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THE EFFECT OF FOOTWEAR ON THE MECHANICS OF THE LOWER BACK DURING TREADMILL RUNNING

by

Jeffrey Ray McClellan

Bachelor of Science
Brigham Young University
2008

A thesis submitted in partial fulfillment
of the requirements for the

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Department of Kinesiology and Nutrition Sciences
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University of Nevada, Las Vegas
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Abstract

The Effect of Footwear on the Mechanics of the Lower Back During Treadmill Running

by

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Little is known regarding the effect that footwear cushioning can have on the mechanics of the low back. The purpose of this study was to 1) determine the material characteristics of a minimalist running shoe tested with and without a commercially available shoe insole, 2) determine if there are differences in lower back or knee kinematics when minimalist shoes are worn with and without a shoe insole during treadmill running, and 3) determine if there are differences in levels of muscle activation when minimalist shoes are worn with and without a shoe insole during treadmill running. Following the receipt of informed consent 10 subjects (age 33.3±13.0 years, height 168.5±9.8 cm, mass 64.5±13.5 kg) ran on the treadmill while wearing a minimalist running shoe with (IN) and without (OUT) added cushioning. Following determination of each subjects preferred running speed, a running warm-up was performed while wearing the test shoes during IN and OUT. Study subjects were then instrumented with surface electrodes on the left erector spinae, rectus abdominis, and biceps femoris, while also being instrumented with electrogoniometers placed over the lumbar spine, and the lateral side of the left knee. Subjects ran on the treadmill for two minutes at their preferred speed after which data collection took place for an additional 45 seconds with

shoe condition order being counterbalanced. The first ten running strides were extracted for analysis. Muscle activity and kinematics were extracted using a telemetry system for electromyography (TeleMyo 2400T, G2; Noraxon USA Inc. Scottsdale, AZ; 1500Hz), with the fully rectified, normalized signal from the surface electrodes being used to calculated average muscle activity for the erector spinae (ES), the rectus abdominis (RA), and the biceps femoris (BF) during the stance phase of running, using peak extension from the knee electrogoniometer to determine stance. Kinematic analysis was performed using the knee and back electrogoniometers which included calculating knee range of motion (KnROM), knee angle at the moments of peak extension (KnExt) and peak flexion (KnFlx), low back range of motion (BaROM), and average flexion/extension of the low back (BaPos). Following subject testing, Paired T-tests (α=0.05) were performed to compare the test conditions. Impact testing of the test shoes was also performed at the heel (HL) and forefoot (FF) of all shoes during IN and OUT using a mechanical impact tester (Exeter Research Inc. Brentwood, NH; 3000Hz). Testing followed a modified American Standard for Testing Materials (ASTM) test procedure (ASTM F-1614). A missle head (mass 8.5kg; diameter 45mm) was dropped from a height of 50 mm with twenty pre-impacts being performed, followed by data being collected during ten test impacts. Peak acceleration (PA) and peak pressure (PP) were extracted from test results, and Independent T-tests (α=0.01) were used to separately compare HL and FF during IN and OUT while also comparing IN and OUT during HL and FF. Results for impact testing showed differences between HL and FF during IN for all variables, with differences between IN and OUT being observed during HL and FF for all variables. Results for KnFlx showed increases in maximum knee flexion when cushioned inserts were placed in the shoes (32.2±4.7° with inserts vs. 30.3±5.5° without inserts). These results suggest that differences in shoe cushioning material do not significantly affect mechanics of the low back during running, although implications for knee stiffness do exist.

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Many thanks should also be given to Altra® Footwear who kindly donated the shoes that were used for this study.

Most of all, I would like to thank my beautiful wife for her help and support as I worked towards completion of my thesis and masters studies. Whether it was acting as a pilot research subject, or helping me instrument myself as I ran through the test protocol, Kimia was always there to provide help and support.

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Chapter 1

Introduction

Running is a widespread form of physical activity participated in frequently by roughly 18 million people in the United States, with participation rates growing steadily (Running USA, 2011). Injuries among runners have been found to occur fairly regularly, with 47 percent of runners reporting some sort of injury during a two year period (Jacobs & Berson, 1986). Research has been performed which attempts to determine the nature and cause of various injuries. Injury to the knee has been reported as the most prevalent form of injury experienced by runners (Jacobs & Berson, 1986; James, Bates, & Osternig, 1978; Taunton, Ryan, Clement, McKenzie, Lloyd-Smith, Zumbo, 2003). Injury to the low back, while not as common has still been reported to comprise 5-15 percent of running injuries (Jacobs & Berson, 1986; Taunton et al. 2003). Attempts have also been made to aid in the prevention and treatment of injuries with the prescribed use of running shoes or orthoses comprising two methods of attempted injury prevention/rehabilitation (James et al. 1978).

While the prevalence of low back injuries among runners is fairly small, the prevalence of low back pain is a widespread problem in our society, with 80% of injuries to the low back being classified as non-specific or of unknown origin (Njoo & Van der Does, 1994). Low back injuries of all types have been classified as one of the most costly types of injury commonly found in industrialized countries (Vogt, Pfeifer, Portscher, & Banzer, 2001), affecting between 70-85% of all people at some point in their lives (Andersson, 1999). As such, understanding the nature and cause of low back pain with a focus on understanding non-specific low back pain is desirable. Research into the mechanics of the low back is necessitated in order to help develop this understanding. Despite this need to understand low back mechanics, few research studies have been performed which examine the mechanics of the lower back during running. The studies performed have focused on low back kinematics (Hart, Kerrigan, Fritz, & Ingersoll, 2009; Levine, Colston, Whittle, Pharo, & Marcellin-Little, 2007; Schache, Blanch, Rath, Wrigley, & Bennell, 2002; Seay, Van Emmerik, & Hamill, 2011), lumbo-sacral forces (Seay, Selbie, & Hamill, 2008), low back muscle activity and accelerometry (Ogon, Aleksiev, Spratt, Pope, & Saltzman, 2001), and joint stiffness when non-specific low back pain is present (Hamill, Moses, & Seay, 2009).

Of studies performed that examined low back mechanics only one examined the effect of the material properties of footwear cushioning on low back mechanics (Ogon et al. 2001). None of the studies combined the use of kinematics and electromyography to study function of the low back. As such, the effect that a running shoe's material can have on either the development or treatment of low back pain is unknown. Additionally, recent literature has called into question the evidence surrounding the general prescription of running footwear for the treatment or prevention of injuries (Richards, Magin, & Callister, 2009). In recent years the popularization of minimalist footwear designed to simulate running barefoot has led to additional questioning among the general public regarding the efficacy of injury prevention resultant to wearing traditional running shoes. In order to determine the efficacy of injury prevention due to use of different types of footwear, longitudinal studies must be performed which examine runners across time who wear the various types of footwear in training.

Discussion regarding the use of minimalist footwear or barefoot running has also taken place as a result of research findings that indicate barefoot or minimally shod runners contact the ground with the foot in a more plantar flexed position than their shod counterparts, initially striking the ground with the forefoot versus the heel, while demonstrating lower peak ground reaction forces (Lieberman, Venkadesan, Werbel, Daoud, D'Andrea, Davis, Mang'eni, & Pitsiladis, 2010). As the only previous study to examine low back mechanics and the material properties of footwear constrained foot ground contact to heel only during barefoot and shod running, it is impossible to extend results to a population that does not consciously change foot strike patterns, or who contact the ground with a forefoot or midfoot strike pattern. Due to the lack of research studies addressing footwear material, low back muscular activity, and low back kinematics it would be advantageous to perform research that helps to develop a better understanding of the effect that the material properties of footwear can have in order to come to a better understanding of low back function during running.

Study Purpose

In order to better understand the relation of footwear cushioning to low back mechanics, the purpose of this study was to 1) determine the material characteristics of a minimalist running shoe tested with and without a commercially available shoe insole, 2) determine if there are differences in lower back or knee kinematics when minimalist shoes are worn with and without a shoe insole during treadmill running, and 3) determine if there are differences in levels of muscle activation when minimalist shoes are worn with and without a shoe insole during treadmill running.

Limitations/Delimitations

1) No constraints were made regarding foot alignment at ground contact and some subjects may have adopted different running styles in the various footwear conditions.

2) Subjects may have had different amounts of experience with barefoot/minimalist running which may have influenced running kinematics.

3) As study participants ran at a preferred speed there were variations in speed among participants.

4) As a minimalist shoe was used with and without an added insert, and no standard running footwear was used, study results do not necessarily reflect the differences between minimalist footwear and standard footwear.

5) Study trials were completed during treadmill running, and some differences may exist if the procedure were to be completed during over-ground running.

6) Analysis of electromyography was constrained to the stance phase of running, thus ignoring any anticipatory effect that footwear material may have on muscle activity.

7) Both men and women were used as study subjects, ignoring any gender differences that may exist in relation to muscle activity or running kinematics.

Definitions

Biceps Femoris – The muscle on the posterior and lateral side of the thigh which is involved with knee flexion, and lateral rotation.

Erector Spinae – A muscle group in the back that runs along the length of the spine, with origins at the ilium, the sacrum, and the lumbar vertebrae. It is involved with spinal flexion/extension, and lateral bending.

Fully Rectified EMG – A process used to reduce and analyze EMG data where the absolute values for all data points are returned.

Insole – A material placed in a shoe that may be designed to improve comfort, decrease injury risk, eliminate odor, etc.

Last – The form upon which a shoe is designed, affecting shoe proportions, and being specific to shoe size, heel height, and potentially the gender of the wearer.

Maximal Voluntary Contraction (MVC) – An isometric contraction performed maximally in an effort to determine the peak level of muscle activity for a given muscle.

Midsole – The material in a shoe that lies between the shoe sole and the shoe insole with the general purpose to absorb impact shock.

Minimalist Footwear – Footwear which is designed to provide a minimal level of support for the foot while still providing protection against ground hazards.

Pressure – The amount of force that is exerted by an object over a given area of space. Calculated as Pressure = F/A.

Rectus Abdominis – A muscle running vertically along the abdomen which originates at the pubis, aiding in trunk stabilization, and trunk flexion.

Surface Electromyography (EMG) – The process through which electrodes attached to the skin are used to obtain an electrical signal from the underlying muscle.

Chapter 2

Review of Related Literature

Overground and Treadmill Running Kinematics

Running is an exercise activity regularly participated in by roughly 18 million people annually in the United States (Running USA, 2011). Due to the nature and constraints of modern society limitations have been placed on the practicality of running overground for many who wish to run as a form of exercise, leading some to choose treadmill running as their main mode of exercise. Approximately six percent of runners perform the majority of their training on the treadmill with the rest running primarily outdoors (Taunton et al. 2003). As many runners participate in both forms of running exercise, it is important to describe similarities and differences between the two forms of running while examining running kinematics as a whole.

Treadmill and overground running.

In general, both treadmill and overground running kinematics have been found to be very similar (Williams, 1985; Pink, Perry, Houglum, & Devine, 1994). At moderate speeds there are very few differences between the two mediums, however some research studies have reported a slightly higher cadence and decreased stride length while running on the treadmill (Elliott & Blanksby, 1976; Riley, Dicharry, Franz, Della Croce, Wilder, & Kerrigan, 2008). Of note is that Riley et al. (2008) reports that half of study subjects did not change cadence or stride length during treadmill running while Elliott and Blanksby (1976) only observed significant differences at speeds greater than 4.8m/s. Increased maximum and decreased minimum angles of knee flexion have also been reported when comparing treadmill and overground running kinematics (Riley et al. 2008), with less vertical displacement being recorded on the treadmill as well (Pink et al. 1994). With the exception of these minor differences there are relatively few changes that occur between treadmill and overground running kinematics while additional factors such as footwear (Bishop, Fiolkowski, Conrad, Brunt, & Horodyski, 2006; Hardin, van den Bogert, & Hamill, 2004; Lee Y, Kim YK, Kim YH, Kong, & Lee K, 2011; Lieberman et al. 2010; Lohman, Balan Sackiriyas, & Swen, 2011), and speed (Bishop et al. 2006; Mann & Hagy, 1980; Pink et al. 1994), can play a role in observable kinematic differences.

Footwear related kinematic changes.

Footwear has been shown to influence running kinematics to varying degrees depending on the particular shoe being used. For example, Hardin et al. (2004) found differences in the velocity of ankle dorsiflexion when footwear of different hardness was worn, while also observing greater levels of knee and hip extension at foot contact when running over a hard surface. Lee et al. (2011) noted that individuals running in dress shoes demonstrated increased levels of ankle dorsiflexion on ground contact in comparison to individuals running in standard running shoes. It was thought that individuals increased ankle dorsiflexion in order to reduce the likelihood of slipping (Lee et al. 2011). Increased ankle dorsiflexion among shod vs. barefoot runners has also been observed when running in standard running shoes, with barefoot runners demonstrating a greater total range of motion at the ankle during running at 3.6m/s, with no difference being recorded at a slower velocity of 2.2m/s (Bishop et al. 2006). This is in agreement with research on minimalist footwear that found runners using minimalist shoes generally used a midfoot or forefoot strike pattern with which to contact the ground, vs. heel contact employed by runners wearing standard running shoes (Lieberman et al. 2010; Lohman et al. 2011), while also demonstrating the dual effect that speed and footwear can have on running kinematics.

Kinematic changes related to speed.

