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Greater Breast Support Reduces Common Biomechanical Risk Factors Associated with Anterior Cruciate Ligament Injury

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GREATER BREAST SUPPORT REDUCES COMMON BIOMECHANICAL RISK FACTORS ASSOCIATED WITH ANTERIOR CRUCIATE LIGAMENT INJURY

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PREFACE

The findings from this thesis will be submitted for publication to *Human Movement Science* and the format of this journal is presented in the Abstract and Chapter III. The formatting of this portion of the document is therefore reflective of the submission requirements of this *Human Movement Science*.

ABSTRACT

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To examine the effects of breast support on trunk and knee joint biomechanics in female collegiate athletes during a double-limb landing task.

Methods: Fourteen female athletes completed five landing in three different sports bra conditions: no support, low support, and high support. 3D kinematics and ground reaction forces were recorded simultaneously. Visual 3D was used to calculate trunk and knee joint angles and moments. Custom software determined discrete trunk and knee joint variables. A repeated measures analysis of covariance with post-hoc t-tests compared landing biomechanics by condition.

Results: Greater breast support was associated with reductions in knee flexion and knee valgus angles as well as increases in knee varus moments. Greater breast support was associated with greater trunk flexion angles at initial contact and greater peak trunk flexion angles.

Conclusions: Lower levels of breast support are associated with knee joint and trunk biomechanical profiles suggested to increase ACL injury risk.

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ABBREVIATIONS

ACL	Anterior Cruciate Ligament
MCL	Medial Collateral Ligament
PCL	Posterior Cruciate Ligament
LCL	Lateral Collateral Ligament
GRF	Ground Reaction Force
IC	Initial Contact
INI	Instant 100 ms After Initial Contact
PRE	Prior to Experimental Testing
CON	Control Support Condition
LOW	Low Support Condition
HIGH	High Support Condition
T6	Sixth Thoracic Vertebra
T12	Twelfth Thoracic Vertebra

CHAPTER I: INTRODUCTION

Female participation in both high school and collegiate sports has increased dramatically, since the early 1970s. The most common sports females participate in are soccer and basketball and thus, these two sports also see the highest incidence of anterior cruciate ligament (ACL) injuries (Joseph, et al., 2013; NFHS, 2016). Specifically, female athletes are up to eight times more likely to experience an ACL injury compared to their male counterparts in the same sport (Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001). Furthermore, female athletes are still at a greater risk of ACL injuries than males when controlling for additional risk factors such as age and level of play (Joseph, et al., 2013; Renstrom, et al., 2008). Common movements athletes may experience in multidirectional sports are landing and cutting. These two specific movements can result in a combination of valgus loading, external tibial rotation, and knee hyperextension with internal tibial rotation, which places great stress on the ACL, possibly leading to injury (Whiting & Zernicke, 1998). ACL injuries can be costly as well as detrimental to an athlete's career and life long physical wellbeing (Fleming, Hulstyn, Oksendahl, & Fadale, 2005; Joseph, et al., 2013). Therefore, there is an increased need to understand the factors underlying ACL injuries in female athletes.

A plethora of research has focused on biomechanical differences between the sexes that result in greater ACL injuries in female athletes. Female biomechanical differences during a 60-centimetrer double-landing task includes greater peak ankle dorsiflexion, peak foot pronation, and peak knee valgus angles (Kernozek, Torry, H, Cowley, & Tanner, 2005). Similarly, even double-limb landing from 40-centimeter still results in females exhibiting significantly greater peak knee valgus angles as well as peak vertical GRFs, than males (Pappas, Hagins, Sheikhzadeh, Nordin, & Rose, 2007). Female biomechanics are also significantly different than males during cutting tasks including greater peak knee abduction angles and greater peak ankle eversion angles (Ford, Myer, Toms, & Hewett, 2005). Additionally, during cutting, females also experience greater trunk side flexion range of motion, greater peak knee valgus, and greater knee and hip flexion range of motion (Pappas, Shiyko, Ford, Myer, & Hewett, 2016). These previously researched biomechanical differences place females at a greater risk of an ACL injury.

Trunk biomechanics is another factor that can contribute to ACL injury risk. Position of the trunk during landing and cutting tasks can cause significant changes in lower extremity biomechanics. Landing with greater trunk flexion results in increased peak trunk, hip, and knee flexion angles as well as decreases in peak vertical GRFs (Blackburn & Padua, 2008, 2009). Additionally, landing with trunk flexion strategy, as opposed to a trunk extension strategy, decreases average and peak knee anterior shear forces. This decrease in knee anterior shear forces is a result of increased hamstring muscle force, which limits the quadriceps from anterior translation of the tibia relative to the femur (Kulas, Hortobagyi, & Devita, 2010). Similarly, landing with a moderate amount of trunk lean results in decreases in ACL forces and strains, as well as increases in hamstring muscle force (Kulas, Hortobagyi, & DeVita, 2012). While trunk biomechanics and lower extremity differences between sexes have been linked to an increase risk of ACL injury, few known studies have determined if female breast size and sports bra support affect trunk and lower extremity biomechanics, especially during landing and cutting tasks. This would further explain why females experience higher rates of ACL injuries than males.

Female breasts have limited intrinsic support, which results in large breast displacement, during sports activities (Gaskin, Peoples, & McGhee, 2020; McGhee & Steele, 2020; J. Scurr, White, & Hedger, 2009; J. Scurr, White, Milligan, Risius, & Hedger, 2011; J. C. Scurr, White, & Hedger, 2011). To control for this limited intrinsic support, females use extrinsic support, typically in the form of sports bras, to decrease exercise induced breast pain (Brisbine, Steele, Phillips, & McGhee, 2020; McGhee & Steele, 2020; Risius, Milligan, Berns, Brown, & Scurr, 2017). Different types of sports bras provide different levels of support. Females, depending on breast size, can experience displacements as high as 20 centimeters while running with no support. With increasing support, there are significant reductions in displacement (J. C. Scurr, et al., 2011). Vertical displacement, as opposed to anteroposterior and mediolateral displacement, accounts for a majority of breast displacement (J. Scurr, et al., 2009; J. C. Scurr, et al., 2011). Breast support has been found to create significant changes in running biomechanics including peak pelvis rotation, pelvis range of motion, vertical trunk oscillation, peak trunk rotation, and trunk range of motion as well as peak torso yaw, torso pitch and yaw, and torso range of motion in women with a D- and D+ -cup breast size. (Milligan, Mills, Corbett, & Scurr, 2015; Risius, et al., 2017). Additionally, breast support has also been found to create significant changes in breast-body time lag during running (Risius, et al., 2017). However, the majority of previous research that has investigated the biomechanical effects of different levels of breast support worn during exercise has been limited to running and has focused on upper extremity and trunk biomechanics, rather than lower limb biomechanics. Further, this research has primarily investigated large breasted females with a breast size of a D-cup.

In conclusion, breast support has been shown to cause significant changes in trunk biomechanics in females with larger breast sizes that may also result in changes in lower extremity biomechanics. These biomechanical changes are suggested to have a significant impact on the amount of ACL stress that occurs at the knee. Therefore, the purpose of this study is to determine the effect of breast size and sports bra support on ACL stress during a double-limb landing task.

