The Influence of Self-Prescribed Knee Bracing on Walking in Healthy Adults

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THE INFLUENCE OF SELF-PRESCRIBED KNEE BRACING ON WALKING IN HEALTHY ADULTS

A thesis
presented in partial fulfillment of requirements
for the degree of Master of Science
in the Department of Health, Exercise Science
and Recreation Management
The University of Mississippi

By
Lauren A. Luginsland
May 2018
ABSTRACT

The knee is one of the most commonly loaded anatomical joints in the human body due to maintaining an upright posture. Increased joint loading can lead to various injuries which can either be treated surgically or non-surgically including immobilization by bracing. Numerous studies have looked at functional knee bracing post-operative surgery creating a gap within the literature analyzing non-operative functional knee braces within a healthy population. Commonly prescribed functional knee braces such as the Ossur Rebound® have been prescribed by physicians, but also available over-the-counter for self-prescription to assist with injury prevention, stabilization, and protection. Previous studies have looked at non-operative functional knee bracing on football players and runners; however, its effect on the average middle-age individual who use it for support and preventative measures has yet to be determined. Therefore, the purpose of this study was to evaluate how a self-prescribed, over-the-counter knee brace with no formal instruction impacts gait kinematics and kinetics. Nineteen healthy participants (10 females and 9 males) completed the study. A pre- and post-intervention design was used for the study. A 3D motion capture system was used to collect twenty (10 no-brace, 10 braced) initial gait passes for each participant which was duplicated upon the participants returning to the lab after wearing a knee brace for eight consecutive hours during a work day. A linear mixed model ANOVA with fixed factors of time (4) and condition (2) which employed unstructured covariance to examine possible differences in joint angles, joint moments, and time to peak joint moments analyzed kinematic and kinetic variables of interest. If main effect significance was found, a Bonferroni post-hoc adjustment compared simple main effects. The
findings from this study can be used as a series of recommendations for future research studies as well as future clinical recommendations. Knee braces, especially the Ossur Rebound, are devices that can serve a variety of conditions of the knee. While wearing a knee brace during walking, there is an increased internal load present on the ankle, knee, and hip joint. Therefore, self-prescribing is not appropriate for an individuals without a diagnosed pathology.
ACKNOWLEDGEMENTS

First and foremost, this thesis would not have been possible without the guidance and the help of many influential and important individuals. Their contributions led me to the preparation and completion of this study. My sincerest gratitude to each of my committee members: Dr. Martha Ann Bass, Dr. Chip Wade, and Dr. Dwight Waddell. My thesis would not have been possible if not for their instruction, advice, and guidance throughout this process. Thank you for pushing me through adverse times and never letting me become complacent.

A special thank you to Dr. Scott Breloff, my previous advisor, for introducing me to the topic of biomechanics. His constant support and consideration have been the key driver of my academic and research success. Thank you for always encouraging me to reach my true potential while also continuously assisting with my understanding of biomechanical and research design questions.

A big thank you to Ms. Dale, Mr. Jenkins, and Ms. Delene for the laughs and encouragement throughout my time at Ole Miss as well as printing an obscene amount of documents for me over the year.

I would like to extend my deepest appreciation, and most sincere thank you to all of my loved ones; parents, family, friends who have supported me across the hundreds of miles away from home. Thank you for supporting me through the ups and downs, all in order to chase a dream of becoming a PhD, one day. And finally, I will be forever grateful for the strong friendships and relationships developed over my two years in Oxford, MS.
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CHAPTER I

INTRODUCTION

The knee is continuously at an increased risk for injury since it is one of the most commonly loaded anatomical joints in the human body due to maintaining an upright posture (Pereira, Anand, Rajendran, & Wood, 2007). The National Survey of Ambulatory Surgery states the number of arthroscopic knee procedures has increased 49% between 1996 and 2006 with the largest increase observed in middle-aged patients. During 2006, nearly 984,607 of the 6.3 million (16%) orthopedic surgical procedures were arthroscopic procedures on the knee (Kim, Bosque, Meehan, Jamali, & Marder, 2011). Roughly 60% of the surveys completed by the American Academy of Orthopedic Surgeons recommend their patients wear a brace post-surgery for at least the first 6 weeks (Marx, Jones, Angel, Wickiewicz, & Warren, 2003). There is a large market for knee braces with more than 5 million sold in the United States in 2011. This demand is expected to yield $1.2 billion in revenues by 2018 according to Opportunities and Challenges in the US Market for Orthopedic Braces and Supports (2012).

Pain and injury at the knee can arise from a multitude of factors including but not limited to being overweight, poor muscle flexibility, lack of strength in the pelvis, trunk, and hip areas, participation in certain sports/positions, and previous injuries that went untreated or did not heal properly (Powers 2010; Mayo Clinic 2017). Excess weight causes added stress on the joint during activities of daily living in addition to poor muscle flexibility and poor strength which may influence the areas surrounding the knee, ultimately redirecting the exerted forces. Sport-specific movements such as pivoting, running, and jumping, increase the loading and
redistribution of forces increasing the risk for injury (Mayo Clinic 2017). In addition, anterior cruciate ligament (ACL) injuries are one of the most common knee injuries among various populations and age groups. The annual ACL injury rate has been estimated at approximately 200,000 with 100,000 ACL reconstruction surgeries completed (American Academy of Orthopedic Surgeons 2009) significantly affecting gait patterns. It has been shown that reconstructive surgery can cause an individual three months to achieve gait patterns comparable to before their injury occurring (Ferber, Osternig, Woollacott, Wasielewski, & Lee, 2002). Extensive evidence demonstrates that some individuals adjust their gait to compensate for instability, pain or a neuromuscular pathology (Andriacchi & Alexander, 2000). Those who have a knee injury can develop adaptive changes in their gait which may be analyzed through kinematic and kinetic measurements (Berchuck, Andriacchi, Bach, & Reider, 1990; Prodromos, Andriacchi, & Galante, 1985; Schipplein & Andriacchi, 1991).

Mechanical instability of the knee, defined as laxity and excessive joint motion due to structural damage of the supporting ligamentous tissues (Hughes & Rochester, 2008), is a principal pathology evaluated by clinicians. Previous literature (Huston & Wojtys, 1996; Rosene & Fogarty, 1999; Rozzi, Lephart, Gear, & Fu, 1999; Trimble, Bishop, Buckley, Fields, & Rozea, 2002) has shown that females tend to have more anterior knee laxity than males which reveals structural differences of the knee based on gender. Hormone receptors such as estrogen and progesterone are present in the ACL in humans (Liu et al., 1996) (Sciore, Smith, Frank, & Hart, 1997) (Shultz, Sander, Kirk, & Perrin, 2005). Therefore, a female’s knee laxity depends on their hormone changes during their menstrual cycle. A study compared the laxity differences of the knee of females across their menstrual cycle and compared the results to the laxity of males. The study showed that by day 5 of menses, days 3-5 close to ovulation, day 1-4 of the luteal phase
days 1, 2, 4, and 5 of the “late” luteal phases elicited an increase in knee laxity compared to males (Shultz et al., 2005). Previous literature has compared knee joint motions between males and females during athletic tasks such as running, side-cutting, and cross-cutting (Malinzak, Colby, Kirkendall, Yu, & Garrett, 2001). Results of the study showed that women had decreased knee flexion angles and greater knee valgus angles with a greater percent of quadriceps activation with a lesser hamstring activation during the stance phase (Malinzak et al., 2001).

Biological age has also been shown to play a role in structural knee changes demonstrated by an increase in age leading to defects such as thinning of the cartilage (Ding, Cicuttini, Scott, Cooley, & Jones, 2005). Devita and colleagues (2000) completed a study which showed that the elderly population had a decrease in joint torques compared to young adults at self-selected walking speeds (DeVita & Hortobagyi, 2000).

Knee injuries can be treated with nonsurgical or surgical treatment. Nonsurgical treatment includes immobilization such as bracing, physical therapy, and anti-inflammatory medications. Arthroscopic surgical procedures are most common and require a small incision. Post-surgery immobilization may be needed (American Academy of Orthopedic Surgeons, 2014). Roughly, 60% of the surveys completed by the American Academy of Orthopedic Surgeons recommend their patients to wear a brace post-surgery for at least the first 6 weeks (Marx, Jones, Angel, Wickiewicz, & Warren, 2003). Many studies have looked at the functional knee braces post-operative surgery (Brandsson, Faxen, Kartus, Eriksson, & Karlsson, 2001) (Birmingham et al., 2008; McDevitt et al., 2004) (Kartus et al., 1997) and literature lacks on the non-operative functional knee braces on the normal population.
The Effect of a Brace on the Knee Joint

Knee braces are common assistive devices used for a variety of applications such as injury prevention, stabilization, and protection (Butler, Evans, Rose, & Patrick, 1983). According to the American Academy of Orthopedic Surgeons, knee braces can be classified into prophylactic, rehabilitation, and functional (Surgeons, 1985). The prophylactic braces are used to reduce and help prevent the severity of damage while rehabilitative braces are used to allow for extra protection of injured knees which have been operative or non-operative, and functional knee braces are used to help provide stability for those who have “unstable” knees (Surgeons, 1985) (Montgomery & Koziris, 1989). Bracing is a common treatment for many individuals such as those who suffer from unicompartmental osteoarthritis of the knee. For these individuals, a valgus brace, a type of unloader functional brace, is a common treatment and has been seen to help with improvements of function as soon as the brace is worn (E. Draper et al., 2000).

Functional braces are used as a form of aid post-injury and are very popular within the athletic populations. These braces can be used to support the knee after the injury occurs and before a surgical procedure or in place of having surgery altogether. Functional braces may also be used post-surgery to assist with the limitation of excessive rotation (Surgeons, 1985). An epidemiology study collected data from registered high school athletic trainers on knee injuries amongst their teams. These injuries can place a physical as well as the economic burden on the athlete and their families. For example, it is estimated that ACL reconstruction surgery can cost over $5000. Knee injuries account for 15% of all high school sports injuries. Roughly, 6.5% of all knee injuries of these athletes occurred while wearing a knee brace. It was reported that 45.4% of athletes were wearing a rigid frame while 39.7% were wearing a knee sleeve. The
other athletes brace type were not report. Evidence for functional knee bracing as preventative physical measures are inconclusive (Swenson et al., 2013).

**Hypotheses**

**Gait Hypothesis**

**Specific Aim 1:**

To investigate the kinematics of gait following knee bracing to evaluate the conflict hypothesis in which individuals who wear knee braces as a protective measure could be exposing themselves to detrimental conditions conflicting with their goal of protection. The biomechanical comparison will be conducted between the normal gait condition [No Bracing (NB)] and after the 8 hours of bracing the comparison condition [Bracing (B)]. The primary mechanism of comparison will be normal gait NB and normal gait NB following an 8-hour brace utilization period.

**H01:** There will be no differences in lower extremity gait kinematics between the NB and B conditions during the stance phase of the gait cycle (0%-62% of gait cycle) of the dominant leg.

**H11:** There will be significant differences in gait kinematics between the NB and B conditions.

Kinematics of human gait in previous literature has been altered with the presence of a knee brace (E. Draper et al., 2000). Over a three month period, there was a significant functional improvement in their gait, both during the stance phase and swing phase. This may suggest that the brace was effective in relieving discomfort and relieving symptoms of antalgic gait (E. Draper et al., 2000). Draper and colleague’s measured gait symmetry on a treadmill instrumented with force plates to gather information on the steps of the individual. The system analyses are based on parameters including heel strike, toe off, stance phase, swing phase, peak forces, as
well as loading and unloading rates. The results were then used to create a ratio known as the symmetry indices (E. R. Draper, 2000).

DeVita and colleagues completed a study examining healthy individuals functional knee brace on running and found that there were a greater extensor torque and power at the hip with a reduced extensor torque and power at the knee compared to running without a brace (De Vita, Torry, Glover, & Speroni, 1996). DeVita and colleagues examined kinematics on an individual during running while the hypothesis aims at individuals walking.

In a previous study conducted with the knee brace, the mean active knee flexion without the knee brace was 122.9±6.17º while 112±4.37º was the mean active knee flexion wearing the Ossur® Hinged Polycentric Short Sleeve brace (Luginsland, 2016). Although it is unexpected that the individual would need to go into the complete active range of motion of knee flexion during their activities of daily living, it is important to notice that in extreme flexion there is a 10º decrease of range of motion while wearing the brace. This decrease must be causing compensation elsewhere on the kinematic chain, possibly in the hip or ankle.

