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A comparative evaluation of the effects of incline on kinematics and muscle function during backward walking

Daniel Brent Jensen

University of Nevada, Las Vegas, dbj460@gmail.com

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A COMPARATIVE EVALUATION OF THE EFFECTS OF INCLINE ON
KINEMATICS AND MUSCLE FUNCTION DURING BACKWARD WALKING

by

Daniel Brent Jensen
Bachelor of Science in Exercise Science
Utah Valley University
2012

A thesis submitted in partial fulfillment
of the requirements for the

Master of Science - Kinesiology

Department of Kinesiology and Nutrition Sciences
School of Allied Health Sciences
Division of Health Sciences
The Graduate College

University of Nevada, Las Vegas
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**Daniel Brent Jensen**

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**Department of Kinesiology and Nutrition Sciences**

Janet Dufek, Ph.D., Committee Chair

John Mercer, Ph.D., Committee Member

James Navalta, Ph.D., Committee Member

Robbin Hickman, Ph.D., Graduate College Representative

Kathryn Hausbeck Korgan, Ph.D., Interim Dean of the Graduate College

**August 2014**
ABSTRACT

A comparative evaluation of the effects of incline on kinematics and muscle function during backward walking

By

Daniel Brent Jensen

Dr. Janet S. Dufek, Examination Committee Chair
Professor of Kinesiology and Nutrition Sciences
University of Nevada, Las Vegas

The purpose of this study was to investigate lower extremity kinematics and muscle activation patterns in inclined backward walking (IBW) versus traditional backward walking (BW). This purpose was achieved by evaluating performance of individuals walking backward on a treadmill at inclines of 0%, 6%, 8%, 10%, and 12%. Eleven participants (24.9 ± 4.8 yrs, 71.9 ± 11.6 kg, 1.7 ± 0.09m) recruited from the UNLV student body went through a familiarization training program before any data were collected. The familiarization training, which took place across 3 different days, served to: 1) help the participants become comfortable and familiar with BW and IBW, and 2) find each participant’s preferred speed of walking at each tested incline. During each training day, participants walked at each incline for five minutes, totaling in 25 minutes of BW and IBW. Participants completed data collection within 1 week of familiarization. Data collection consisted of a warmup followed by attachment of surface EMG electrodes to the right leg to the rectus femoris (RF), tibialis anterior (TA), medial gastrocnemius (MG), and the biceps femoris (BF) muscles. The EMG signals (2000 hz) were recorded with a Noraxon MyoSystem 2000 (Noraxon, Scottsdale, AZ). Maximal voluntary contraction (MVC) exercises for each monitored muscle were performed and recorded. Next, 35 reflective markers were attached (Vicon Fullbody PlugInGait model)
for motion capture (200 hz) with a 12 camera Vicon nexus 3D motion capture system (Vicon Motion Systems Ltd, Oxford, UK). Participants next walked for five minutes at each incline (0%, 6%, 8%, 10%, 12%), presented in random order. Stance and swing phase for the left and right lower extremities were determined from kinematic data and average EMG for each muscle and condition was assessed for each phase. Lower extremity ankle, knee and hip joint kinematic parameters included position, velocity and acceleration at initial contact and range of motion during stance and swing. Step rate was also measured. One way repeated measure ANOVAs were conducted for each dependent variable. Pairwise comparisons were conducted between 0% and each incline ($\alpha=0.05$, Bonferroni correction). Results indicated that as incline increased, preferred speed decreased (0% = 1.00 ± 0.14 m/s, 6% = 0.96* ± 0.15 m/s, 8% = 0.95* ± 0.16 m/s, 10% = 0.92* ± 0.18 m/s, and 12% = 0.91* ± 0.18 m/s). The ankle joint angle was different at 10% (p = .048) and at 12% (p = 0.01). The knee joint angle was different at all inclines (6% - p <0.01, 8% - p = 0.017, 10% - p < 0.01, 12% - p < 0.01). The hip joint angle was different at 10% (p = 0.022) and at 12% (p <0.01). During stance, the ankle joint went through a significantly greater range of motion (ROM) at 10% (p <0.01) and 12% (p = 0.017), the knee joint showed significant differences in ROM across all inclines (6% (p < 0.01), 8% (p = 0.016), 10% (p < 0.01), and at 12% (p < 0.01)), and the hip joint was significantly different at 6% (p < 0.01), 8% (p < 0.01), 10% (p < 0.01), and at 12% (p < 0.01). During swing, only the hip joint demonstrated significantly different ROM at all inclines (6% (p < 0.01), 8% (p <0.01), 10% (p < 0.01) and 12% (p < 0.01)). Step rate was significantly different at 10% (p = 0.022) and 12% (p < 0.01). Only the RF muscle showed any significant changes (8%: p = .039; 12%: p = 0.013). $RF_{swing}$ was significant
at 8% (p = 0.013), 10% (p = 0.017), and 12% (p = 0.013). The observed increased knee extensor activity, increased ROM, with no changes in hamstring or gastrocnemius muscle activity may suggest IBW as a potential intervention for knee rehabilitation.
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CHAPTER 1

Introduction

The world of rehabilitation revolves around improving the quality of life for someone who has been injured. From athletes to the common man, rehabilitation is about returning the injured back to where they were physically before the injury or as close to that condition as possible.

There are many tools used in rehabilitation, and one of those tools is backward locomotion (BL). BL includes walking backward and running backward. BL has been shown to have many beneficial outcomes that are applicable to both the athletic world and the clinical world. For example, BL has been shown to improve an individual’s balance (Hao & Chen, 2011), increase hamstring flexibility (Whitley & Dufek, 2011), and decrease lower back pain (Dufek, House, Mangus, Melcher, & Mercer, 2011).

Backward walking (BW) has also been shown to decrease patellofemoral pain (Flynn & Soutas-Little, 1995; Myatt et al., 1995; Roos, Barton, & van Deursen, 2012). This decrease in patellofemoral pain occurs in part through an increase in knee extensor strength (Threlkeld, Horn, Wojtowicz, Rooney, & Shapiro, 1989). BW has been shown to increase the activity of the knee extensor muscles (Grasso, Bianchi, & Lacquaniti, 1998). That increased muscle activity can lead to an increase in knee extensor strength when incorporated into training regimens.

Recently, inclined backward walking (IBW) has been recommended as a rehabilitation method (Shankar, Bhandiwad, & Pai, 2013). The challenge is that IBW has had very little research done to support using it for rehabilitation. The implementation of such an under-researched modality is risky, and goes against the concept of evidence-
based practice. IBW could result in more damage to current injuries, it could produce a placebo effect without really helping beyond the benefits than traditional BW, or it could truly be beneficial. Further research is needed before IBW is widely used as a modality of rehabilitation.

The differences in BW and forward walking (FW) were examined extensively by Grasso, Bianchi, & Lacquaniti (1998). BW is different from FW primarily because the motion of the foot contact is in reverse order, meaning that the forefoot typically comes in contact with the ground before the heel. The metatarsal joints at the distal end of the foot articulate with the toes to act as a more compliant support surface than the posterior extremity of the tarsus articulating with the leg and shank. The differences in the flexibility of the extremities appeared to change the initial force spike of foot contact, as observed in force-time profiles (Grasso et al., 1998). Other differences observed by Grasso et al. (1998) were the electromyography (EMG) signals. Some muscle groups were triggered reciprocally instead of co-activated in BW compared to FW. The EMG patterns in BW did not resemble the EMG signals in FW and as such imply that FW and BW are more than just a reversal of movement (Grasso et al., 1998).

IBW is starting to make appearances in rehabilitation, but is not strongly evidence-based. Shankar et al. (2013) tested IBW but only at a single inclination of 15 degrees and did not compare their results to walking on flat ground. The researchers found that their inclined retro-walking (backward) was effective in overcoming and reducing the symptoms of knee joint osteoarthritis. Shankar et al. (2013) used the Visual Analog Scale, Western Ontario and McMaster universities (WOMAC) index as well as having their subjects self-report on any experienced pain. Shankar et al. (2013) found the results of the study to be
significant, but kinematic, kinetic, or EMG analysis were not used or included as part of the study. The only other reported data was a comparison between walking speed and distance walked pre-study and post-study. Cipriani, Armstrong, & Gaul (1995) measured IBW but only tested inclines of 5% and 10% and compared the results to those found during traditional BW done at an incline of 0%. The researchers measured lower extremity kinematics as well as EMG comparisons of the tibialis anterior, medial gastrocnemius, rectus femoris, and an unspecified hamstring muscle. In the conclusion, the researchers mentioned that the majority of the significant results were not observed at the 5% incline but only at the 10% incline. The researchers theorized that the majority of the analyzed variables actually changed with some incline between 5% and 10%.

BW has shown many clinically beneficial outcomes when incorporated into a rehabilitation program. Among these benefits is an increase in knee extensor activity which appears to help reduce patellofemoral pain. This increased activity is due to the muscle undergoing greater concentric contractions in order to aid in accelerating the leg backward (Winter, Pluck, & Yang, 1989). Taken together with the recent use of incline combined with backward walking (Shankar et al., 2013), it appears as though a more thorough understanding of the interactive effects on the biomechanics of gait would be beneficial.