Specific Aims:

Aim #1: To determine the effect of breast support level on trunk angles during a double-limb landing task.

Hypothesis #1: We hypothesized that increasing levels of breast support will be associated with greater trunk flexion angles at contact and greater peak trunk flexion angles during the landing task.

Aim #2: To evaluate changes in sagittal and frontal plane knee joint angles and moments in response to increasing levels of breast support during a double-limb landing task.

Hypothesis #2a: We hypothesized that increasing levels of breast support would result in smaller peak knee flexion and peak knee valgus angles while knee joint angles at initial contact would remain unchanged.

Hypothesis #2b: We hypothesized that peak sagittal and frontal plane joint moments would be reduced in response to increased levels of breast support.

CHAPTER II: LITERATURE REVIEW

Female Athletes in Sports

Since 1971, female participation in high school sports have increased twelve-fold. From 1971-72, total female participation in all high school sports was less than 300,000. However, from 2018-19, total female participation in high school sports was greater than 3.4 million (NFHS, 2016). In a 2018-19 National Federation of State High School Associations (NFHS) athletics participation summary survey, volleyball, basketball, and soccer made up the three of the top five most popular girl's programs (NFHS, 2016). Volleyball participation included more than 452,000 female participants, basketball participation included more than 399,000 female participations, and soccer participation included more than 394,000 participants (NFHS, 2016). Since 1983, female participation in Division 1 (D1) collegiate sports have increased three-fold. In 1983, total female participation in D1 sports was less than 27,000. However, in 2019, total female participation in D1 collegiate sports was greater than 85,000 (NCAA, 2019). In a 2018-19 National Collegiate Athletic Association participation survey, total female participation in all three divisions included more than 218,000 athletes (NCAA, 2019).

Importance of Reducing ACL Injuries

Knee injuries account for 60 percent of high school sport related surgeries, and anterior cruciate ligament (ACL) injuries account for more than 50 percent of all knee injuries (Joseph, et al., 2013). An estimated 70 percent of ACL injuries occur during sport related activities and athletes are 7 times more likely to sustain an ACL injury in competition as opposed to practice (Joseph, et al., 2013; Malinzak, et al., 2001). In 1982, an estimated 50,000 ACL injuries occurred in the United States (Malinzak, et al., 2001). Currently, an estimated 250,000 ACL injuries occur in the Unites States, and almost half of these injuries result in reconstruction surgery in the United States per year (Pappas, et al., 2016). Annually, ACL injuries cost nearly $1 billion in the United States (Joseph, et al., 2013). A study of ACL injuries in nine different high school sports across five years determined that 79.6 percent of ACL injuries result in surgery, 46.4 percent of ACL injuries result in season disqualification and 15.4 percent require a three week or longer recovery period (Joseph, et al., 2013). Not only do a majority of ACL injuries result in surgical intervention as well as long-term rehabilitation, but it can also lead to season disqualification and/or the end of a competitive career. Furthermore, ACL injuries increase the risk of early knee osteoarthritis (OA) up to 14 years after injury (Fleming, et al., 2005). OA is the most common type of arthritis (Nordin & Frankel, 2012). It is a progressive disease that results in a gradual softening and disintegrating of articular cartilage (Drake, Drake, & Gray, 2008). OA is most common in several different joints, including the knee joint (Nordin & Frankel, 2012). Knee OA is known as one of the most disabling medical conditions in the world as it can severely affect quality of life well as physical activity level, due to symptoms such as swelling, pain, joint stiffness and instability, and limited range of motion (Murphy, et al., 2008). Decreasing level of physical activity can also result in negative health effects such as increased risk of obesity, diabetes, and heart disease (Bindawas & Vennu, 2015). The risk of knee OA affects an estimated 1 in 2 people, and this risk continues to increase for individuals with a prior history of knee injury (Murphy, et al., 2008).

Anterior Cruciate Ligament Anatomy and Injury

Ligaments are a crucial structure to provide both strength and integrity to the knee (Whiting & Zernicke, 1998). There are four primary ligaments surrounding the knee: lateral collateral, medial collateral, posterior cruciate, and anterior cruciate. The two collateral ligaments are located on the lateral and medial aspect of the knee and are named the lateral collateral ligament (LCL) and the medial collateral ligament (MCL) respectively. The LCL originates on the lateral epicondyle and inserts on the lateral head of the fibula. It is considered to have a more cordlike structure than the MCL, which is larger and forms a more broad, triangular band. There are two parts to the MCL, a deeper, posterior section and a superficial, anterior section. The posterior section has shorter fibers and originates on the medial femoral epicondyle and inserts on the tibial plateau as well as to the joint capsule and medial meniscus. The anterior section has longer fibers and originates on the medial femoral epicondyle and inserts on the medial surface of the shaft of the tibia (Nordin & Frankel, 2012). The two cruciate ligaments are located between the femur and tibia on the anterior and posterior aspect of the knee and are named the anterior cruciate ligament (ACL) and posterior cruciate ligament (PCL) respectively. The ACL originates on the posterior femur on the medial surface of the lateral condyle and inserts on the anterior tibia and slightly lateral to the intercondylar eminence. The PCL originates anteriorly on the lateral surface of the medial femoral condyle and inserts posteriorly on the proximal tibia (Nordin & Frankel, 2012). The PCL has a larger cross-sectional area and results in the ACL being the weaker of the cruciate ligaments (Nordin & Frankel, 2012; Whiting & Zernicke, 1998). Therefore, the ACL is more prone to injuries. The main function of the ACL is to minimize anterior translation of the tibia relative to the femur, while secondary functions of the ACL are to minimize tibiofemoral abduction angle (valgus), adduction angle (varus), and external and internal tibial rotation (Whiting & Zernicke, 1998). Most often, ACL injuries occur due to a combination of valgus loading, external tibial rotation, and knee hyperextension with internal tibial rotation. Typically, valgus loading occurs during specific tasks such as landing and cutting, and with a combination of external tibial rotation, the ACL can experience high magnitude of stress (Whiting & Zernicke, 1998).

ACL injuries occur when the ligament is unable to resist the force applied to it and reaches the failure point. The stress strain curve describes the relationship between stress and strain and can be used to understand how injuries occur (D.W. Powell, 2020). Stress is defined as the force applied per unit area, and strain is defined as the relative change in size in a prescribed direction (Nigg, MacIntosh, & Mester, 2000). As stress begins to increase, strain also increases. Ligaments undergo stress when a force is applied to it, and in order to compensate, the ligament deforms as a result of the force applied. The deformation is temporary, and the ligament is able to return to its original length once the force is removed. This is known as the elastic behavior of the system. However, only so much force can be applied to the ligament, and only so much deformation can occur. It reaches a point that the force applied to the ligament exceeds this elastic behavior, which is called the yield point. Past the yield point, the ligament begins to experience plastic behavior, in which the ligament can no longer return to its original length after a force is applied and instead, results in permeant deformation (D.W. Powell, 2020). This can lead to minor injuries such as sprains. However, if force is continually applied to the ligament once it surpasses the yield point, it can reach the failure point. At this point, the ligament is no longer able to able to withstand any force that is applied to it and fails. This can lead to major injuries, such as ruptures in ligaments and tendons, or fractures in bones.