**Specific Aim 2:**

To investigate the kinetics of gait following knee bracing to evaluate the conflict hypothesis in which individuals who wear knee braces as a protective measure could be exposing themselves to detrimental conditions conflicting with their goal of protection. The biomechanical comparison will be conducted between the normal gait condition [No Bracing (NB)] and after the 8 hours of bracing [Bracing (B)]. The primary mechanism of comparison will be normal gait NB and normal gait NB following an 8-hour brace utilization period.
**H₀₁**: There will be no differences in lower extremity gait kinematics between the NB and B conditions during the stance phase (0%-62%) of the gait cycle on the dominant leg.

**Hₐ₁**: There will be significant differences in gait kinetics between the NB and B conditions.

Previous studies have shown that functional knee bracing did not significantly affect the kinetics of the apparently healthy population (Lu, Lin, & Hsu, 2006) while some studies have shown a significant difference with the ground reaction forces on the braced leg creating an increase in the loading and push-off (Richards, Sanchez-Ballester, Jones, Darke, & Livingstone, 2005).

**Specific Aim 3:**
To investigate the effects of a knee brace on the contralateral lower extremity kinematics and kinetics.

**H₀₁**: There will be no differences in lower extremity gait kinematics between the NB and B conditions during the stance phase of the gait cycle (0%-62% of gait cycle) of the dominant leg.

**Hₐ₁**: There will be significant differences in the contralateral lower extremity kinematics and kinetics between the NB and B conditions.

Previous literature has studied the contralateral healthy knee compared to an injured knee by way of a torn ACL and found that the contralateral leg has a similar risk of an ACL tear (Wright et al., 2007). Other studies have determined that after twelve months there is no greater risk of injury on the contralateral ACL (Salmon, Russell, Musgrove, Pinczewski, & Refshauge, 2005).
Knee Brace Hypothesis

Specific Aim 4:
To investigate the effects of the subject’s normal pre-intervention gait passes (NB) compared to post-intervention gait passes (B) by comparing kinematics via a Motion Capture system and the kinetics via an AMTI force plate to evaluate the conflict hypothesis. Therefore, pre-intervention condition (B) will be compared to the post-intervention (B), and the pre-intervention (NB) will be compared to the post-intervention (NB).

H₀₁: There will be no differences in lower extremity gait kinematics between the NB and B conditions during the stance phase (0%-62%) of the gait cycle of the dominant leg.

Hₐ₁: There will be differences in the pre-intervention gait and post-intervention gait between the 8 hours of pre and post testing measures.

During the toe off phase of a normal gait cycle, the ankle is in plantar flexion. There is a plantar flexor acting internally at the ankle while there is an extensor moment acting on the knee. Once the knee is at 30° flexion, the hip will be a flexor moment (Levangie & Norkin, 2011). At this point of the gait cycle with the brace restricting the flexion of the knee joint, the degree of flexion would decrease causing an increase in the extensor moment to overcome the restriction of the brace. However, it is not well known how these compensatory changes will present when the body is only exposed to self-selected velocity changes and not running when the knee flexion would be greater than 30°.
Operational Definitions

Gait:
Also referred to as walking; convenient means for traveling short distances. In the absence of pathology, gait appears to be coordinated, efficient, and effortless (Perry & Burnfield, 2010).

Phases of Gait:

Initial Contact:
0% to 2% of the gait cycle; the initial foot drop on the floor and the instant reaction to body weight transfer (Perry & Burnfield).

Loading Response:
2% to 12% of the gait cycle; known as the second phase which makes up the initial double stance period. The phases contribute to the initial contact of the foot with the floor and continue until the other limb lifts for the initiation of swing (Perry & Burnfield).

Mid Stance:
12% to 31% of the gait cycle; known as the third phase which makes up the first part of the single limb support task. This phase begins with the other foot lifted and continues until the body weight becomes aligned over the forefoot (Perry & Burnfield).

Terminal Stance:
31% to 50% of the gait cycle; known as the fourth phase which completes the single limb support task. This phase is initiated when the heel rises and continues until the other foot strikes the ground. The body weight shifts to move ahead of the forefoot (Perry & Burnfield).
**Pre-swing:**

50% to 62% of the gait cycle; known as the fifth phase which can be defined as the official end of stance in the second double stance interval of the gait cycle. This phase is initiated when the initial contact of the opposite limb and is completed with the ipsilateral toe-off. The phase initiates the forward progression motion seen in the swing phase (Perry & Burnfield).

**Initial swing:**

62% to 75% of the gait cycle; known as the sixth phase which can be defined as the initiation of the foot being lift from the floor and ending with the swinging foot placed in the opposite position of the stance foot. This initial swing phase is one-third of the entire swing task (Perry & Burnfield).

**Mid swing:**

75% to 87% of the gait cycle; known as the seventh phase which can be defined as the beginning when the swinging foot is opposite the stance limb. The phase will end when the swinging limb is forward, and the hip and knee flexion postures are equal. This phase can be defined as the middle third of the swing period (Perry & Burnfield).

**Terminal swing:**

87% to 100% of the gait cycle; known as the final phase of the swing task. The phase begins with the vertical tibia and is completed when the foot strikes the ground. At this point, the limb advancement task has been completed (Perry & Burnfield).
Knee Structures:

**Menisci:**

The menisci of the knee are made up of the medial meniscus and lateral meniscus. The menisci assist in the tibial plateau into the concavities from the femoral condyles. These structures are key in the distribution of weight-bearing forces, reduction of friction between the tibia and fibula, and absorption of shock, (Levangie & Norkin 2011).

**Medial Collateral Ligaments (MCL):**

Broad, flat ligament. Attaches to the tibia at two points to allow the ligament to provide joint stability in both flexion and extension. When a valgus force is applied to the knee, the MCL provides less than 60% of the restraining force, but when the force is applied to the knee in ~25deg flexion, the MCL applies about 80% of the protective restraining force (Houglum & Bertoti, 2012).

**Lateral Collateral Ligament (LCL):**

Short, cord-like structure which sits outside the joint capsule. Protects the knee in the opposite direction of the MCL. Assists in varus stress protection when the knee is in full extension. As the knee decreases in extension and moves into slight flexion, the LCL provides roughly about 70% protection against varus stress (Houglum & Bertoti, 2012).

**Anterior Cruciate Ligament (ACL):**

Arise from the tibia joint and attached to the distal femur within the knee. ACL longer than PCL and follows a vertical alignment compared to PCL (Houglum & Bertoti, 2012).

**Posterior Cruciate Ligament (PCL):**

Primary restraint to posterior displacement of the tibia (Levangie & Norkin 2011).
**Center of Mass (CoM):**

Center of Mass is defined as the point where the three mid-cardinal planes of the body meet, not necessarily located in the body (Rodgers & Cavanagh, 1984).

**Center of Gravity (CoG):**

Center of Gravity is defined as the point at which the weight force (mg) of a body or system should be applied to a rigid body or system to balance exactly the translational and rotational effects of gravitational forces acting on the components of the body or system. Also known as, the point at which the weight of the body or system can be considered to act (Rodgers & Cavanagh, 1984).

**Kinematics:**

The description of motion (Rodgers & Cavanagh, 1984).

**Kinetics:**

The study of the forces that cause motion (Rodgers & Cavanagh, 1984).

**Line of Gravity (LoG):**

Line of Gravity is defined as the perpendicular line towards the ground from the center of gravity of that particular body (Levangie & Norkin, 2011).

**Base of Support (BoS):**

Human being’s base of support is defined by the area bounded posteriorly by the tips of the heels and anteriorly by a line joining the tips of the toes, and is considerably smaller than the quadruped base of support (Levangie & Norkin, 2011).

**Center of Pressure (CoP):**

Center of Pressure is defined as a quantity, collected by a force platform that describes the centroid of the pressure distribution over a given area (Rodgers & Cavanagh, 1984).
Rigid Body:

A collection of particles occupying fixed locations with respect to each other. Rigid body assumption is that it will not deform under applied forces no matter the size of the force (Rodgers & Cavanagh, 1984).

Linear Motion (translation):

The motion in which all parts of the body travel along parallel paths (Rodgers & Cavanagh, 1984).

Force:

A vector quantity that describes the action of one body acting on another. The action may be direct or indirect. Direct consisting of a foot pressing on the floor and indirect consisting of the gravitational attraction between a body and Earth (Rodgers & Cavanagh, 1984).

Newton’s Laws:

Basis of Newtonian Mechanics. Newton’s 1st law: A body at rest will remain at rest unless acted upon by a net force. Newton’s 2nd law: change in momentum of the body under the action of a resultant force will be proportional to the product of the magnitude of the force and the time. The change in the momentum will be in the same direction of the resultant force. Newton’s 3rd law: the action and reaction are equal and opposite (Rodgers & Cavanagh, 1984).

Kinetic Energy:

Mechanical energy of a body resulting from its motion: translation and/or rotation (Rodgers & Cavanagh, 1984).
Moment of Inertia:
The rotational equivalent of mass in its mechanical effect that is the resistance to a change of state during rotation. Can be expressed mathematically by $I = m*r^2$ (Rodgers & Cavanagh, 1984).

Force Platform:
An electromechanical device that gives electrical signals proportional to the components of the force acting on it. The most common use of a force plate is to measure the reaction forces and center of pressure between the foot and the floor during locomotor activities (Rodgers & Cavanagh, 1984).

Proprioceptive System:
Sensory system which provides body/limb position and contributes to the maintenance of balance; includes input from the musculoskeletal system (muscles, tendons, and joints); sensory receptors such as muscle spindles and Golgi tendon organs that supply information for changes in muscle length and rate of change of muscle length (Kandel, Schwartz, Jessell, Siegelbaum, & Hudspeth, 2000).

Ossur Rebound® Knee Brace:
According to U.S. Patent No. D758598 S1 issued in 2016, Ossur Rebound short sleeve is a knee brace comprising the following characteristics. The purpose of the Ossur Rebound brace is to serve the needs of a broad range of patients. The brace consists of reversible strapping to pull laterally-to-medially, medially-to-laterally, or a mix of both depending on the patients’ needs. The brace has two straps that assist with the control of the anterior-posterior hinge placement. The patella is supported with a removable patella buttress. The popliteal area of the knee is not effected with the brace due to the flexibility of the brace to minimize irritation in this area. The patient population
that the knee brace is indicated for includes mild to moderate ACL and/or PCL, MCL, and LCL instabilities, sprains and strains as well as mild medial/lateral knee instabilities. The brace material is breathable, reduces the heat retention and dries ultrafast due to the company’s 3-layer Cooltech™ fabric (Applegate, 1981).
CHAPTER II

REVIEW OF LITERATURE

The purpose of this study is to examine the effects of knee bracing on the kinematics and kinetics during human gait. This chapter will present relevant literature for gait, including pathological gait associated with knee injuries. This chapter will be divided into three major parts: First, a review of normal gait. The second portion of this review will examine general knee biomechanics, with pathological knee biomechanics, and treatment of knee pathologies. Finally, a discussion of bracing, with a focus on post-operative bracing and in-operative bracing.

Normal Gait

Gait, also referred to as walking, is a repetitive movement of limb sequencing to progress the body forward while also maintaining stability (Perry & Burnfield). As the body moves forward, the one limb which has contact with the ground becomes the source of support while the other limb progresses to become the new source of support. The transfer of body weight from one limb to the other when both feet are in contact with the ground can be referred as a key component to gait (Burnfield, 2010; Perry & Burnfield).

The Rancho Los Amigos Gait Analysis Committee developed generic terminology for the phases of the gait cycle. These phases consist of initial contact, loading response, mid-stance, terminal stance, pre-swing, initial swing, mid-swing, and terminal swing. These phases assist in the completion of weight acceptance, single limb support, and swing limb advancement of the stride (Burnfield, 2010; DeLisa, 2000).
Weight acceptance consists of shock absorption, initial limb stability, and the control of the progression. The initial contact and loading response phases are involved in this task. The single limb support task consists of one foot in swing while the opposite foot remains in contact with the ground. Mid-stance and terminal stance phases are also involved in this task. The swing limb advancement can be defined as the lift of the foot from the single leg support which progresses through the stride and preps for the proceeding stance phase. The task consists of pre-swing, initial swing, mid-swing, and terminal swing (Burnfield, 2010; Perry & Burnfield).

The gait cycle (GC) phases have no definitive starting and ending points. This allows the gait cycle to be defined as heel strike to heel strike or as forefoot strike to forefoot strike but these parameters are dependent on the participants walking pattern. One gait cycle can be termed as a stride (Burnfield, 2010; Perry & Burnfield) which is made-up of two steps. A step (i.e., right heel strike to left heel strike) can be described as the space between each limb (See Appendix).