**Purpose of the Study**

Therefore, the purpose of this study was to investigate the lower extremity kinematics and muscle activation patterns in IBW versus traditional BW. This purpose
was achieved by evaluating performance of individuals walking backward on a treadmill at inclines of 0%, 6%, 8%, 10%, and 12%.

**Research Question**

When walking backward, at what inclines do kinematics and muscle activity patterns differ compared to backward walking over a level surface?

**Significance of the Study**

There has been a paucity of research regarding inclined backward walking. The majority that has been done either used a slight incline of 5% or ramped up to an incline of 10% or steeper. The significance of this study is to expand the current body of literature relative to the biomechanical/functional understanding of IBW. Information gleaned from this study will provide empirical evidence to inform as to potential rehabilitation benefits of IBW.

**Statistical Hypotheses**

Two statistical hypotheses have been developed to address the primary research question. The null (Ho) hypotheses to be tested and alternative (Ha) hypotheses are:

- **Ho**\(_1\): There will be no differences in joint kinematics between 0% and each IBW.
- **Ha**\(_1\): There will be differences in joint kinematics between 0% and each IBW.
- **Ho**\(_2\): There will be no differences in lower extremity muscle activity between 0% and each IBW.
- **Ha**\(_2\): There will be differences in lower extremity muscle activity between 0% and each IBW.
**Limitations**

1) Footwear was not controlled or restricted beyond “closed toed shoes must be worn”. This may contribute to variability among subjects.

2) Regardless of individual subject dominant limb, the right leg was used for EMG measurement and analysis.

3) A treadmill was used for walking. Different results may have been observed while walking over ground.

**Delimitations**

1) Footwear was not controlled or restricted beyond “closed toed shoes must be worn”. This allowed the participants to walk more naturally.

2) Each individual participant established a preferred speed for each incline. This provides for a more clinically appropriate interpretation of the results.

3) A treadmill was used for walking. Most clinicians and rehabilitation centers have and use treadmills, making the current study more clinically applicable.

All participants completed a backward walking familiarization protocol prior to data collection. This reduced the potential short-term accommodation effects and/or learning from the interpretation of results.
CHAPTER 2

Review of Related Literature

Backward Walking

Backward walking (BW) is a method of locomotion that often is used by coaches and rehabilitation clinicians because of the health benefits that are associated with walking backwards. From an athletic standpoint, BW can increase foot coordination as well as overall endurance (Terblanche, Page, Kroff, & Venter, 2005). BW can also be used to improve cardiorespiratory fitness as well as body composition (Flynn, Connery, Smutok, Zeballos, & Weisman, 1994; Irving et al., 2008; Myatt et al., 1995; Terblanche et al., 2005). From a rehabilitation standpoint, BW has been shown to decrease patellofemoral joint compressive forces and reduce patellofemoral pain (Boucher, King, Lefebvre, & Pepin, 1992; Flynn & Soutas-Little, 1995; Roos et al., 2012; Sussman, Alrowayeh, & Walker, 2000). BW may be used to increase hamstring flexibility (Whitley & Dufek, 2011) as well as decrease back pain (Dufek et al., 2011; Dufek, Mercer, Aldridge, Melcher, & Gouws, 2009; Hodges & Richardson, 1999; Kim, Park, & Shim, 2010). Where poor balance is an issue, BW may be implemented in order to improve balance performance (Dufek et al., 2009; Hao & Chen, 2011; Threlkeld et al., 1989; Y. R. Yang, Yen, Wang, Yen, & Lieu, 2005).

With BW as widely used as it is, it is important to understand how BW is accomplished. Winter, Pluck, and Yang set out to determine if backward walking was a reversal of forward walking (Winter et al., 1989). With the use of electromyography (EMG) equipment, force platforms, and reflective markers, 6 subjects were tested in both the forward and backward directions. Joint angles, joint moments, muscle activity levels,
and muscle power production were all analyzed from the collected data. The first analysis reversed the backward walking so a visual comparison between forward and backward could be conducted. Even using experienced observers, no visual difference between the two different conditions was found (Winter et al., 1989). A kinematic analysis of the joint angles was conducted. With the hip and ankle joints, joint range of motion and joint angles were not significantly different across the conditions. The EMG signals showed that for the most part, going backward is simply a temporal reversal of the muscle contractions of forward locomotion (Winter et al., 1989).

Walking uphill backwards has intrigued some researchers (Cipriani et al., 1995; Shankar et al., 2013). The researchers set out to analyze and compare backward walking across 3 different inclines, 0%, 5%, and 10%. The equipment used included EMG equipment and reflective markers for sagittal plane motion analysis. (Cipriani et al., 1995). The EMG signals were normalized against data collected from walking forward at an incline of 0%. For analysis, the researchers looked at 4 different phases of gait – initial contact, midstance, heel-off, and swing phase. It was found that at 5% incline, there was little to no change in the EMG data or kinematic data when compared to going at 0%. At the 10% grade, all of the EMG signals were found to be significantly different from the 0% grade (Cipriani et al., 1995). For the kinematic analysis, the differences varied by the phase of gait. At initial contact, ankle dorsiflexion and knee flexion were different. During midstance and during the swing phase, knee flexion was different, and during heel-off, ankle plantar flexion was different (Cipriani et al., 1995). The researchers suggest that due to their results, inclined backward walking might further
enhance the rehabilitative benefits of backward walking found when walking on flat ground.

**Patellofemoral Pain**

The knee is generally considered to be one of the most unstable joints in the body. The instability of the knee joint allows for the potential for many different injuries or problems. Patellofemoral pain (PFP) is a common injury for the knee, especially for athletes. PFP accounts for about 25% or knee injury complaints amongst athletes (Taunton et al., 2002). While not limited to athletes, it is one of the most common injuries in athletics, especially amongst female athletes (Tumia & Maffulli, 2002). Women are twice as likely to develop patellofemoral pain than men.

PFP can be caused by many different mechanisms. Increased patellofemoral joint compression forces can be a major contributor to PFP (Roos et al., 2012) and the consistent overloading of the patellofemoral joint can lead to such chronic injuries like osteoarthritis (Buckwalter & Brown, 2004). Another mechanism for PFP is a misalignment of the patella inside the trochlear groove (Grelsamer, 2000). According to Grelsamer, one of the groups of people with PFP are those individuals who feel their patella slip within the trochlear groove. This subluxation of the patella can be the result of a few different mechanisms. One possible cause could be that the vastus medialis oblique can vary in its position/orientation or could be poorly synched with the surrounding muscles when undergoing contraction. It is also possible that the trochlear groove could also be misshaped or that the patella rests too distally or too proximally relative to the trochlear groove (Grelsamer, 2000).
The treatments for PFP can range from anti-inflammatory drugs to surgical intervention (Grelsamer, 2000). Ideally, the pain management should be as non-invasive as possible. BW has been shown to be effective at reducing PFP while remaining non-invasive. In a study analyzing patellofemoral joint compression forces (PFJCF), the effects exerted on the knee were analyzed in both backward running and forward running (Roos et al., 2012). While not the first to analyze PFJCF, the researchers set out to correct what they viewed as methodological flaws from previous studies. Roos et al. set out to correct what they viewed as methodological flaws in research conducted by others.

One study that Roos et al. were critical of was conducted by Flynn and Soutas-Little. In that study (Flynn & Soutas-Little, 1995), the PFJCFs of backward running were compared to those of forward running but the speed of the backward running was substantially different than the speed of the forward running. The researchers allowed the subjects to self-select the pace that was used in the backward direction and a different pace for the forward direction. The study found that the quadriceps were active substantially longer while moving backward compared to forward (Flynn & Soutas-Little, 1995). Despite the extra muscular activity, it was also discovered that the PFJCF loading occurred at a later point of the stance phase when going backwards. Flynn and Soutas-Little concluded that backward running, when done under supervision, could serve as a viable rehabilitation modality that would be fairly easy to supplement (Flynn & Soutas-Little, 1995).

Another study critiqued by Roos et al. was conducted by Sussman, Alrowayeh, and Walker. The researchers decided to test each subject, regardless of direction, at the
same speed (Sussman et al., 2000). In order to avoid subject injury, a slow running pace was selected. Unlike other studies, the PFCJF’s showed no differences between forward and backward when the speed was kept slow. Despite no difference in the compression forces, Sussman et al. found that when going backwards the knee exhibited a substantially greater knee joint range of motion. It was also found that when going backwards, the individual will undergo greater muscle activity in the leg, demonstrate increased leg extensor strength, and a minimized vertical ground reaction force.

Roos et al. controlled the speed at which their subjects underwent the test, although the speed used was significantly faster than the speed used by Sussman et al. (2000). Unlike Sussman et al. (2000), the PFJCFs of backward running were significantly different than those of forward running despite having similar directional speeds. However, the results were not consistent with every subject (Roos et al., 2012). It was concluded that backward running does in fact have lower PFJCFs than forward running and is a viable rehabilitation modality for reducing PFP. It was also concluded that with lower PFJCFs, backward running may prevent overloading of the patellar tendon and help ward off chronic injuries like osteoarthritis (Roos et al., 2012).