Timing of ACL Injuries

ACL rupture occurs when the stress experienced by the ACL exceeds the failure load of the tissue (Nordin & Frankel, 2012). Given the viscoelastic properties of biological materials, the rate of load application alters the tissue's stress-strain response to the applied load (Neumann textbook). As such, the duration between initial contact and the instant at which ACL is ruptured becomes an important factor in evaluating the window of analysis for studies investigating biomechanical factors influencing ACL injury. An observational inspection of thirty-nine individual basketball videos revealed that estimated time of ACL ruptures occurred between 17 ms and 50 ms following initial contact (Krosshaug, et al., 2007). Similarly in another video analysis of ten female team handball and basketball athletes, ACL injury occurred approximately 40 ms after initial contact during either a cutting or single-limb landing maneuver (Koga, et al., 2010). In another study during stimulated single-limb jump landing, peak ACL relative strain values occurred between 30 ms and 40 ms (Withrow, Huston, Wojtys, & Ashton-Miller, 2006). In a more recent study, Bates et al. (2020) used thirty-five lower-extremity cadaveric specimens to measure exact timing of peak ACL strain relative to initial contact during a simulation imitating a drop-landing from 31-cm. Bates et al. (2020) also applied different loads to the lower extremity specimens to create a subthreshold protocol. These loads placed the specimen at baseline, moderate, high, and very high-risk profiles of injury, based upon knee abduction moment, anterior tibial shear, and internal tibial rotation moments. There was no significance of risk profiles to the timing of peak strain following initial contact. The time to peak ACL strain occurred from 48 to 61 ms after initial contact, and the mean peak ACL strain prior to ACL failure occurred at 53 ms after initial contact. This study suggests that non-contact ACL injuries occur between 0 and 61 ms after initial contact (Bates, Schilaty, Ueno, & Hewett, 2020).

Female Athletes and ACL Injury Rate

One major risk factor for ACL injuries is sex. Female athletes are 4 to 6 times more likely to experience a major knee injury (Joseph, et al., 2013). Additionally, female athletes are up to 8 times more likely to experience an ACL injury compared to their male counterparts in the same sport (Malinzak, et al., 2001). Rate of ACL injuries are also dependent upon type of sport. In a High School Sports-Related Injury Surveillance Study from 2007-12, a total of 91,002 high school female athletes experienced an ACL injury across nine different sports (Joseph, et al., 2013). Across those 91,002 ACL injuries, 53.2 percent of ACL injuries occurred during soccer, 26.5 percent occurred during basketball, and 8.8 percent occurred during volleyball. Girls playing soccer were twice as likely to sustain an ACL injury than any other sport, while girls playing soccer or basketball were four times as likely to sustain and ACL injury than softball or volleyball (Joseph, et al., 2013). Age also plays a role in the chances of experiencing an ACL injury. Young females ages 15 to 19 are more at risk for ACL reconstruction surgeries than males of the same age. This age group also has the most number of cases of ACL reconstruction surgeries regardless of sex (Renstrom, et al., 2008). Level of play, which is also dependent upon age, is also a risk factor for ACL injuries. Compared to males, females below collegiate level are four times as likely to suffer an ACL injury while playing basketball, whereas females in the collegiate level are only three times as likely to suffer an ACL injury while playing basketball. However, compared to males, females regardless of collegiate level are still twice as likely to suffer an ACL injury while playing soccer (Renstrom, et al., 2008). Type of contact leading to the injury is also a risk factor. A non-contact mechanism of ACL injury is the most common in sports (Renstrom, et al., 2008). Up to 80 percentage of ACL injuries occur due to non-contact mechanisms as opposed to contact mechanisms (Boden, Dean, Feagin, & Garrett, 2000)

Sex Differences in Lower Extremity Biomechanics

This increase in ACL injury rate among female athletes can be attributed to lower extremity biomechanical differences between males and females. During double limb drop landings from sixty centimeters, females exhibit greater peak ankle dorsiflexion, peak foot pronation, and peak knee valgus angles than males. Females also exhibit greater peak vertical ground reaction forces (GRFs) (Kernozek, et al., 2005). Even during a double limb drop landing from forty centimeters, females still exhibit greater peak knee valgus angles and peak vertical GRFs than males. When increasing the demand from bilateral to a unilateral drop landing from forty centimeters, peak knee valgus angles continued to increase. This change in demand places females at an even greater risk of injury as it brings them closer to the threshold of injury with increased biomechanical changes in valgus (Pappas, et al., 2007). Furthermore, during a simulated *in vitro* jump-landing study, if the knee is experiences compressive loading while in both valgus and flexion, rather than just flexion, peak ACL relative strain is 30 percent larger, further increasing the risk of ACL injury (Withrow, et al., 2006). During unanticipated side-step cutting, females exhibit greater knee abduction angles at initial contact as well as greater peak ankle eversion angles during stance phase compared to males. Greater knee valgus as well as ankle eversion may be possible explanations for increased ACL injuries in females. Greater knee valgus prior to the cutting maneuver may be due to altered muscular control of the hip and knee which can place greater load upon the knee and ACL. Greater ankle eversion may be attributed to greater tibial internal rotation (Ford, et al., 2005). Furthermore, during a study of 721 female basketball, volleyball, and soccer athletes in a side cutting task, sixty percent are profiled as having high risk biomechanical deficits that puts them at a greater risk of an ACL injury. Of the sixty percent, fourteen percent had a ligament dominance deficit. These individuals had greater knee valgus range of motion and greater peak knee valgus moment, which may put these individuals at the highest risk for ACL injury. Another twenty-two percent, of the sixty percent, had a trunk-leg-ligament deficit. These individuals had greater trunk side flexion range of motion, greater peak knee valgus, and greater knee and hip flexion range of motion as well as ligament dominance deficits. This suggests that trunk biomechanics may also be a risk factor for ACL injuries (Pappas, et al., 2016).

Role of Trunk Biomechanics

Trunk biomechanics may also play a role in female ACL injury rates. During a sixty-centimeter vertical drop-landing task, forty individuals that landed with a greater peak trunk flexion angle were more likely to exhibit significant increases in peak hip flexion angles and peak knee flexion angles. This increase in flexion angles of the trunk, hip, and knee may possibly decrease the risk of an ACL injury (Blackburn & Padua, 2008). Similarly, during a sixty-centimeter double-limb vertical drop-landing task, individuals landing in a more flexed landing position, versus preferred landing position, exhibit increases in peak trunk, hip, and knee flexion angles, as well as a decrease in quadricep activity and decrease in peak vertical GRFs. This decrease in GRFs with greater trunk flexion may be indicative of decreases in ACL loads (Blackburn & Padua, 2009). During a forty-five-centimeter single-leg landing task, individuals landing with more of a forward trunk lean strategy are more likely to exhibit greater plantar-flexor flexion moments, less knee-extensor moment and greater hip-extensor moments. This suggests that landing with a forward trunk lean results in less quadricep muscle force and greater hamstring muscle force at the knee (Shimokochi, Yong Lee, Shultz, & Schmitz, 2009). During a forty-five-centimeter double-limb landing task with increased trunk load, individuals that exhibit a more trunk extensor dominant landing strategy, versus trunk flexor dominant landing strategy, had significant increases in peak and average knee anterior shear forces. The increase in knee anterior shear forces was due in part to a significant decrease in hamstring muscle force, as opposed to an increase in hamstring muscle force which would limit knee anterior shear force (Kulas, et al., 2010). Even during a single-leg squat task, individuals that exhibit a moderate amount of trunk lean, versus minimal amount of forward trunk lean, experience significantly lower peak and average ACL forces and strains. A moderate amount of trunk lean during the task was also found to have higher hamstring muscle forces (Kulas, et al., 2012). Again, limiting the amount of knee anterior shear forces and decreasing the risk of ACL injury.