The GC can be divided into the swing (~40% of GC) and stance (~60% of GC) periods. The swing period can be defined as the time in which the foot is in the air and the period begins as soon as the foot releases off the floor, usually referred to as toe-off. The stance period is the time in which the foot remains on the ground, and the GC usually begins with the initial contact of the foot on the ground (heel strike). Stance can be expressed throughout the GC in different types. The first instance of stance would be referred to as the initial double stance where there is loading on one limb while both feet are in contact with the ground. Following the initial stance, the leg that was previously loaded can now be classified as the single-leg stance phase. Terminal double limb stance then occurs preceding single-leg stance which can be defined as the floor contact by the contralateral limb that continues until the ipsilateral lower extremity toe-off. Single leg stance directly follows until double limb stance takes place initiating the new cycle.
The speed and length of stance are proportional, therefore, as the speed increases, the length of single stance increases and the double stance intervals decrease. Consequently, as the speed decreases, the length of single decreases and the double stance intervals increase (Burnfield, 2010; Perry & Burnfield).

**Normative Kinematic Joint Angles**

Kinematic data can be extremely meaningful regarding understanding locomotion. The process occurs in the supraspinal centers and involves the interaction between the central nervous system (CNS), the peripheral nervous system (PNS), and the musculoskeletal system (Enoka 1988). Vaughan and colleagues (1992) summarized gait into events starting with the activation of the CNS, the gait signals sent to the PNS, activation of tension in the muscles, generation of forces and moments across the joints, regulation of these via the skeletal segments, movement of the segment, and the generation of the ground reaction forces (Vaughan, Davis, & O’connor, 1992). Therefore, kinematic data can represent the position of the human body and offer a glimpse of the communication between the nervous and musculoskeletal system that are working together during gait. The trajectories of each joint are important to track during gait to determine deviations and hypothesize about pathologies. For example, tracking the trajectory of the hallux becomes an important determinate in the prediction of falls research. While the trajectories are important to study, the joint angles are of primary importance in gait analysis. Each joint displaces a relative joint angle between the adjacent joints (Levangie & Norkin, 2011).

Kuster and colleagues summarized the angular position of the lower extremity joints during leveled gait. The ankle position at heel strike, toe-off, and dorsiflexion can be defined as $2^\circ \pm 4.4$, $-15^\circ \pm 7.3$, and $11^\circ \pm 4.5$, respectively. The knee position at flexion during swing,
position at heel strike, flexion at stance, and extension at stance can be defined as \(83^\circ \pm 4.0, 11^\circ \pm 4.5, 24^\circ \pm 8.5\), and \(11^\circ \pm 5.0\), respectively. The hip at flexion during the swing phase, position at heel strike, and the extension in the late stance phase can all be defined as \(26^\circ \pm 6.5, 23^\circ \pm 8.0, -8.5^\circ \pm 4.8\), respectively (Kuster, Sakurai, & Wood, 1995). Kuster and colleagues results compare closely with other literature such as Vaughan (1992) and Winter (1990) (Vaughan et al., 1992), (D. A. Winter, 1991).

Kadaba and colleagues (1989) determined the mean and standard deviation (SD) of multiple correlations (CMC) for lower extremity joint angle in healthy individuals. CMC is a measure of similarity of the waveforms and related to the range of motion. The results have shown that the coefficient of multiple correlations for normal subjects at the knee joint was similar between both extremities: \(0.994 \pm 0.005\) (Kadaba et al., 1989). These results have been validated by various researchers (Kadaba et al., 1989; Roislien et al. 2012). The joint angles during the stride phase do not vary much and have not been observed to change with various cadences (D. A. Winter, 1983).

**Normative Kinematics of the Ankle Joint**

**Sagittal Plane**

According to Winter and colleagues, the joint angles in normal gait during level walking are as follows: initial contact \(0^\circ\) at \(0\%\) of the gait cycle (heel strike), \(5^\circ\) plantar flexion at \(10\%\) of the gait cycle when the foot is flat on the ground, \(5^\circ\) dorsiflexion at \(30\%\) of the gait cycle (mid-stance), \(0^\circ\) at \(40\%\) of the gait cycle (heel off), \(20^\circ\) plantar flexion at \(60\%\) of the gait cycle (toe-off), \(10^\circ\) plantar flexion during the initial swing of the gait cycle, \(0^\circ\) at the mid-swing of the gait cycle, and \(0^\circ\) at the terminal swing of the gait cycle (D. Winter, 1995).
Normative Kinematics of the Knee Joint

Kinematics of the Tibiofemoral Joint

Clinically speaking, the knee joint has two degrees of freedom consisting of flexion-extension and axial rotation. Knee flexion in a healthy individual varies between 120° and 150° with the average degree of flexion being 135°. If the knee is flexed at 90°, knee extension may be affected due to the flexibility of the hamstring muscles (Houglum & Bertoti, 2012). The axes of flexion and extension at the knee can be classified as the joint line above the femoral condyles. The axial rotation occurs in the transverse plane during knee flexion. The transverse rotation takes place nearest the longitudinal axis located medial to the intercondylar ridge of the tibia (lateral condyle rotate about the medial condyle). Therefore, lateral rotation is twice as large as medial rotation (Houglum & Bertoti, 2012).

Kinematics of the Patellofemoral Joint

The patellofemoral joint and tibiofemoral joint are related in function. The patellofemoral joint moves with respect to the movement of the tibiofemoral joint. If one of the following joints are restricted, knee mobility is also affected. When the lower extremity is in an open chain position, the tibia progresses forward as the femur remains stable and the patella glides over the condyles of the femur. When the lower extremity is in a closed chain position, the femur moves and the condyles of the femur glide at the surface of the patella.

During knee extension, the patella stays within the intercondylar groove at the proximal end. As the knee moves into flexion, the patella contacts the femur at about 25° of flexion. At 90° knee flexion, there is a maximal contact between the patella and femur. As the flexion increases in degrees, the contact becomes less, and the posterior portion of the patella remains in contact with the condyles of the femur (Houglum & Bertoti, 2012)
Sagittal Plane Marker Data

According to Winter and colleagues, the joint angles in normal gait during level walking are as follows: initial contact $0^\circ$ at 0% of the gait cycle (heel strike), $15^\circ$ flexion at 10% of the gait cycle when the foot is flat on the ground, $5^\circ$ flexion at 30% of the gait cycle (mid-stance), $0^\circ$ at 40% of the gait cycle (heel off), $30^\circ$ flexion at 60% of the gait cycle (toe-off), $60^\circ$ flexion during the initial swing of the gait cycle, $30^\circ$ flexion at the mid-swing of the gait cycle, and $0^\circ$ at the terminal swing of the gait cycle (D. Winter, 1995).

Normative Kinematics of the Hip Joint

Sagittal Plane

According to Winter and colleagues, the joint angles in normal gait during level walking are as follows: initial contact $20^\circ$ at 0% of the gait cycle (heel strike), $15^\circ$ flexion at 10% of the gait cycle when the foot is flat on the ground, $0^\circ$ flexion at 30% of the gait cycle (mid-stance), $10^\circ$-$20^\circ$ hyperextension at 40% of the gait cycle (heel off), $10^\circ$-$20^\circ$ hyperextension at 60% of the gait cycle (toe-off), $20^\circ$ flexion during the initial swing of the gait cycle, $30^\circ$ flexion at the mid-swing of the gait cycle, and $30^\circ$ at the terminal swing of the gait cycle (D. Winter, 1995).

Normative Kinetics of gait

Kinetics of gait are most often analyzed with the link-segment model and inverse dynamic approach. The link-segment model determines the forces and moments acting on each joint for the outcome (gait) to be completed (Levangie & Norkin, 2011). A moment can be defined as the turning effect of a force (Spencer, 1967). The moment pattern during the stride phases at all speeds vary the least in the ankle joint (D. A. Winter, 1983). Although these findings have been observed in normal gait and motor behavior, there is no literature on the use of any knee brace for an eight-hour period and its impact on cadence and moment force patterns.
Previous studies have supported that the moments at the knee and hip joints are highly variable compared to the ankle at different cadences. The cadences of individuals may alter due to their musculoskeletal pain. Consequently, the variability decreases as the cadence increases (D. A. Winter, 1983). Kuster and colleagues summarized the net joint moment maxima of the lower extremity joints during gait. Their findings of each joint are as follows: ankle 1.14 ±0.17, knee 1.20±0.50, and hip 0.98±0.55 (Kuster et al., 1995).

Vaughan and colleagues determined the three-dimensional resultant joint forces and moments at the joints of the lower extremity of human gait at roughly the 0.3-second mark. The right extremity of the individual hip, knee, and ankle were -10N in the medial/lateral direction, -34N in the anterior/posterior direction, -674 in the proximal/distal direction, -76N in the medial/lateral direction, 85N in the anterior/posterior direction, -599 in the proximal/distal direction, and -45N in the medial/lateral direction, -568N in the anterior/posterior direction, 57N in the proximal/distal direction, respectively. In the left extremity of the individual hip, knee, and ankle were 6N in the medial/lateral direction, -38N in the anterior/posterior direction, -111 in the proximal/distal direction, 1N in the medial/lateral direction, 33N in the anterior/posterior direction, -32N in the proximal/distal direction, 1N in the medial/lateral direction, -6N in the anterior/posterior direction, -9N in the proximal/distal direction, respectively (Vaughan et al., 1992). Mediolateral forces were determined by the forces which act on the mediolateral axis of the proximal segment. Anterior/posterior forces were determined by the forces which act on the axis perpendicular to the mediolateral/longitudinal axes while the proximal/distal forces can be determined by the forces which act on the longitudinal axis of the distal segment (Vaughan et al., 1992).
Pathological Gait

In pathological gait, the overall pathology may be the same between patients, but the deviations may be expressed in different volumes as well as the severity of the pathology. For example, a patient who suffers from osteoarthritis may feel discomfort during activity while another patient who suffers from the same condition may not feel as much discomfort. The pathology of gait can be expressed in five different categories consisting of deformity, muscle weakness, sensory loss, pain, and impaired motor control (Burnfield, 2010).

As per Burnfield (2010), deformity exists when the tissues at the joint deform disrupting the “passive mobility” of the area to remain normal posture (Burnfield, 2010). The most common deformity is a contracture which is a structural alteration of the fibrous connective tissue within the muscles, ligaments, or joint capsule. Muscle weakness can be defined as the “insufficient muscle strength to meet the demands of walking. Sensory loss can be defined as the “impaired proprioception effecting the position of their hip, knee, ankle, or foot and the type of contract with the floor.” Sensory loss may affect the patient in a multitude of ways especially with the transfer of their body weight. Patients who suffer from any type of sensory loss walk “very slow and cautious.” To compromise for the patient’s proprioception inability, they refer to their visual system to help them determine where they are visually located in space.

The center of gravity (center of mass) motion during level walking as per Detrembleur (2000) can be compared to the motion of an inverted pendulum where at each the center of gravity shifts from behind to in front of the point of contact of the foot on the ground (Detrembleur, Van den Hecke, & Dierick, 2000). The goal of individuals is to expend the least amount of energy as possible in order to conserve kinetic energy of the progressive motion and the potential energy of the center of gravity (Detrembleur et al., 2000).
According to previous literature, gait can be divided into normal and abnormal with abnormal further divided into neurological or non-neurological. Neurological gait deviations can be defined as unsteady, neuropathic, spastic, ataxic, hemiparetic, frontal and parkinsonian while non-neurological gait deviations are caused by arthritis, peripheral vascular disease, cardiac diseases, and injury. Deviations can be graded based on severity ranging from mild, moderate, or severe (Verghese et al., 2006).

Ankle and Foot Pathology

At the initial contact phase of the gait cycle, if and when the forefoot obtains contact with the ground before the heel, the normal gait cycle may become disrupted and can cause interference with the heel rocker, forward progression of the tibia, and the shock absorption at the knee. The delay of heel contact may be detrimental in the heel rocker and forward progression. This gait deviation differs from forefoot contact in that the heel does come in contact with the ground during the gait cycle phases. Although the heel is not the initial contact point, the forefoot comes in contact with the ground first followed by the heel with the unloading first taking place on the forefoot (Burnfield, 2010).

Knee Deviations

Common knee dysfunctions are most prominent in the sagittal plane. There are various gait deviations. Limited knee flexion can be defined as a less than normal ability of knee flexion. This would reduce the shock absorption as well as reduce the demand on the quadriceps. Limited knee flexion may also enhance the occurrence of foot drag during the swing phase. Knee hyperextension occurs when the knee is positioned more posterior to the neutral placement. As the knee is positioned into hyperextension, there is a reduction of need on the quadriceps. During the phases of weight acceptance, there is a decrease in the shock absorption which may
interfere with the sequential phases (Burnfield, 2010). Extensor thrust is rapid posterior movement into extension which is responsible for reducing the demand of the quadriceps at the knee joint.