Further evidence that running speed affects running kinematics was provided by Bishop et al. (2006) as he found that runners demonstrated a greater range of motion at the knee joint when speeds were slow (2.2m/s) while demonstrating a lower range of motion when speeds were faster (3.6m/s). Results from other studies are not in agreement with those described by Bishop et al (2006), as others noted that there was increased knee flexion at ground contact when speed increased (Mann & Hagy, 1980), and increased knee flexion during the middle and late portions of the swing phase of running (Pink et al. 1994) which would subsequently lead to a higher range of motion at faster speeds (Mann & Hagy, 1980; Pink et al. 1994). Additionally, differences in hip (Mann & Hagy, 1980; Pink et al. 1994) and ankle (Mann & Hagy, 1980) kinematics were noted as an outcome of speed. Mann and Hagy (1980) described the hip as displaying decreased levels of extension and increased levels of flexion during sprinting, while the ankle was in a more plantar flexed position at ground contact but experienced a lesser magnitude of plantar flexion during sprinting. Pink et al. (1994) also observed a significant increase in hip flexion as a product of speed, while significant differences in ankle kinematics were not observed, and hip extension was shown to increase.

Differences between these studies are likely related to the speeds used as the slower of two speeds used by Mann and Hagy of 5.4m/s (1980) was faster than the fastest speed used by Pink et al. (1994) or Bishop et al. (2006), while the slowest speed used by Bishop et al. (2006) of 2.2m/s was only slightly faster than speeds commonly used in some studies to test subjects while walking (Seay et al. 2011). Additional findings showed that during the stance phase of sprinting only knee flexion occurred (Mann & Hagy, 1980), as opposed to distinct periods of knee flexion and extension being observable during sub maximal running (Mann & Hagy, 1980; Pink et al. 1994).

Body segment range of motion during running.

In addition to identifying differences in running kinematics that may occur as a function of speed, footwear, etc., there seem to exist standard ranges in which segment angles may fall during running. In his classic review, Williams (1985) reported that maximum angles of thigh flexion during the swing phase of running will vary from 25°-59° when measured at the thigh relative to a vertical plane at 90° to the right horizontal, with variations primarily being dependant on running velocity. During stance the thigh begins to extend prior to toe off, with levels of extension reached between 24° and 32°, and maximum extension being reached just following toe off, again using the same reference frame as used previously (Williams, 1985). Knee angle, calculated as the angle between the leg and the extension of the thigh has been shown to vary between 21°-30° at ground contact with maximum angles of extension occurring just prior to ground contact, while maximum angles of flexion will reach as high as 120° and as low as 30° depending on speed (Williams, 1985). Maximum extension of the knee has been further tested, with testing indicating that ground contact occurs approximately 20ms after maximum extension is reached during treadmill running, while toe-off occurs approximately 5ms before a second extension peak is reached (Fellin, Rose, Royer, & Davis, 2010). When measuring the ankle angle as the angle between the foot and the extension of the leg, angles of dorsiflexion have been reported between 84° and 101° at ground contact, while angles of plantarflexion have been reported between 59° and 75° at toe off (Williams, 1985).

While most research examining running kinematics has examined function of the hips, knees, and ankles, there is additional research that focuses on the kinematics of the trunk and pelvis (Hart et al. 2009; Levine et al. 2007; Schache et al. 2002; Seay et al. 2011). Coordination between the lumbar spine and the pelvis has been found to be very high, displaying a high correlation for anterior-posterior and lateral rotation (Schache et al. 2002; Seay et al. 2011). A forward lean of the trunk and anterior tilt to the pelvis have also been observed when subjects are running (Levine et al. 2007; Schache et al. 2002; Williams, 1985), with increased forward lean becoming apparent as a result of increases in speed (Seay et al. 2011), surface inclination (Levine et al. 2007), and following fatigue (Hart et al. 2009). Range of motion of the trunk and pelvis has also been found to be higher when subjects are running as opposed to walking (Levine et al. 2007; Seay et al. 2011), with average range of motion varying from approximately 10°-21° (Levine et al. 2007), and an average angle at the lower back between 22° (Schache et al. 2002) and 26° (Levine et al. 2007). Average inclination of the forward lean of the trunk has been reported between 4°-7° at both foot strike and toe off, with maximum levels reaching approximately 12°-13° during the middle of stance (Williams, 1985).

Muscle Activity During Running

While kinematic descriptors of running are important and not to be discounted, it is also important to note the effect that running can have on the electrical activity of various muscle groups, when tested using surface electromyography. For some uses surface electromyography can be difficult to quantify, as there is a large amount of inter-subject variability for different lower extremity muscle groups, although inter-subject variability for the onset of muscle activation during various running gait phases remains low (Guidetti, Rivellini, & Figura, 1996). The signals received through use of surface electromyography must also be normalized in some fashion, with Soderberg and Knutsen (2000) advocating the adoption of study subjects performing maximum voluntary isometric contractions in order to normalize signals. The use of maximum voluntary isometric contractions is highly repeatable, but more variable than other forms of signal normalization such as using 70% of peak running speed or a sprint test for normalization (Albertus-Kajee, Tucker, Derman, Lamberts, & Lambert, 2011). While the use of maximum voluntary isometric contractions is repeatable and is the suggested form of normalization (Soderberg and Knutsen, 2000), it is not a good indicator of the highest possible levels of muscle activation that can be produced as research has indicated that when effort is high during running, the electromyographic signal will often reach levels higher than that produced by the maximum voluntary isometric contraction (Kyröläinen, Avela, & Komi, 2005).

Timing of muscle activation.

As mentioned previously, the differing muscle groups of the lower extremity have been found to display low variability in the onset of muscle activation during various gait phases (Guidetti et al., 1996). Several researchers have successfully identified the basic timing of muscle firing patterns during running (Flynn & Soutas-Little, 1993; Guidetti et al., 1996; Mann, Moran, & Dougherty, 1986). The most total muscle activity tends to occur during the stance phase of running, and during the terminal portion of the swing phase just prior to foot ground contact (Guidetti et al., 1996; Mann et al., 1986). During the middle portion of the swing phase when the contralateral foot has made contact with the ground the rectus femoris (Chumanov, Heiderscheit, & Thelen, 2011; Guidetti et al., 1996; Kyröläinen et al., 2005), erector spinae, and tibialis anterior (Guidetti et al., 1996; Mann et al., 1986) have shown additional periods of muscle activity, although some have only noted the increased rectus femoris activation at high speeds (Chumanov et al., 2011; Kyröläinen et al., 2005).

All lower extremity muscles tend to be activated at ground contact (Guidetti et al., 1996; Kyröläinen et al., 2005; Mann et al., 1986), with the noted exception that at times the biceps femoris has been found to activate only following the initiation of the stance phase (Flynn & Soutas-Little, 1993). Muscles of the lower extremity have also been found to remain activated for approximately 60-80% of the stance phase of running (Flynn & Soutas-Little, 1993). Guidetti et al. (1996) found that there are generally two distinct peaks in the electromyographic signal that is received from the muscle, occurring just prior to and following ground contact. The peak times of muscle activation that occurred tend to follow a distinct pattern, as the biceps femoris has the tendency to reach peak activation levels the earliest before, and the latest following ground contact, with the erector spinae similarly reaching its peak activation time the latest prior to, and the soonest following ground contact (Guidetti et al., 1996).

Effect of speed on muscle activation.

As speed has been shown to affect muscle activation of the rectus femoris in the studies by Kyröläinen et al. (2000) and Chumanov et al. (2011), likewise the rectus abdominus has been shown to display increased levels of activity just prior to and following toe-off during sprinting (Mann et al., 1986). Additionally, the relative timing of peak gastrocnemius muscle activation appears to shift somewhat towards toe off during sprinting (Chumanov et al., 2011). In terms of the average magnitude of muscle activity recorded at various speeds, it has been found that all of the major lower extremity muscle groups tend to display increased muscle activity as speed increases (Liebenberg, Scharf, Forrest, Dufek, Masumoto, & Mercer, 2011). These results remained consistent even when subject weight was reduced to levels equaling 60% of subjects normal body weight (Liebenberg et al. 2011). During sprinting the greatest magnitude of muscle activation tends to shift towards the later stages of swing rather than stance, as the biceps femoris, medial hamstrings, and vastus lateralis have been shown to exhibit greater average levels of muscle activation during terminal swing than during stance (Chumanov et al., 2011).

Effect of footwear on muscle activity.

The use of footwear or orthoses is another factor which can potentially affect muscle activity during running as evidenced by a review of literature which cites many studies where differences between footwear/orthoses use and a control have been recorded (Murley, Landorf, Menz, & Bird, 2009). When comparing running with and without orthoses, differences have been noted in activation of the biceps femoris, and tibialis anterior (Nawoczenski & Ludewig, 1999), in addition to the vastus medialis, medial gastrocnemius, and peroneus longus (Kelley, Girard, & Racinais, 2011), with the general results indicating that muscular activity tends to increase in the more anterior portions of the leg when wearing orthoses, while other muscle groups tend to exhibit decreased activation (Kelley et al., 2011; Nawoczenski & Ludewig, 1999). Mündermann, Wakeling, Nigg, Humble, and Stefanyshyn (2006) were in partial agreement with this observation, recording increased levels of peroneus longus and tibialis anterior muscle activation, but also recording increased biceps femoris activation, versus decreased muscle activation recorded by Nawoczenski & Ludewig (1999).

Von Tscharner, Goepfert, and Nigg (2003) found that when running either barefoot or in shoes, the timing of tibialis anterior muscle activity prior to heel-contact was the same; however, the timing of post heel-contact muscle activity demonstrated a muscular delay as a result of wearing shoes. Bird, Bendrups, and Payne (2003) reported an earlier onset of erector spinae activation and a later onset of gluteus medius activation when bilateral heel lifts were worn during walking. These results were substantiated by Lee, Jeong, and Freivalds (2001) who found that when high heels are worn during walking, erector spinae muscle activation is increased. Lee et al. (2001) suggests that these changes could potentially lead to the development of low back pain. It is unknown if altered muscle activation patterns would remain consistent if these studies were carried over from walking to running. Further study on erector spinae muscle activation was performed by Ogon et al. (2001), who studied the effect of different shoe and material conditions on electromyography of the lower back. Ogon et al. (2001) determined that in a barefoot condition muscle activation of the erector spinae is initiated sooner following heel strike than in a shod condition, while the time from peak lumbar acceleration to peak muscle activation is longer in a barefoot condition.

Low Back Pain

An overuse injury can be defined as pain or discomfort attributed to running that can potentially result in limitations being placed on running speed, distance, duration, or frequency (Hreljac, 2004). Among runners, between 25 and 65 percent tend to be diagnosed with overuse injuries over the course of a given training program (Taunton et al. 2003) with between 5-15 percent of injuries being said to originate in the lower back (Jacobs & Berson, 1986; Taunton et al. 2003). Thus, as there are approximately 18 million people in the United States that regularly run for fitness (Running USA, 2011) it can be estimated that between 0.23-1.76 million runners will suffer from some sort of injury of the lower back during a given training program. It is unknown how many additional persons may suffer from mild low back pain that would not be classified as an injury based upon the inclusion criteria for different research studies. Among low back injuries, approximately 80% are of unknown origin with resultant pain being classified as non-specific low back pain (Njoo & Van der Does, 1994). Low back injuries of known or unknown origin can be classified as one of the most costly types of injury commonly found in industrialized countries (Vogt et al. 2001) while back problems are the most common form of impairment among the young and middle aged (Andersson, 1999).

Changes in gait kinematics resulting from low back pain.

While the origin of most low back injuries is unclear some basic differences exist between those suffering from chronic non-specific low back pain compared to healthy controls. When walking speed is controlled, those suffering from chronic non-specific low back pain have the tendency to employ a walking strategy in which smaller and faster strides are taken comparative to a healthy population (Keefe & Hill, 1985;

Khodadadeh & Eisenstein, 1993). It is thought that these changes to normal gait may be employed by those with low back pain as a protective mechanism to prevent or lessen additional pain from occurring (Vogt et al. 2001), although not all research has noted differences in these stride parameters (Hanada, Johnson, & Hubley-Kozey, 2011). Further studies have revealed that in addition to stride length decreases among those suffering from low back pain, spinal joint accelerations decrease across the frontal, and sagittal planes during walking (Moe-Nilsson, Ljunggren, & Torebjörk, 1999; Vogt et al. 2001). It is thought that reduced proprioception of the lower back, which has been found among those with low back pain (Gill & Callaghan, 1998) may contribute to increased variability of joint movement, as any perturbations in feedback mechanisms could disrupt lower back coordination, thus leading to increased variability (Vogt et al. 2001). This hypothesis is supported by research which recorded increased gait variability among those with low back pain (Papadakis, Christakis, Tzagarakis, Chlouverakis, Kampanis, Stergiopoulos, & Katonis, 2009). While movement variability increases it is important to note that total range of motion in the lumbar region remains relatively equal during walking for those with and without low back pain, and as maximum range of motion is much higher than that employed during walking it becomes difficult to use kinematics as a descriptor for low back pain (Vogt et al. 2001). As the total range of motion during running is more than double that of walking (Levine et al. 2007), the use of kinematics is more useful during running than during walking, with studies reporting differences in running kinematics between those with low back pain and controls (Schache et al. 2002; Seay et al. 2011). A kinematic relationship has also been established between running and low back pain. Lower extremity joint stiffness, which has been linked with the development of low back pain (Voloshin & Wosk, 1982), at the knee increased with a decrease in joint range of motion for a population suffering from low back pain (Hamill et al. 2009).