Recently, inclined backward walking (IBW) has been utilized as a rehabilitation modality (Shankar et al., 2013). With a treadmill set at a 15 degree slope, patients with clinically diagnosed knee osteoarthritis underwent a 10 day IBW treatment program. With use of the Western Ontario and McMaster (WOMAC) survey, it was concluded that inclined backward walking did significantly reduce the symptoms of osteoarthritis (Shankar et al., 2013). However, the WOMAC survey is a self-report survey that relies upon subjects truthfully answering the questions. Shankar et al. did not use any
kinematic, kinetic, or electromyography (EMG) analysis so the results of the study are arbitrary and difficult to apply toward scientific understanding.

Sloped walking (uphill and downhill) was analyzed by Lay, Hass, & Gregor (2007). The subjects were instrumented with EMG sensors and reflective markers and walked across a force-plate instrumented incline, with the slope of the incline set to a steep grade of ±39% (±21°). The researchers set out to analyze the joint kinetics and EMG activity while focusing on the hip and knee joints. It was concluded that when joint power absorption increased, the measured muscle activity levels were all related to the joint moment patterns. On the flipside, when more power was needed, the only changes in joint moments occurred around the primary joint. All other muscle activity levels remained un-phased. Despite having reflective markers, joint range of motion wasn’t analyzed (Lay et al., 2007). The researchers also did not make any statements to the effects on PFP or use of inclined locomotion as a rehabilitation modality.

The significance of these studies is that while backward locomotion has been shown to require significantly more energy than forward locomotion, it has been shown to decrease PFJCFs as well as decrease PFP. However, the effects of using an incline for such rehabilitation efforts remain mostly untested and further research is needed to help establish that relationship. While reducing PFP is one of the primary clinical applications for BW, it is not the only clinical reason for implementing a BW program.

Hamstring Flexibility and Reduction in Back Pain

Despite modern technology and diagnostic methods, specific pathologies and the mechanisms of lower back pain (LBP) remain relatively unknown. In up to 80% of individuals reporting to have lower back pain, modern diagnostics remain unable to link
the pain to any specific pathology (Foster, 2007). Pain can always be debilitating. When in pain, individuals are affected physically and mentally, and LBP is no exception. In athletics, the intensity of the pain being experienced does play a significant role in determining if the athlete continues to train or sees out some form of pain management. Effective, non-invasive modalities are preferred and new methods of rehabilitation are always being investigated by those responsible for the athlete’s recovery.

BW may provide a much needed, non-invasive rehabilitation modality that would be easily implemented by any athletic trainer or healthcare provider. It has been shown (Masumoto et al., 2009; Masumoto, Takasugi, Hotta, Fujishima, & Iwamoto, 2005, 2007) that when walking backwards in a water environment, metabolic costs were greater than when going forward (Masumoto et al., 2009; Masumoto et al., 2005, 2007). The same was true for core muscle activity. Other studies have also demonstrated the increased metabolic costs of going backward and the improved body composition that accompanies a backward training program (Flynn et al., 1994; Irving et al., 2008; Myatt et al., 1995; Terblanche et al., 2005). An increase in core muscle strength has been suggested as a crucial part in rehabilitation efforts in alleviating LBP (Hodges & Richardson, 1999).

Backed with this understanding, Dufek et al. (2011) investigated BW’s effectiveness in alleviating LBP as well as identifying any specific characteristics of BW that prove to be the most effective (Dufek et al., 2011). After 3 weeks of BW exercises, the control group and the treatment group both saw significant increases in gait velocity, low back range of motion, and stride parameters of stride length and stride rate. It was argued that the increased hip extension found with BW served to help unload the compressive forces found within the spinal discs, helping to reduce LBP (Bates,
Morrison, & Hamill, 1986; Leteneur, Gillet, Sadeghi, Allard, & Barbier, 2009). Overall, it was found that the implementation of BW does support the theory that BW helps reduce LBP while at the same time improving low back range of motion.

Hamstring flexibility is another characteristic that has an impact on LBP. There is a relationship between hamstring flexibility and LBP (Phalen & Dickson, 1961). When hamstring flexibility increases, the amount of LBP that is experienced decreases. If the hamstrings remain tight and non-flexible, LBP becomes worse. Whitley and Dufek (2011) set out to analyze the effects of BW on hamstring flexibility. It was determined that BW does cause a significant increase in hamstring flexibility. Since the subject population was healthy and free of LBP, it was concluded that BW does increase hamstring flexibility and that, in theory, helps reduce LBP. However, since none of the subjects had LBP, the researchers were unable to state any direct relationships between LBP symptoms and BW (Whitley & Dufek, 2011).

**Balance**

Falling remains a major issue for individuals of varying ages, young and old (Hao & Chen, 2011). In children, falling remains a major contributor to unintentional injuries as well as morbidity and mortality (Donroe, Gilman, Brugge, Mwamburi, & Moore, 2009; Jiang et al., 2010; Love, Tepas, Wludyka, & Masnita-Iusan, 2009). Amongst the elderly, falling can cause major medical issues as well as societal challenges. In adults older than 65 years of age, more than 25% will fall at least once every year (F. Yang, Anderson, & Pai, 2007). Finding easy methods of improving balance, especially within populations that traditionally struggle with balance control, is crucial for rehabilitation efforts. Some exercise that have been used in the past include Tae Kwon Do or Tai Chi
Chuan (Cromwell, Meyers, Meyers, & Newton, 2007; Wong & Lan, 2008; F. Yang et al., 2007), but those exercises can be difficult to learn for those who struggle with intellectual disabilities, coordination disorders, or are physically incapable of enduring such physical activity. One of the main benefits of using BW is that it is a much easier task to teach and implement than martial arts. It is also significantly less physically demanding than learning martial arts, allowing more people with a much greater spectrum of physical limitations to participate. It also allows those of all ages to exercise.

BW has been used in various studies as a treatment to improve balance (Dufek et al., 2009; Hao & Chen, 2011; Threlkeld et al., 1989; Zhang, Lin, Yuan, & Wu, 2008). In every one of those studies, the implementation of BW as an exercise caused a significant improvement in the subject’s balance performance, regardless of the population being tested. The populations tested between those studies include individuals who suffered strokes, school aged boys, 18 year old women, college aged adults, and healthy older adults.

In all of those studies, BW or backward running, was implemented differently and for differing durations between testing. Dufek et al used a BW intervention that had the subjects walk backwards for 10-15 minutes a day, 3 days a week for 4 weeks (Dufek et al., 2009). Hao & Chen had their subjects walk backward for 25 minutes a session. They had 2 sessions a week for a total of 12 weeks (Hao & Chen, 2011). Threlkeld et al. implemented a progressive backward running program. During the first two weeks, 5% of the subject’s forward running was to be replaced with backward running. After two weeks, the percentage of backward running was increased each week until 30% of their running was in the backward direction. The 30% backwards program was then
maintained for the last weeks, making the training program 8 weeks in total (Threlkeld et al., 1989). It seems that regardless of implementation style or the population that is being tested, the use of backward locomotion causes a significant improvement in balance.

**Summary**

BW is used in rehabilitation due to several rehabilitative benefits. BW has been shown to decrease patellofemoral pain (Roos et al., 2012), increase hamstring flexibility (Whitley & Dufek, 2011), decrease lower back pain (Dufek et al., 2011), and improve balance (Hao & Chen, 2011). While some studies have examined inclined backward walking (Cipriani et al., 1995; Shankar et al., 2013), more in-depth research needs to be conducted to truly understand the benefits of IBW. The effects of IBW on patellofemoral joint compressive forces are unknown. While the current study will not be directly testing patellofemoral joint compressive forces, those forces have been linked to quadriceps muscle activity (Boucher et al., 1992), and the current study is analyzing the effects of IBW on quadriceps muscle activity.

With inclined backward locomotion, there are a few main questions and/or concerns that need to be addressed. A possible concern with IBW research is choosing an appropriate incline to test. The American Disabilities Act has set a standard for all handicap ramps to be no greater than a slope with a ratio of 1” rise for every 12” of ramp (Act, 2010). That translates into an incline grade of 8.3%. Cipriani et al. (1995) found very little difference between 0% inclined BW and 5% inclined BW. It wasn’t until they surpassed the ADA’s guideline and tested at 10% that most of their significant results were found (Cipriani et al., 1995). Lay et al., (2007) went even further and tested inclines up to ±39%. Logically, if an incline is steep enough, then significant differences
will be found when compared against 0% grades. What appears to be lacking from the current collection of literature is what happens at inclines between 5% and 10%. This current study set out to establish the kinematic differences of backward walking on a flat surface and between inclined backward walking on slopes of 6%, 8%, 10%, and 12%. Differences of muscle activity will also be measured across the different conditions. The purpose of this study was to investigate the lower extremity kinematics and muscle activation patterns in IBW versus traditional BW. This purpose was achieved by evaluating performance of individuals walking backward on a treadmill at inclines of 0%, 6%, 8%, 10%, and 12%.
CHAPTER 3

Methodology

Subjects

Eleven subjects (24.9 ± 4.8 yrs, 71.9 ± 11.6 kg, 1.7 ± 0.09m) were recruited from the UNLV student body. They signed an Institutional Review Board (IRB) approved informed consent form prior to participation. The disqualification criteria included having any pain in the ankle, knee, or hip joints, or any previous injury that had required surgery in any of those joints. None of the recruited subjects were disqualified.