During the weight acceptance and stance phases, the increase in knee flexion increases the demand on the quadriceps. In the terminal swing phase, the excessive knee flexion may shorten the step length as well as interrupting forward progression. Excess contralateral knee flexion occurs when there is a “greater-than-normal” knee flexion in the contralateral knee joint. This type of gait deviation in the knee lowers the body’s COM and usually lengths the reference limb for the compensation (Burnfield, 2010). Wobble is a sagittal gait deviation at the knee when there is a rapid alternation of flexion and extension. This decreases progression, increases energy costs by the constant rapid movements, and decreases the stability of the patient (Burnfield, 2010). Coronal (frontal) plane gait deviations are also common, especially noticeable at the stance phases of the gait cycle at the knee joint. Excessive abduction may also be referred to as valgus when there is an excessive lateral deviation of the distal tibia from the neutral position of the center of the knee. Effects on stability may cause compensation up or down the kinetic chain compensation and could be accompanied by pain. Excessive adduction, varus, occurs when there is an excessive medial deviation of the distal tibia from the neutral position of the center of the knee. The significance are common to excessive abduction, the severity of the adduction may cause limited stability, compensation, and pain (Burnfield, 2010).

**Hip Deviations**

Hip deviations are very common in all three planes of motion. Some deviations that take place in the sagittal plane include excess flexion, limited flexion, and past retract. Excessive flexion can be defined as more than normal flexion at the hip joint at any phase of gait.
Functionally, there is usually an increased demand on the extensors of the hip and the quadriceps during the stance phase, but there may also be compensation in the trunk area to overcome this deviation. Limited flexion can be defined as less than normal flexion in the hip joint for any phase of the GC. The limited hip flexion may affect knee flexion and ankle flexion during weight acceptance phases of the GC. During the swing phase, limited flexion deviation may cause the step length to shorten as well as affect overall progression of the gait cycle with an interference of foot clearance. Past retract can be defined as “an observable forward and backward movement of the thigh during terminal swing.” This deviation ensures knee extension in the terminal swing phase when there is inadequate activity in the quadriceps. There may be an impact on the forward progression due to the impedance of the step length due to the involuntary deviation (Burnfield, 2010).

The frontal plane deviations focus on the alignment of the thigh (femur) on the lateral/medial placement. Excess adduction refers to the greater than normal adduction for any phase of the GC. The hip joint deviation of excessive adduction may cause stability issues as well as issues with the progression of the GC and foot clearance. Excess abduction is another common deviation in the frontal view of the hip joint. This deviation can be defined as a greater than normal abduction for any phase of the GC. Due to the deviation during stance phase of gait, there exists a wider base of support, and in the swing phase, there is a decrease in the length of the limb for the forward progression during foot clearance (Burnfield, 2010).

The transverse plane deviations are any rotation of the limb greater than 10° of total displacement. The excess external rotation deviation of the hip joint can be defined as “greater-than-normal external rotation for a particular phase.” The deviation impacts the stance phase by changing the base of support to “toe-out” position. This position could be causing an increase of
stress on the ligament of the hip and knee during stance. On the other hand, excess internal
rotation can be defined as a “greater-than-normal internal rotation for a particular phase.”

Compared to the external rotation, internal rotation deviation of the hip joint influences the base
of support to the “toe-in” position. Unlike the external rotation, internal rotation deviation may
place more stress on the lateral ligaments of the knee and hip joints during the stance phase
(Burnfield, 2010).

**Importance of the Knee**

Although deviations in gait are susceptible at any joint in the lower extremity, the knee is
important to observe. The knee joint is located between the hip and ankle joints on the lower
extremity’s kinematic chain. Individuals all have anatomical and structural differences in the
lower extremity, and these differences are most prominent between males and females
(Hutchinson & Ireland, 1995). Hormonal effects play a role on the ligaments of the knee and the
laxity and flexibility and may be correlated with subluxations of the patella as well as ligament
sprains, especially in females (Hutchinson & Ireland, 1995).

Anatomical differences exist between individuals, especially between sexes. Alignment
of the lower extremity has been shown to contribute directly to the force distribution and strain
on the knee joint structures. For example, females on average have a lower center of mass, wider
pelvis area, shorter legs, and a greater possibility of genu valgum (Hunter, Andrews, Clancy, &

The quadriceps angle (Q-angle) for individuals may influence the forces and strain on the
knee joint and since the knee has fewer degrees of freedom compared to the ankle and hip. Q-
angle can be described as the angle measured at the anterior-superior iliac spine and the patella.
As the Q-angle increases, there are more lateral forces active within the quadriceps. The Q-angle
may be associated with malalignment of the patellofemoral joint and push the joint into lateral tracking causing additional knee pain. Some specific differences between individuals that affect the knee include alignment of the lower extremity, muscularity, torsion, varus and valgus at the ankle, pronation, and supination (Hutchinson & Ireland, 1995). The femoral notch size and shape can differ between individuals regarding shape and width. The shape and width usually coincide with the amount of surrounding tissue able to restrain anterior as well as displacement forces which may put the individual more at risk for injury. (Hutchinson & Ireland, 1995).

Previous literature has demonstrated that prophylactic ankle braces affect normal biomechanics such as functional performance and range of motion as well as kinematics and kinetics. Previous literature has concluded that alterations in normal biomechanics (kinematics/kinetics) at the ankle due to an ankle brace could shift the unloading force to the knee, therefore, increasing the risk of injury in the knee (Venesky et al., 2006). Although studies have previously analyzed ankle bracing of other joints of the lower extremity, the protocols rarely analyze gait of the subject. In many circumstances, many studies include unilateral and bilateral drop landings.

Some previous prophylactic knee brace studies have shown that overall performance may or not be affected depending on the activity. The study also claims that the trial learning effect of wearing the brace did not affect the results of the participants (Sforzo, Chen, Gold, & Frye, 1989). The effect of a knee brace on other joints of the lower extremity is not clearly understood, especially in healthy normal population. Previous literature focuses on knee bracing and sport (Hewson JR, Mendini, & Wang, 1986) (Rovere, Haupt, & Yates, 1987) (Najibi & Albright, 2005) (Pietrosimone, Grindstaff, Linens, Uczekaj, & Hertel, 2008) and literature lacks on knee bracing in regular activities of daily living. There seems to be more research on bracing of the
ankle (Mickel et al., 2006) (Callaghan, 1997) (Barkoukis, Sykaras, Costa, & Tsorbatzoudis, 2002) (Olmsted, Vela, Denegar, & Hertel, 2004) (Pedowitz, Reddy, Parekh, Huffman, & Sennett, 2008) (Hopper, McNair, & Elliott, 1999) than the knee joint which is equally important (Polk 2017).

**Pathological Knee Biomechanics:**

According to an epidemiology study of knee injuries, a ten-year documentation report of patients and their diagnoses were taken within athletic populations. The most common knee injuries were lateral collateral ligament (LCL), medial collateral ligament (MCL), anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), lateral meniscus (LM), and medial meniscus (MM) (Majewski, Susanne, & Klaus, 2006). The literature lacks on the normal population injury documentation of patients. Therefore, the athletic population injury documentation will be referenced.

ACL injuries are one of the most common knee injuries among populations and age groups. Previous literature shows that gait analysis data with an ACL deficient population prior to and following a three month reconstructive surgery showed significant functional differences compared to an uninjured control group. The results suggest that surgical repair of the ligament significantly affects lower extremity gait patterns and the re-adjustment back to pre-injury gait pattern takes longer than three months (Ferber et al., 2002). The subjects in the ACL reconstruction group of the study had about $3^\circ$ greater flexion at the knee joint and hip joint compared to the pre-surgery values of $5^\circ$ more flexed than the control groups (Ferber et al., 2002). A similar study reported that subjects with ACL reconstruction surgery had approximately $10^\circ$ greater flexion at the hip joint and knee joint after the three weeks of surgery,
but there were no differences observed roughly six months after compared to the control group (Devita, Hortobagyi, & Barrier, 1998).

**Common Treatment Post-Operative**

Braces are a common and necessary treatment for post-operative rehabilitation. The post-operative ACL braces are used for restriction and the re-development of normal range of motion, knee stability, and protection (Chaitanya & Kumar, 2015; Saka, 2014). Roughly 60% of the surveys completed by the American Academy of Orthopedic Surgeons recommend their patients wear a brace post-surgery for at least the first six weeks (Marx et al., 2003). Bracing post-surgery has become a very controversial topic within the literature. Studies have begun to compare rehabilitation with and without the knee brace, and there have been mixed results. For example, literature has shown no evidence that wearing a brace had any effect on the rehabilitation process including pain, range of motion, stability of the graft, and protection from further injury (Wright & Fetzer, 2007) (Andersson, Samuelsson, & Karlsson, 2009).

**Non-Operative Functional Knee Bracing & Purpose**

Many studies have looked at the functional knee braces post-operative surgery (Brandsson et al., 2001) (Birmingham et al., 2008; McDevitt et al., 2004) (Kartus et al., 1997) and literature lacks on the non-operative functional knee braces on the normal population. A few studies focus on the non-operative treatment of complete tears of the knee structures (Indelicato, 1983) as well as athletic populations such as high school football players (Jones, Henley, & Francis, 1986). This is the first study to look at the differences between testing pre- and post- an 8-hour intervention to compare gait characteristics. Previous studies focus on the immediate effects of bracing and fail to examine possible alterations that may occur a stimulated work day.
CHAPTER III

METHODS

The purpose of this study was to examine the effects knee bracing has on gait kinematics and kinetics over the course of a typical, allotted 8-hour work day.

Participants

Twenty eight healthy individuals (14 males and 14 females) between the ages of 18 and 35 were recruited to participate in the study. In fulfillment of the Institutional Review Board, written informed consent was obtained for each subject. Exclusionary criteria included any lower body musculoskeletal injuries, ligament injury or reconstructive surgery, neurological abnormalities including vestibular diseases and any difficulties that may interfere with activities of daily living.

Instrumentation

All testing took place at the Applied Biomechanics Laboratory (ABL) in the Department of Health, Exercise Science and Recreation Management at The University of Mississippi. The Vicon Nexus (Oxford, UK) 3D motion capture system with 8 wall mounted infra-red T-series cameras were used to collect and analyze kinematic gait data at a frequency of 120Hz. A Helen-Hayes plug-in gait lower body model marker set was used for each participant.

The ground reaction moments (GRM) were collected using Bertec (Bertec Corporation, Columbus, OH) and Advanced Mechanical Technology, Inc (AMTI) (176 Waltham Street, Watertown, MA 02472, USA) force plates that were embedded in the vinyl floor. Force plate
data were collected at a frequency of 1080 Hz and was synchronized with the Vicon motion capture system. Force plate positioning allowed for the right leg to strike the AMTI force plate and the left leg to strike the Bertec force plate.

**Experimental Conditions**

Participants were tested in the two experimental conditions: a normal no-brace (NB) control and a braced (B) condition wearing an Ossur knee brace. Both conditions were completed during visit one and repeated during visit two after an eight-hour intervention period. Testing took place on a flat laboratory surface located within the capture volume.

**Experimental Procedures**

All participants visited the ABL at The University of Mississippi twice on the same day: once in the early morning and the other eight hours later in the early afternoon/early evening.

Testing Day Part I: The first visit consisted of familiarization and pre-intervention measurements. Each participant signed The University of Mississippi approved informed consent. After paper-work completion, participants had their anthropometric measurements (height, mass, leg lengths, upper leg circumference, lower leg circumference) taken and began the experimental session. Participants were fitted with a knee brace according to the manufacturers recommendation of size and fit characteristics and were asked to walk around to determine their comfort and willingness to complete the study.

Gait data were recorded and analyzed via a Vicon Nexus (Oxford, UK) 3D motion capture system. The subject changed into spandex clothing and was prepared for walking gait trials. Retroreflective markers were placed on anatomical landmarks in accordance with the lower body plug-in-gait model from the Helen Hayes marker system. After completion of marker placement, a static capture was taken to define the coordinates of each marker. Participants then performed
twenty, 10 NB followed by 10 B, pre-intervention gait passes in the sneaker that the subject wore over the course of the entire day. A gait pass was classified as a participant walking from one end of the video capture volume to the other. Next, the markers were removed from the subject, and they were asked to change into their work clothes. The subject was then given a pedometer to track their steps and then went about the workday while wearing the knee brace on their dominant knee (determined by asking which foot they would use to kick a ball. All participants analyzed in the study were right leg dominant). Participants then returned to the ABL promptly in 8 hours.

Testing Day Part II: Eight hours after the placement of the knee brace participants returned to ABL and were prepped for twenty, 10 B followed by 10 NB, post-intervention gait trials. The gait passes were conducted in the sneaker that the subject wore throughout the day.