Muscle activation among those with low back pain.

Additional research has focused on the muscle activation of those who do and do not suffer from low back pain. Those suffering from low back pain have demonstrated higher levels of muscle activity at rest (Jones, Henry, Raasch, Hitt, & Bunn, 2012), and during movement (Arab, Ghamkhar, Emami, & Nourbakhsh, 2011; Hanada et al. 2011; Jones et al. 2012; Van der Hulst, Vollenbroek-Hutten, Rietman, & Hermens, 2010; Van Dieen, Cholewicki, & Radebold, 2003; Wilson, Madigan, Davidson, & Nussbaum, 2006) than those not suffering from low back pain. Synchronous coactivation of the muscles of the core has also been found to be disturbed in a population suffering from chronic low back pain, as healthy individuals tend to display a synchronous muscle activation pattern for all core muscles, while this is not evident among those with low back pain (Hubley-Kozey & Vezina, 2002). The erector spinae muscle group has been shown to demonstrate increased muscle activation levels in a population suffering from low back pain (Hanada et al. 2011; Jones et al. 2012). It is thought that this occurs as the individual with low back pain subconsciously attempts to maintain stability in order to avoid injury, leading to greater levels of muscle activation and stiffening of the trunk (Jones et al. 2012).The hamstrings also tend to display higher activation levels among those with low back pain although more research is necessary to establish this relationship (Arab et al. 2011). Muscle activity of the rectus abdominis has been shown to have mixed results when comparing those with and without low back pain (Hanada et al. 2011; Jones et al. 2012; Van der Hulst et al. 2010; Wilson et al. 2006).

When comparing those with low back pain and healthy controls following fatigue, it has been found that healthy individuals tend to demonstrate increased muscle activation of the rectus abdominis (Wilson et al. 2006). This is believed to occur concurrently with an increase in trunk flexion and a decrease in lumbar lordosis in a healthy population, while populations demonstrating low back pain demonstrate increased lumbar lordosis and increased trunk extension (Hart et al. 2009). Of note is that when walking (Van der Hulst et al. 2010), and when performing movement perturbations (Jones et al. 2012) in a non-fatigued state those with low back pain have been documented to experience increased activation of the rectus abdominis. This conflicts with the analysis performed by Hanada et al. (2011), where higher levels of rectus abdominis activation were documented among an asymptomatic control group during normal gait. A possible explanation for these conflicting results could stem from the differences in study design, as Jones et al. (2012) and Van der Hulst et al. (2010) studied muscle activity during movement perturbations and gait, without fatiguing subjects, while Wilson examined muscle activation following fatigue. In studying muscle activity in older adults it is also possible that age may have factored into observable differences between research by Hanada et al (2011) and that of other authors, while the notation by Jones et al. (2012) of higher rectus abdominis activation among those with low back pain was in direct response to a backwards perturbation, which is opposite that which would occur during gait.

Lumbar lordosis.

Previously it was noted that lumbar lordosis increased in a non-healthy population following fatigue, while the opposite occurred in a healthy population (Wilson et al. 2006). In general, lumbar lordosis is highest during quiet standing, with slightly higher levels of lumbar lordosis during running vs. walking (Levine et al. 2007). Changes in surface grade from uphill to downhill have been shown to produce decreases in lumbar lordosis during both walking and running (Levine et al. 2007).

Increased levels of lumbar lordosis during lifting have been linked with increased levels of muscle activation in various muscles of the lumbar spine, and increased stability (Arjmand & Shirazi-Adl, 2005). Some researchers believe that increased lordosis is employed by those with low back pain as a protective mechanism in order to avoid additional back pain (Hart et al. 2009). The increased muscle activation could potentially lead to increased fatigability of the muscle groups of the lumbar spine (Arjmand & Shirazi-Adl, 2005; Hart et al. 2009), while low muscular endurance of the lumbar spine has been linked with the development of low back pain (Biering-Sorensen, Thomsen, & Hilden, 1989). A more lordotic posture has also been shown to result in increased compressive and shear forces during loading (Arjmand & Shirazi-Adl, 2005).

Supporting the hypothesis that high fatigability can lead to low back pain development is research that found that two types of strength programs aimed to increase strength in both the dorsal and ventral muscles of the trunk, thus reducing fatigability, will lead to decreases in pain for those suffering from low back pain (Franca, Burke, Hanada, & Marques, 2010), in addition to modeling research which found that spinal stability increased as a result of increasing intra-abdominal pressure (Stokes, Gardner-

Morse, & Henry, 2011). It should be noted that the modeling research did not find voluntarily increasing muscle activation at specific muscles to be effective in increasing intra-abdominal pressure, although increased muscle activation of the obliques and empos te g abdominis did lead to a mild increase in stability (Stokes et al. 2011). These results indicate that more research needs to be performed in this area in order to establish a firmer relationship between muscle strength/endurance and low back pain, while also leading to the potential but as yet undetermined hypothesis that while it may be beneficial to those with low back pain to employ a strategy to increase lumbar lordosis during gait, it may be unadvisable for healthy individuals whom have never experienced low back pain to do so.

The question generally left unanswered by current research on low back pain is whether or not the common abnormalities demonstrated by those with low back pain precede or follow the emergence of said pain (Vogt et al. 2001). Seay et al. (2011) indicated that even following the resolution of low back pain, individuals continue to display movement patterns that are similar to those suffering from pain. These results do not give strength to the adoption of either argument. Therefore it is unknown if the adoption of gait characteristics, or muscle activation, similar to that employed by those with low back pain would lead to the development of low back pain in a healthy population.

Running shoes

Over the past three decades, modern running shoes or orthotics have been commonly prescribed by medical practitioners as aids in helping to prevent running injuries (Johnston, Taunton, Lloyd-Smith, & McKenzie, 2003). Proponents of the use of modern running shoes suggest that shoes can reduce injury rates by reducing impact forces during running (Richards et al. 2009). Recent literature calls into question this practice, suggesting that the prescription of running shoes is not evidence based, and calling for further evaluation of footwear and running injuries (Richards et al. 2009). While opinions may be mixed regarding the use of footwear, documented differences can be seen when runners are tested with or without shoes. As can be seen from the previously reviewed literature, the modern running shoe has been linked with many biomechanical changes during running, ranging from kinematic changes (Bishop et al. 2006; Hardin et al. 2004; Lee et al. 2011; Lieberman et al. 2010; Lohman et al. 2011) to changes in muscular activity (Bird et al. 2003; Kelley et al. 2011; Mündermann et al. 2006; Nawoczenski & Ludewig, 1999; Von Tscharner et al. 2003). Studies have also been performed which have evaluated the effect of footwear other than running shoes on running kinematics/kinetics (Lee et al. 2011), and on injury risk resultant to wearing shoes with high heels (Lee et al. 2001).

Surface stiffness.

Leg stiffness resultant to running on surfaces of varying degrees of stiffness has been examined. A well established finding of this line of research has determined that as surface stiffness increases, leg stiffness decreases (Bishop et al. 2006; Divert, Baur, Mornieux, Mayer, & Belli, 2005; Ferris, Liang, & Farley, 1999; Hardin et al. 2004), although Divert et al. (2005) recorded somewhat conflicting results as in this study shoe surface stiffness increased across time leading to decreases in leg stiffness, while runners in a barefoot condition recorded higher values for leg stiffness than while shod despite running on the supposedly stiffer surface of the treadmill bed. This is in agreement with results from De Wit, De Clercq, & Aerts, (2000) who also recorded higher levels of leg stiffness during barefoot running, but in contrast to results from Bishop et al. (2006) who recorded lower levels of leg stiffness during barefoot hopping.

Further research has demonstrated similar results during hopping (Farley & Morgenroth, 1999), while research examining head and trunk mechanics has measured increased levels of stiffness when surface stiffness decreases during walking (Nadeau, Amblard, Mesure, & Bourbonnais, 2003). Findings of various studies vary in relation to specific variables that are primarily thought to affect stiffness, with hip (Hardin et al. 2004), knee (Hardin et al. 2004; McMahon, Valiant, & Frederick, 1987), and ankle (Bishop et al. 2006; Farley & Morgenroth, 1999; Hardin et al. 2004) kinematics being thought to play a role in the development of leg stiffness. Changes in knee and hip kinematics as a result of surface stiffness come due to increased extension at ground contact when running over a stiffer surface, in addition to higher peak angular velocities being recorded at the hip and knee (Hardin et al. 2004). Changes in ankle kinematics as a result of surface stiffness included increased peak angular velocity when the surface was of a stiffer material, and increased angle of plantarflexion at toe-off (Hardin et al. 2004), with empos te ankle dorsiflexion at ground contact being recorded when running in a shod condition as compared with running barefoot (Bishop et al. 2006).

Shoe material.

While material stiffness and material hardness are somewhat interrelated it is uncommon for both variables to be tested concurrently when performing research, with some exceptions noted (Divert et al. 2005). Material hardness has most often been tested during running in relation to the use of custom insoles that are inserted into the running shoe (Chen, Nigg, & de Koning, 1994; Nigg, Herzog, & Read, 1988; Ogon et al. 2001). Additional research has not specifically examined shoe material, but has tested variables across time as related to shoe age or amount of shoe mileage (Kong, Candelaria, & Smith, 2008; Rethnam & Makwana, 2011).

Research examining material hardness has examined materials of different hardness ranging from Shore 9.5 to Shore 34 (Nigg et al. 1988; Ogon et al. 2001), while another study used the more subjective insole comfort, as described by study participants, with which to differentiate various insoles (Chen et al. 1994). Nigg et al. (1988) found that variations in insole hardness caused no significant differences related to vertical impact forces, nor were systematic kinematic differences observed when shoes were worn with the various insoles. This is in general agreement with Ogon et al. (2001) who recorded few differences in relation to muscle activity and accelerometry of the lower back, despite recording general trends supporting faster muscular onset with increasing insole hardness when comparing running with or without various insoles. The lack of systematic, significant differences when comparing insole materials in these studies can be attributed in part to energy storage/return relative to shoes and the lower extremity as Shorten (1993) reported that the energy return gained from a running shoe can affect the kinematics of the ankle and foot, but will provide a much smaller potential change than that which can be obtained from passive energy transfer or strain energy in the lower extremity.

In studies examining new and old footwear, it was found that runners wearing old shoes recorded lower values for plantar pressure than runners wearing new shoes (Rethnam & Makwana, 2011). Further studies have also reported that stance time increases, forward lean decreases, maximum dorsiflexion decreases, and plantar flexion at toe-off increases when subjects run in old running shoes (Kong et al. 2008). No differences were noted during the studies in relation to the properties of the various shoe materials, indicating that shoe cushioning properties were fairly independent of material (Rethnam & Makwana, 2011). Researchers in these studies conclude that the body adapts to changes in shoe material across time in order to maintain a constant external load on the system (Kong et al. 2008), while also suggesting that shoes should be given a substantial breaking in period prior to running in them (Rethnam & Makwana, 2011).

Summary

As footwear has been shown to cause changes in low back muscle activity (Bird et al. 2003; Ogon et al. 2001), with changes in low back kinematics being seen as a product of speed (Seay et al. 2011), surface inclination (Levine et al. 2007), and fatigue (Hart et al. 2009), the relevance of performing additional research examining low back function under various shoe conditions is warranted. Given additional research that involves the effect of footwear on leg stiffness (Bishop et al. 2006; Divert et al. 2005; Hardin et al. 2004), with further research recording differing values of leg stiffness among those suffering from low back pain (Hamill et al. 2009), the relevance of performing this type of research is substantiated further.

Chapter 3

Methods

The purpose of this study was threefold, to determine the material characteristics of a minimalist running shoe tested with and without a commercially available shoe insert (1), and to determine if there are differences in kinematics (2) or muscle activity (3) when the minimalist shoe is worn with and without the same commercially available insert during treadmill running. The study protocol was determined and performed in accordance with this purpose, being performed as follows:

Subjects

Subjects were ten individuals (men n = 4, women n = 6, age 33.3 ± 13.0 years, height 168.5 ± 9.8 cm, mass 64.5 ± 13.5 kg) who were recruited from the Las Vegas area community by word of mouth. Inclusion criteria for participation stipulated that subjects must have had no history of low back pain, must have had no known health problems related to a leg length discrepancy, were able to comfortably fit into the designated shoes in either the men's or the women's sizes, were comfortable running on a treadmill for up to 10 minutes, and performed aerobic exercise at least three times per week. Study participants reported running an average of 44.4 km (±22.2 km) per week while running 4.7 days (±1.5 days) per week. Exclusion criteria for this study stipulated that subjects must not be pregnant. All subjects provided written informed consent prior to performing data collection (Appendix I).

Instrumentation

Footwear.

This study used two specific running shoes for data collection: 1) Altra Adam™ (Figure 1) primarily used by male participants, and 2) Altra Eve™ (Figure 2) primarily used by female participants. A single commercially available running insole, the Spenco® PolySorb® Crosstrainer (Figure 3) was used for the study. The Altra Adam™, and Eve™ are both classified as minimalist running shoes with a reported outsole thickness of 3.4mm and no midsole. The shoes are packaged with a rubber insole reported to be 3mm, which can be placed in the shoes if desired. While there are some changes in the last of the different shoes, the overall differences between the shoes are minimal. The Spenco® PolySorb® Crosstrainer is a shoe insole designed to add cushioning and support in place of the insoles that come with a standard pair of athletic shoes. The construction of the insole consists of a polyurethane arch and heel support that covers the bottom of the insole until just past the arch. Styrene Butadiene Rubber covers the additional space on the bottom of the insole from the end of the arch to the toes. The center of the insole consists of a cushioning material which covers the length of the insole, and which is also placed in the center of the heel in the polyurethane arch and heel support. The insole is covered by a fabric top cloth. Performance conditions for the study included: 1) running in the Adam™ or the Eve™ without any inserts being placed in the shoes, and 2) running in the Adam™ or the Eve™ when the Spenco® PolySorb® Crosstrainer and 3mm rubber insole were placed in the shoes.