Female Breast Anatomy and Research

Breasts are a combination of fatty adipose and glandular tissues located on the anterior portion of the trunk. Muscles such as the pectoralis major and minor, serratus anterior, and rectus abdominus lie underneath the breasts (Drake, et al., 2008). Because of this, breast motion is not actively controlled by muscles, making breasts a passive tissue that is only supported by connective tissue. Females with a D-cup bra size can have breasts that weight an estimated 920 grams (~2 pounds) (Turner & Dujon, 2005). Breast mass and breast extension, vertical nipple displacement, have a significant and positive correlation during running in a no support condition (J. Scurr, et al., 2011). Due to a lack of intrinsic support and dependent on mass and size, breasts can have large ranges of motion. During walking with no bra support, women with a D-cup breast size experience vertical displacement as high as 4.2 ± 1.0 centimeters. However, there is also a significant effect of speed on breast displacement. During running with no bra support, women with a D-cup breast size experience displacement as high as 15.2 ± 4.2 centimeters (J. C.

Scurr, et al., 2011). During both walking and running, vertical displacement accounts for an estimated 52 percent of total breast displacement (J. Scurr, et al., 2009; J. C. Scurr, et al., 2011). Breast displacement also occurs in the mediolateral and anteroposterior direction, with breast moving in a figure-eight trajectory (J. Scurr, et al., 2009). Breasts also experience breast-body time lag, which is determined as the time between sternal notch of the trunk and nipple reaching max superior-inferior displacement. There is a significant reduction in high time lag during flight phase of the gait cycle as sports bra support increased (Risius, et al., 2017).

To counteract the lack of intrinsic support and large ranges of motion, females often require the application of extrinsic support, typically in the form of sports bras. However, even with the use of sports bras, female athletes often experience breast discomfort and pain. In a survey of 540 elite female athletes, 44 percent of the participants had reported experiencing exercise-induced breast pain. Of those athletes, 37 percent reported their breast pain to interfere with their ability to train, and 32 percent claimed their breast pain was severe enough to negatively effect performance (Brisbine, et al., 2020). In another survey of 504 elite female athletes, 32 percent had experienced a direct contact induced breast injury. Of those athletes injured, 21 percent, approximately one in five, perceived their injury to negatively affect performance, and only 10 percent reported their injury to a coach or medical professional. Furthermore, 9 percent reported modifying their movement in order to limit and prevent injury (Brisbine, Steele, Phillips, & McGhee, 2019). Regardless of whether the discomfort or pain is exercise or contact induced, changes in movement patterns seeking to limit and/or prevent breast discomfort, pain, or injury may place athletes at an exaggerated risk of lower extremity injury by altering lower extremity biomechanics. There are also differences in levels of breast support provided by sports bras. Changes in support from low support to high support have significant effect on willingness to exercise as well as breast comfort (non-painful vs painful) (Risius, et al., 2017). As vertical breast extension increased there was a significant increase in breast pain (J. Scurr, et al., 2011). Furthermore, changes in support can also affect lower and upper extremity biomechanics. During treadmill running, there were significant changes in peak pelvis rotation, pelvis range of motion, vertical trunk oscillation, peak trunk rotation, and trunk range of motion between the no bra, low support, and high support bra conditions (Risius, et al., 2017). Also during treadmill running there were significantly greater differences in peak clockwise torso yaw, peak pelvic right obliquity, peak pelvic anti-clockwise rotation as well as significantly greater differences in range of motion in torso pitch and yaw, pelvis rotation, and upper arm extension from the low support to high support sports bra (Milligan, et al., 2015).

While research has determined significant changes in running biomechanics, few studies have focused on multidirectional tasks and athletes would often experience, such as landing and cutting. Previous research has determined that breast support has a significant effect on trunk biomechanics during running, and trunk biomechanics has a significant effect on lower extremity biomechanics ultimately creating increased ACL stress during landing. Therefore, the purpose of this study is to determine if changes in breast support significantly effects ACL stress during a double-limb landing task.

CHAPTER III: MANUSCRIPT

Greater breast support reduces common biomechanical risk factors associated with anterior cruciate ligament injury

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Manuscript in preparation for Human Movement Science

ABSTRACT

To examine the effects of breast support on trunk and knee joint biomechanics in female collegiate athletes during a double-limb landing task.

Methods: Fourteen female collegiate athletes completed five trails of a double-limb landing task in each of three different sports bra conditions: no support (CON), low support (LOW), and high support (HIGH). Three-dimensional kinematics (250 Hz) and ground reaction forces (1250 Hz) were recorded simultaneously. Visual 3D was used to calculate trunk segment and knee joint angles and moments. Custom software (MATLAB 2021a) was used to determine discrete values for trunk segment and knee joint variables. A repeated measures analysis of covariance with post-hoc paired samples t-tests were used to evaluate the effect of breast support on landing biomechanics.

Results: Increasing levels of breast support were associated with reductions in peak knee flexion and peak knee valgus angles as well as increases in peak knee extension and varus moments. Increasing levels of breast support were associated with greater trunk flexion angles at initial contact and greater peak trunk flexion angles.

Conclusions: Lower levels of breast support are associated with knee joint and trunk biomechanical profiles suggested to increase ACL injury risk.

1. Introduction

Landing tasks in multidirectional sports result in a variety of lower extremity injuries for both males and females. However, female athletes have a greater prevalence of traumatic knee injury than males (Arendt & Dick, 1995; NFHS, 2016). Specifically, female athletes are up to eight times more likely to experience an anterior cruciate ligament (ACL) injury than their male counterparts in the same sport (Arendt & Dick, 1995; NFHS, 2016).

One of the reasons for this increase in ACL injury rate within the female population can be attributed to differences in lower extremity biomechanics between females and males. During a double limb drop landing task at forty centimeters, females exhibit greater peak knee valgus angles and peak vertical ground reaction forces (GRFs) compared to males (Pappas, et al., 2007). Furthermore, as the demand of the task increases to a double-limb landing from sixty centimeters, females still exhibit peak knee valgus angles and peak vertical GRFs, as well as the addition of greater peak ankle dorsiflexion, and peak foot pronation than males (Kernozek, et al., 2005). During unanticipated side-step cutting, females exhibit greater knee abduction angles at initial contact as well as greater peak ankle eversion angles during stance phase compared to males. Greater ankle eversion angle may be attributed to greater tibial internal rotation and greater knee valgus prior to cutting may place greater load upon both the knee and ACL, therefore, increasing the risk of injury (Ford, et al., 2005). These lower extremity biomechanical differences, during both landing and cutting, may help explain the increased incidence of ACL injuries for female athletes.