**Data Analysis**

Peak joint angles, peak moments, and time to peak of the ankle, knee, and hip of the legs were evaluated in the frontal, sagittal, and transverse planes from right heel strike to right toe-off as well as left heel strike to left toe-off. Joint moments were normalized for body weight and reported in terms of a step, the stance phase of the gait cycle. The joint moment peak was calculated at the first positive peak moment over the course of a step. The first peak moment was calculated due to previous literature findings that in knee OA patients, there was an observed increase in the 1st peak moment (Shull et al., 2013). The value would assist in revealing the brace impact on the initial loading phase.

**Statistical Analysis**

Linear mixed model ANOVA’s with fixed factors of time [Morning No Brace (MNB), Morning Brace (MB), Afternoon Brace (AB), and Afternoon No Brace (ANB)] and condition [(No Brace (NB) and Brace (B))] employing unstructured covariance were used to examine possible
differences in joint angles, joint moments, and time to peak joint moments. This model takes into account that each person provided data for two conditions. If a statistically significant interaction was found, simple effects were examined using pairwise comparisons with Bonferroni adjustment. Otherwise, main effects were examined in a similar fashion. Values are reported as estimated marginal means and standard errors. All statistical analyses were performed using SPSS 23. Statistical significance was set a priori at alpha = 0.05.
CHAPTER IV

RESULTS

Participant Information:

Twenty-eight healthy adults (14 females and 14 males) completed the study successfully. Processing restriction resulted in 19 subjects being used in the data analysis (10 females and 9 males). The data was analyzed according to the cardinal and rotational planes frontal, sagittal, and transverse; X, Y, and Z, respectively.

ANgLES:

Joint Angles: Hip - Braced and Unbraced Leg

Main effects of both time (F(3, 18.0) = 5.329, P=.008) and condition (F(1,18.0) = 24.495, P <.0005) were present. There was an overall increase in the range of motion (ROM) from MNB to AB (+3.595 (1.081) °, P = .023) measurement. The braced leg had a greater ROM than the unbraced leg (+3.862(0.780) °, P<.0005). The analysis of peak hip angle in the Y-axis when the unbraced leg completed a step showed no significant time by condition interaction (F(3, 18.0) = 2.283, P = .114). There was no significant main effect of condition (F(1,18) = 1.144, P = .299). There was a significant main effect of time (F(3,18.0) = 4.734, P = .013) but following a Bonferroni adjustment, these differences were insignificant. Peak hip angle in the Z-axis when the unbraced leg completed a step showed neither a significant time by condition interaction (F(3, 18.0) = 3.034, P =.056), nor main effects of time (F(3,18.0) = .816), P = .502) or condition (F(1,18.0)= 3.176, P=.092). There was no change regardless of time or condition.
In comparison, peak hip angle in the X-axis when the braced leg completed a step showed a significant time by condition interaction (F(3, 18.0) = 8.012, P = .001). However, following pairwise comparisons with a Bonferroni adjustment this interaction was shown to be insignificant. There was no significant main effect of condition (F(1,18) = 0.017, P = .897). There was a statistically significant main effect of time (F(3,18.0) = 4.514, P = .016). Pairwise comparisons with Bonferroni adjustment revealed that at AB there was a greater ROM than the MB (+3.486 (1.159)°, P = .045). When the braced leg completed a step, it showed no significant time by condition interaction (F(3, 18.0) = 1.449, P = .262) in the Y-axis. There was no statistically significant main effect of condition (F(1,18) = 0.839, P = .490), but there was a statistically significant main effect of condition (F(1,18.0) = 8.603, P = .009), which may influence the braced leg appearing to have less ROM than the unbraced leg (-1.518(0.518)°). Analysis of peak hip angle in the Z-axis when the braced leg completed a step showed no significant time by condition interaction (F(3, 18.0) = 2.747, P = .073). There was no significant main effect of time (F(3,18) = 0.973, P = .427), but there was a significant main effect of condition (F(1,18.0) = 20.307, P < .0005) which may influence the braced leg appearing to have lesser ROM than the unbraced leg (-8.816(1.956)°).

Overall, when the unbraced leg took a step, there was an increase ROM about the X-axis between the MNB and AB conditions. The braced leg had an increase in ROM than the unbraced leg. On that contrary, when the braced leg took a step, there was greater ROM about the X-axis at the AB condition than MB condition. There was less ROM about the unbraced leg in the Y-axis as well as Z-axis.
Joint Angles: Knee - Braced and Unbraced Leg

Analysis of peak knee angle in the X-axis showed a significant time by condition interaction ($F(3, 18.0) = 9.317, P = .001$). However, following pairwise comparisons with a Bonferroni adjustment, this interaction was insignificant. There was a significant main effect of condition ($F(1,18) = 318.314, P < .0005$) and of time ($F(3,18.0) = 10.468, P < .0005$). Pairwise comparisons with a Bonferroni adjustment revealed a reduced ROM in the morning NB compared to the morning braced time point ($-2.439(.437)^\circ$). The unbraced leg showed no significant time by condition interaction ($F(3, 18.0) = 1.901, P = .166$) in the Y-axis. There was no significant main effect of time ($F(3,18) = 0.406, P = .751$), but there was a statistically significant main effect of condition ($F(1,18.0) = 5.037, P = .038$), which may influence the braced leg appearing to have slightly greater ROM than the unbraced leg ($-2.878(1.028)^\circ$).

Analysis of peak knee angle in the Z-axis when the unbraced leg completed a step showed neither a significant time by condition interaction ($F(3, 18.0) = 1.980, P = .153$), nor main effects of time ($F(3,18.0) = 0.956, P = .435$) or condition ($F(1,18.0)= 0.047, P=831$). There was no change regardless of time or condition.

In comparison, peak knee angle in the X-axis when the braced leg completed a step showed a significant time by condition interaction ($F(3, 18.0) = 28.714, P < .0005$). However, this interaction was shown to be insignificant following pairwise comparisons. There was a significant main effect of condition ($F(1,18) = 225.487, P < .0005$) as well as a significant main effect of time ($F(3,18.0) = 15.706, P < .0005$). Pairwise comparisons with Bonferroni adjustment revealed a lesser ROM in the morning no brace condition compared to the morning braced condition time point ($-3.311(.568)^\circ$), as well as a lesser ROM in the afternoon no brace condition.
compared to the afternoon, braced time point (\(-2.881(0.674)^{\circ}\)). Overall, the braced leg had a lesser ROM than the unbraced leg (\(-20.050(1.335)^{\circ}, P<0.0005\)).

Analysis of peak knee angle in the Y-axis when the braced leg completed a step showed no significant time by condition interaction (F(3, 18.0) = 2.930, P = 0.062). There was no significant main effect of time (F(3,18) = 2.442, P = 0.098). There was a statistically significant main effect of condition (F(1,18.0) = 49.081, P < 0.0005), revealing that the braced leg had a lesser ROM than the unbraced leg (\(-12.28(1.753)^{\circ}\)). Analysis of peak knee angle in the Z-axis when the braced leg completed a step showed no significant time by condition interaction (F(3, 18.0) = 2.768, P = 0.072). There was no significant main effect of time (F(3,18) = 1.794, P = 0.184). There was a statistically significant main effect of condition (F(1,18.0) = 7.804, P = 0.012), revealing that the braced leg had a lesser ROM than the unbraced leg (\(-4.67(1.674)^{\circ}\)).

To summarize, when the unbraced leg took a step there was less ROM in the no brace condition compared to the morning brace condition about the X-axis. There was a greater ROM in the unbraced knee compared to the braced knee in the Y-axis. On the contrary, when the braced leg took a step, there was a smaller ROM in the MNB compared to the MB as well as a smaller ROM in the ANB compared to the AB. Overall, there was less ROM at the braced leg compared to the unbraced leg about the X-axis. The braced leg showed to have a smaller ROM than the unbraced leg about the Y-axis and Z-axis.

**Joint Angles: Ankle - Braced and Unbraced Leg**

Analysis of peak ankle angle in the X-axis showed a significant time by condition interaction (F(3, 18.0) = 8.297, P = 0.001). The within and between time measures revealed no difference in the unbraced leg at any time point. The morning NB and afternoon NB time points revealed no difference, as well as the morning B and afternoon B, revealed no difference. The
only difference is when the brace was worn which attributes to the increase in ROM in the X-axis at the ankle. The within time and between condition measures revealed morning no brace and afternoon no brace were not significantly different in conditions. The morning brace and afternoon brace had a difference between the braced and unbraced legs. This may suggest an increase in ROM of the braced leg at the ankle in the X-axis.

Ankle angle in the Y-axis showed a significant time by condition interaction ($F(3, 18.0) = 3.255, P = .046$). However, following pairwise comparisons with Bonferroni adjustment, this interaction was found to be insignificant. There was no significant main effect of condition ($F(1,18) = 0.277, P = .654$). There was a statistically significant main effect of time ($F(3,18.0) = 5.498, P = .007$). The pairwise comparisons suggest a ROM difference between morning no brace and morning braced time periods (-1.559(.520) °).

There was no significant main effect of condition ($F(1,18) = 1.963, P = .178$). There was a statistically significant main effect of time ($F(3,18.0) = 17.954, P < .0005$). Pairwise comparisons show a reduced range of motion in the morning no brace time period compared to the morning brace (-4.108(.960) °, $P = .003$), afternoon brace (-6.374(1.362) °, $P = .001$), and afternoon no brace condition (-5.179(1.665) °, $P = .036$).

Analysis of peak ankle angle in the X-axis when the braced leg completed a step showed a significant time by condition interaction ($F(3, 18.0) = 11.209, P < .0005$). The within condition between time points analysis revealed no difference on the unbraced leg at any time point. The right leg in the morning NB condition had a decrease in ROM than the morning B and afternoon B time points but no difference in afternoon NB time points. The no braced conditions and the braced conditions only difference was when the brace was worn there was an increase in ROM. The within time point and between condition measures showed that MNB time period was not
different, while the right leg in the MB condition had an increase in ROM than the left leg. At the ANB time point the contralateral leg seems as if compensation took place due to no difference after the eight hour intervention braced or unbraced between the legs. The Y-axis when the braced leg completed a step showed a significant time by condition interaction (F(3, 18.0) = 3.547, P = .035). However, following pairwise comparisons with a Bonferroni adjustment, this interaction was shown to be insignificant. There was a no significant main effect of condition (F(1,18) = 1.153, P = .297). There was a statistically significant main effect of time (F(3,18.0) = 5.676, P = .006). However, following pairwise comparisons with a Bonferroni adjustment these values were insignificant.

Analysis of peak ankle angle in the Z-axis when the braced leg completed a step showed a significant time by condition interaction (F(3, 18.0) = 6.823, P =.003). However, following pairwise comparisons, this interaction was shown to be insignificant. There was significant main effect of condition (F(1,18) = 5.625, P = .029). There was a statistically significant main effect of time (F(3,18.0) = 17.656, P < .0005). Pairwise comparisons with a Bonferroni adjustment may suggest a lesser range of motion in morning no brace time period compared to the morning brace (-3.964(.967) °, P = .004), afternoon brace (-6.891(1.436) °, P = .001), and afternoon no brace condition (-5.328(1.737) °, P = .040). Overall, the braced leg had a lesser ROM than the unbraced leg (-4.041(1.704) °).

Overall, when the unbraced leg took a step, there was a finding about the X-axis revealing that there was an increase ROM at the ankle in the braced leg. In the Y-axis, there was less ROM from the MNB to the MB conditions, and in the Z-axis there was less ROM when comparing the MNB to the MB, MNB to MNB to AB, and MNB to ANB. On the contrary, when the braced leg took a step, there was a decrease ROM from the MB to the AB time points, but
there was no difference seen between the MB and afternoon NB conditions. Lastly, there was less ROM in the MNB compared to MB, MNB compared to AB, and MNB compared to ANB about the Z-axis.

MOMENTS:

A similar linear mixed model with fixed factors of time and condition was employed to examine the possible changes in joint moments following unaccustomed knee brace wear across an entire day.

Joint Moments: Hip - Braced and Unbraced Leg

Analysis of hip moments in the X-axis revealed a statistically significant condition by time interaction (F(3,18)=3.294, \(P = .044\)). Following pairwise comparisons, this interaction was found to be statistically insignificant. There were main effects of both time (F(3,18)=9.042, \(P = .001\)) and condition (F(1,18)=34.286, \(P < .0005\)). Examination of the time effect found that the unbraced hip moments at the first measurement, MNB, were significantly lower than the braced hip moments at both the morning (mean difference: -1.551, \(P = .015\)) and afternoon (mean difference: -1.654, \(P = .023\)) measurements, but were not different from the afternoon, unbraced measurement (mean difference: -1.276, \(P = .312\)). Examination of the condition determined that the unbraced hip had an overall lower joint moment than the braced hip (mean difference: -7.568, \(P < .0005\)).

