC	Mean		1251.878749	35.56420406
	95% Confidence Interval for Mean	Lower Bound	1176.844837	
		Upper Bound	1326.912660	
	5% Trimmed Mean		1242.504621	
	Median		1228.427821	
	Variance		22766.627	
	Std. Deviation		150.8861392	
	Minimum		1064.564085	
	Maximum		1607.927718	
	Range		543.3636337	
	Interquartile Range		231.7280627	
	Skewness		.734	.536
	Kurtosis		.091	1.038

Descriptives

			Statistic	Std. Error
Q3_Pelvis Jerk	Mean		-979.582838	507.4268360
	95% Confidence Interval for Mean	Lower Bound	-2050.15988	
		Upper Bound	90.99420547	
	5% Trimmed Mean		-959.985849	
	Median		-711.556120	
	Variance		4634675.890	
	Std. Deviation		2152.829740	
	Minimum		-6576.18152	
	Maximum		4264.270040	
	Range		10840.45156	
	Interquartile Range		2267.412450	
	Skewness		-.259	.536
	Kurtosis		3.358	1.038

Descriptives

			Statistic	Std. Error
Q3_Trunk Jerk	Mean		-536.452560	956.8369463
	95% Confidence Interval for Mean	Lower Bound	-2555.20205	
		Upper Bound	1482.296935	
	5% Trimmed Mean		-403.597213	
	Median		-439.814220	
	Variance		16479664.95	
	Std. Deviation		4059.515359	
	Minimum		-10916.0342	
	Maximum		7451.732800	
	Range		18367.76696	
	Interquartile Range		3925.574510	
	Skewness		-.262	.536
	Kurtosis		2.231	1.038

Model Summary

R	R Square	Adjusted R Square	Std. Error of the Estimate
.192	.037	-.091	157.634

The independent variable is Q3_Pelvis Jerk.

ANOVA

	Sum of Squares	df	Mean Square	F	Sig.
Regression	14306.239	2	7153.120	.288	.754
Residual	372726.419	15	24848.428		
Total	387032.659	17			

The independent variable is Q3_Pelvis Jerk.

Coefficients

	Unstandardized Coefficients		Standardized Coefficients	t	Sig.
	B	Std. Error	Beta		
Q3_Pelvis Jerk	-.004	.021	-.056	-.191	.851
Q3_Pelvis Jerk ** 2	-3.058E-6	.000	-.214	-.731	.476
(Constant)	1264.354	42.817		29.529	.000

Model Summary

R	R Square	Adjusted R Square	Std. Error of the Estimate
.370	.137	.022	149.225

The independent variable is Q3_Trunk Jerk.

ANOVA

	Sum of Squares	df	Mean Square	F	Sig.
Regression	53012.615	2	26506.307	1.190	.331
Residual	334020.044	15	22268.003		
Total	387032.659	17			

The independent variable is Q3_Trunk Jerk.

Coefficients

	Unstandardized Coefficients		Standardized Coefficients	t	Sig.
	B	Std. Error	Beta		
Q3_Trunk Jerk	.014	.009	.371	1.489	.157
Q3_Trunk Jerk ** 2	9.915E-7	.000	.198	.795	.439
(Constant)	1243.558	39.990		31.097	.000

Research Question 4

Descriptives

			Statistic	Std. Error
HandVel_BC	Mean		1236.994153	47.96761627
	95% Confidence Interval for Mean	Lower Bound	1138.202998	
		Upper Bound	1335.785308	
	5% Trimmed Mean		1236.860021	
	Median		1228.427821	
	Variance		59823.197	
	Std. Deviation		244.5878114	
	Minimum		540.4760410	
	Maximum		1898.591945	
	Range		1358.115904	
	Interquartile Range		277.5421625	
	Skewness		-.052	.456
	Kurtosis		3.152	.887

Descriptives

			Statistic	Std. Error
Q4_Load Time_%swing	Mean		44.01677273	1.558774032
	95% Confidence Interval for Mean	Lower Bound	40.80641752	
		Upper Bound	47.22712795	
	5% Trimmed Mean		43.72806211	
	Median		44.14774835	
	Variance		63.174	
	Std. Deviation		7.948219208	
	Minimum		32.45132069	
	Maximum		61.65431260	
	Range		29.20299191	
	Interquartile Range		10.64108903	
	Skewness		.407	.456
	Kurtosis		-.326	.887