Figure 1: Altra Adam™


Figure 2: Altra Eve™

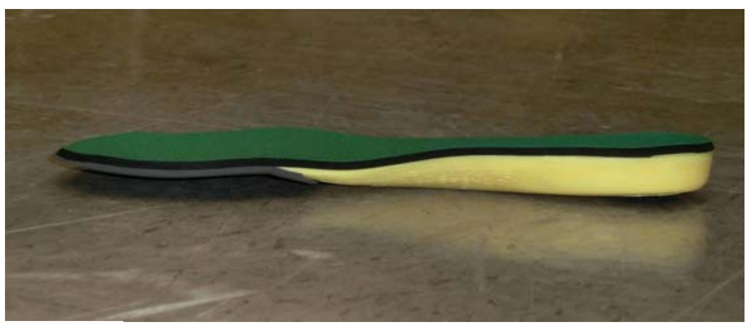


Figure 3: Spenco® PolySorb® Crosstrainer

Impact testing system.

Shoe impact testing was performed after completion of the performance conditions, using a mechanical impact tester (Exeter Research Inc. Brentwood, NH) (Figure 4) used jointly with Impact Plus (version 3.0) software. The impact tester measures different variables related to dropping an instrumented missle head with a mass of 8.5 kg onto the surface of a given object. Study dependent variables included: peak acceleration (g’s) of the missle head, and peak pressure (Kpa) which are both measured during impact testing. Heel to forefoot drop height was also measured (mm) by first determining material thickness (mm) which was measured by adjusting the height of the missle head to the material height followed by lowering the missle head to the test platform and then calculating the difference. This was performed at the heel and forefoot of the shoes in order to determine heel to forefoot drop height, with average differences of 0.7 mm (5.4 mm heel; 4.6 mm forefoot) when no insoles were placed in the shoes, and 9.0 mm (23.1 mm heel; 14.1 mm forefoot) when insoles were present. Material thickness as measured is accurate to within 0.5mm.

Figure 4: Impact Tester

Electromyography and kinematics.

Subject electromyography and kinematics were obtained by first cleaning the electrode placement sites with alcohol pads, abrading the skin, and if necessary removing any hair. Electrode placement then occurred, with dual electrodes (Ambu Blue Sensor N; Ambu Inc. Ballerup, DK) being placed on the left side of the body and an inter-electrode distance of 25mm at each muscle site. Muscle sites which were instrumented included the rectus abdominis, with the electrode center being placed at equal height and 30mm lateral to the navel (Figure 5), the erector spinae, with electrode placement 30mm lateral to the third lumbar vertebrae (Figure 6), and on the belly of the biceps femoris muscle (Figure 7), with a single electrode being placed on the posterior superior iliac spine for grounding purposes (Figure 6). Subjects were also instrumented with two electrogoniometers (Biometrics, Ltd. Gwent, UK; 1500Hz), with one being placed on the lumbar spine (Model G150B) with the bottom of the electrogoniometer being placed at the sacrum (Figure 6), while the other was placed on the lateral side of the left knee (Model G150) with the center of the electrogoniometer being placed over the knee joint axis (Figure 7). Leads from a telemetry system (TeleMyo 2400T, G2; Noraxon USA Inc. Scottsdale, AZ; 1500Hz) were attached to all electrodes, while electrogoniometers were also connected to the telemetry system. All leads and electrogoniometers were adhered to the subject's skin using the minimal amount of adhesive tape necessary to prevent tension being placed on the leads or electrodes during running. Electrogoniometers were attached to the skin following the obtaining of a zero offset in the data acquisition software where the current signal from the electrogoniometer was set as zero with subjects standing in a relaxed position, and electrogoniometers being placed in a neutral position on a flat countertop.

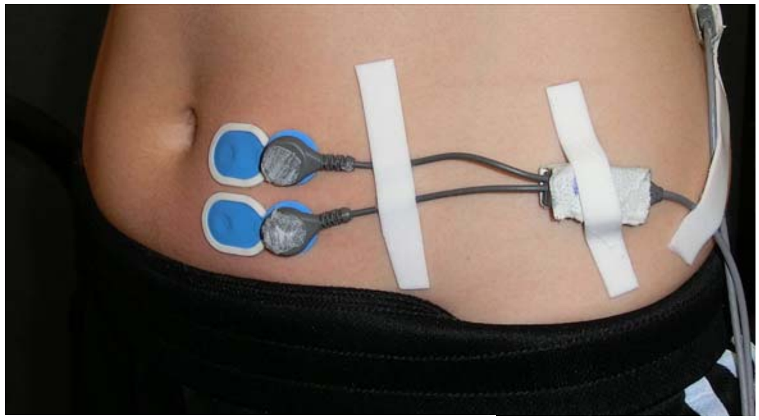


Figure 5: Rectus Abdominis electrodes

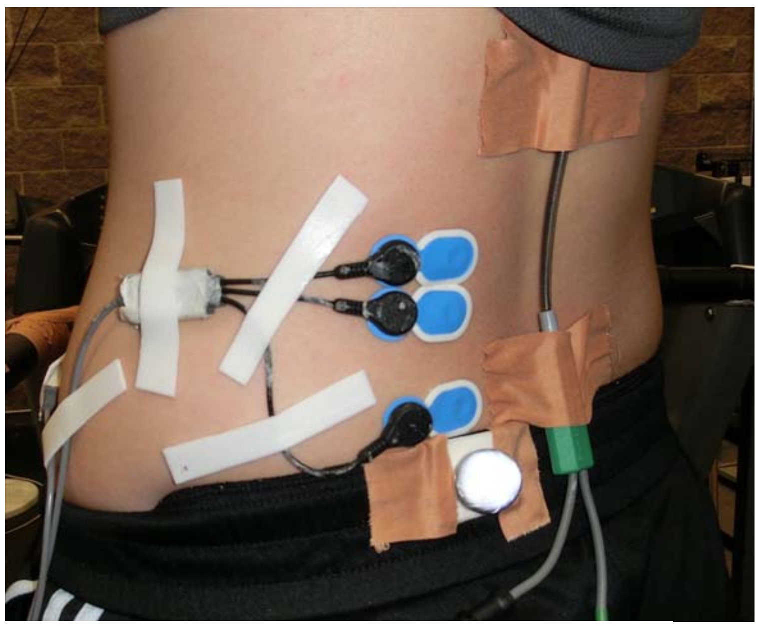


Figure 6: Lumbar Electrogoniometer, Erector Spinae electrodes, and grounding electrode

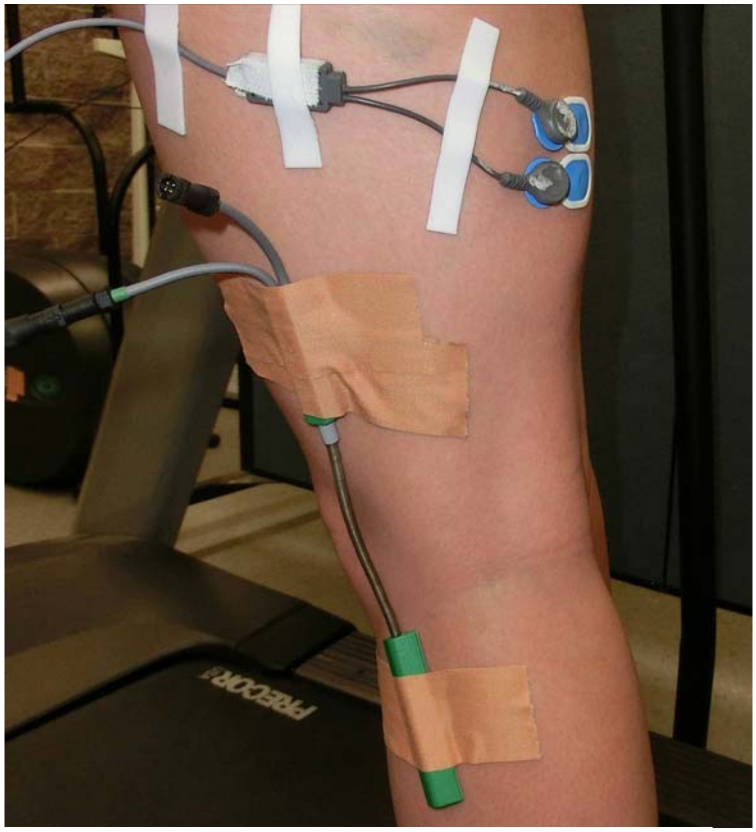


Figure 7: Knee Electrogoniometer, and Biceps Femoris electrodes

Treadmill running.

All running during test conditions, including any necessary warm-up was performed on a treadmill (C966; Precor, Woodinville, WA). Preferred running speed was determined blindly by the individual participants, who were instructed to choose a speed at which they would run if they were to run continuously for 30 minutes at an easy effort.

Test Procedure

Running protocol.

Following receipt of written informed consent as approved by the Institutional Review Board at the University of Nevada, Las Vegas, and verification that inclusion criteria had been met, subjects were measured for height, weight, and bilateral functional leg length which was measured from the umbilicus to the medial malleolus. Subjects were then given the opportunity to warm up on the treadmill, and preferred speed was determined. Following warm up and determination of preferred speed, subjects ran in the test shoes under each condition to become accustomed with running in the test shoes, which was then followed by instrumentation of all electrodes. Maximum voluntary isometric contractions were then performed lasting five seconds each for all three muscle groups being tested.

The maximum voluntary isometric contraction of the rectus abdominis was obtained by having subjects perform an abdominal crunch, holding the position once the shoulders were lifted off of the table. The maximum voluntary isometric contraction of the erector spinae was obtained by having subjects lie with their upper extremity hanging off of a table, and instructing them to raise their chest above the level of the table and extend their arms out in front of their heads. The maximum voluntary isometric contraction of the biceps femoris was obtained by having subjects attempt to flex their leg at the knee while resistance was applied to the heel preventing any movement from taking place.

Prior to and following performance of the maximum voluntary isometric contractions a zero offset was obtained for all signals, with electrogoniometer instrumentation occurring after the zero offset was obtained the second time. Subjects then ran on the treadmill at their preferred speed while wearing the test shoes in each condition. Condition order was counterbalanced, with running taking place for two minutes, followed by data being collected for an additional 45 seconds.

Shoe impact testing.

Following subject data collection, the mechanical impact tester was used to test the material properties of the footwear. Women's sizes 6-9 of the Eve™ and men's sizes 8-11 of the Adam™ were impact tested (3000hz), with testing being performed at the heel and the forefoot of both shoes (Figures 8 and 9). Testing was performed twice as all shoes were impact tested with and without the added inserts. During both heel and forefoot testing under all conditions an additional piece of hard rubber was placed underneath the surface of the shoe in order to avoid damaging the mechanical impact tester when the shoe was tested without the added inserts. Testing followed a modified American Standard for Testing Materials (ASTM) test procedure (ASTM F-1614). A load of 8.5 kg with a missle head diameter of 45 mm was dropped from a height of 50 mm resulting in 5 ± 0.5 Joules of energy at impact, with twenty pre-impacts being performed to condition the material, followed by data being collected during ten test impacts.

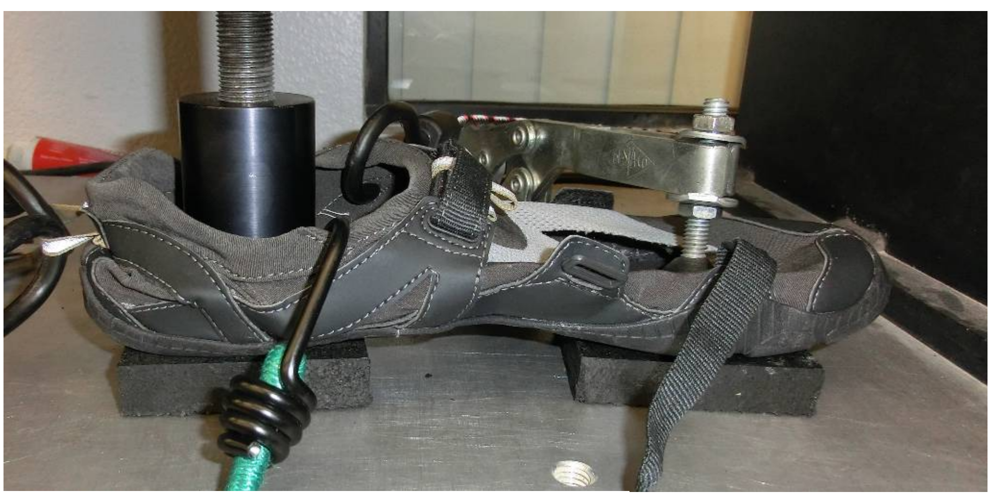


Figure 8: Impact testing at the heel of a test shoe


Figure 9: Impact testing at the forefoot of a test shoe

Data Reduction

Running data.