Another reason for this increase in ACL injury rate within the female population can be attributed to trunk biomechanics. During a sixty-centimeter vertical double-limb drop-landing task, individuals that landed with greater trunk flexion angles also exhibited greater hip and knee flexion angles (Blackburn & Padua, 2008, 2009). In addition, individuals landing with greater trunk, hip, and knee flexion angles also experienced a decrease in quadricep activity and increase in hamstring muscle force (Blackburn & Padua, 2009; Kulas, et al., 2010). This increase in hamstring muscle force is suggested to limit knee anterior shear force when landing with greater trunk flexion, as opposed to landing with greater trunk extension (Kulas, et al., 2010). Even when the demand of the task is decreased to a single-limb squat task, individuals with moderate amount of trunk lean, as opposed to minimal amount of trunk lean, still experienced higher hamstring muscle forces as well as significantly lower peak and mean ACL forces and strains (Kulas, et al., 2012). While trunk biomechanics may play a role in increased ACL stress and increased ACL injury, these studies do not compare trunk biomechanical differences between females and males.

Female breasts are a passive tissue that are only supported by connective tissue (Gaskin, et al., 2020). Because of this, breast have limited intrinsic support and often require the use of extrinsic support, typically in the form of sports bras especially during sports activities. Without the use of sports bras and sufficient breast support, females can experience increased levels of embarrassment, a decreased willingness to exercise, and increased levels of breast discomfort or pain (Risius, et al., 2017). By wearing sports bras and sufficient support, females can control for vertical, anteroposterior, and mediolateral breast displacement (J. Scurr, et al., 2009; J. C. Scurr, et al., 2011). Additionally, breast support has been found to create significant changes in running biomechanics including peak pelvis rotation, pelvis range of motion, vertical trunk oscillation, peak trunk rotation, and trunk range of motion as well as peak torso yaw, torso pitch and yaw, and torso range of motion. (Milligan, et al., 2015; Risius, et al., 2017). However, a majority of breast support in sports movement research has limited focus to upper extremity and trunk biomechanics specifically during running. Further, this research has primarily investigated large breasted females with a breast size of a D-cup.

While previous literature has determined that both lower extremity and trunk biomechanics can increase the risk of ACL injuries, limited research has determined if insufficient breast support can alter lower extremity and trunk biomechanics, possibly further increasing the risk of ACL injuries. Therefore, the purpose of this study is to determine the effect of sports bra support on trunk and knee joint biomechanics in female collegiate athletes during a single- and double-limb landing task.

2. Methods

2.1 Participants

An *a prior* power analysis was conducted based on findings from previous preliminary data. Using an effect size of 0.40, an alpha level of 0.05 and power (1-β) of 0.80, it was determined that a total sample size of 12 will provide sufficient statistical power for the study. However, a total of 14 participants were recruited due to two of the participants not completing the control condition. Inclusion criteria included (1) 18-25 years of age, (2) former (<2 years) or current female collegiate athlete, (3) self-reported bra size of B-D cup, (4) no history of prior breast surgeries (reduction or implants), (5) free from a recent history of musculoskeletal injuries (within the past six months), and (6) free from any history of ACL injuries.

2.2 Experimental Equipment

Participants were asked to wear spandex shorts and their preferred athletic shoes for testing. Ground reaction forces (GRFs) and three-dimensional kinematics were recorded simultaneously using a 10-camera motion capture system (250 Hz, Qualisys AB, Goteburg, Sweden) and two force platforms (1500 Hz, AMTI Inc., Watertown, MA, USA) embedded in the laboratory floor. The skeleton was modeled using 14 mm retro-reflective markers and included trunk and pelvis, as well as left and right thigh, shank, and foot segments. Retro-reflective markers were placed bilaterally on the participant's lower extremity and trunk in order to measure individual segment motion during the double-limb landing task. The pelvis, thigh, and shank were tracked using rigid clusters of four 14 mm retroreflective markers. The rearfoot was tracked using three individual 14 mm retroreflective markers placed over the superior, inferior and lateral calcaneus. The trunk was defined using individual markers placed over the left and right acromion processes and the right and left iliac crests. The trunk segment was tracked using individual markers placed on the skin over the superior sternum, the spinous process of the first thoracic vertebra (T1), the left and right transverse processes of the sixth thoracic vertebrae (T6), the left and right transverse processes of the twelfth thoracic vertebra (T12) and the anterior portion of the 10th osteochondral junction. Breast motion was tracked using individual markers placed over the superior sternum and left and right nipples. Anatomical markers were placed over the left and right iliac crest, and trochanters. Anatomical markers were also be placed over the medial and lateral femoral epicondyles, medial and lateral malleoli, and the first and fifth metatarsal heads. After a standing calibration, anatomical markers were removed leaving only the tracking markers on the breast, trunk, pelvis, thigh, shank, and rearfoot.

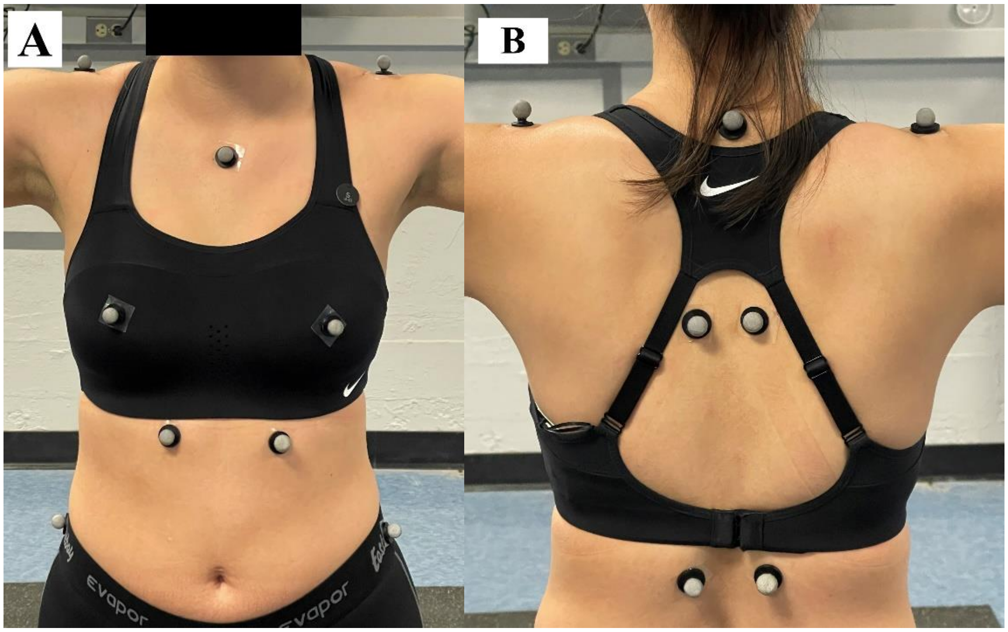

Figure 1: Image of retroreflective marker locations used to define and track the trunk segment and breasts.