Analysis of hip moments in the Y-axis revealed no statistically significant condition by time interaction (F(3,18)=0.885, \(P = .467\)), main effect of condition (F(1,18)=0.018, \(P = .894\)), or main effect of time (F(3,18)=1.523, \(P = .243\)).
Analysis of hip moment in the Z-axis revealed no statistically significant condition by time interaction (F(3,18) = 2.592, P = .085) and no main effect of condition (F(1,18)=.116, P = .737). However, there was a main effect of time (F(3,18)=6.951, P=.003). Overall, there was a time difference at morning no brace compared to afternoon braced (mean difference: -.474, P=.024) measurements, as well as afternoon, braced compared to afternoon no brace (mean difference: .361, P=.011).

In summary, the hip moment about the X-axis was lower at MNB than MB and AB. Overall, there was a lower joint moment about the unbraced leg’s hip than the braced leg’s hip. The Z-axis revealed slight significant differences from MNB compared to AB as well as AB compared to ANB.

**Joint Moments: Knee - Braced and Unbraced Leg**

Analysis of knee moments in the X-axis revealed a statistical significance for both time (F(3,18)=9.625, P = .001) and condition (F(1,18)=77.942, P < .0005). The time effect found that the morning unbraced knee moments were significantly lower than the braced morning conditions (mean difference: -1.615, P = .001). There was a statistically significant difference found between the afternoon braced and the afternoon no braced (mean difference: 1.087, P = .035) measurements. Examination of the condition effect determined that the unbraced knee, overall, had a lower joint moment than the braced knee (mean difference: -11.106, P < .0005).

Analysis of knee moments in the Y-axis revealed no statistically significant condition by time interaction (F(3,18)=0.486, P = .696), main effect of condition (F(1,18)<0.0005, P = .993), or main effect of time (F(3,18)=0.333, P = .802).

Analysis of knee moments in the Z-axis revealed no statistically significant condition by time interaction (F(3,18)=2.909, P = .063). There were main effects of both time (F(3,18)=4.196,
Examination of the time effect found that the afternoon braced condition was statistically significantly higher than the afternoon no braced condition moments (mean difference: .195, $P = .018$) measurements. After pairwise comparisons, the unbraced leg, overall, had a lower joint moment than the braced leg at the knee (mean difference: -.807, $P < .0005$).

Overall findings of the knee moments reveal that about the X-axis the unbraced leg had lower moments at the MNB than during the MB conditions. The unbraced knee had a lower joint moment than the braced knee. About the Z-axis, the ANB had a lower moment than the AB condition.

**Joint Moments: Ankle - Braced and Unbraced Leg**

Analysis of ankle moment in the X-axis revealed no statistically significant condition by time interaction ($F(3,18) = 1.718, P = .199$) and no main effect of time ($F(3,18)=1.070, P = .386$). However, there was a main effect of condition ($F(1,18)=31.802, P<.0005$. Overall, there was a condition difference between the braced and unbraced leg (mean difference: 3.139, $P<.005$) measurements.

Ankle moments in the Y-axis revealed no statistically significant condition by time interaction ($F(3,18) = .411, P = .747$) and no main effect of time ($F(3,18)=.553, P = .653$). However, there was a main effect of condition ($F(1,18)=118.673, P<.0005$. Overall, there was a condition difference between the braced and unbraced leg (mean difference: 3.558, $P<.005$) measurements.

Analysis of ankle moments in the Z-axis revealed no statistically significant condition by time interaction ($F(3,18)=0.301, P = .825$), main effect of condition ($F(1,18)=1.555, P = .228$), or main effect of time ($F(3,18)=1.486, P = .252$).
Overall, there was a difference between the braced and unbraced leg about the X-axis. The braced leg had a greater moment than the unbraced leg. There was a larger moment in the braced leg than the unbraced leg about the Y-axis, as well.

**TIME TO PEAK MOMENTS:**

A similar linear mixed model with fixed factors of time and condition was employed to examine the possible changes in time to peak of joint moments following unaccustomed knee brace wear across an entire day.

**Time To Peak Joint Moments: Hip - Braced and Unbraced Leg**

Analysis of time to peak of the hip moments in the X-axis revealed no statistically significant condition by time interaction (F(3,18)=3.123, \( P = .052 \)) or main effects for time (F(3,18)=0.659, \( P = .588 \)). However, there was significance found for condition (F(1,18)=162.678, \( P < .0005 \)). Examination of the condition effect determined that the unbraced hip, overall, had a smaller time to peak joint moment than the braced hip (mean difference: -12.245, \( P < .0005 \)).

Analysis of time to peak of the hip moments in the Y-axis revealed no statistically significant condition by time interaction (F(3,18)=1.322, \( P = .298 \)), main effect of condition (F(1,18)=0.918, \( P = .351 \)), or main effect of time (F(3,18)=1.638, \( P = .298 \)).

Analysis of time to peak hip moment in the Z-axis revealed no statistically significant condition by time interaction (F(3,18) = 0.565, \( P = .645 \)) and no main effect of time (F(3,18)=.877, \( P = .471 \)). However, there was a main effect of condition (F(1,18)=8.737, \( P = .008 \)). Overall, there was a condition difference in the unbraced leg compared to the braced leg (mean difference: -13.704, \( P = .008 \)) measurements.

In summary, the unbraced hip took less time to reach peak moment than the braced hip in both the X-axis and Z-axis.
Time To Peak Joint Moments: Knee - Braced and Unbraced Leg

Analysis of time to peak of the knee moment in the X-axis revealed no statistically significant condition by time interaction (F(3,18) = 3.070, P = .054) and no main effect of time (F(3,18)=2.468, P < .0005. However, there was a main effect of condition (F(1,18)=46.895, P<.0005. Overall, there was a condition difference between the unbraced and braced leg (mean difference: 47.888, P<.0005) measurements.

Analysis of time to peak of the knee moment in the Y-axis revealed no statistically significant condition by time interaction (F(3,18) = 2.112, P = .134) and no main effect of time (F(3,18)=1.656, P = .212. However, there was a main effect of condition (F(1,18)=57.062, P<.0005. Overall, there was a condition difference between the unbraced and braced leg (mean difference: -33.896, P<.0005) measurements.

Analysis of time to peak of the knee moment in the Z-axis revealed no statistically significant condition by time interaction (F(3,18) = 1.641, P = .215) and no main effect of time (F(3,18)=1.826, P = .179. However, there was a main effect of condition (F(1,18)=11.408, P=.003. Overall, there was a condition difference between the unbraced and braced leg (mean difference: -16.529, P=.003) measurements.

In summary, the unbraced leg took less time to reach peak than the braced leg for both the Y-axis and Z-axis. The unbraced leg took longer to reach time to peak moment than the braced leg about the X-axis.

Time To Peak Joint Moments: Ankle - Braced and Unbraced Leg

Time to peak of the ankle moments in the X-axis revealed no statistically significant condition by time interaction (F(3,18)=2.029, P = .146). There were main effects of both time (F(3,18)=3.177, P = .049) and condition (F(1,18)=85.751, P < .0005). The Bonferroni correction
revealed that there was no statistically significant main effect for time. Examination of the condition effect showed that the unbraced leg had a smaller time to peak at the ankle compared to the braced condition (mean difference: -2.263, \( P < .0005 \)) measurements.

Analysis of time to peak of the ankle moment in the Y-axis revealed no statistically significant condition by time interaction (\( F(3,18) = 1.639, P = .216 \)) and no main effect of time (\( F(3,18) = 2.249, P = .118 \)). However, there was a main effect of condition (\( F(1,18)=5.478, P=.031 \)). Overall, there was a condition difference between the unbraced and braced leg (mean difference: -13.621, \( P=.031 \)) measurements.

Analysis of time to peak of the ankle moments in the Z-axis revealed no statistically significant condition by time interaction (\( F(3,18)=0.166, P = .918 \)) nor main effect of time (\( F(3,18)=1.205, P = .336 \)). However, there was a main effect of condition (\( F(1,18)=105.977, P<.0005 \)). The unbraced condition had a greater time to peak than the braced condition (mean difference: 12.139, \( P<.0005 \)).

In conclusion, the unbraced leg took less time to obtain peak at the ankle in both the X-axis and Y-axis but took longer to reach peak in the unbraced condition compared to the braced condition about the Z-axis.
CHAPTER V
DISCUSSION

Previous literature has investigated functional knee braces on post-operative surgery as well as within athletic populations. Nevertheless, the literature lacks input on non-operative functional knee brace studies on a healthy population that uses a brace for support and preventative measures. The purpose of this study was to determine variability in gait characteristics after an 8-hour knee brace intervention. During the four conditions [morning no brace (MNB), morning brace (MB), afternoon brace (AB), and afternoon no brace (ANB)] lower extremity joint angles, joint moments, and time to peak joint moments were analyzed.

As variability in joint angles are an important aspect when detailing human locomotion, I investigated the impact bracing has on lower extremity joint angles at the hip. When the unbraced leg took a step, there was a 4° increase in flexion from the MNB time point to the AB time point while the braced leg had an overall increase in ROM of roughly 4°. This may suggest that the unbraced leg throughout a workday compensates due to the brace on the opposite leg. The results coincide with Devita and colleagues’ work on healthy individuals and functional knee braces which showed that the knee brace caused an individual to use greater extensor torques at the hip and ankle to account for less mobility at the knee (Devita et al., 1996).

When the braced leg took a step, the AB had a greater flexion ROM than the MB by roughly 3°. The abduction increased roughly 2° in the braced leg compared to the unbraced leg.
The brace leg had less rotation about the Z-axis than the unbraced leg by roughly 8°. Although statistically significant, there may not be a functional significance due to the small deviation from the norm. Functional significance takes into account the minimal clinically important difference (MCID) that represents the smallest change of a value in an outcome measure that would be beneficial as well as the minimal detectable change (MCD) which is the measurement required to exceed measurement variability. In other words, MCD would be the smallest change in an outcome measure that would be considered “real.” The MCID would always be a larger value than the MCD (Tilson et al. 2010). Therefore, this suggests that if there is functional significance with the increased abduction, the difference between the conditions may be due to the 8 hour wear time of the brace causing a disturbance moving up the kinematic chain (Powers, 2010). Similarly, when the braced leg took a step, it experienced more external rotation of roughly 9°. During the loading response of the GC, the hip is flexed, slightly adducted, and internally rotated (Powers, 2010). The addition of the knee brace may force the gluteus medius and minimus to abduct the leg to compensate for the width of the brace to prevent irritation on the contralateral leg knee area and allow for clearance.

The knee angles are important to analyze due to the influence of knee flexion and extension. There was a 2° change in the extension angle at the knee between the MNB and MB conditions suggesting the brace may decrease ROM and cause slight compensation due to the material of the brace restricting the joint. In addition, the braced leg adducted roughly 3° more than the unbraced leg which may be due to the marker placement. With the placement of the reflective markers having to be placed on the lateral condyle which was covered by the brace, the marker appears to place the knee into an adducted state causing it to be a limitation within the study. On the other hand, Winter et al. reported minor variation in joint kinematics at different
cadences and slight increases in the peak knee flexion during the loading response phase (Winter et al. 1984).

When the braced leg took a step, there was a 3° increase in extension between the MNB to the MB as well as the ANB to AB condition. Similarly, the braced leg had 12° less ROM about the Y-axis than the unbraced leg and had 5° less rotation in the Z-axis compared to the unbraced leg suggesting the brace itself made the knee more extended due to its restrictive properties.

Overall, knee flexion and the velocity at which it occurs is an important factor for the swing phase of normal gait (Mochon and McMahon, 1980; Mena et al., 1981; Piazza and Delp, 1996). The knee flexion and the velocity at which it occurs are contributors to stiff-knee gait. Limited knee flexion and excessive knee extension moment during the swing phase are common signs of stiff-knee gait (Goldberg et al., 2003; Goldberg et al., 2004). The increase in knee extension with the Ossur knee brace present becomes worrisome from a clinical perspective due to the possible implication it has on the swing phase leading to stiff-knee gait characteristics. Future research in the swing phase of the GC will need to be completed to conclude this speculation.

When the unbraced leg took a step, there was increased flexion at the ankle. The MB and AB leg were significantly different from one another indicating the ankle of the braced leg needed to compensate for the brace on the knee. There was roughly a 2° difference about the Y-axis between the MNB and MB time periods suggesting the ankle is forced into more abduction due to the possible unfamiliarity of the brace. In addition, there was an increase in external rotation between the MNB and the MB condition of 4° as well as a difference between MNB to AB of roughly 6° and an increase in external rotation from the MNB to the ANB condition of
roughly 5°. Due to similar angular values, this may suggest that although a statistical significance was found, there may not be a functional at the ankle joint.