Data from electromyography and kinematic data collection were reduced, with analysis being performed through the use of custom computer programs written using the MatLab computer programming language (The Math Works Inc., Natick, MA) (Appendix II). Average muscle activity from the rectus abdominis, erector spinae, and biceps femoris were calculated by fully rectifying the signal, with data normalization occurring by calculating the greatest one second average for each muscle when performing maximum voluntary isometric contractions and by relating muscle activity to 100% of the maximum voluntary isometric contraction. Further data reduction took place by smoothing signal data with a fourth order Butterworth filter (250hz). Data from the stance phase of ten consecutive running strides were extracted with the instances when peak knee extension occurred, as measured by the electrogoniometer instrumented to the left knee, being used to approximate the timing of the stance phase of running.

Kinematic data obtained from the electrogoniometers instrumented to the knee and lumbar spine were analyzed by extracting stance data from the same ten consecutive running strides used for analysis of electromyography. Sagittal plane range of motion, and average position of the lumbar spine during stance were calculated using the electrogoniometer placed over the lumbar spine. Values for peak flexion, peak extension, and range of motion during stance were calculated in the sagittal plane using the electrogoniometer placed over the knee.

Shoe impact testing.

Peak acceleration and peak pressure were determined during impact testing of the heel and the forefoot of the shoes during both shoe conditions, while shoe material thickness was measured prior to performing test impacts. The Impact Plus 3.0 software calculated values for all variables. Peak acceleration was determined using the first central difference method with the formula $a_i = (v_{i+1} - v_{i-1})/(t_{i+1} - t_{i-1})$ and was subsequently converted to units of gravity (g). Velocity was calculated with the formula $v_i = (x_{i+1} - x_{i-1})/(t_{i+1} - t_{i-1})$. Peak pressure was determined by using the equation Kpa = F/A, where force in newtons was calculated using the formula $\Sigma F = m * a$, and area was calculated based upon the diameter of the missile head. The data from all ten test impacts was then returned and averaged in order to prepare data for analysis.

Statistical Analysis

Running data.

Nine dependent variables were analyzed, with average muscle activation during stance for the rectus abdominis, erector spinae, and biceps femoris being analyzed using electromyography, while low back range of motion, average position of the low back, knee range of motion, peak knee extension, and peak knee flexion were analyzed using electrogoniometers. Paired T-tests ($\alpha = 0.05$) were performed for all variables in order to determine differences between the two shoe conditions.

Shoe impact testing.

Two dependent variables were analyzed, including peak pressure, and peak acceleration. Independent T-tests were performed for each separate shoe condition comparing heel and forefoot impact characteristics without comparing conditions. Conditions were then compared by performing Independent T-tests comparing impact characteristics when shoes were worn with and without the added inserts, with separate analysis being performed based upon impact location (heel or forefoot). An alpha level of 0.01 was used for all statistical analysis.

Chapter 4

Results

In order to better understand the relation of footwear cushioning to low back mechanics, the purpose of this study was to 1) determine the material characteristics of a minimalist running shoe tested with and without a commercially available shoe insole, 2) determine if there are differences in lower back or knee kinematics when minimalist shoes are worn with and without a shoe insole during treadmill running, and 3) determine if there are differences in levels of muscle activation when minimalist shoes are worn with and without a shoe insole during treadmill running. The study was designed as such in order to come to a better understanding of the relation of footwear cushioning to low back mechanics during running, as little research has been performed which examines the relationship between footwear cushioning and low back mechanics during running, even though it has been reported that 70-85% of the population will experience an injury to the low back at some point in their lives (Andersson, 1999).

Shoe Impact Testing Results

All impact testing was performed with the test shoes being placed on top of a piece of black rubber in order to avoid damaging the impact tester. Observable differences for peak acceleration were recorded when impact test results were compared with and without the use of shoe inserts at both the heel ($t(7.3)=83.5$, $p<0.001$) and the forefoot ($t(7.2)=18.9$, $p<0.001$) of the respective shoes. Similarly, values for peak pressure were also significantly different when shoes were compared with and without inserts at the heel ($t(7.3)=83.3$, $p<0.001$), and the forefoot ($t(7.2)=18.9$, $p<0.001$). When impact test results were compared between the heel and the forefoot of the shoes without inserts being placed therein, no significant differences were observed for peak acceleration (t(14)=0.554, p=0.589) or for peak pressure (t(14)=0.534, p=0.601). Test results comparing the heel and forefoot when inserts were placed in the shoes resulted in significant observable differences for acceleration (t(14)=-36.2, p<0.001) and pressure (t(14)=-36.1, p<0.001). Results of impact testing can be viewed in Table 1.

Table 1 Impact Testing Data

		Heel		Forefoot		
		Acceleration (g)	Pressure (kPa)	Acceleration (g)	Pressure (kPa)	
Without Inserts	mean	23.88	1251.2	23.86	1250.4	
	std	0.06	2.9	0.07	3.6	
With Inserts	mean	11.51	603.3	20.22	1059.8	**
	std	0.42	21.8	0.54	28.3	
		*	*	*	*	

* = differences between test conditions (with or without inserts) are significant (p<0.001)

** = differences between test location (heel or forefoot) are significant (p<0.001) for all variables

Running Results

No statistically significant differences were identified following analysis of muscle activity (Table 2) with average activity of the erector spinae (t(9) = -0.017; p = 0.987), rectus abdominis (t(9) = 0.814; p = 0.437), and biceps femoris (t(8) = -2.011; p = 0.079) all recording p-values of greater than 0.05. Results for one subject were excluded from analysis for the biceps femoris as an adequate maximum voluntary isometric contraction was not obtained. Analysis of kinematic variables resulted in significant observable differences between conditions for maximum knee flexion during stance (t(9) = 3.560; p = 0.006) while no statistically significant differences were identified for additional kinematic variables (Table 3) including mean back angle (t(9) = -0.554; p =

0.593), back range of motion (t(9) = -1.489; p = 0.171), maximum knee extension (t(9) = 1.743; p = 0.115), and knee range of motion (t(9) = -1.598; p = 0.144).

Table 2		**Muscle Activity**		
		ES EMG (%)	RA EMG (%)	BF EMG (%)
Without Inserts	mean	15.4	9.7	35.9
	std	7.1	3.7	19.0
With Inserts	mean	15.4	9.3	41.0
	std	5.6	2.9	23.9

Table 3		**Kinematics**				
		Mean Back Angle (deg)	Back ROM (deg)	Max Knee Extension (deg)	Max Knee Flexion (deg)	Knee ROM (deg)
Without Inserts	mean	26.7	10.2	2.7	30.3	27.6
	std	9.9	3.8	6.9	5.5	8.5
With Inserts	mean	27.0	11.0	3.6	32.2	28.6
	std	10.4	3.4	6.7	4.7	7.4
					*	
* = differences between test conditions (with or without inserts) are significant (p<0.05)						

Chapter 5

Discussion

Discussion of Results

Results of impact testing identified differences between the test shoes when measured with and without the use of an added insert, while also displaying differences between the heel and the forefoot of the test shoes when they were tested with the added inserts. Results of muscle activity and kinematics resulted in significant differences being observed between conditions for maximum knee flexion, with no additional differences being observed. While few significant differences were observed among the different variables during treadmill running, trends were observed for knee kinematics and for muscle activity at the rectus abdominis and biceps femoris which warrant further explanation. Individual subject data for maximum knee extension revealed that 7 of 10 participants experienced greater knee extension at ground contact when wearing the shoes without the added inserts (Figure 10). Similarly, individual subject data indicated that 8 of 10 subjects had a greater range of motion at the knee when wearing the shoes with the inserts (Figure 11). When observing results for muscle activity, 7 of 10 subjects recorded greater values for rectus abdominis activity when wearing the shoes without the inserts added (Figure 12), while 8 of 9 subjects recorded values for biceps femoris activation that were lower without the added inserts (Figure 13).

Figure 10: Average Maximum Knee Extension

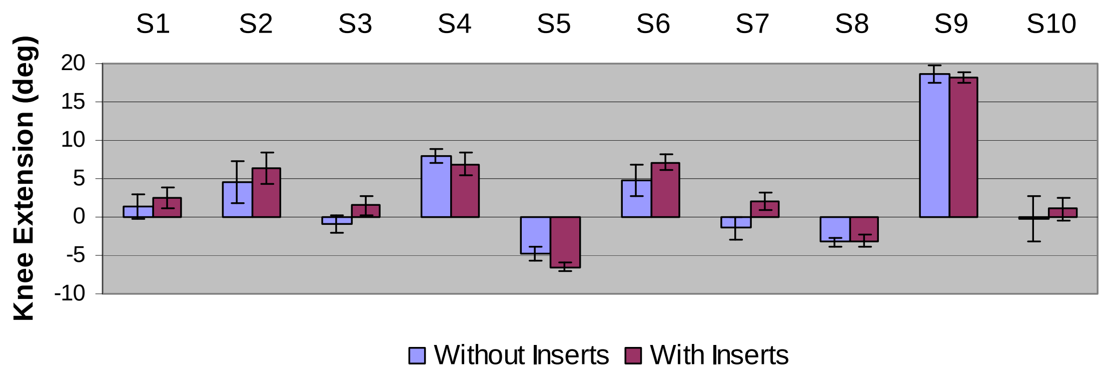


Figure 11: Average Knee Range of Motion

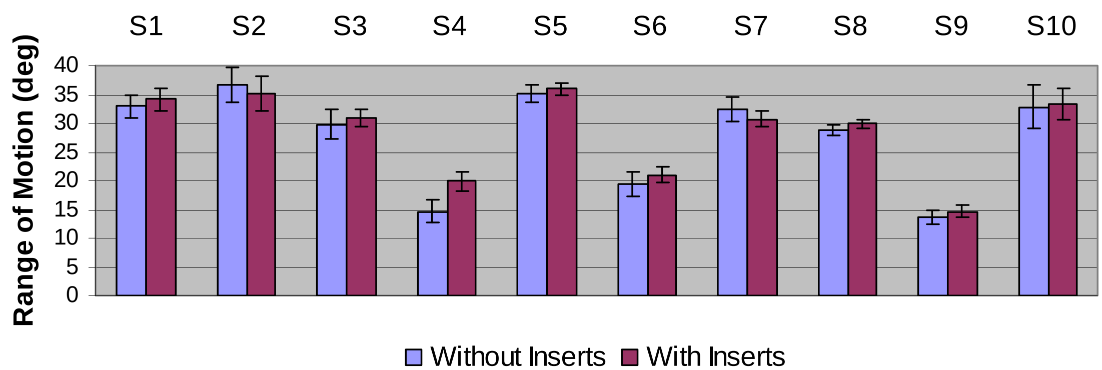


Figure 12: Average Rectus Abdominis Muscle Activity

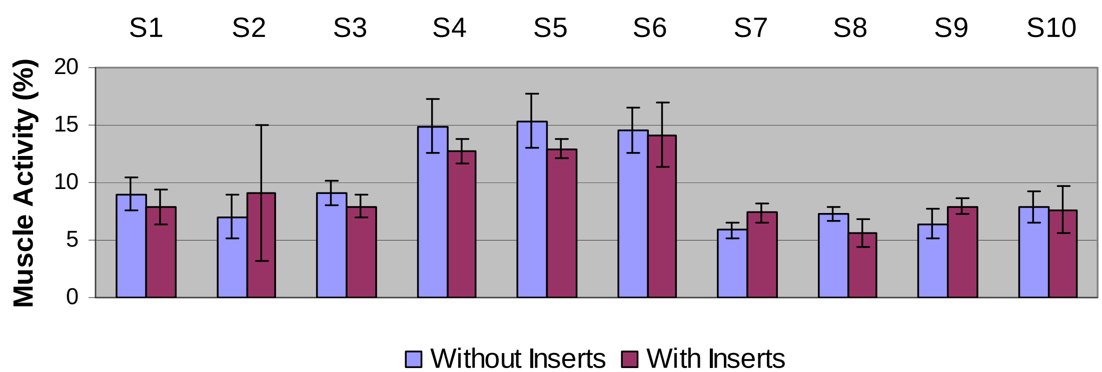
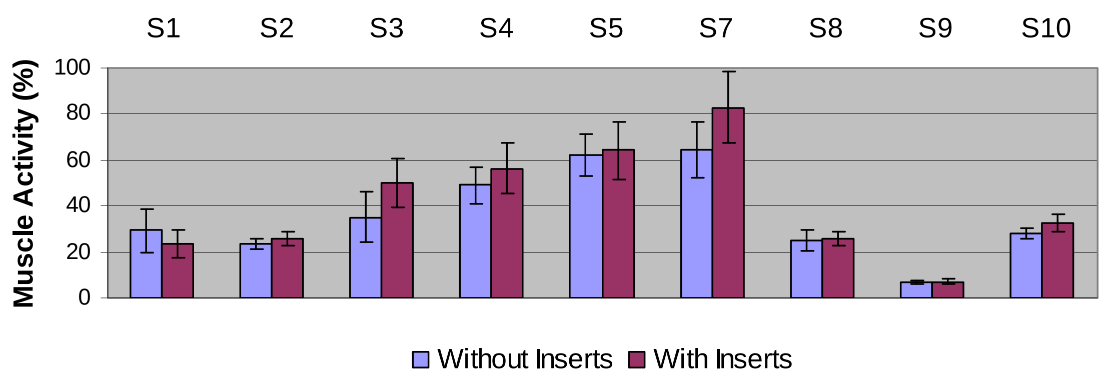


Figure 13: Average Biceps Femoris Muscle Activity



Muscle activity.