2.3 Experimental Protocol

Participants visited the Musculoskeletal Analysis Laboratory (MAL) once for examination and testing. Participants were screened for inclusion criteria, completed a written Physical Activity Readiness Questionnaire (PAR-Q), and provided written informed consent. Each testing session occurred in the following order: (1) measurement of anthropometric variables (age, height (cm), weight (kg), bust size (cm), and rib cage size (cm)), (2) warm-up exercises, (3) placement of measurement sensors, and (4) completion of the dynamic testing protocol. Prior to dynamic testing, the participants were also be asked a series of questions regarding the date of their last menstruation and use of oral contraceptives, as well as breast discomfort prior to collection and following each sports bra condition (control, low support, and high support). Previous research has determined that time of menstruation can affect breast discomfort as well as lower extremity biomechanics and ACL injury risk (Balachandar, Marciniak, Wall, & Balachandar, 2017; Shultz, Kirk, Johnson, Sander, & Perrin, 2004; Wojtys, Huston, Boynton, Spindler, & Lindenfeld, 2002). The dynamic testing protocol consisted of two different dynamic movement tasks, including single- and double-limb landing, in three different support conditions, including low support, high support, and control (no support).

The protocol was completed in three different sports bra conditions: low support, high support, and control condition. The low support conditions (LOW) required the participant to wear a sports bra that is described by the manufacturer as having "light" support for low-impact workouts. The low support sports bras offered the breasts limited support. The low support sports bra was the Nike Indy (Nike Inc., Beaverton, OR, USA). The fabric of the sports bra includes a body and lining made of 88 percent recycled polyester and 12 percent spandex, center back mesh and bottom hem made of 81 percent nylon and 19 percent spandex, elastic made 84 to 85 percent nylon and 15 to 16 percent spandex, interlining made of 80 percent polyester and 20 percent spandex, pad top fabric and pad back fabric made of 100 percent polyester, and pad made of 100 percent polyurethane. The high support condition (HIGH) required the participant to wear a sports bra that is described by the manufacturer has having their "highest" level of support with a compressive feel for minimal bounce. The high support sports bras offered the breasts the maximum amount of support. The high support sports bra was the Nike Alpha (Nike Inc., Beaverton, OR, USA). The fabric of the sports bra includes a body and back lining insets made of 79 percent nylon and 21 percent spandex, mesh and mesh lining made of 81 percent nylon and 19 percent spandex, pad made of 100 polyurethane, and pad back fabric made of 100 percent polyester. The control condition (CON) required the participant to complete the protocol bare chested with no sports bra and no breast support. The control condition was optional for participants. The purpose of the control condition is to compare data from previous studies to the current study. Low and high support sports bra sizes was provided to the participant based on fitting described by the manufacturer. The order of the low and high support condition was randomized while the control condition was completed last.

The protocol consisted of a double-limb landing task in which required the participant to step-off of a 40-cm box and land bilaterally with one foot on each force platform. A successful trail was characterized by the participant landing from the box with simultaneous left and right ground contacts with one foot on each of the two force platforms. The participants completed a total of five successful trials. The participants were allowed to familiarize themselves with the landing task until they reported their comfort. The protocol was repeated in each support condition: LOW, HIGH, CON. Prior to the beginning of testing (PRE) and following the completion of each experimental condition, participants reported their level of breast discomfort using a visual analog scale with values ranging from 1 (very severe pain) to 5 (no pain) (Brisbine, et al., 2019, 2020).

2.4 Data Analysis

Landing data were analyzed from initial contact (IC) to an instant 100 milliseconds after contact (INI). The energy absorbed during this period has been associated with injury biomechanics (Norcross, Blackburn, Goerger, & Padua, 2010) and includes the period in which the ACL is most likely to experience significant injury (Bates, et al., 2020; Koga, et al., 2010; Krosshaug, et al., 2007). IC was determined as the instant at which vertical GRF exceeds a threshold of 20 N and remained above this threshold for a period greater than 0.10 s. Visual 3D (C-Motion Inc., Bethesda, MD, USA) was used to create a six degree-of-freedom kinematic model as well as filter kinematic and GRF data. Retroreflective marker trajectories and GRF data were filtered using a fourth-order, zero-lag Butterworth lowpass filter with cutoff frequencies of 10 Hz and 40 Hz, respectively (Smith, Paquette, Harry, Powell, & Weiss, 2020). Sagittal and frontal plane knee joint angles and moments as well as sagittal plane trunk segment angles were calculated using Visual3D. Custom software (MatLab 2021a, MathWorks, Natick, MA) was used to identify discrete data points for knee joint angles and moments as well as trunk segment angles.

2.5 Statistical Analysis

A 1 x 3 (task by support level) repeated measures analysis of covariance (ANCOVA) was conducted for each dependent kinematic and kinetic variable to determine the effect of breast support level on knee joint biomechanics and trunk kinematics when adjusted for breast size. In the presence of a significant interaction, post-hoc pairwise comparisons were performed to determine source of the significant interaction. A Holm-Bonferroni Correction was performed to adjust the level of significance for multiple comparisons (Holm, 1979). To conduct this correction, the p-values for post-hoc pairwise comparisons were placed in ascending order (from smallest to largest) and compared to the adjusted level of significance. As three paired samples t-tests were performed, significance for the first post-hoc comparison was set at $p < 0.017$ ($p < 0.05/3$) while significance for the second post-hoc comparison was set at $p < 0.025$ ($p < 0.05/2$) and significance for the third post-hoc comparison was set at $p < 0.05$ ($p < 0.05/1$). The sequential adjustment of the p-value is designed to reduce the risk of Type I error associated with multiple comparisons while also maintaining sufficient statistical power. To evaluate the effect of breast support on breast discomfort, a 1 x 4 repeated measures analysis of variance (ANOVA) was conducted using the Likert Scale breast discomfort data. Post-hoc paired samples t-tests with

Holm-Bonferroni correction were performed to determine the source of significance if a significant main effect of breast support was found. Significance for omnibus testing was set at $p < 0.05$ while post-hoc alpha levels were adjusted as previously described. All statistical comparisons were conducted using SPSS (IBM, Armonk, New York).

3. Results

3.1 Participants

Table 1 presents a summary of participant anthropometrics. Participants had an average age of 20.9 (± 1.7) years, average height of 170.1 (± 6.4) cm, average weight of 63.8 (± 6.9) kg, average bust circumference of 83.9 (± 2.4) cm, and average rib cage circumference of 74.3 (± 3.1) cm. No comparisons were made between individuals of different breast sizes.

Table 1. Participant anthropometric values including age, height, weight, bust circumference and rib cage circumference. Presented as mean ± SD.

Group	**N**	**Age (yrs)**	**Height (cm)**	**Weight (kg)**	**Bust (cm)**	**Rib Cage (cm)**
B-Cup	6	20.8 ± 1.6	172.7 ± 7.0	65.8 ± 8.1	83.3 ± 2.9	75.1 ± 3.9
C-Cup	3	21.0 ± 2.0	169.4 ± 4.7	60.0 ± 6.2	82.7 ± 2.8	74.2 ± 3.4
D-Cup	5	21.0 ± 2.1	165.6 ± 3.4	65.6 ± 6.4	85.3 ± 1.9	73.4 ± 3.0
Total	14	20.9 ± 1.7	170.1 ± 6.4	63.8 ± 6.9	83.9 ± 2.4	74.3 ± 3.1

3.2 Breast Displacement and Breast Discomfort

Increasing levels of breast support were associated with reductions in vertical breast motion (Table 2) during the double-limb landing task for the left ($F = 3.0$, $p < 0.001$) and right breasts ($F = 3.4$, $p < 0.001$). Breast displacement was greater in the CON compared to LOW ($p < 0.001$) and HIGH ($p < 0.001$) breast support conditions while breast displacement was also greater in the LOW compared to HIGH support conditions (p < 0.001) for both the right and left breasts.