Instrumentation

A 12 camera (MX-T40-S cameras) Vicon Nexus 3D motion capture system (Vicon Motion Systems Ltd., Oxford, UK), recording at 200 Hz, was used to capture all movement. The subjects were instrumented with 35 reflective markers per the Vicon Fullbody Plug-In-Gait Model (Vicon Motion Systems Ltd., Oxford, UK). Electromyography (EMG) signals, measures of muscle activation, were collected from the right rectus femoris muscle (RF), the right tibialis anterior muscle (TA), the right medial gastrocnemius muscle (MG), and the right biceps femoris muscle (BF) at 2000 Hz using a Noraxon MyoSystem 2000 (Noraxon, Scottsdale, AZ). A Precor C956 treadmill (Precor, Woodinville, WA) was used for the walking measurement sessions. To help aid in the tracking of the reflective markers, the side handrails were removed from the treadmill. A single square wave device was hardwired into both systems to enable data synchronization between the Vicon Nexus and the Noraxon MyoSystem. To ensure subject safety, a harness attached overhead to a safety bar was worn at all times the subject was on the treadmill.
Protocol

Following granting of written consent to participate, the subjects were familiarized with the five different walking conditions (BW, IBW 6%, IBW 8%, IBW 10%, and IBW 12%). A familiarization training program of three sessions was implemented. These familiarization training sessions served two main objectives. The first objective was to assure that each subject was familiar and comfortable with IBW. The second objective of the training sessions was to find a preferred speed of locomotion at each incline.

During familiarization, before getting on the treadmill, each subject was given oral instructions on finding their preferred speed. After the instructions were given, each subject was secured into the safety harness and positioned on the treadmill. Preferred speed was then determined by increasing the treadmill speed until the subject was not able to walk comfortably. Once the preferred speed was achieved, the treadmill was brought back to a standstill (zero velocity). This process was repeated two more times and the average of the three trials determined to be the preferred speed at that grade. This process was repeated at 0% and each subsequent test incline. The subjects then walked backward for five minutes at each incline starting with the null incline (0%) and progressed in order to the 12% incline. At each incline, the previously mentioned protocol for determining a preferred speed was used. During all training sessions, the subjects were blinded to the speed of the treadmill. All three familiarization sessions were performed on separate days within the span of 1 week. The day of data collection took place within 1 week of completion of the third familiarization training session.
On the day of data collection, the subjects first underwent a warm-up of up to 5 minutes on the treadmill at their preferred backward speed, as determined during the familiarization sessions. After the warm-up, surface EMG electrodes were attached to the RF, TA, MG, and the BF. Prior to EMG electrode placement, each subject’s skin was prepared by shaving, cleaning, and abrading the skin over the desired muscle bellies. After the EMG leads were attached, maximal voluntary contraction (MVC) exercises for each monitored muscle were performed and recorded. The MVCs were collected while the subject was at rest and were done one muscle at a time. For the RF, the subject underwent knee flexion. While flexed, the researcher held onto the subject’s shin and prevented the subject from extending the knee. The MVC for the BF were obtained while the subject had their knee extended. The researcher held onto the subject’s calf and prevented the subject from flexing their knee. The MVC for the TA was acquired while the subject had their foot flat on the ground. The researcher placed their hands on the superior aspect of the subject’s foot and prevented the subject from putting their foot through dorsiflexion. For the MG MVC, the subjects started in a standing position with their feet flat on the ground. The researcher placed their hands on the subject’s shoulders and prevented the subject from performing a heel-raise with their right foot. After the MVCs were collected, the subject was placed into the safety harness, following which 35 reflective markers were applied to the subject according to the Vicon Fullbody PlugInGait protocol. Five different angles of incline were used; a null incline of 0%, along with inclines of 6%, 8%, 10%, and 12%. On the day of data collection, the order of the inclines was randomized. Each walking condition lasted for 5 minutes. The motion capture data and EMG data were recorded for the entire 5 minutes of each condition.
Thirty seconds of data were extracted from the full data set at the 4 minute mark. From that 30 seconds, data were extracted for 10 full strides and the average was used to represent the participant’s performance.

**Data Analysis**

After the data collection was completed, the kinematic data were filtered with a 4th order zero-lag Butterworth filter with a cutoff frequency of 6Hz. The EMG data and kinematic data were then imported into MATLAB for data analysis. Within MATLAB, the EMG data were filtered using a 4th order zero-lag butterworth filter with a cutoff frequency of 200 Hz. To analyze the EMG data, right foot contact with the treadmill was be used to determine step cycles. Following the protocol validated by Zeni, Rihards, & Higginson (2008), the heel markers were tracked to determine stance and swing phase. Using data from a motion-capture system, Zeni et al. (2008) used the anterior and posterior movements of the heel marker to determine foot contact. Foot contact was determined to be between the maximum displacement of the heel marker in the anterior direction and the maximum displacement of the heel marker in the posterior direction. Zeni et al. (2008) validated this method of determining foot contact across over 200 trials through the use of an instrumented treadmill. Following the protocol of Zeni et al. (2008), stance phase and swing phase for both feet were determined. With use of MVCs, the EMG data during stance phase ($RF_{stance}$, $BF_{stance}$, $MG_{stance}$, $TA_{stance}$) as well as during swing phase ($RF_{swing}$, $BF_{swing}$, $MG_{swing}$, $TA_{swing}$) were analyzed using root mean squared (RMS) values normalized to the MVC values for comparison between the different inclines. Position of the ankle joint (IAJ), knee joint (IKJ), and hip joint (IHJ) were evaluated at foot contact. Instantaneous angular velocity ($AJV$, $KJV$, $HJV$), and
instantaneous angular acceleration (AJ, KJ, HJ) were also computed for foot contact. The velocity and acceleration values were calculated using the first central difference method. In addition joint ROM (AJ_{stance}, KJ_{stance}, HJ_{stance}) across the entire support phase and across the swing phase (AJ_{swing}, KJ_{swing}, HJ_{swing}) were evaluated. The inclination of the foot (IF) at contact was also measured and analyzed. The analysis also included a time comparison of the subject’s step rate (right foot contact to left foot contact). The dependent variables analyzed were RF_{stance}, RF_{swing}, BF_{STANCE}, BF_{SWING}, MG_{STANCE}, MG_{SWING}, TA_{STANCE}, TA_{SWING}, IAJ, IKJ, IHJ, AJ_{stance}, KJ_{stance}, HJ_{stance}, AJ_{swing}, KJ_{swing}, HJ_{swing}, AJ_{stance}, KJ_{stance}, HJ_{stance}, IF.

**Statistical Analysis**

One-way repeated measures ANOVAs were conducted for each dependent variable. Statistical significance was set at $\alpha = 0.05$. Follow-up *post-hoc* tests were run as appropriate (with Bonferonni corrections) to identify the source of any observed significant differences between the 0% incline and the other tested inclines.
CHAPTER 4

Results

The purpose of this study was to investigate the lower extremity kinematics and muscle activation patterns in IBW versus traditional BW. This purpose was achieved by evaluating performance of individuals walking backward on a treadmill at inclines of 0%, 6%, 8%, 10%, and 12%.

All dependent variables were measured on all inclines (0%, 6%, 8%, 10%, and 12%). The results from inclines 6%, 8%, 10%, and 12% were all compared a priori to the 0% incline. The subjects walked for five minutes at each incline at a pre-determined preferred speed specific to that incline. The preferred speed values by subject at each incline are presented in Table 1. It was found that walking speed at all inclines was significantly different from the 0% speed (6% - p < 0.01, 8% - p < 0.01, 10% - p < 0.01, 12% - p < 0.01).

Table 1. Preferred walking speed (m/s) per subject at each incline.

<table>
<thead>
<tr>
<th>Subject</th>
<th>0%</th>
<th>6%</th>
<th>8%</th>
<th>10%</th>
<th>12%</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.89</td>
<td>0.85</td>
<td>0.80</td>
<td>0.80</td>
<td>0.76</td>
</tr>
<tr>
<td>2</td>
<td>1.21</td>
<td>1.16</td>
<td>1.12</td>
<td>1.07</td>
<td>1.03</td>
</tr>
<tr>
<td>3</td>
<td>0.85</td>
<td>0.80</td>
<td>0.80</td>
<td>0.76</td>
<td>0.76</td>
</tr>
<tr>
<td>5</td>
<td>0.94</td>
<td>0.89</td>
<td>0.89</td>
<td>0.85</td>
<td>0.85</td>
</tr>
<tr>
<td>6</td>
<td>0.76</td>
<td>0.72</td>
<td>0.67</td>
<td>0.58</td>
<td>0.58</td>
</tr>
<tr>
<td>7</td>
<td>1.16</td>
<td>1.16</td>
<td>1.16</td>
<td>1.16</td>
<td>1.16</td>
</tr>
<tr>
<td>8</td>
<td>0.89</td>
<td>0.89</td>
<td>0.85</td>
<td>0.85</td>
<td>0.85</td>
</tr>
<tr>
<td>9</td>
<td>1.16</td>
<td>1.16</td>
<td>1.16</td>
<td>1.16</td>
<td>1.16</td>
</tr>
<tr>
<td>10</td>
<td>1.07</td>
<td>0.98</td>
<td>0.98</td>
<td>0.94</td>
<td>0.89</td>
</tr>
<tr>
<td>11</td>
<td>1.03</td>
<td>0.94</td>
<td>0.94</td>
<td>0.89</td>
<td>0.89</td>
</tr>
<tr>
<td>12</td>
<td>1.03</td>
<td>1.03</td>
<td>1.03</td>
<td>1.03</td>
<td>1.03</td>
</tr>
</tbody>
</table>

Mean 1.00 ± 0.14 0.96* ± 0.15 0.95* ± 0.16 0.92* ± 0.18 0.91* ± 0.18

Speed Reduction 0% -3.7% -5.3% -8.1% -9.4%

* = Significantly different from 0% at p < 0.05
Abbreviations used below are defined in Appendix B, Table 4.