Changes in muscle activation and onset of muscle activity of the erector spinae have been recorded previously as a result of wearing heel lifts (Bird et al. 2003) or high heel shoes (Lee et al. 2001). The current research did not further substantiate the results of Lee et al. (2001) finding no significant differences in muscle activation between running conditions despite recording a greater heel to forefoot drop height when inserts were placed in the shoes (9.0±1.1mm vs. 0.7±0.5mm) while also recording a net heel height that was greater by 17.8mm on average. As the onset of muscle activity was not examined in this experiment it is unknown if results would have been in agreement with those recorded by Bird et al. (2003). It is possible that muscle activity showed no significant differences between shoe conditions due to the comparatively small change in heel height relative to that which exists due to the use of high heel shoes while also utilizing a shoe material that is compressive, unlike that of a high heel shoe. Additionally it is noted that muscle activity in the current study was only examined during the stance phase of gait, while Lee et al. (2001) determined average muscle activity across an entire stride. As the swing phase was excluded from the current analysis, any anticipatory effects that may exist relative to the use of different types of footwear can not be determined, potentially explaining the lack of significant differences in muscle activation observed in this study.

Results also differ in relation to those reported by Nadaeu et al. (2003) who reported increased stiffness of the trunk during walking on a foam pad versus the bare floor. It is possible that the necessary muscular stimulus from running itself was greater than any differences related to the material placed in the shoes. This thought is partially substantiated by Shorten (1993), who reported that energy transfer from shoes was modest in comparison to that which occurred due to passive energy transfer within the body. It is also possible that changes in trunk stiffness during running may be related to causes other than increases in trunk muscle activation.

Despite the lack of statistically significant group results, it was noted that the majority of study participants increased muscle activation at the rectus abdominis when inserts were not worn in the shoes, while most participants also experienced reduced levels of biceps femoris activation during this same condition. The lack of significant results may be due in part to the low number of study participants as variability among subjects was high, especially in relation to muscle activity recorded at the biceps femoris as results for this variable did approach significance ($p = 0.079$). Despite the observed trends this still seems somewhat unlikely as variability was also high among subjects with observable differences being relatively small. Muscle activity related to foot strike at ground contact is of additional interest as it was observed that the three subjects who recorded increased rectus abdominis activity when inserts were placed in the shoes all changed footstrike at ground contact, running with a heel strike pattern with inserts added to the shoes, and running with a midfoot or forefoot strike pattern when inserts were not present. This trend was not observed when study subjects did not change footstrike pattern at ground contact from one condition to the next, as all seven of these subjects recorded increased rectus abdominis activity when inserts were not placed in the shoes. These tendencies among study participants lead to the potential hypothesis that greater biceps femoris activation is required to stabilize the body when running on a softer surface, with increased rectus abdominis activation resulting in greater stability of the trunk when running on a harder surface. As results of this study showed no differences between conditions, this potential hypothesis would need to be re-examined, by controlling for foot strike at ground contact and recruiting a greater number of study participants, in order to determine its validity.

Study design also makes it impossible to determine what type of effect that the different types of footwear would have on muscle activity when examined for differences across longer periods of time either during a single bout of running, or during a training cycle. These types of studies could potentially shed more light on the findings of the current study, as study participants would be given the opportunity to develop long term muscular adaptations to the cushioning in the various shoes. When examining rectus abdominis muscle activation it may also be beneficial to perform future running trials at faster speeds, as research has indicated that the rectus abdominis does not activate at high levels unless the speed is relatively fast (Mann et al. 1986).

Kinematics.

While no changes in kinematics were observed as a result of footwear condition, there did seem to be a trend toward greater knee extension being observed at ground contact. If true, this would indicate that running in footwear containing minimal cushioning could potentially result in greater extension being maintained throughout the stance phase of gait even if the range of motion is similar as it was shown in this study that maximum knee flexion was lower, and thus extension was greater, when inserts were not placed in the shoes. This thought would be in agreement with previous research that found that knee extension at ground contact increased as surface stiffness increased (Hardin et al. 2004). Bishop et al. (2006) found similar results to those observed in this study, with no significant results being seen for knee extension at ground contact when running barefoot vs. shod. It was postulated that the impact force of ground contact is absorbed primarily by the ankle during barefoot running (Bishop et al. 2006). Additionally, while the changes in knee flexion/extension led to no significant differences being observed for knee range of motion among subjects in this study, eight of the subjects decreased knee range of motion when running in the shoes without the added inserts. This was in general agreement with results from Bishop et al. (2006) who similarly found a decreased range of motion at the knee during barefoot running vs. shod running, while displaying no statistical significance between conditions. This would indicate that knee stiffness was potentially greater in the condition where inserts were not added to the shoes. This could have been related to the unfamiliarity of minimalist running shoes to the individuals who took part in the current study as none had previously run in minimalist footwear, with the only barefoot running reported having taken place on grass surfaces. Any changes due to shoe unfamiliarity were undoubtedly small as all kinematic differences were insignificant. Based upon these results, it seems reasonable to suggest that the surface material in running footwear is not the main determinant of knee stiffness. This result can be partially explained by research from Farley & Morgenroth (1999) who determined that during hopping, ankle stiffness is the primary determinant of leg stiffness, with further substantiation of results provided by Bishop et al. (2006). This could lead to the potential hypothesis that stiffness at the knee would be affected less than stiffness at the ankle when modifications to footwear are made. Lower extremity joint stiffness has been linked with the development of low back pain (Voloshin & Wosk, 1982) and additional research has found increased knee joint stiffness in those with low back pain (Hamill et al. 2009). When referring to the results of these previous studies, while also referring to results of the present study where range of motion at the knee was not significantly different, it seems more likely that if low back pain developed following modifications being made to footwear, it would likely occur prior to the development of knee stiffness rather than low back pain developing due to the emergence of increased knee stiffness. Further research is warranted to substantiate such claims.

It is also worth noting that proponents of minimalist footwear advocate the adoption of a forefoot running style in which that portion of the foot contacts the ground first at ground contact, while runners who wear standard running shoes have the tendency to make initial ground contact with the rearfoot rather than the forefoot (Lieberman et al. 2010). Among the participants of this study seven ran with a habitual rearfoot strike pattern to their gait, with only three of the seven subjects switching to a midfoot strike (where the heel and forefoot make ground contact simultaneously) or forefoot strike pattern. In order for future studies to reach conclusions in regards to the effect on lower extremity knee joint stiffness due to running in shoes with minimal or standard amounts of cushioning it will be necessary to control for foot placement at ground contact, especially when considering the trend for rectus abdominis muscle activity that was observed in this study when subjects did change their foot strike pattern at ground contact.

In relation to lower back kinematics, no trends were observed relative to the average angle of the low back or range of motion. It was previously reported that the body adjusts when running over different surfaces in order to keep a constant external load on the system (Kong et al. 2009), while the main determinant of leg stiffness during hopping is the ankle (Farley & Morgenroth, 1999). Based upon these studies and the results of the current study, it can be hypothesized that the majority of kinematic adaptations related to shoe cushioning occur in the lower extremity which would lead to more minute, or non-existent kinematic differences being observable in the low back. As no differences were seen between conditions for low back kinematics in this study, there appear to be no advantages or disadvantages to low back kinematics when running in shoes with either minimal or greater amounts of cushioning.

Conclusions

During treadmill running peak knee flexion during stance increases following the addition of a cushioned insert to a shoe with minimal cushioning. While it is possible that some additional differences would have become apparent if a larger sample size had been used, or if foot strike at ground contact would have been controlled for, additional differences in muscle activity and kinematics of the trunk and lower extremity as a result of wearing shoes with different cushioning properties did not exist in this study. This leads to the conclusion that when performing a single bout of treadmill running changing the cushioning material of a shoe will not provide a differing stimulus to the low back, leading to no differences in low back muscle activity or kinematics. Additionally, despite the differences observed for maximum knee flexion there does not appear to be any advantage, in relation to low back muscle activity or kinematics, to wearing footwear of either minimal or greater cushioning during a single bout of treadmill running when the duration of time spent running is relatively short in duration.

Appendix I

Informed Consent

INFORMED CONSENT

Department of Kinesiology and Nutrition Sciences

TITLE OF STUDY: The Effect of Footwear on Mechanics of the Lower Back During Treadmill Running

INVESTIGATORS: J. McClellan, J.S. Dufek, Ph.D.

CONTACT PHONE NUMBER: J. McClellan, 702-573-8169

Purpose of the Study

You are invited to participate in a research study. The purpose of the study is to investigate the effects of an athletic shoe, under different conditions, on the mechanics of the lower back during human treadmill running.

Participants

You are being asked to participate in the study because you are an apparently healthy individual between the ages of 18-50, with no health related problems resultant to leg length discrepancy, and you are not pregnant. In addition, you are able to fit comfortably into the available test shoes. It is also expected that you voluntarily perform exercise at least three times per week, and are able to run unassisted on a treadmill for up to 10 minutes, with rest.

Procedures

If you volunteer to participate in the study, you will be asked to do the following: 1) allow your age, height, weight, and bilateral leg length to be measured and recorded, 2) allow for instrumentation using standard equipment used to test muscle activity, determine joint angles, and perform video capture (this will include shaving and cleaning/abrading the electrode placement sites), 3) perform a standard running warm-up on the treadmill in order to determine your preferred speed, 4) run on a treadmill for up to 10 minutes at your preferred speed while wearing the test shoe with and without a commercial insole at which time video, joint angles, and muscle activity will be recorded.

Benefits of Participation

There may be no benefits to you as a participant in this study. However, you may gain some information about running shoes or inserts relative to personal preferences such as comfort, etc. The study may also benefit research in general by adding to the available literature concerning lower back mechanics.

Risks of Participation

There are risks involved in all research studies. This study may include only minimal risks. It is possible that you may trip or fall during treadmill running. We will ensure that the treadmill bed is clean and free of hazards. Also, you may potentially feel sore or acquire a running related injury. The electromyography sensors may also cause skin irritation or discomfort upon removal, which may last for a few days after participating in the study.

Cost/Compensation

There will be no financial cost to participate in this study. The study will take between 60-90 minutes of your time. You will not be compensated for your time.

Contact Information

If you have any questions or concerns about the study, you may contact Jeffrey McClellan at 702-573-8169. 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 research 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 only be visible from the neck down in the camera field of view.

______________________________ ______________

Signature of Participant Date

____

Participant Name (Please Print)