Correlations

		HandVel_BC	Q4_Load Time_% swing
Pearson Correlation	HandVel_BC	1.000	-.208
	Q4_Load Time_%swing	-.208	1.000
Sig. (1-tailed)	HandVel_BC	.	.155
	Q4_Load Time_%swing	.155	.
N	HandVel_BC	26	26
	Q4_Load Time_%swing	26	26

Research Question 5

Descriptives

			Statistic	Std. Error
HandVel_BC	Mean		1236.994153	47.96761627
	95% Confidence Interval for Mean	Lower Bound	1138.202998	
		Upper Bound	1335.785308	
	5% Trimmed Mean		1236.860021	
	Median		1228.427821	
	Variance		59823.197	
	Std. Deviation		244.5878114	
	Minimum		540.4760410	
	Maximum		1898.591945	
	Range		1358.115904	
	Interquartile Range		277.5421625	
	Skewness		-.052	.456
	Kurtosis		3.152	.887

Descriptives

			Statistic	Std. Error
Avg Pelvis % acc phase	Mean		58.39896588	2.794011430
	95% Confidence Interval for Mean	Lower Bound	52.64459162	
		Upper Bound	64.15334013	
	5% Trimmed Mean		58.55126910	
	Median		61.78030120	
	Variance		202.969	
	Std. Deviation		14.24671880	
	Minimum		30.35535536	
	Maximum		83.36320191	
	Range		53.00784656	
	Interquartile Range		20.28002629	
	Skewness		-.193	.456
	Kurtosis		-.835	.887

Descriptives

			Statistic	Std. Error
Avg Torso % acc phase	Mean		81.97506173	2.396602506
	95% Confidence Interval for Mean	Lower Bound	77.03916647	
		Upper Bound	86.91095698	
	5% Trimmed Mean		82.42371579	
	Median		82.07315279	
	Variance		149.336	
	Std. Deviation		12.22032294	
	Minimum		55.98778999	
	Maximum		98.92473118	
	Range		42.93694120	
	Interquartile Range		17.88372160	
	Skewness		-.543	.456
	Kurtosis		-.618	.887

Model Summary[b]

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	Change Statistics					Durbin-Watson
					R Square Change	F Change	df1	df2	Sig. F Change	
1	.401[a]	.161	.088	233.5659410	.161	2.208	2	23	.133	2.168

a. Predictors: (Constant), Avg Torso % acc phase, Avg Pelvis % acc phase

b. Dependent Variable: HandVel_BC

Correlations

		HandVel_BC	Avg Pelvis % acc phase	Avg Torso % acc phase
Pearson Correlation	HandVel_BC	1.000	-.379	-.316
	Avg Pelvis % acc phase	-.379	1.000	.538
	Avg Torso % acc phase	-.316	.538	1.000
Sig. (1-tailed)	HandVel_BC	.	.028	.058
	Avg Pelvis % acc phase	.028	.	.002
	Avg Torso % acc phase	.058	.002	.
N	HandVel_BC	26	26	26
	Avg Pelvis % acc phase	26	26	26
	Avg Torso % acc phase	26	26	26

Variables Entered/Removed[a]

Model	Variables Entered	Variables Removed	Method
1	Avg Torso % acc phase, Avg Pelvis % acc phase[b]	.	Enter

a. Dependent Variable: HandVel_BC

b. All requested variables entered.

ANOVA[a]

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	240859.815	2	120429.907	2.208	.133[b]
	Residual	1254720.122	23	54553.049		
	Total	1495579.937	25			

a. Dependent Variable: HandVel_BC

b. Predictors: (Constant), Avg Torso % acc phase, Avg Pelvis % acc phase

Coefficients[a]

Model		Unstandardized Coefficients		Standardized Coefficients	t	Sig.	95.0% Confidence Interval for B		Correlations			Collinearity Statistics	
		B	Std. Error	Beta			Lower Bound	Upper Bound	Zero-order	Partial	Part	Tolerance	VIF
1	(Constant)	1790.231	317.842		5.632	.000	1132.725	2447.737					
	Avg Pelvis % acc phase	-5.044	3.891	-.294	-1.297	.208	-13.093	3.004	-.379	-.261	-.248	.710	1.408
	Avg Torso % acc phase	-3.155	4.536	-.158	-.696	.494	-12.538	6.228	-.316	-.144	-.133	.710	1.408

a. Dependent Variable: HandVel_BC