**Kinematics**

Mean and standard deviation values for all kinematic dependent variables are presented in Table 2. There were no significant differences found for AJV, KJV, HJV, AJA, KJA, HJA, FI, AROM\textsubscript{swing}, & KROM\textsubscript{swing}. IAJ showed no significant differences at the 6% incline or the 8% incline. IAJ did show a significantly different angle at 10% (p = 0.048) and at 12% (p = 0.01). IKJ showed significant differences at all inclines (6% - p < 0.01, 8% - p = 0.017, 10% - p < 0.01, 12% - p < 0.01). IHJ did not show significant differences at 6% or at 8%, but did show significant differences at 10% (p = 0.022) and at 12% (p < 0.01). AJRom\textsubscript{stance} showed no significant differences at 6% or at 8%, but did at 10% (p < 0.01) and 12% (p = 0.017). KJRomp\textsubscript{stance} showed significant differences at 6% (p < 0.01), 8% (p = 0.016), 10% (p < 0.01), and at 12% (p < 0.01). HJRomp\textsubscript{stance} was significantly different at 6% (p < 0.01), 8% (p < 0.01), 10% (p < 0.01), and at 12% (p < 0.01). AJRom\textsubscript{swing} and KJRomp\textsubscript{swing} did not show any significant differences. HJRomp\textsubscript{swing} showed significant differences at 6% (p < 0.01), 8% (p < 0.01), 10% (p < 0.01) and 12% (p < 0.01). StepRate did not show significant differences at 6% or at 8%, but did show significant differences at 10% (p = 0.022) and at 12% (p < 0.01).
Table 2. Kinematic Average and Standard Deviation Values by Incline

<table>
<thead>
<tr>
<th>Inclines</th>
<th>0%</th>
<th>6%</th>
<th>8%</th>
<th>10%</th>
<th>12%</th>
</tr>
</thead>
<tbody>
<tr>
<td>Foot Inclination (deg)</td>
<td>47.4 ± 10.2</td>
<td>48.2 ± 12.7</td>
<td>45.9 ± 8.8</td>
<td>45.6 ± 9.4</td>
<td>47.4 ± 10.4</td>
</tr>
<tr>
<td>Ankle Angle (deg)</td>
<td>82.3 ± 6.8</td>
<td>80.8 ± 6.2</td>
<td>80.6 ± 6.9</td>
<td>8.0* ± 6.0</td>
<td>79.6* ± 6.2</td>
</tr>
<tr>
<td>Knee Angle (deg)</td>
<td>156.6 ± 7.3</td>
<td>150.9* ± 7.7</td>
<td>149.7* ± 9.1</td>
<td>8.3</td>
<td>145.3* ± 8.5</td>
</tr>
<tr>
<td>Hip Angle (deg)</td>
<td>182.7 ± 8.0</td>
<td>181.8 ± 9.0</td>
<td>181.0 ± 9.3</td>
<td>9.3</td>
<td>179.0* ± 9.0</td>
</tr>
<tr>
<td>Ankle Joint Velocity (deg/sec)</td>
<td>-210.4 ±</td>
<td>-202.5 ±</td>
<td>-189.1 ±</td>
<td>-200.5 ±</td>
<td>178.2* ± 9.0</td>
</tr>
<tr>
<td>Knee Joint Velocity (deg/sec)</td>
<td>145.9 ± 51.5</td>
<td>134.2 ± 52.2</td>
<td>136.8 ± 49.9</td>
<td>45.7</td>
<td>136.2 ± 55.4</td>
</tr>
<tr>
<td>Hip Joint Velocity (deg/sec)</td>
<td>15.9 ± 18.1</td>
<td>7.9 ± 23.2</td>
<td>8.8 ± 19.5</td>
<td>6.5 ± 26.6</td>
<td>13.0 ± 25.3</td>
</tr>
<tr>
<td>Ankle Joint Acceleration (deg/sec/sec)</td>
<td>866.4 ±</td>
<td>1181.1 ±</td>
<td>3628.3 ±</td>
<td>2092.1 ±</td>
<td>1671.8 ±</td>
</tr>
<tr>
<td>Knee Joint Acceleration (deg/sec/sec)</td>
<td>-1208.2 ±</td>
<td>-1105.6 ±</td>
<td>-1157.1 ±</td>
<td>-1358.7 ±</td>
<td>-1689.8 ±</td>
</tr>
<tr>
<td>Hip Joint Acceleration (deg/sec/sec)</td>
<td>-1411.2 ±</td>
<td>-1298.2 ±</td>
<td>-1429.5 ±</td>
<td>-1648.7 ±</td>
<td>-1648.7 ±</td>
</tr>
<tr>
<td>Ankle ROM – Support (deg)</td>
<td>26.1 ± 5.1</td>
<td>27.6 ± 5.1</td>
<td>28.3 ± 5.9</td>
<td>6.0</td>
<td>30.6* ± 4.6</td>
</tr>
<tr>
<td>Knee ROM – Support (deg)</td>
<td>26.7 ± 6.3</td>
<td>32.0* ± 5.6</td>
<td>33.1* ± 6.0</td>
<td>8.6</td>
<td>36.5* ± 6.1</td>
</tr>
<tr>
<td>Hip ROM – Support (deg)</td>
<td>30.8 ± 4.5</td>
<td>27.1* ± 3.6</td>
<td>25.7* ± 4.9</td>
<td>5.7</td>
<td>22.9* ± 4.2</td>
</tr>
<tr>
<td>Ankle ROM – Swing (deg)</td>
<td>14.2 ± 4.2</td>
<td>12.0 ± 4.0</td>
<td>13.0 ± 4.3</td>
<td>13.3 ± 4.9</td>
<td>15.5 ± 5.4</td>
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<tr>
<td>Knee ROM – Swing (deg)</td>
<td>42.9 ± 9.4</td>
<td>45.6 ± 8.5</td>
<td>46.0 ± 8.5</td>
<td>50.5 ± 9.4</td>
<td>48.9 ± 8.3</td>
</tr>
<tr>
<td>Hip ROM – Swing (deg)</td>
<td>32.0 ± 4.7</td>
<td>28.1* ± 4.2</td>
<td>46.7* ± 4.9</td>
<td>6.0</td>
<td>24.4* ± 4.3</td>
</tr>
<tr>
<td>Step Rate (Hz)</td>
<td>.57 ± .05</td>
<td>.56 ± .06</td>
<td>.55 ± .07</td>
<td>.54* ± .08</td>
<td>.53* ± .07</td>
</tr>
</tbody>
</table>

* = Significantly different from 0% at p < .05;

EMG

Mean and standard deviation values or all EMG dependent variables are presented in Table 3. BF<sub>stance</sub>, MG<sub>stance</sub>, TA<sub>stance</sub>, BF<sub>swing</sub>, MG<sub>swing</sub>, and TA<sub>swing</sub> all showed no significant differences. RF<sub>stance</sub> did not show significant differences at 6% or 10% but did
at 8% (p = .039) and at 12% (p = 0.013). RF_{stance} showed no significant differences at 6% but did at 8% (p = 0.013), 10% (p = 0.017), and at 12% (p = 0.013).

Table 3. EMG Average and Standard Deviation Values by Muscle and Incline

<table>
<thead>
<tr>
<th>Inclines</th>
<th>0%</th>
<th>6%</th>
<th>8%</th>
<th>10%</th>
<th>12%</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>(%MVC)</td>
<td>(%MVC)</td>
<td>(%MVC)</td>
<td>(%MVC)</td>
<td>(%MVC)</td>
</tr>
<tr>
<td>RF-Support</td>
<td>42.3 ± 27.3</td>
<td>57.2 ± 50.8</td>
<td>66.2* ± 50.4</td>
<td>67.6 ± 61.8</td>
<td>94.2* ± 70.1</td>
</tr>
<tr>
<td>BF-Support</td>
<td>37.9 ± 24.3</td>
<td>33.0 ± 25.1</td>
<td>48.5 ± 55.2</td>
<td>41.7 ± 29.4</td>
<td>31.1 ± 13.4</td>
</tr>
<tr>
<td>MG-Support</td>
<td>89.4 ± 76.7</td>
<td>88.1 ± 58.4</td>
<td>86.4 ± 64.5</td>
<td>218.7</td>
<td>143.9 ± 127.5</td>
</tr>
<tr>
<td>TA-Support</td>
<td>52.3 ± 16.5</td>
<td>71.9 ± 45.9</td>
<td>66.9 ± 41</td>
<td>74.2 ± 43.0</td>
<td>76.6 ± 40.1</td>
</tr>
<tr>
<td>RF-Swing</td>
<td>35.3 ± 28.1</td>
<td>46.4 ± 34.1</td>
<td>54.9* ± 47.7</td>
<td>92.6* ± 87.6</td>
<td>75.0* ± 60.2</td>
</tr>
<tr>
<td>BF-Swing</td>
<td>25.2 ± 18.8</td>
<td>35.5 ± 30.3</td>
<td>35.2 ± 30.7</td>
<td>40.0 ± 41.0</td>
<td>32.1 ± 18.2</td>
</tr>
<tr>
<td>MG-Swing</td>
<td>72.3 ± 47.3</td>
<td>71.1 ± 58.1</td>
<td>72.6 ± 58.2</td>
<td>239.2</td>
<td>117.3 ± 103.9</td>
</tr>
<tr>
<td>TA-Swing</td>
<td>49.7 ± 26.5</td>
<td>56.8 ± 34.3</td>
<td>61.2 ± 30.1</td>
<td>48.9 ± 33.3</td>
<td>58.3 ± 36.5</td>
</tr>
</tbody>
</table>