Table 2. Average vertical breast displacement in the CON, LOW and HIGH support conditions during the double-limb landing task. Displacements are presented in cm. Presented as mean ± SD.

Breast	Control	Low	High	F-Value	P-Value
Left	4.4 ± 1.9	3.0 ± 1.0 [a]	2.4 ± 0.8 [a,b]	3.0	< **0.001**
Right	4.3 ± 2.0	3.1 ± 1.1 [a]	2.4 ± 1.0 [a,b]	3.4	**< 0.001**

Note: [a] – denotes significant difference compared to CON support condition; [b] – denotes significant difference compared to the LOW support condition.

Greater levels of breast support were associated with lower levels of reported breast discomfort (p < 0.001). Post-hoc comparisons revealed that breast discomfort was significantly greater following the CON compared to PRE condition (p < 0.001; PRE: 4.93 ± 0.18; CON: 3.77 ± 1.03) while no differences in breast discomfort were reported between the PRE and LOW (p = 0.062; LOW: 4.67 ± 0.56) or HIGH (p = 0.423; HIGH: 4.97 ± 0.13) conditions. Breast discomfort was significantly greater in the CON compared to LOW (p < 0.001) and HIGH conditions (p < 0.001). Further, the LOW support condition was associated with greater breast discomfort than the HIGH support condition (p = 0.043).

3.3 Knee Joint Angles

At IC, level of sports bra support was not associated with changes in knee flexion angles for either left (F = 1.25; p = 0.166) or right (F = 1.42; p = 0.146) limbs. Moreover, no effect of sports bra support was observed for knee joint valgus angles for either left (F = 0.60; p = 0.284) or right (F = 0.65; p = 0.284) limbs.

At INI, level of sports bra support was associated with altered knee joint flexion angles for both left (F = 3.40; p = 0.029) and right (F = 6.94; p= 0.008) limb (Table 3). Pairwise comparisons revealed no differences in knee flexion angles at INI between the CON and LOW conditions (p = 0.370) or the LOW and HIGH conditions (p = 0.167) while CON condition was associated with greater knee flexion angles than the HIGH condition (p = 0.039). For the right limb, knee flexion angles at INI were smaller in the LOW (p = 0.009) and HIGH conditions (p = 0.019) compared to the CON condition. However, no differences were observed between the LOW and HIGH conditions (p = 0.493).

Table 3. Knee joint kinematics and kinetics during the double-limb landing task. Presented as mean ± SD.

Limb	Condition	Flexion Angle at IC (°)	Valgus Angle at IC (°)	Flexion Angle at INI (°)	Valgus Angle at INI (°)
Left	Control	19.2 ± 4.4	-0.4 ± 3.9	68.8 ± 4.3	-5.1 ± 6.9
	Low	20.4 ± 6.9	0.5 ± 2.9	67.6 ± 7.0	-2.0 ± 6.1 [a]
	High	17.9 ± 4.7	0.7 ± 2.8	66.2 ± 4.7[a]	-0.2 ± 6.0 [a]
	p-value	0.166	0.284	**0.029**	**0.003**
Right	Control	19.4 ± 4.8	-0.7 ± 2.6	69.0 ± 4.9	-6.5 ± 5.3
	Low	18.3 ± 5.9	0.6 ± 3.2	66.3 ± 5.8[a]	-2.1 ± 6.7 [a]
	High	18.5 ± 5.4	0.9 ± 2.0	66.3 ± 5.5[a]	-0.4 ± 4.2 [a]
	p-value	0.146	0.284	**0.008**	**0.011**

Note: [a] – denotes significant difference compared to CON support condition; [b] – denotes significant difference compared to the LOW support condition.

Knee valgus angles at INI (Table 3) were altered by increasing levels of sports bra support for both left (F = 11.01; p = 0.003) and right (F = 11.0; p = 0.011) limb. The CON condition was associated with greater knee valgus angles than either the LOW (Right: p = 0.003; Left: p = 0.002) or HIGH conditions (Right: p = 0.003; Left: p = 0.001). No differences in knee valgus angles were observed between the LOW and HIGH conditions (Right: p = 0.362; Left: p = 0.355).

3.4 Knee Joint Moments

Level of sports bra supports had no effect on peak knee joint moments for left limb (F = 0.96; p = 0.216). However, level of sports bra support altered peak knee extension moments in the right-limb (F = 4.22; p = 0.026). However, pairwise comparisons revealed no differences between the individual sports bra support conditions (CON-LOW: p = 0.330; CON-HIGH: p = 0.144; LOW-HIGH: p = 0.321).

Peak varus moments were increased with greater levels of breast support during the double-limb landing task (Table 4). In the left limb, peak knee varus moments increased with increasing breast support (F = 3.91; p = 0.033). Post-hoc comparisons revealed greater knee varus moments in the LOW (p = 0.046; p = 0.013) and HIGH compared to CON conditions while the HIGH condition was also associated with greater peak knee varus moments than the LOW condition (p = 0.006). In the right limb, increasing levels of breast support were associated with greater peak knee varus moments (F = 4.00; p = 0.038). Pairwise comparisons revealed no differences between the CON and LOW conditions (p = 0.051) while the CON condition was associated with smaller peak knee varus moments than the HIGH support condition (p = 0.021) while the LOW support condition was associated with smaller peak knee varus moments than the HIGH support condition (p = 0.011).

Table 4. Peak knee extension and varus moments during the double-limb landing task. Presented as mean ± SD.

Limb	Condition	Peak Extension Moments (Nm/kg)	Peak Varus Moments (Nm/kg)
Left	Control	2.0 ± 0.3	-0.1 ± 0.2
	Low	2.1 ± 0.3	0.1 ± 0.2 [a]
	High	2.1 ± 0.4	1.1 ± 0.2 [a,b]
	p-value	0.216	**0.033**
Right	Control	2.1 ± 0.2	-0.3 ± 0.2
	Low	2.2 ± 0.2	0.2 ± 0.3
	High	2.2 ± 0.3	1.4 ± 0.2 [a,b]
	p-value	**0.026**	**0.038**

Note: [a] – denotes significant difference compared to CON support condition; [b] – denotes significant difference compared to the LOW support condition.

3.4 Trunk Angles

At initial contact, increasing levels of breast support were associated with greater trunk flexion (Table 5; F = 4.59; p = 0.024). Post-hoc analyses revealed no differences in trunk flexion angles between the CON and LOW support conditions (p = 0.142) while trunk flexion angles were greater in the HIGH compared to CON (p = 0.006) and LOW support conditions (p = 0.020). Similarly, increasing levels of breast support were associated with greater trunk flexion at INI (F = 15.3; p = 0.001). Pairwise comparisons demonstrated that trunk flexion angles were greater in the LOW (p = 0.001) and HIGH conditions (p = 0.001) compared to CON condition while trunk flexion angles were greater in the HIGH compared to LOW support conditions (p = 0.003).

Table 5. Average trunk angles at IC and at INI during the double-limb landing task. Angles are presented in degrees (°). Presented as mean ± SD.