When the braced leg took a step, there was a difference at the ankle joint when the right leg became unbraced. The findings suggest that there was more extension than at the MB and AB conditions. This may indicate that the brace caused a kinematic disturbance and forced the ankle joint to extend. Due to these alterations, there was less ROM about the Z-axis in the MNB compared to the MB by roughly 3°. The MNB compared to the AB there was less ROM of about 7°. In addition, there was less rotation in the MNB time point than the ANB by roughly 5°. This difference may be explained by the overall wear time of the knee brace. The varying degrees in the morning between brace wear and NB wear suggests the brace influenced this change of angle. The afternoon difference may be explained by the length of the intervention; over the course of 8 hours, the body made various adaptations to the brace that did not occur in the few minutes it took between the MNB to MB conditions. Overall, the braced leg had a more external rotation of roughly 4° than the unbraced leg at the ankle.

**Joint Moments**

Joint moments are vital to understanding the internal load distribution during human gait. The lower extremity joint moments are important to analyze due to the possible compensation down the kinematic chain. First, analyzing the hip, with the knee brace on the dominant leg, revealed a statistically significant difference between conditions in the X-axis showing an overall lower joint moment than the braced hip. The unbraced leg’s hip moment at the MNB was lower than the MB and AB time points. About the Z-axis, MNB compared to AB had a lesser moment, as well as AB compared to ANB. The time it took to reach the peak may have been influenced by the condition effect. The unbraced hip had a smaller time to peak joint moment about the X-
axis and Z-axis than the braced leg which may be due to the kinetic chain disturbance caused by the brace as well as the possible influence it had on gait velocity. Previous studies have concluded there to be an increase in hip extension during increased gait velocity (Kerrigan et al. 2001). Gait velocities were not reported in this study, and therefore, it is unclear if the hip differences were due to gait velocity changes between conditions or due to the intervention of the study.

The braced knee joint sustains a large internal loading due to the limitation of functional degrees of freedom. The results show that when there was no knee brace present, the joint moment in the X-axis had lower moments at the MNB than the braced leg at the MB. There was a significant difference between the AB and ANB with the braced leg showing a higher moment. Overall, the unbraced knee had a lower joint moment than the braced knee. Across the three planes of motion, there were condition differences between the unbraced and braced leg. The braced leg had an increase in the overall time it took to reach the 1st positive peak (Figure 4 &5). Within the current literature, there is no consistency. Studies by Baliunas et al. and Messier et al. showed a decreased midstance knee angle while Stauffer et al. and others did not. The current study aligns with the findings of Baliunas et al. that the knee angle and peak knee moment may be correlated. The subjects that walked with a higher or lower knee flexion angle also reported a higher or lower knee moment, respectively.

In normal gait, 60-80% of the total intrinsic compressive load transmitted across the knee is in the medial compartment (Andriacchi 1995). A deviation from the norms may be due to the compressive characteristics of the knee brace narrowing the joint space. The moments have been shown to be strongly related to the magnitude of compressive loads on the knee (Schipplein and Andriacchi 1991).
Lastly, the ankle results show an overall condition difference (B vs. NB) across all axes of rotation. The X-axis, as well as Y-axis, showed a significant increase in the calculated joint moment when the brace was present. There was a condition difference between the braced and unbraced leg. The unbraced leg tended to elicit a smaller time to peak than the braced conditions about the X-axis and Y-axis suggesting that the unbraced condition took less time to reach peak due to stabilizing the joint more rapidly. About the Z-axis the unbraced leg had a greater time to peak than the braced condition. In addition, alterations in walking speed have previously been shown to affect gait variables, such as joint moments in human gait (Zeni et al. 2011). These alterations may alter the moment results due to the slight variations in gait speed. For example, if the subjects gait velocity significantly decreased when wearing the knee brace, this may suggest an error in the results due to the limitation of not reporting velocity.

**Conclusion:**

The knee sustains the highest percentage of injuries of the lower extremities, especially within physically active individuals (Powers et al., 2010). From surgical procedures to rehabilitation, the high percentage of orthopedic injuries has led to an economic burden to patients and providers. At least a quarter of adults 64 years of age and younger with private health insurance are concerned about the costs of a major unexpected medical expense such as surgery or chronic illness. Roughly 20% refuse to see a physician when they are ill due to financial concerns, even when they have insurance (Olen, 2017). This may suggest, with arthroscopic knee procedures on the rise, patients desire a quick fix to assist with pain management before opting for a surgical procedure.

The findings from this study can be used as a series of recommendations for future research studies as well as future clinical recommendations. Knee braces, especially the Ossur
Rebound, are devices that can serve for a variety of conditions to aid in protection and prevention. Hence, the following recommendations can be made from the findings of this study for individuals in need of a knee brace. While wearing a knee brace during walking, there is an increased internal load present on the hip, knee, and ankle of the braced leg. Previous literature has shown that functional knee bracing decreased internal loads within a pathological population, such as ACL-deficient populations (Devita et al., 1996). In healthy individuals, without a diagnosed pathology, there was an unintended result of joint load increase. Therefore, self-prescribing a functional brace is not appropriate for healthy populations. Prescriptions and referrals from a doctor are necessary.

These results are also useful for the wearer to account for other aspects of their gait. For example, if an Ossur Rebound knee brace is worn, flip-flops should not be worn in addition to the brace since previous studies have found that flip-flops may reduce stride length, increase ankle angles, and increase the internal joint load compared to a traditional sneaker (Shroyer et al. 2010). By wearing appropriate footwear, such as a sneaker, additional internal loads on the lower extremity joints may be prevented when wearing a knee brace.

The results of the study are solely based off of gait passes that may also suggest these internal loads to be exacerbated during a high load condition such as running. For example, running increases the demands on lower extremity joint loading with velocity alterations (Zeni et al. 2011). If significant findings are found in gait, then it can be hypothesized that a high joint load, such as running, would amplify the findings. In addition, the redistribution of moments may suggest a reorganization of motor control to adjust for the knee brace. Further research is needed to investigate these motor behavior alterations to assist in the development of knee braces. Overall, self-prescribing a knee brace could potentially lead to an increase in lower
extremities injuries due to the increased internal loads at the hip, knee, and ankle, therefore, functional braces are not recommended on a healthy population.
LIST OF REFERENCES


Winter et al reported minor variation in joint kinematics at increasing cadences, except for slight increases in the peak knee flexion during the loading response phase.


APPENDICES
APPENDIX A: TABLES
Table 1: Knee Brace Characteristics

<table>
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*Circumference measurements taken 15cm above mid-patella

Table 2: Participant Characteristics and Anthropometrics

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Table 3: Mean Angle in Lower Extremity at Left Leg Heel Strike

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<th>Mean Angle in Lower Extremity</th>
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<th>MB</th>
<th>AB</th>
<th>ANB</th>
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<td><strong>Left Leg</strong></td>
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<tr>
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<td>28.38±5.82</td>
<td>28.16±4.37</td>
<td>30.66±6.59</td>
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<td>Hip Internal/External Rotation (°)</td>
<td>10.55±3.75</td>
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<td>Hip Abduction/Adduction (°)</td>
<td>17.87±5.85</td>
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<td><strong>Right Leg</strong></td>
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Table 4: Mean Angle in Lower Extremity at Right Leg Heel Strike

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<td><strong>Left Leg</strong></td>
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<td>Hip Flexion/Extension (°)</td>
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<td>31.32±8.61</td>
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Table 5: Mean Moment in Lower Extremity

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Table 6: Mean Time to Peak Moment in Lower Extremity

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<tr>
<th>Mean Time to Peak Moments in Lower Extremity (Percent of Step)</th>
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<th>AB</th>
<th>ANB</th>
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<td><strong>Left Leg</strong></td>
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</tr>
<tr>
<td>Hip Flexion/Extension</td>
<td>5.01±3.23</td>
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<td><strong>Right Leg</strong></td>
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<tr>
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<td>68.24±25.18</td>
<td>71.93±21.69</td>
<td>55.97±30.02</td>
</tr>
<tr>
<td>Knee Abduction/Adduction</td>
<td>33.82±29.11</td>
<td>27.56±25.02</td>
<td>24.55±19.61</td>
<td>31.72±24.83</td>
</tr>
<tr>
<td>Ankle Flexion/Extension</td>
<td>81.51±1.02</td>
<td>81.83±1.19</td>
<td>81.48±1.39</td>
<td>80.44±4.52</td>
</tr>
<tr>
<td>Ankle Internal/External Rotation</td>
<td>36.32±27.58</td>
<td>44.67±25.29</td>
<td>48.38±27.99</td>
<td>42.79±24.83</td>
</tr>
<tr>
<td>Ankle Abduction/Adduction</td>
<td>75.13±4.75</td>
<td>75.22±6.53</td>
<td>73.72±9.25</td>
<td>73.59±6.19</td>
</tr>
</tbody>
</table>
APPENDIX B: FIGURES
Figure 1. Lower Body Plug-in-Gait Marker Placement from the Helen Hayes marker system

Figure 2. Ossur Rebound® knee brace
Figure 3. Gait Cycle as defined by Jacqueline Perry
Figure 4: Left Leg Joint Moments – Morning, No Brace Condition
Figure 5: Left Leg Joint Moments – Morning, Brace Condition
Figure 6: Left Leg Joint Moments – Afternoon, No Brace Condition
Figure 7: Left Leg Joint Moments – Afternoon, Brace Condition
Figure 8: Right Leg Joint Moments – Morning, No Brace Condition
Figure 9: Right Leg Joint Moments – Morning, Brace Condition
Figure 10: Right Leg Joint Moments – Afternoon, Brace Condition
Figure 11: Right Leg Joint Moments – Afternoon, No Brace Condition
Figure 12: Mean Angle at Hip during Left Step

Figure 13: Mean Angle at Knee during Left Step

Figure 14: Mean Angle at Ankle during Left Step
**Figure 15:** Mean Angle at Hip during Right Step

**Figure 16:** Mean Angle at Knee during Right Step

**Figure 17:** Mean Angle at Ankle during Right Step
**Figure 18:** Mean Moment on the Hip

![Figure 18](image)

**Figure 19:** Mean Moment on the Knee

![Figure 19](image)

**Figure 20:** Mean Moment on the Ankle

![Figure 20](image)
Figure 21: Time to Peak Moment at Hip

Figure 22: Time to Peak Moment at Knee

Figure 23: Time to Peak Moment at Ankle
APPENDIX C: CONSENT FORM
Consent Form

Study Title: The Influence of Self-Prescribed Bracing of the Knee on Walking in Healthy Adults

Investigator
Lauren A. Luginsland, B.S.
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240 Turner Hall
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University, MS 38677
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Martha Ann Bass, Ph.D.
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232 Turner Hall
University of Mississippi
University, MS 38677
(662) 801-6865
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By checking this box I certify that I am 18 years of age or older.

The purpose of this study
We want to know whether a person alters their walking when wearing a knee brace over an 8-hour time period.

What you will do for this study
1. You will come to the Applied Biomechanics Laboratory (ABL) located in the Turner Center at The University of Mississippi twice during the testing day (once in the morning for approximately 1 hour and once 8 hours post your initial visit for approximately 30 minutes). For female participants, you will be asked about the start date of your last period.

2. Upon agreement to participate, your height, weight, leg length, and leg width will be measured. You will be asked if a ball came rolling towards you what leg would you kick it with in order to determine your dominant leg.

3. If you have not already done so, you will be asked to change into compression shorts which will be provided by the researchers at the time of testing. You will be asked to wear your own shoes for the entire duration of the study.

4. In order to track body motion, retro-reflective markers will be placed on anatomical landmarks of the lower body. Markers will be applied to your skin, feet/shoes and clothing via double-sided adhesive tape. Immediately following, you will stand up straight with your arms outstretched and palms facing forward so the motion capture system can capture a static trial.

5. You will then walk 10 times across the capture zone (approximately 30 feet) with each trial separated by a minute. After the 10 passes, you will put on the knee brace and a new static trial will be taken on the motion capture system. You will then walk 10 times
across the capture zone with each trial separated by a minute. After the 10 passes, you will have the markers removed and you can proceed with your everyday activities while wearing the brace. The brace will be worn all day and you will return to the laboratory in 8 hours. When you return, you will have the retro-reflective markers re-placed on you and you will do the 10 passes with the brace on with each pass separated by 1 minute. After the ten passes, you will take the brace off and complete 10 passes each separated by 1 minute.

**Time required for this study**
This study will take about 1 hour for the first session in the morning and 30 minutes for the second session during the day – for a total of 1.5 hours.