* = Significantly different from 0% at p < .05;

In summary, the significant kinematic differences include the IAJ (significant at 10% and 12%), IKJ (significant at 6%, 8%, 10%, and 12%), and the IHJ (significant at 10% and 12%). All three lower extremity joints exhibited a significant difference in range of motion across the stance phase (AJRom_{stance} at 10% and 12%, KJRom_{stance} at all inclines, and HJRom_{stance} at 10% and 12%). During the swing phase, only the hip joint experienced a greater range of motion (HJRom_{swing} increased at 6%, 8%, 10%, and 12%). The step rate also increased at 10% and 12% inclines. Significant EMG differences were only found within the rectus femoris muscle, during both stance and swing phase (RF_{stance} increased at 8% and 12%, and RF_{swing} increased at 8%, 10%, and 12%). Overall, these results suggest that incline does influence lower extremity kinematics during backward gait.
CHAPTER 5

Discussion

Backward walking (BW) has shown many clinically beneficial outcomes when incorporated into a rehabilitation program. Among these benefits is an increase in knee extensor activity which appears to help reduce patellofemoral pain (Lay, Hass, & Gregor, 2007). This increased activity is due to the muscle undergoing greater concentric contractions in order to aid in accelerating the leg backward (Winter Pluck, & Yang, 1989). Recently, inclined backward walking (IBW) has become a popular rehabilitation modality but has been lacking the necessary research support. The purpose of this study was to investigate the lower extremity kinematics and muscle activation patterns in IBW versus traditional BW. This purpose was achieved by evaluating performance of individuals trained in backward walking performing backward on a treadmill at inclines of 0%, 6%, 8%, 10%, and 12%.

From a kinematic perspective, there were several observed differences between IBW and BW. These included the joint angles of the ankle, knee, and hip at foot contact. The ankle and hip positions at contact did not differ significantly from BW until the 10% incline, but the knee demonstrated significant changes with as little as 6% incline. As previously reported (Cipriani, Armstrong, & Gaul, 1995), the joint angles of the ankle, knee, and hip angle all changed as the incline increased. Cipriani et al. (1995) suggested that these variables did change but due to the methodology employed it was unclear at what incline those joint angles showed significant changes. The current study indicated that the ankle and hip joints did not show any significant changes in their position at contact until the 10% incline. The initial knee angle being significantly different at all
tested inclines also confirms the results reported by Cipriani et al. (1995) who found the knee angle to be different at a 5% incline.

There were observed changes in lower extremity joint range of motion (ROM) among conditions. During the stance phase, the ankle joint experienced a greater ROM at the steeper inclines of 10% and 12%. The ROM experienced by the hip joint and the knee joint became significantly different at every incline evaluated. During the swing phase, only the hip showed significant differences in ROM, also across all inclines. Winter, Pluck, and Yang (1989) suggested that the lower extremity joint movement patterns during BW are similar to a reversal of forward locomotion joint movement patterns. It is important to note that only the movement patterns of BW seem to be a reversal of forward locomotion, as observed by Grasso, Bianchi, & Lacquaniti (1998). Grasso, et al. (1998) found that the EMG signals of BW are not a mirror of forward locomotion EMG. The increased ROM experienced in the current study, especially at the hip, seem to follow the findings of Leroux, Fung, and Barbeau (2002). While this study did not implement BW or IBW, it did investigate postural strategies in adapting to uphill and downhill inclinations on a treadmill (-10%, -5%, 0%, 5%, and 10%). Leroux et al. (2002) found that at -10% incline, the hip became decreasingly flexed and the knee experienced greater flexion. At the -10% incline, it was also noted that stride length decreased. Leroux et al. (2002) suggested that those changes, especially at the hip, were key in facilitating power generation as well as task adaptation (walking at that slope). The authors concluded that postural adaptations are specific to the task. Drawing upon the conclusions of both Winter et al. (1989) and Leroux et al (2002), joint ROM during IBW at a 10% incline would approximately match joint ROM while walking forward at a
- 10% incline. In the current study, the increase in ankle, knee, and hip ROM across the different inclines appear to be postural adaptations to IBW.

The only other kinematic variable that showed significant differences was that of step rate, which was significantly faster than BW at the 10% and 12% inclines. The results show that since a preferred speed specific to each individual incline was used, the preferred speed at all inclines was slower than the preferred speed at 0%. A faster step rate could be an indication of more time spent in double support as an adaptation to the increased incline of walking. As mentioned previously, none of the analyzed lower extremity joints experienced a difference in angular velocity or in angular acceleration at contact. It is felt that the lack of any differences in velocity or acceleration is due to the fact that the subjects determined a different self-selected pace for each specific incline. The allowance for a different speed at each incline would have a direct impact on the angular velocities and angular accelerations being experienced at each joint. Cipriani et al. (1995) had the subjects select a single preferred speed to walk during a flat condition, and then that same speed for all inclined conditions (Cipriani et al., 1995). Lay, Hass and Gregor (2007) did not control for walking speed. Following the protocol, the participants in the current study determined a preferred speed to walk at specific to each incline, potentially allowing each participant to have five different preferred speeds. Table 1 shows the mean preferred speed at each incline as well as how much the preferred speed at each respective incline changed compared to the preferred speed at a 0% incline. Individual participant preferred speed values are given in Table 1.

It should be noted that the participants in this study were healthy and had been screened for lower extremity injury or past surgeries that could affect normal gait
characteristics. Despite their healthy state, the participants of the current study demonstrated a decrease in preferred speed at steeper inclines when given the opportunity to choose a preferred speed at each incline, as is seen in Table 1. Clinicians seeking to establish an IBW rehabilitation program might experience a greater reduction in preferred speed with their patients who will not be as healthy or injury free as the participants in the current study.

Cipriani et al. (1995) analyzed lower extremity kinematics as well as the muscle activity patterns of the rectus femoris, tibialis anterior, medial gastrocnemius, and medial hamstring muscle at inclines of 0%, 5%, and 10%. They concluded that the majority of the significant differences found occurred at the 10% incline, but due to the differences in inclines it was possible that performance might have changed at some incline between 5% and 10%. Their testing protocol did not examine midpoint inclines. The current study set out to fill that gap by testing at inclines of 6%, 8%, 10%, and 12%. In the current study, activity of the rectus femoris, during stance and swing phase, was observed to increase starting at the 8% incline. Unlike Cipriani et al. (1995), the current study did not find any other increases EMG as incline increased. The lack of increased activity of the biceps femoris muscle found in the current study is in harmony with the findings of Lay et al. (2007). A possible explanation for why the current study or the study done by Lay et al. (2007) did not find increased activity in the biceps femoris is that as the incline increased, the speed of locomotion decreased. The current study allowed the subjects to establish a preferred speed for each incline and it was found that as the incline increased, the preferred speed slowed down. The Lay et al. study (2007) did not control for or measure walking speed, so it is likely those participants also slowed down as the incline
got more severe. One possible explanation for the increase in EMG observed by Cipriani et al (1995) is that they evaluated the separate phases of the stance phase (initial contact, midstance, and heel-off). The only muscle they found to be significant at all three of those sub-phases was the tibialias anterior muscle. The medial gastrocnemius, rectus femoris, and the medial hamstring muscle were all significant at some of those subphases, but not all. Out of the analyzed muscles and across the analyzed subphases of gait, Cipriania et al (1995) found that the hamstring muscle had the fewest incidences of increased activity. Another possible cause for the EMG significance found by Cipriani et al (1995) centers around the familiarization training done by their subjects. The familiarization training only had the subjects walk backward on flat ground, but did not have them do any inclined backward walking until the day of data collection. This protocol helped their subjects become familiar with task of BW but did not remove the novelty of IBW. It is possible that when accomplishing such a novel task, the subjects experienced a substantial increase in muscle activity compared to what a trained individual would experience (Callewaert, Boone, Celie, De Clercq, & Bourgois, 2013). The current protocol included a training program in backward walking prior to data collection. Thus, the current study results are unique from data presented in the literature.