Appendix II

Matlab

```
%This program was written to process the EMG data collected
%for Jeff’s thesis during treadmill running.
%3/21/2012
%modified dwr_2010.m

clear;
clc;
warning off

%Identify critical variables
   %Program parameters
   subjects_to_process     = 1;    %process one subject at a time
   conditions_to_process   = 1;
   trials_to_process       = 1;

   start_with_subject      = 1;
   start_with_condition    = 1;
   start_with_trial        = 1;

   %parameters for my_save
   avg_file       = [‘s’ int2str(start_with_subject) ‘c’ int2str(start_with_condition)
‘avgemg.txt’];
   rms_file       = [‘s’ int2str(start_with_subject) ‘c’ int2str(start_with_condition)
‘rmsemg.txt’];
   output_file     = [‘s’ int2str(start_with_subject) ‘junk3.txt’];       %output file name
   extraoutput_file   = [‘s’ int2str(start_with_subject) ‘c’ int2str(start_with_condition)
‘kindata.txt’];       %for positions of BF

   %parameters for my_fopen
   directory   = [‘c:\Biomech\jeff\LBR’];  %directory where data are located
   headers     = 5;                        %number of headers in *.asc
   columns     = 7;                        %number of columns
   rows        = inf;

   %muscles
   musonecol   = 2;    %Lumbar Erector Spinae
   mus2col     = 3;    %Rectus Abdominis
   mus3col     = 4;    %Biceps Femoris
   backelgon   = 5;    %Back Elgon
   muscol      = [musonecol mus2col mus3col];
   kneelgon   = 6;    %Knee Elgon

   %for normalization: identify the time (in secs) to calculate average
   %the program (lbropennormalize2012) will determine the greatest 1-second average
   %for each muscle from the MVC files
   avgtime = 1;
```

```
   %EMG collection details
   fs         = 1500;        %sample frequency
   fc         = 300;         %cutoff frequency
   fcgen      = 500;         %cutoff frequency for general smoothing routine
   EMGwin     = 15;          %time to calculate average EMG across
   precision  = 4;           %number of decimals

   %plot window size
   number_of_mins  = 10;
   number_of_peaks = 21;
   window_size    = 15;             %in seconds
   window_size    = window_size*fs;   %in row numbers
   search_size    = 100;            %points to search

   %plot info
   labelone = [‘ES’];
   label2 = [‘RA’];
   label3 = [‘BF’];
   label4 = [‘BaElg’];
   label5 = [‘KnElg’];

   %counter
   rownumber  = 0;

%=================================================================
%      MAIN PROCESSING
%=================================================================

for s = start_with_subject☹start_with_subject + subjects_to_process – 1)

   %reset output data
   avgout = [];
   rmsout = [];

for c = (start_with_condition☹start_with_condition + conditions_to_process -1 ))
for t = (start_with_trial☹start_with_trial + trials_to_process -1 ))

   %open a file
   %EMG data have DC bias removed
   [musone, mus2, mus3, baelg, knelg, inputfile, outputfile] = lbropen2012(s,c, t,
directory, columns, headers, musonecol, mus2col, mus3col, backelgon, kneelgon);

   if window_size > length(musone)
      window_size = length(musone);
   end
```

```
    %group data
    musdata = [musone, mus2, mus3];
    elgondata = [baelg, knelg];

%==============================================================

    %general smooth routine and normalize data
    %   Returns ‘musdatanorm’ that are normalized to %MVC
    %   ‘musdata’ are still raw data (DC bias removed)
    lbrsm2012

%==============================================================

    %run subroutine to smooth data, identify cycles, extract, interpolate and create
ensemble plot
    %calls the following functions:
    %       lbr_cycles2012
    %       lbr_kincycles2012
    %       lbr_extract2012
    %       lbrmean2012
    lbrsub2012

%==============================================================

    %calcualte average and RMS EMG data
    lbremga2_2012

%==============================================================

end %end trial
    rownumber = rownumber+1;

    %for average and RMS EMG processing
    allout(rownumber, ☺ = [s c avgout]; %removed SF from output
    rmsallout(rownumber, ☺ = [s c rmsout]; %removed SF from output

    clear musdata musdatanormsm elgondata;

end %end conditions

end %end subjects

%for average and RMS EMG processing
     %header: subject condition trial averageVL averageRF averageBF averageGA
    my_save([directory ‘\output’], avg_file, allout, precision);
```

```
    %header: subject condition trial rmsVL rmsRF rmsBF rmsGA
    my_save([directory '\output'], rms_file, rmsallout, precision);

fprintf(1,'\nFinished processing.\n\n');

fclose('all');
%clear

%open a file for lbr project

function [musonedata, mus2data, mus3data, baelg, knelg, inputfile, outputfile] =
lbropen2012(s,c, t, directory, peakcol, headers, musonecol, mus2col, mus3col, backelgon,
kneelgon);

%create s?c?t? filename
subj = int2str(s);
cond = int2strI;
tri = int2str(t);

f_name = ['s' subj 'c' cond 't' tri];
fprintf(1,'\n'); fprintf(1,f_name); fprintf(1,'\n');

%create filenames
inputfile   = [f_name '.txt'];
outputfile  = [f_name '.out'];

%open a file using 'my_open' function
data = my_fopen(directory, inputfile, peakcol, inf, headers);

musonedata = data(:,musonecol);
mus2data = data(:,mus2col);
mus3data = data(:,mus3col);
baelg = data(:,backelgon);
knelg = data(:,kneelgon);

%remove DC bias
musonedata = musonedata – mean(musonedata);
mus2data = mus2data – mean(mus2data);
mus3data = mus3data – mean(mus3data);

%function:  my_fopen
%this function will run the commonly used commands to open a file.
%called as:
%   data = my_fopen(directory, filename, columns, rows, headers)
%where
```

```
%   directory   = location of file
%   filename    = name of file with extension
%   columns     = number of columns
%   rows        = number of rows
%   headers     = number of headers to get rid of

function tempdata = my_fopen(my_dir, file__name, columns, rows, headers);

   %my_dir = data directory
   %file__name = filename with extension
   %columns = number of columns
   %headers = number of headers to discard

   %set up commands for eval function
   %change to working directory
   eval([‘cd ‘ my_dir ‘;’]);

   %open the file
   %create substrings
   c = ‘fid=fopen(‘’’;
   d = ‘’’,’’rt’’);’;

   %create filename
   file_name = [c, file__name, d];

      %open peak input file
   eval(file_name);

      %check to see if the open was successful
         if fid == -1
      clc
            message = [‘The filename ‘ file__name ‘ does not exist in directory ‘ my_dir];
            error(message);
               fprintf(1,’\n\n’);
         end


   %get rid of headers
   for h = 1:headers
      fgets(fid);
   end

   %read in data
   A = fscanf(fid, ‘%f’, [columns rows]);
   tempdata = A’;
```

```
%close files
fclose('all');

%normalize data
%routine called via lbr_2012

%general smoothing routine
for I = 1:3

   [musdata(:,i)] = my_filt(musdata(:,i), fcgen, fs, 1); %smoothed with fc of 500hz
   musdatanorm_sm(:,i) = my_filt(musdata(:,i),fc,fs,1); %smoothed with fc of 300hz

end

%normalize data
%there are three muscles to process
for I = 1:3

   %create max file name
   if I == 1
      maxfilename = ['s' int2str(s) 'mvces.txt'];

   elseif I == 2

      maxfilename = ['s' int2str(s) 'mvcra.txt'];

   elseif I == 3

      maxfilename = ['s' int2str(s) 'mvcbf.txt'];

   end

   %Normalize data to peak 1-sec average
   %only do this if it is the first file processed for a subject
   if rownumber == 0
      [norm] = lbropennormalize2012(s,c, t, directory, columns, headers, muscol(i), fs,
maxfilename, avgtime);
   end

   %normalize data
   musdatanorm(:,i) = musdata(:,i)./norm*100;
   musdatanormsm(:,i) = musdatanorm_sm(:,i)./norm*100;

end
```

```
%return to lbr_2012.m

%Fourth Order Zero lag Butterworth Filter
%Function called as:
%[smooth_data] = my_filt(rawdata, fc, fs, type)
%where
%fc = cutoff frequency
%fs = sample frequency
%type = type of filter
%   1 = low pass filter
%   2 = high pass filter
%==================================================

function [smoothed_data] = my_filt(raw_data,fc, fs, type)

warning off;

%calculate wn
wn = 2*fc/fs;

%calculate butterworth coefficients (2nd order)
if type == 1
  [B,A]=butter(2,wn);
end
if type == 2
  [B,A]=butter(2,wn,'high');
end

%calculate smoothed data using a zero-phase lag routine
smoothed_data=filtfilt(B,A,raw_data);

warning on;

%open a file for LBR project
%called via lbrsm2012.m

function [MUSnorm] = lbropennormalize2012(s,c, t, directory, peakcol, headers,
MUScol, fs, maxfilename, avgtime);

close(gcf)

%create s?c?t? filename
subj = int2str(s);
cond = int2strI;
tri = int2str(t);
```

```
%------------------------------------------------------
fprintf(1,'\n'); fprintf(1,maxfilename); fprintf(1,'\n');

%create filenames
inputfile   = [maxfilename '.txt'];

%open a file using 'my_open' function
data = my_fopen(directory, maxfilename, peakcol, inf, headers);

temptime = 1:length(data);

musdata = data(:,MUScol);
musdata = musdata – mean(musdata);  %remove DC bias
musdata = abs(musdata);             %full wave rectify

for I = 1:length(musdata)-(fs*avgtime)

   tempmean(i) = mean(musdata(i:i+fs*avgtime));

end

MUSnorm = max(tempmean);

plot(temptime, musdata)
hold on
musnormplot = ones(1,length(musdata))*MUSnorm;
plot(temptime, musnormplot, 'r-')

if MUScol == 2
   title('ES MVC')

elseif MUScol == 3
   title('RA MVC')

elseif MUScol == 4
   title('BF MVC')

end

ylabel('EMG')
xlabel('Time')

pause
close(gcf)
%------------------------------------------------------
```

```
%called via lbr_2012
%smooth, normalize data, extract data
%
   %plot data
   lbrplot2012(musdatanorm_sm, elgondata, labelone, label2, label3, label4, label5,
[‘Smoothed Data’], window_size);

   %find peaks
   [baelgaverage maxbaelg minbaelg rombaelg cycles] =
lbr_cycles2012([0:1/fs☹length(knelg)-1)/fs], knelg, baelg, fs, 5, 150, inputfile,
number_of_peaks);

   %find peaks, then calculate kinematic data variables
   [knmin knminstdv knmax knmaxstdv knrom knromstdv knflexvel knflexvelstdv] =
lbr_kincycles2012([0:1/fs☹length(knelg)-1)/fs], knelg, fs, 10, 100, inputfile,
number_of_mins);

   %calculate values for back kinematics
   lbrmean2012

   %data to save
   extraoutputdata = [baelgmean’ baelgstd’ meanmaxbaelg’ stdmaxbaelg’ meanminbaelg’
stdminbaelg’ meanrombaelg’ stdrombaelg’ knmin’ knminstdv’ knmax’ knmaxstdv’
knrom’ knromstdv’ knflexvel’ knflexvelstdv’];

   %save subject data
   my_save([directory ‘\output’], extraoutput_file, extraoutputdata, precision);

   clear extraoutputdata;

   %return to lbr_2012

function lbrplot2012(musdata, elgondata, labelone, label2, label3, label4, label5, plottitle,
window_size);

   %plot raw data
   subplot(5,1,1)
      plot(musdata(1:window_size,1))
      ylabel(labelone)
      title(plottitle)
   subplot(5,1,2)
      plot(musdata(1:window_size,2))
      ylabel(label2)
   subplot(5,1,3)
      plot(musdata(1:window_size,3))
      ylabel(label3)
```

```
   subplot(5,1,4)
      plot(elgondata(1:window_size,1))
      ylabel(label4)
   subplot(5,1,5)
      plot(elgondata(1:window_size,2))
      ylabel(label5)

   pause
   close(gcf)

%This routine is written to identify a number of peaks in a data set based upon
%selecting the first 4 peaks.
%time = time column
%data = data column
%fs = sample rate
%plotsec = the number of seconds to plot to identify the first 4 peaks
%searchwindow = number of points to search around
%filename = name of file being processed
%numberofpeaks = number of peaks to pull from the data set

function [baelgaverage maxbaelg minbaelg rombaelg cycles] = lbr_cycles2012(time,
data, moredata, fs, plotsec, searchwindow, filename, numberofpeaks)

         %------------------------------------
         %      Identify max positions
         %------------------------------------

         close(gcf)

         %plot first few seconds of position data
         plot(time(1:plotsec*fs), data(1:plotsec*fs))
         xlabel('time (s)')
         ylabel('knee elgon (units)')
         temptitle = [filename];
         title(temptitle)
         hold on

         %identify four peaks
         fprintf(1,'\nClick on four peaks.')

         for I = 1:4

            [p(i), ppos(i)] = lbrfindmin(data, searchwindow, fs);
            plot(time(ppos(i)), data(ppos(i)), 'ro')
            plot(time(ppos(i)), data(ppos(i)), 'r.')
```

```
end

%calculate interval using different combinations of peaks
tempsf(1) = (ppos(4)-ppos(1))/3;
tempsf(2) = (ppos(3)-ppos(1))/2;
tempsf(3) = (ppos(2)-ppos(1))/1;
tempsf(4) = (ppos(4)-ppos(2))/2;
tempsf(5) = (ppos(4)-ppos(3))/1;
tempsf(6) = (ppos(3)-ppos(2))/1;

%average SF
sf = mean(tempsf);

%use this to predict future peaks knowing where the first one occurs
peak(1) = data(ppos(1));
interval = sf;

close(gcf)

%plot all data points after determining position
plot(time(1:plotsec*fs*2), data(1:plotsec*fs*2))
ylabel('data (units)')
xlabel('time (s)')
temptitle = [filename];
title(temptitle)
hold on

%plot first point
plot(time(ppos(1)), data(ppos(1)), 'ro')
plot(time(ppos(1)), data(ppos(1)), 'r.')

for I = 2:numberofpeaks

   %find min
   startsearch = round(ppos(i-1) + 0.5*interval);
   endsearch   = round(startsearch + interval);

   [temp,    empos] = min(data(startsearch:endsearch));

   %adjust position
   ppos(i) =    empos + startsearch – 1;

   %plot
   plot(time(ppos(i)), data(ppos(i)), 'ro')
   plot(time(ppos(i)), data(ppos(i)), 'r.')
```

```
        end

        hold off

        %return positions of peaks
        cycles = ppos;

        %calculate values for the back elgon
        for I = 1:numberofpeaks-1
           start = cycles(i);
           finish = cycles(i+1);
           baelgaverage(i) = mean(moredata(start:finish)); %mean for each individual
           cycle
           maxbaelg(i) = max(moredata(start:finish)); %max for each cycle
           minbaelg(i) = min(moredata(start:finish)); %min for each cycle
           rombaelg(i) = (maxbaelg(i) – minbaelg(i)); %rom for each cycle
        end

        pause

%This file was written in order to calculate elgon values for Jeff's thesis including:
%knee flexion velocity
%range of motion of the knee
%maximum and minimum values for knee flexion/extension

%called from lbrsub2012

function [knmin knminstdv knmax knmaxstdv knrom knromstdv knflexvel knflexvelstdv]
= lbr_kincycles2012(time, data, fs, plotsec, searchwindow, filename, numberofmins)


        %--------------------------------------
        %      Identify max positions
        %--------------------------------------

        close(gcf)

        %plot first few seconds of position data
        figure('position',[30 50 1200 800]);
        plot(time(1:plotsec*fs), data(1:plotsec*fs))
        xlabel('time (s)')
        ylabel('knee elgon (units)')
        temptitle = [filename];
        title(temptitle)
        hold on
```

```
%identify locations for peak knee extension and for peak knee flexion
fprintf(1,'\nClick on peak extension for ten knee peaks.\n')

for I = 1:numberofmins

   %identify ten locations for peak knee extension
   [p(i), ppos(i)] = lbrfindmin(data, searchwindow, fs);
   plot(time(ppos(i)), data(ppos(i)), 'ro')
   plot(time(ppos(i)), data(ppos(i)), 'r.')
   extmax(i) = data(ppos(i));
end

fprintf(1,'\nClick on peak flexion for ten knee peaks.')

for I = 1:numberofmins
   %identify ten locations for peak knee flexion
   [p(i), pos(i)] = lbrfindpeak(data, searchwindow, fs);
   plot(time(pos(i)), data(pos(i)), 'bo')
   plot(time(pos(i)), data(pos(i)), 'r.')
   flexmax(i) = data(pos(i));
end

hold off
close(gcf)

%calculate average values for knee extension and flexion
knmin = mean(extmax);
knminstdv = std(extmax);
knmax = mean(flexmax);
knmaxstdv = std(flexmax);

for I = 1:numberofmins
   %find range of motion
   ROM(i) = (flexmax(i) – extmax(i));
end

knrom = mean(ROM);
knromstdv = std(ROM);

%find knee flexion velocity
for I = 1:numberofmins
   flextime(i) = ((pos(i)-ppos(i))*(1/fs));
   flexvel(i) = (ROM(i)/flextime(i));
end

knflexvel = mean(flexvel);
```

```
        knflexvelstdv = std(flexvel);

        pause

%Function lbrfindmin.m
%Locates minimum value and position relative to data size
%Important: The function requires that the x axis is time.
%The peakpos returned is position number (not time).
%The function includes a call to ginput for one click.

Function [peak, peakpos] = lbrfindmin(data, searchwindow, fs);

[xpos, ypos] = ginput(1);
xpos = round(xpos*fs);

start = xpos-searchwindow;
if (start<1)
  start=1;
end

peak = min(data(start:xpos+searchwindow));

temppeakpos = find(data(start:xpos+searchwindow)==peak)

temppeakpos(5)=0;

peakpos = temppeakpos(1);

peakpos = peakpos+(start)-1;

%Function lbrfindpeak.m
%Locates peak value and position relative to data size
%Important: The function requires that the x axis is time.
%The peakpos returned is position number (not time).
%The function includes a call to ginput for one click.

Function [peak, peakpos] = findpeak(data, searchwindow, fs);

[xpos, ypos] = ginput(1);
xpos = round(xpos*fs);

start = xpos-searchwindow;
if (start<1)
  start=1;
end
```

```
peak = max(data(start:xpos+searchwindow));

temppeakpos = find(data(start:xpos+searchwindow)==peak)

temppeakpos(5)=0;

peakpos = temppeakpos(1);

peakpos = peakpos+(start)-1;

%Calculate back kinematics for LBR
%called from lbrsub2012

%for I = 1:2:number_of_peaks-1    %extracting only odd numbered cycles
   baelgmean    = mean(baelgaverage(1:2:number_of_peaks-1));
   baelgstd     = std(baelgaverage(1:2:number_of_peaks-1));

   meanmaxbaelg = mean(maxbaelg(1:2:number_of_peaks-1));
   stdmaxbaelg  = std(maxbaelg(1:2:number_of_peaks-1));

   meanminbaelg = mean(minbaelg(1:2:number_of_peaks-1));
   stdminbaelg  = std(minbaelg(1:2:number_of_peaks-1));

   meanrombaelg = mean(rombaelg(1:2:number_of_peaks-1));
   stdrombaelg  = std(rombaelg(1:2:number_of_peaks-1));
%end
    close(gcf)

%Function: my_save(directory, filename, data, precision)
%This function will save data to a specified file with a specified precision

function my_save(directory, filename, data, precision)

   %initialize variable
   all_column_info = [];

   %change directory
      temp = pwd;
      eval(['cd ' directory]);

   %open the file to write to
      fid=fopen(filename, 'w');

   %make quote notation
      q='''';
```

```
%check the size of the data array
   [rows columns] = size(data);

%Create the necessary write commands

   column_precision = int2str(precision);
   column_info = ['%5.' Column_precision 'f'];

   for I = 1:columns
      all_column_info = [column_info ' ' all_column_info];
   end

%transpose the output data array because the print command writes
%column 1, then column 2, ...
   data=data';

%create command line
   print_command = ['fprintf(fid,' q all_column_info '\n' q ', data);'];

%save data
   eval([print_command]);

%close file
   fclose(fid);

%change back to original directory
   eval(['cd ' temp]);
```

```
%calculate means and RMS for time period
%called via lbr_2012
%DC bias already removed

   %check to make sure data set is long enough
   if EMGwin*fs > length(musdatanormsm)
      winend      = length(musdatanormsm);
   else
      winend      = EMGwin*fs;
   end

   %calculate average across each cycle in the data (e.g. during stance only)
   for o = 1:number_of_peaks-1
      start(o) = cycles(o);
      finish(o) = cycles(o+1);

      for I = 1:3
         tempavgmus(o,i)   = mean(abs(musdatanormsm((start(o):finish(o)),i)));
```

```
        temprmsmus(o,i)   = rms(abs(musdatanormsm((start(o):finish(o)),i)));
    end
end

for I = 1:3
    avgmus(i) = mean(tempavgmus(1:2:number_of_peaks-1,i));
    rmsmus(i) = mean(temprmsmus(1:2:number_of_peaks-1,i));
end

avgout          = [avgout avgmus];
rmsout          = [rmsout rmsmus];

%return to lbr_2012
```