Event	CON	LOW	HIGH	p-value
IC	-0.5 ± 2.5	0.1 ± 2.5	0.8 ± 2.4 [a,b]	**0.024**
INI	-1.4 ± 1.8	-0.2 ± 2.3 [a]	0.7 ± 2.4 [a,b]	**0.002**

Note: [a] – denotes significant difference compared to CON support condition; [b] – denotes significant difference compared to the LOW support condition.

4. Discussion

The purpose of the current study was to determine the effects of breast support level on knee joint and trunk biomechanics in female collegiate athletes during a double-limb landing task. The major findings of this study were that increasing breast support were associated with smaller peak knee flexion angles, greater peak knee extension moments, smaller peak knee valgus angles and greater peak knee varus moments. Further, greater breast support was also associated with greater trunk flexion at initial contact and greater peak trunk flexion during the first 100 ms following ground contact.

Knee joint flexion is a major contributor to load attenuation during a landing task. The current findings demonstrated that greater levels of breast support were associated with reduced peak knee flexion and knee flexion excursions. Moreover, no differences in peak knee extension moments were observed between the breast support conditions. In the absence of reductions in knee extension moments, the observed reductions in knee flexion excursions would be associated with greater knee joint stiffness and greater joint loading. Higher joint stiffness values have been previously associated with greater loading rates (Butler, Crowell, & Davis, 2003; D. W. Powell, Paquette, & Williams, 2017; Williams, Davis, Scholz, Hamill, & Buchanan, 2004) and greater peak vertical ground reaction forces (Arnwine & Powell, 2020; Butler, et al., 2003; D. W. Powell, et al., 2017; Williams, et al., 2004), each of which is associated with an increased risk of musculoskeletal injury (Whiting & Zernicke, 1998). Due to the short duration of the analysis window following initial contact, the biomechanics of the landing task were the result of the predicted mechanical requirement of the landing task and were not a feedback dominant motor pattern. Evidence has demonstrated that long latency reflex control (involving sensory processing by supraspinal structures) of lower limb muscle activation presents with latencies greater than 100 ms (Tsuda, Ishibashi, Okamura, & Toh, 2003; Tsuda, Okamura, Otsuka, Komatsu, & Tokuya, 2001). Therefore, we propose that the greater knee flexion excursions observed in the low breast support conditions (CON and LOW) were associated with a predictive motor control pattern selected to increase lower limb compliance and reduce accelerations of the passive breast tissue during the landing task. The assertion that lower extremity biomechanics were altered in response to breast motion and to limit discomfort is supported by the breast discomfort data which demonstrates that the low support conditions (CON and LOW) were associated with greater discomfort than the HIGH support condition as well as the pre-testing period (PRE). Therefore, we postulate that to reduce breast motion and breast discomfort in the low breast support conditions (CON and LOW), female athletes implemented a predictive movement pattern characterized by greater knee flexion and a more compliant lower extremity.

A consequence of greater knee flexion to increase limb compliance during landing is an expansion of the available knee joint range of motion in the frontal and transverse planes (Nordin & Frankel, 2012). The current data demonstrated that in the low breast support conditions (CON and LOW), peak knee valgus angles were greater than in the HIGH breast support condition. Greater knee valgus during a landing task has been associated with reduced neuromuscular control and a greater risk of ACL injury (Hewett, et al., 2005; Kernozek, et al., 2005; Pappas, et al., 2007). Though the differences in average peak knee valgus between breast support conditions were small (~3° - 4°), research has suggested that deviations in frontal plane knee joint angle as small as 2° can result in meaningful reductions in the external load required to rupture the ACL (Chaudhari & Andriacchi, 2006). The mechanical effect of greater knee valgus angles is supported by the current findings which demonstrated increased knee varus moments in the greater breast support conditions.

Trunk motion has been suggested to modify knee joint biomechanics during load attenuation tasks including single leg squatting and landing tasks (Blackburn & Padua, 2008, 2009; Kulas, et al., 2010, 2012). The current findings revealed that greater breast support was associated with greater initial and peak trunk flexion angles. It is postulated that the movement pattern adopted during the HIGH support sports bra condition represents a greater number of successful movement strategies available to the athlete by which to complete the landing task. These participants selected a movement pattern associated with reduced quadriceps and greater hamstring contributions to the landing task (Blackburn & Padua, 2008, 2009), decreasing the risk of ACL injury. Previous research has demonstrated that a moderate forward trunk lean was associated with lower peak ACL forces and strains as well as reduced knee anterior shear forces compared to a minimal forward trunk lean during single-leg squats and double limb landing (Kulas, et al., 2010, 2012). Functionally, the hamstrings muscle group acts to protect the ACL by limiting anterior translation of the tibia on the femur. Moreover, an intrinsic ACL-hamstrings reflex pathway exists to provide active, muscular support to an ACL that is experiencing strain (Tsuda, et al., 2001). The findings of the current study demonstrate that greater breast support was associated with increased trunk flexion angles at initial contact as well as peak trunk flexion angles. Therefore, these data suggest that the high breast support condition was associated with trunk biomechanics that are indicative of a lower risk of ACL injury compared to low breast support conditions (CON or LOW).

While the current study presents novel findings pertaining to the influence of breast support on knee joint and trunk biomechanics, the authors acknowledge several limitations of the current study. One limitation of the current study is the homogenous, “small breasted” nature of the population recruited for participation in this study. The participants in the current study self-identified their bra size as being between B- and D-cup size, though, previous research investigating breast pain and breast biomechanics has only included women with large breast sizes (Milligan, et al., 2015; Risius, et al., 2017; J. Scurr, et al., 2009; J. Scurr, et al., 2011; J. C. Scurr, et al., 2011). It is possible that the relatively small breast sizes of the women included in the current study resulted in limited effects of breast motion on lower extremity joint kinematics and kinetics. Evidence supporting this limitation includes the small differences in vertical breast motion in the LOW compared to HIGH support sports bra conditions, suggesting that the participants may not have had sufficient breast mass to find differences between the LOW and HIGH sports bra conditions. However, it is known that a vast majority of elite athletes have breast sizes within the range included in the current study. Brisbine et al (2020) reported a mean bra size of 540 national or international female athletes was 32B (US) while more than 75% of this sample of elite athletes was not considered “large breasted”. Therefore, we feel that the current sample of participants represents the physique of the “average” elite female athlete and better represents the effect of sports bra support on trunk and lower extremity biomechanics. A second limitation of the current study is the relatively small sample size. However, a power analysis based on preliminary data revealed a sample size of 12 participants would present sufficient power to find differences in knee joint biomechanics resulting from altered breast support. However, the small sample size may limit generalizations of the current findings to the population as a whole. Despite the small sample size, several variables in the current study were found to be significantly changed by greater breast support.

Conclusion

Greater breast support was associated with a multi-joint biomechanical adaptation characterized by reduced knee flexion, reduced knee valgus and greater trunk flexion angles. These movement profiles are associated with lower risks of traumatic knee injury suggesting that breast support is an important consideration for optimal sport performance. Future research should expand the current analysis to investigate altered contributions of the ankle and hip joint as well as the influence of tri-axial trunk motion on lower limb biomechanics during single limb tasks. Moreover, lower extremity stiffness and its interaction with trunk biomechanics should also be investigated.

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