The current study differs in protocol from the study done by Cipriani et al (1995) in at least two ways. These protocol differences could help explain the differences in the results between this current study and the study done by Cipriani et al (1995). One difference is that in the current study, the entire stance phase was analyzed as a whole. The averaging of the entire stance phase together could contribute to the lack of significance found in medial gastrocnemius muscle or in the tibialis anterior muscle.
Another difference in protocol deals with the implemented familiarization training. The current study called for the subjects to walk backward for five minutes per incline at each incline for three familiarization days. Thus, each subject walked backward for 25 minutes, 20 minutes of which was IBW. This familiarization training is more thorough than that found in other IBW studies (Cipriani et al., 1995; Lay et al., 2007; Minetti & Ardigo, 2001) and was intended to allow each subject to become fully familiar, comfortable, and trained with the IBW task.

Within the current study, a possible explanation for the lack of some EMG significant results could stem from external factors such as gravity and a moving treadmill belt. The nature of BW and IBW is that the initial foot contact occurs slightly posterior to the individual’s projected COM and is contacted with the forefoot. Stance phase ends when the foot leaves the treadmill belt which when going backward is anterior to the subject’s position. When going backward, the end of stance phase occurs when the heel leaves the ground. When walking backward on a treadmill, once stance phase begins, the treadmill belt will pull the foot through stance phase until the subject picks up their foot. This mechanical pull caused by the treadmill could also help explain some of the lack of differences in some of the EMG signals.

Possible Clinical Relevance

Clinical benefits of BW previously reported include improvement in balance (Hao & Chen, 2011), an increase hamstring flexibility (Whitley & Dufek, 2011), and a decrease lower back pain (Dufek, House, Mangus, Melcher, & Mercer, 2011). BW has also been shown to decrease patellofemoral pain (Flynn & Soutas-Little, 1993; Myatt et al., 1995; Roos, Barton, & van Deursen, 2012; Shankar, Bhandiwad, & Pai, 2013).
According to Threlkeld et al. (1989), the decrease in patellofemoral pain is partially due to an increase in knee extensor strength. BW has been shown to have an increased amount of knee extensor muscle activity (Grasso et al., 1998). That increased muscle activity can lead to an increase in knee extensor strength when incorporated into training regimens. Through the current study, as well as the work of Cipriani et al. (1995), some characteristics about IBW have become fairly clear. A major difference between BW and IBW is that with IBW, the ankle, knee, and hip joints all undergo a greater ROM. Along with the increased ROM, both studies agree that the rectus femoris experiences an increase in activity. With the rectus femoris being a primary knee extensor muscle, the increased activity of the rectus femoris could lead toward increase in knee extensor strength.

Threlkeld et al. (1989) found that with an increase in knee extensor strength, a decrease in patellofemoral pain is highly likely to be experienced. BW has been shown to reduce patellofemoral compressive forces, also reducing patellofemoral pain (Flynn & Soutas-Little, 1995; Myatt et al., 1995; Roos et al., 2012). The combination of a reduction in patellofemoral compressive forces and the potential increase in knee extensor strength should result in a decrease in patellofemoral pain. The effects of IBW could prove beneficial in a rehabilitation program, however, training studies may be needed to determine clinical relevance. Any rehabilitation program would need to be implemented under the care of the appropriate professionals and customized to the meet the needs of the injured or rehabilitating patient.

It should be noted that the population tested in this study was a healthy population. The results of this study would have differed greatly if an injured population
would have been tested. It is commonly accepted that injuries can potentially affect the kinematics of walking as well as muscle activity patterns. Likely, a healthy population will respond to exercises and treatment differently than an unhealthy population.

Task specific exercises are widely accepted as a preferred component of rehabilitation and are also notably used in sport applications. Within these frameworks, it is also important to understand that task specific exercises are not the only component of rehabilitation or training. BW and IBW are not considered task specific exercises. Rather, BW and IBW are general exercises that provide a stimulus to work a broader spectrum of muscles. Task specific exercises are useful, but when done exclusively, there may be an increased chance of overuse injuries. General exercises that work a broader spectrum of muscles do not apply as much stress to any one specific muscle or muscle group. Backward locomotion has been shown to have many different benefits, from improving balance (Dufek, Mercer, Aldridge, Melcher, & Gouws, 2009; Hao & Chen, 2011) to decreasing patellofemoral pain (Flynn & Soutas-Little, 1993; Lay et al., 2007; Myatt et al., 1995; Roos et al., 2012). The general nature of backward locomotion exercises is what helps generate the wide range of benefits found with backward locomotion.

Conclusion

Walking backward has been shown to be beneficial in rehabilitation settings. Despite there being very little research to support it, IBW has also recently started to become popular in rehabilitation settings. The purpose of this study was to investigate the lower extremity kinematics and muscle activation patterns in IBW versus traditional BW. This purpose was achieved by evaluating performance of individuals walking
backward on a treadmill at inclines of 0%, 6%, 8%, 10%, and 12%. It was hypothesized that IBW would result in greater knee extensor EMG activity than traditional BW. It was also hypothesized that the knee would experience a greater range of motion (ROM). Given the results, each null hypothesis was rejected. IBW was found to change the angles of the ankle, knee, and hip joints at initial contact, as well as put those joints through an increased range of motion. It was also found that IBW has a significant influence on the activity of the rectus femoris muscle. The rectus femoris muscle acts as both a knee extensor as well as a hip flexor, and given that both joints experience an increased range of motion, it logically follows that the rectus femoris would become more active. From a rehabilitative perspective, IBW looks to have potential for rehabilitation of the knee joint, but further study is needed to determine clinical relevance. With the increased knee extensor activity, increased range of motion, and no significant increases in hamstring muscle activity patterns or gastrocnemius activity patterns, IBW has the potential to have an even greater impact on knee rehabilitation techniques. However, further research is warranted.

**Recommendations**

Future research should examine three-dimensional kinematics of IBW, as there is a paucity of information describing IBW in all planes. Understanding what kinematic differences exist between IBW and BW in the coronal or transverse plane would help provide a better foundation for clinicians looking to implement IBW in rehabilitation efforts. Future studies should also look into the clinical relevance of IBW as a rehabilitative technique. The analysis of shock attenuation at impact would also help provide valuable information for clinicians. With IBW being used as a rehabilitative
modality, especially for knee injuries, it would be important to understand if IBW has differing vertical accelerations of the leg at contact than BW.
APPENDIX A

IRB Form

UNLV

INFORMED CONSENT

Department of Kinesiology and Nutrition Sciences

TITLE OF STUDY: A comparative evaluation of the effects of incline on kinematics and muscle function during backward walking

INVESTIGATORS: J.S. Dufek, Ph.D., D.B. Jensen, Austin Coupe, K.R. Bartel

CONTACT PHONE NUMBER: J.S. Dufek, Ph.D., 702.895.0702

Purpose of the Study

You are invited to participate in a research study. The purpose of this study is to investigate the lower body joint angles and muscle activation patterns in inclined backward walking versus walking backward across flat ground.

Participants

You are being asked to participate in the study because you are a healthy individual between the ages of 18-30 years. You will be excluded from this study if you have any pain in your ankle, knee, or hip joints, or if you have had any ankle, knee or hip injuries that have previously required surgery. You will also be excluded from this study if you are a pregnant female who is past the 1st trimester. Stage of pregnancy is self-reported.

Procedures

If you volunteer to participate in this study, you will be asked to arrive at the Sports Injury Research Center rested, properly nourished, and hydrated on the days of familiarization as well as the day of testing. On each day, you will complete a self-directed warm up, which may include walking, jogging, and stretching. For safety reasons, anytime you are walking backward on a treadmill, you will be wearing a safety harness. There will be 3 familiarization training days. On each of those days, you will walk become familiar with walking backwards as well as become familiar with wearing a safety harness. You will also be utilizing that time in establishing your preferred speed for walking backward at each of the 5 different inclines (0%, 6%, 8%, 10%, and 12%). This familiarization happens by having you walk backward on a treadmill, starting with the 0% incline at the slowest speed the treadmill allows for (1 mph). As you become more comfortable with walking backward, that speed is increased until the point where walking is no
longer at a comfortable speed. You will do this for each of the 5 inclines. On the day of actual testing and after the warm up, 4 small patches of skin on the right leg will be prepared for EMG sensor placement through means of shaving, cleaning, and abrading the skin. The EMG sensors will then be placed in the prepared spots. Following that, reflective markers will be adhered to various bony landmarks in order to allow for motion capture of your backward walking. Once the preparations are complete, you will walk backward at your previously determined backward walking speed at each of the 5 inclines for 5 minutes per incline, with the option to sit and rest in-between each incline.

**Benefits of Participation**

There may be no direct benefit to you as a participant in this study. However, the data gathered from your participation will be used to help provide insight into the effects of inclined backward walking and its use as a potential rehabilitation technique.

**Risks of Participation**

There are risks involved in all research studies. This study may include only minimal risks. It is possible that you might experience delayed discomfort or acquire a muscle strain as a result of your physical performance. It is also possible but unlikely that you may trip/fall during testing. A safety harness will be worn to ensure your safety in case of such an incident. It is unlikely that injury will occur as the walking trials will occur in a controlled environment, with warm up during testing. It is possible that a participant may experience a localized skin irritation from the skin preparations for the EMG sensors. The EMG sensors are latex free, but it is also possible that they may cause mild skin irritation. Slight discomfort may be experienced by wearing the safety harness. All adhesive tape will be latex free, but some skin irritation may still present itself in those areas that require any application of tape.