**Possible risks from your participation**
Although highly unlikely, walking a total of 40 passes in the laboratory as well as walking throughout the day may result in a muscle or joint injury. You may experience some muscle soreness 24 to 48 hours after testing, but this should go away within a few days. Skin irritation and chafing may occur from wearing the brace.

**Benefits from your participation**
You should not expect benefits from participating in this study. However, you might experience insight into the potential benefits of utilizing a specific knee brace and how that may alter your walking.

**Confidentiality**
Any information obtained from or for this research study will be kept as confidential (private) as possible. You will not be identifiable by name or description in any reports or publications about this study. The records identifying your name will be (1) stored in a locked cabinet and/or in a password-protected computer file, (2) kept separate from the rest of the research records, and (3) be accessible to only the researchers listed on the first page of this form and their staff.

**Right to Withdraw**
You are not required to participate in this study. If you decide to participate, but later change your mind, you can withdraw at any time. There are no penalties or consequences of any kind if you decide to withdraw. Please inform the principal investigator that you will not be returning to the study. The participation in this study may be terminated at any time by the investigators if they believe that it is in their best interest to do so or if they fail to follow the study procedures.

**Compensation for Illness or Injury**
I understand that I am not waiving any legal rights or releasing the institution or their agents from liability from negligence. I understand that in the event of physical injury resulting from the research procedures, The University of Mississippi does not have funds budgeted for compensation for 1) lost wages, 2) medical treatment, or 3) reimbursement for such injuries. The University will help, however, obtain medical attention which I may require while involved in the study by securing transportation to the nearest medical facility.
IRB Approval
This study has been reviewed by The University of Mississippi’s Institutional Review Board (IRB). The IRB has determined that this study fulfills the human research subject protections obligations required by state and federal law and University policies. If you have any questions or concerns regarding your rights as a research participant, please contact the IRB at (662) 915-7482 or irb@olemiss.edu.

Please ask the researcher if there is anything that is not clear or if you need more information. When all your questions have been answered, then decide if you want to be in the study or not.

Statement of Consent
I have read the above information. I have been given an unsigned copy of this form. I have had an opportunity to ask questions, and I have received answers. I consent to participate in the study. Furthermore, I also affirm that the experimenter explained the study to me and told me about the study’s risks as well as my right to refuse to participate and to withdraw.

Signature of Participant __________________________ Date ___________

Printed name of Participant ______________________________________ Date ___________

NOTE TO PARTICIPANTS: DO NOT SIGN THIS FORM IF THE IRB APPROVAL STAMP ON THE FIRST PAGE HAS EXPIRED
VITA

LAUREN A. LUGINSLAND, BS
3 English Court
Bridgewater, NJ 08807
Email: lluginsland@gmail.com
Cell Phone: (908) 528-4402

EDUCATION
The University of Mississippi, Oxford, MS  Anticipated: 5/2018
Master of Science in Exercise Science, Concentration: Biomechanics
Thesis Title: “The Influence of Self-Prescribed Bracing of the Knee on Walking in Healthy Adults”

The University of Scranton, Scranton, PA  5/2016
Bachelor of Science in Exercise Science, Minor: Psychology, Concentration: Nutrition
Research Project: “Influence of a Knee Brace on Ipsilateral Hip Kinematics of Various Slopes”

PROFESSIONAL EXPERIENCE
The University of Mississippi, Oxford, MS  8/2016-Present
Graduate Assistant/Instructor
• Primary instructor for undergraduate lecture, laboratory, and exercise leisure courses
  o Biomechanics, Introduction to Exercise Science, First Aid/CPR, Weight Lifting, Jogging
• Undergraduate Group Advisor
  o Advise undergraduate students with scheduling of academic courses
  o Department of Health, Exercise Science, and Recreation Management

The University of Scranton, Scranton, PA  8/2014-5/2016
Teacher Assistant
• Teacher assistant for undergraduate lecture course and laboratory
  o Biomechanics of Human Movement

RESEARCH EXPERIENCE
NIOSH and CDC, Morgantown, WV  Anticipated: 5/2018 – 8/2018
Oak Ridge Institute for Science and Education (ORISE) program
• Guest Researcher

The University of Mississippi, Oxford, MS  8/2016-Present
Research Assistant in the Applied Biomechanics Laboratory
• Prepare, set-up, and calibrate equipment
• Apply marker set to subjects
- Collect and reduce data
- Assisted with studies on various populations such as Collegiate Division I Softball & Baseball teams

**The University of Scranton, Scranton, PA** 1/2015-5/2016

**Research Assistant in the Human Movement and Ergonomics Laboratory**
- Collect and reduce data
- Prepare, set-up, and calibrate laboratory and equipment
- Equipment responsibility: motion capture system, force plates, EMG system
- Assisted with various studies such as: blood glucose levels influence on balance, Olympic lifts, dual-task activities effect on gait, different footwear on inclined surfaces, psychological stress on gait

**RESEARCH FUNDING/TRAVEL GRANTS**
- The University of Mississippi Department of Health, Exercise Science, and Recreation Management Travel Grant (2017 & 2018)
- The University of Mississippi Graduate School Travel Grant (2017)
- The University of Scranton Office of Research and Sponsored Programs Travel Grant (2016)
- The University of Scranton Panuska College of Professional Studies Travel Grant (2016)
- The University of Scranton Office of Research and Sponsored Programs Travel Grant (2015)

**PEER-REVIEW MANUSCRIPTS**

**ABSTRACTS**


Hill CM, Wilson SJ, Mouser JG, Williams CC, Luginsland LA, Donahue PT, Chander H. Impact Of


Williams, CC; Gdovin JR; Wilson SJ; Hill CM; Donahue, PT; Luginsland LA; Eason JD; Yarbrough AL; Wade C; Garner JC. Examining Changes in Bat Angle at Ball Contact in Collegiate Softball Players over a Fall Softball Season. Southeast Chapter of the American College of Sports Medicine (SEACSM) Annual Meeting, Chattanooga, Tennessee February 15-17th.


RESEARCH COLLABORATIONS
Mississippi State University, Department of Kinesiology
Missouri State University, Department of Kinesiology

TEACHING EXPERIENCE
Undergraduate Instructor:
**ES 447: Biomechanics of Human Movement Laboratory**
An introductory course that exposes students to the concepts of mechanics as they apply to human movement- particularly those pertaining to exercise, sport, and physical activity.
Students obtain the understanding of the mechanical and anatomical principles that govern human motion.

HP 203: Red Cross Responding to Emergencies (First Aid/CPR)
This course acts as an introduction to safety instruction and practices as prescribed in the American Red Cross standards and advanced courses. Students learn how to prevent accidents and to care for individuals in the event of an emergency. New Methods of accident prevention, first aid techniques, and cardiopulmonary resuscitation (CPR) are taught with a hands-on approach.

ES 100: Introduction to Exercise Science
An introduction course to exercise science, with an emphasis on career planning and student development. This course provides an overview of the field of exercise science, its development, professional activities, and sub-disciplines.

EL 151: Weight Lifting
This course is an interactive introductory course to weight lifting. The course covers exercises and skills associated with weight lifting and provides the students with the knowledge to develop their own weight lifting program.

EL 156: Jogging
An introduction to the essential fundamentals and techniques of jogging. The course is interactive and designed with a lecture and lab component to develop the knowledge of the benefits of jogging and different training regimens.

Teaching Assistant:
EXSC 313: Biomechanics of Human Movement
An introduction course to the principles and analysis of biomechanics, emphasizing the contribution of biomechanics to understanding human movement, and develops an understanding of mechanical and anatomical concepts related to human performance in various biomechanics disciplines.

TECHNIQUES/SKILLS
Biomechanical Analysis Equipment:
- 12 camera Motion Analysis system with Cortex version 3.6 Motion Capture system (Motion Analysis Corporation, Santa Rosa CA, USA)
- 8 Vicon M2 cameras with a Vicon 612 Datastation (Vicon, Oxford, UK)
- 8 channel Wireless EMG System (Noraxon, Scottsdale AZ, USA)
- 4 force plates (Advanced Mechanical Technology, Inc., Watertown MA, USA)
- 2 Canon GL-2 video cameras (Canon, Lake Success NY, USA)
- Retroreflective markers (3M, St. Paul MN, USA)
- Power Plate Whole Body Vibration Platform (Performance Health Systems, Northbrook IL, USA)
- Biodex System 3 (Biodex Medical Systems Inc., Shirley, NY, USA)
- Neurocom® Equitest
RESEARCH INTERESTS

- Gait Analysis: healthy and pathological gait
- Kinematics and kinetic alterations during gait with restrictive apparatus
- Footwear effect on gait
- Effects on balance during footwear and surface alterations
- Lower body kinematics during activities of daily living

PROFESSIONAL LICENSES AND CERTIFICATES

- First Aid/CPR/AED Instructor
  - Issued by the American Red Cross Expires: April 2019
- Blood borne Pathogens Training
  - Issued by the American Red Cross Expires: June 2019
- Adult/Pediatric First Aid
  - Issued by the American Red Cross Expires: June 2019
- Adult/Pediatric CPR
  - Issued by the American Red Cross Expires: June 2019
- Adult/Pediatric AED
  - Issued by the American Red Cross Expires: June 2019

MEMBERSHIPS

- Member of Gamma Beta Phi
  - The University of Mississippi Chapter
- Honors Society Member at The University of Mississippi
- Student Member of the American Society of Biomechanics
  - Presented at the 41st conference in Boulder, CO
  - Presented at the 40th conference in Raleigh, NC
  - Attended the 39th conference in Columbus, OH
- American College of Sports Medicine
  - Member of the South East Chapter
  - Abstract submitted for the annual conference in February 2018
  - Presented research poster at the annual conference in February 2017
- University of Scranton Women’s Crew Team
  - Member from August 2012-August 2015
- University of Scranton Exercise Science Club
  - Member from August 2012-May 2016
  - Participated in events (Powerlifting Competition, 5k races & local Healthy Heart Fair)
- University of Scranton Physical Therapy Club
  - Member from August 2012- May 2016
  - Participated in events (Wheelchair basketball: raise awareness for spinal cord injuries)
- University of Scranton Circle K
  - Member from August 2014-2015
  - Participated in the Street Sweep to clean the streets of the community
- University of Scranton Habitat for Humanity
  - Member from August 2014-2015
  - Raised awareness for homelessness
CLEARANCES
ACT 33 – Pennsylvania Child Abuse History Clearance Received: Fall 2014
ACT 34 - Pennsylvania State Police Criminal Record Check Received: Fall 2014
ACT 73 – FBI Criminal Background Check Received: Fall 2014

HONORS/AWARDS
Honors Society (Gamma Beta Phi)
Graduate Assistantship in the Department of Health, Exercise Science, and Recreation Management
(University of Mississippi)
16th Annual Celebration of Student Scholars
(University of Scranton)
The University of Scranton Dean’s List
The Xavier Grant
The Ellie Schifano Memorial Scholarship
The Panther Athletic Club Award
Knights of Columbus #5959 Scholarship

COMMUNITY OUTREACH
National Biomechanics Day 2017 (April 2017)
Full responsibility to plan and organize students, transportation, and schedule of the events for the students of Oxford High School, MS.

University of Success
Assisted local, economically disadvantaged, and/or students who are underrepresented in higher education who will be first-generation college bound students, to learn though activities held in the Human Ergonomics Movement Laboratory.

National Biomechanics Day 2016
The University of Scranton Human Movement and Ergonomics laboratory was open to the public, such as the students of Wallenpaupack Area High School, PA, in hopes for them to learn more about biomechanics and everyday implications through our labs instruction.

Science Fair Judge (Oxford Intermediate School, MS)
United Neighborhood Center (UNC)
Jewish Community Center (JCC)
Gino Merli Veterans Center, Scranton PA
Rebuilding Together (Miami, FL and Hopewell, VA)
Arrow Physical Therapy (Somerville, NJ)
Somerset Medical Physical Therapy (Bridgewater, NJ)

RELEVANT COURSEWORK
The University of Mississippi
ES 514 Applied Electromyography (Spring 2017)
ENGR 598 Special Projects in Engineering – Programming (Spring 2017)
ES 651 Integrative Calculus (Fall 2017)
ES 612 Instrumentation and Analysis in Biomechanics (Spring 2017)
ES 512 Foundations of Biomechanics (Fall 2016)
ES 609 Motor Control and Learning (Fall 2016)
ES 611 Exercise Physiology (Fall 2016)

The University of Scranton
EXSC 382 Gait Analysis (Fall 2015)
EXSC 482 Research Methods in Biomechanics and 3D Rotations (Spring 2016)
EXSC 313 Biomechanics of Human Movement and Sport (Spring 2014)
EXSC 448 Research Methods (Spring 2016)