**Cost / Compensation**

There will be no financial cost to you to participate in this study. The familiarization training sessions will require approximately 30 minutes per day across 3 days for a total of 90 minutes spent on familiarization. The day of actual testing will require approximately 60-120 minutes by the time preparation, verbal instructions, rest periods, and data collections are completed. In total, each participant will require an approximate 150-210 minutes.

**Contact Information**

If you have any questions or concerns about the study, you may contact Dr. Janet Dufek at 702.895.0702. For questions regarding the rights of research subjects, any complaints or comments regarding the manner in which the study is being conducted you may contact the UNLV Office of Research Integrity – Human Subjects at 702-895-2794 or toll free at 877-895-2794 or via email at IRB@unlv.edu.

**Voluntary Participation**

Your participation in this study is voluntary. You may refuse to participate in this study or in any part of this study. You may withdraw at any time without prejudice to your
relations with the university. You are encouraged to ask questions about this study at the beginning or any time during the study.

**Confidentiality**

All information gathered in this study will be kept completely confidential. No reference will be made in written or oral materials that could link you to this study. All records will be stored in a locked facility at UNLV for 3 years after completion of the study. After the storage time the information gathered will be destroyed.

**Participant Consent:**

I have read the above information and agree to participate in this study. I am at least 18 years of age. A copy of this form has been given to me.

_________________________________________________________  ________________
Signature of Participant                                          Date

_________________________________________________________
Participant Name (Please Print)

**Audio/Video Taping**

This study involves audio/video taping. It is my understanding that I will appear within the field of view of the camera.

_________________________________________________________  ________________
Signature of Participant                                          Date

_________________________________________________________
Participant Name (Please Print)

*Participant Note: Please do not sign this document if the Approval Stamp is missing or is expired.*
## APPENDIX B

**Definitions of All Dependent Variables and Abbreviations**

Table 4: Definition of Dependent Variables, abbreviations, and definitions.

<table>
<thead>
<tr>
<th>Dependent Variables</th>
<th>Abbreviations</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Kinematics</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Foot Inclination</td>
<td>FI</td>
<td>Inclination of the foot segment at contact</td>
</tr>
<tr>
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<td>IAJ</td>
<td>Ankle angle at contact</td>
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<td>IKJ</td>
<td>Knee angle at contact</td>
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<td>Instantaneous Hip Angle</td>
<td>IHJ</td>
<td>Hip angle at contact</td>
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<td>KJA</td>
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<td>Hip Accel</td>
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<td>Ankle joint Range of Motion during stance</td>
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<td>Hip ROM Swing</td>
<td>HJRom&lt;sub&gt;swing&lt;/sub&gt;</td>
<td>Hip joint Range of Motion during swing</td>
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<td>StepRate</td>
<td>Time between right foot contact and left foot contact</td>
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<td>RMS score during stance phase normalized to Rectus Femoris MVC</td>
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<td>B&lt;sub&gt;fstance&lt;/sub&gt;</td>
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<td>Short Form</td>
<td>RMS Score During Swing Phase Normalized To MVC</td>
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<td>RMS score during swing phase normalized to Biceps Femoris MVC</td>
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<td>RMS score during swing phase normalized to Tibialis Anterior MVC</td>
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APPENDIX C
Lower Extremity Joint Kinematics by Participant Across Levels of Incline

Figure 1: Foot Inclination Angle at Initial Contact by Participant

Figure 2: Ankle Angle at Contact by Participant
Figure 3: Knee Angle at Contact by Participant

Figure 4: Hip Angle at Initial Contact by Participant
Figure 5: Angular Velocity at the Ankle Joint at Initial Contact by Participant

Figure 6: Angular Velocity at the Knee Joint at Initial Contact by Participant
Figure 7: Angular Velocity at the Hip Joint at Initial Contact by Participant

Figure 8: Angular Acceleration at the Ankle Joint at Initial Contact by Participant
Figure 9: Angular Acceleration at the Knee Joint at Initial Contact by Participant

![Knee Acceleration Graph](image)

Figure 10: Angular Acceleration at the Hip Joint at Initial Contact by Participant

![Hip Acceleration Graph](image)
Figure 11: Ankle Joint Range of Motion during Stance by Participant

Figure 12: Knee Joint Range of Motion during Stance by Participant
Figure 13: Hip Joint Range of Motion during Stance by Participant

Figure 14: Ankle Joint Range of Motion during Swing by Participant
Figure 15: Knee Joint Range of Motion during Swing by Participant

Figure 16: Hip Joint Range of Motion during Swing by Participant
Figure 17: Step Rate by Participant
APPENDIX D

Lower Extremity Joint Kinematics by Group Across Levels of Incline

Figure 18: Foot Inclination Angle at Initial Contact by Group

Figure 19: Ankle Angle at Initial Contact by Group

* = Significantly different from 0% at $p < 0.05$
Figure 20: Knee Angle at Initial Contact by Group

![Knee Angle Graph]

* = Significantly different from 0% at p < 0.05

Figure 21: Hip Angle at Initial Contact by Group

![Hip Angle Graph]

* = Significantly different from 0% at p < 0.05
Figure 22: Angular Velocity at the Ankle Joint at Initial Contact by Group

![Ankle Velocity Chart]

Figure 23: Angular Velocity at the Knee Joint at Initial Contact by Group

![Knee Velocity Chart]
Figure 24: Angular Velocity at the Hip Joint at Initial Contact by Group

![Hip Velocity Graph]

Figure 25: Angular Acceleration at the Ankle Joint at Initial Contact by Group

![Ankle Acceleration Graph]
Figure 26: Angular Acceleration at the Knee Joint at Initial Contact by Group

Figure 27: Angular Acceleration at the Hip Joint at Initial Contact by Group
Figure 28: Ankle Joint Range of Motion during Stance by Group

* = Significantly different from 0% at $p < 0.05$

Figure 29: Knee Joint Range of Motion during Stance by Group

* = Significantly different from 0% at $p < 0.05$
Figure 30: Hip Joint Range of Motion during Stance by Group

* = Significantly different from 0% at p < 0.05

Figure 31: Ankle Joint Range of Motion during Swing by Group
Figure 32: Knee Joint Range of Motion during Swing by Group

**Knee ROM - Swing**

![Knee ROM - Swing](image)

* = Significantly different from 0% at p < 0.05

Figure 33: Hip Joint Range of Motion during Swing by Group

**Hip ROM - Swing**

![Hip ROM - Swing](image)

* = Significantly different from 0% at p < 0.05
Figure 34: Step Rate by Group

* = Significantly different from 0% at p < 0.05
APPENDIX E

Lower Extremity Muscle Activity by Participant Across Levels of Incline

Figure 35: Rectus Femoris Muscle Activity during Stance by Participant

Figure 36: Biceps Femoris Muscle Activity during Stance by Participant
Figure 37: Medial Gastrocnemius Muscle Activity during Stance by Participant

Figure 38: Tibialis Anterior Muscle Activity during Stance by Participant
Figure 39: Rectus Femoris Muscle Activity during Swing by Participant

Figure 40: Biceps Femoris Muscle Activity during Swing by Participant
Figure 41: Medial Gastrocnemius Muscle Activity during Swing by Participant

Figure 42: Tibialis Anterior Muscle Activity during Swing by Participant
APPENDIX F

Lower Extremity Muscle Activity by Group Across Levels of Incline

Figure 43: Rectus Femoris Muscle Activity during Stance by Group

* = Significantly different from 0% at p < 0.05

Figure 44: Biceps Femoris Muscle Activity during Stance by Group
Figure 45: Medial Gastrocnemius Muscle Activity during Stance by Group

![Medial Gastrocnemius - Support](image)

Figure 46: Tibialis Anterior Muscle Activity during Stance by Group

![Tibialis Anterior - Support](image)
Figure 47: Rectus Femoris Muscle Activity during Swing by Group

* = Significantly different from 0% at p < 0.05

Figure 48: Biceps Femoris Muscle Activity during Swing by Group
Figure 49: Medial Gastrocnemius Muscle Activity during Swing by Group

![Medial Gastrocnemius - Swing]

Figure 50: Tibialis Anterior Muscle Activity during Swing by Group

![Tibialis Anterior - Support]
REFERENCES


VITA

Daniel B. Jensen

Master’s Student, Kinesiology - Biomechanics

University Of Nevada, Las Vegas
Las Vegas, NV  89154
(801) 602-3149

EDUCATION

M.S.  University of Nevada, Las Vegas  (Proposed) Summer 2014
Kinesiology, Biomechanics (Emphasis)

B.S.  Utah Valley University, Orem, UT  Spring 2012
Exercise Science and Outdoor Recreation

Academic Experience:

Graduate Assistantship, University of Nevada, Las Vegas  2013-2014
Teaching Assistant
Undergraduate Biomechanics Lab (Kin 346)
Head Teaching Assistant (Kin 346)

RESEARCH

PUBLICATIONS

Biomechanical and Physiological Responses to Prolonged Interactive Video Game Play (Contributing Author, Publication date pending)

Effects of Different Backpacks and Selected Gait Parameters (Lead Author, Accepted for Publication, date pending)

PRESENTATIONS

Refereed Podium Presentation

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Newport
Beach, USA, October 2013.

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**Non-Refereed Poster Presentation**

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Study. 6th Interdisciplinary Research Scholarship Day, University of Las Vegas, NV,
April 2013