Appendix III

Data Collection Sheet

Footwear – Low Back data collection sheet

Informed Consent?
Warm up?
Shoes fit?
Prior history of low back pain?
Data sampling rate at 1500hz?

Date:____________________

Subject:__________________ Gender:___________________

DOB:_____________________

Start Condition:___________

Shoe Size:
With Insert:__________

Without Insert:___________

Leg Length (cm): Right:______________ Left:______________

Height (cm):_________________

Mass (kg):_________________

Preferred Speed (mph):________________

Weekly Running Mileage:______________

Running Frequency (times/week):_____________

History of minimalist/barefoot running and running experience:

Notes:
C1 – without insert (note footstrike pattern) –

C2 – with insert –

Appendix IV

Individual Subject Data

Subject	1
Birthday	6/2/1993
Age	18
Gender	Female
Height (cm)	152.9
Mass (kg)	44.7
R Leg Length (cm)	84.5
L Leg Length (cm)	84
Preferred Speed (m/s)	2.73
Weekly Mileage (km)	80.5
Running Frequency (days/wk)	6
Running Experience (yrs)	2
Barefoot/Minimalist Running Experience	barefoot strides on grass, running in track flats
Footstrike C1	rearfoot
Footstrike C2	rearfoot

	C1		C2	
	Mean	Std	Mean	Std
ES EMG (%)	9.4232	1.7318	8.8213	1.5326
RA EMG (%)	8.9916	1.4804	7.8572	1.4995
BF EMG (%)	29.4802	9.4641	23.7301	6.1609
Average Back Position (deg)	28.3093	1.647	30.0119	1.5738
Back ROM (deg)	13.1263	1.0995	12.8349	1.0025
Max Knee Extension (deg)	1.3724	1.6104	2.4268	1.3582
Max Knee Flexion (deg)	34.293	1.085	36.5863	1.1699
Knee ROM (deg)	32.9206	1.9046	34.1596	1.9221

Subject	2
Birthday	9/21/1993
Age	18
Gender	Male
Height (cm)	181.3
Mass (kg)	66.7
R Leg Length (cm)	102.1
L Leg Length (cm)	102.5
Preferred Speed (m/s)	4.20
Weekly Mileage (km)	64.4
Running Frequency (days/wk)	7
Running Experience (yrs)	6
Barefoot/Minimalist Running Experience	barefoot strides on grass, running in Nike Free and track flats
Footstrike C1	midfoot
Footstrike C2	rearfoot

	C1		C2	
	Mean	Std	Mean	Std
ES EMG (%)	6.4211	2.3307	7.7729	2.2157
RA EMG (%)	7.0325	1.8744	9.0595	5.866
BF EMG (%)	23.1848	2.3341	25.7794	3.1235
Average Back Position (deg)	31.7017	1.836	31.346	1.1226
Back ROM (deg)	17.7868	2.0626	15.9882	1.1039
Max Knee Extension (deg)	4.6146	2.6961	6.461	2.0584
Max Knee Flexion (deg)	41.2657	0.8764	41.5987	1.8233
Knee ROM (deg)	36.6511	2.8956	35.1377	3.061

Subject	3
Birthday	2/18/1970
Age	42
Gender	Female
Height (cm)	170.1
Mass (kg)	57.6
R Leg Length (cm)	96
L Leg Length (cm)	96
Preferred Speed (m/s)	2.86
Weekly Mileage (km)	53.1
Running Frequency (days/wk)	6
Running Experience (yrs)	29
Barefoot/Minimalist Running Experience	beach running, barefoot strides on grass, running in track flats
Footstrike C1	forefoot
Footstrike C2	forefoot

	C1		C2	
	Mean	Std	Mean	Std
ES EMG (%)	12.0231	4.2901	13.9469	2.3022
RA EMG (%)	9.1104	1.0119	7.9535	0.9444
BF EMG (%)	34.9346	11.0578	50.2094	10.5987
Average Back Position (deg)	30.0644	1.2566	31.4074	0.9816
Back ROM (deg)	6.7393	1.7569	9.2685	1.559
Max Knee Extension (deg)	-0.8951	1.1012	1.4981	1.26
Max Knee Flexion (deg)	28.8292	1.8615	32.3696	1.2248
Knee ROM (deg)	29.7242	2.5497	30.8715	1.5389

Subject	4
Birthday	9/14/1991
Age	20
Gender	Female
Height (cm)	155.3
Mass (kg)	48.2
R Leg Length (cm)	86
L Leg Length (cm)	87.5
Preferred Speed (m/s)	1.61
Weekly Mileage (km)	48.3
Running Frequency (days/wk)	6
Running Experience (yrs)	4
Barefoot/Minimalist Running Experience	ran barefoot as a child, but not for the past 10 years
Footstrike C1	midfoot
Footstrike C2	midfoot

	C1		C2	
	Mean	Std	Mean	Std
ES EMG (%)	13.6416	1.6173	13.7239	3.1748
RA EMG (%)	14.893	2.3112	12.7396	1.0543
BF EMG (%)	48.9066	8.2211	56.4351	11.3226
Average Back Position (deg)	35.2915	2.3545	38.8859	0.9837
Back ROM (deg)	10.97	1.21	11.2526	1.2426
Max Knee Extension (deg)	7.9022	0.9355	6.8648	1.4862
Max Knee Flexion (deg)	22.4916	1.6949	26.8035	0.9736
Knee ROM (deg)	14.5894	1.9388	19.9387	1.6412

Subject	5
Birthday	10/17/1972
Age	39
Gender	female
Height (cm)	168.9
Mass (kg)	67.3
R Leg Length (cm)	98
L Leg Length (cm)	98.3
Preferred Speed (m/s)	2.9057665
Weekly Mileage (km)	40.233675
Running Frequency (days/wk)	4
Running Experience (yrs)	6
Barefoot/Minimalist Running Experience	5 months wearing lightweight trainers
Footstrike C1	rearfoot
Footstrike C2	rearfoot

	C1		C2	
	Mean	Std	Mean	Std
ES EMG (%)	14.0613	1.2394	15.8021	2.5445
RA EMG (%)	15.3503	2.3169	12.9451	0.8281
BF EMG (%)	62.1023	8.9271	64.0886	12.4781
Average Back Position (deg)	30.5863	0.761	30.8994	1.2016
Back ROM (deg)	14.2522	1.2027	16.1133	1.0024
Max Knee Extension (deg)	-4.7711	0.9242	-6.5475	0.5791
Max Knee Flexion (deg)	30.3486	1.1466	29.3904	0.9473
Knee ROM (deg)	35.1197	1.5564	35.9379	1.099

Subject	6
Birthday	8/5/1963
Age	48
Gender	Male
Height (cm)	180.1
Mass (kg)	82.7
R Leg Length (cm)	103.3
L Leg Length (cm)	102.9
Preferred Speed (m/s)	3.00
Weekly Mileage (km)	16.1
Running Frequency (days/wk)	3
Running Experience (yrs)	8
Barefoot/Minimalist Running Experience	none. Retrained gait from rearfoot to forefoot strike
Footstrike C1	forefoot
Footstrike C2	forefoot

	C1		C2	
	Mean	Std	Mean	Std
ES EMG (%)	29.8009	5.24	24.284	4.2501
RA EMG (%)	14.5326	2.0168	14.1179	2.7905
BF EMG (%)	161.7067	30.4779	149.2973	23.2651
Average Back Position (deg)	21.6061	1.0718	20.452	0.676
Back ROM (deg)	6.4533	1.4618	5.9563	0.9385
Max Knee Extension (deg)	4.7253	2.0691	7.0899	1.0002
Max Knee Flexion (deg)	24.1563	2.079	28.0717	1.351
Knee ROM (deg)	19.431	2.1923	20.9818	1.3814

Subject	7
Birthday	10/2/1982
Age	29
Gender	Female
Height (cm)	172.6
Mass (kg)	63.2
R Leg Length (cm)	97.5
L Leg Length (cm)	97.8
Preferred Speed (m/s)	2.59
Weekly Mileage (km)	24.1
Running Frequency (days/wk)	3
Running Experience (yrs)	7
Barefoot/Minimalist Running Experience	cooldown barefoot on grass after track workouts
Footstrike C1	forefoot
Footstrike C2	rearfoot

	C1		C2	
	Mean	Std	Mean	Std
ES EMG (%)	8.828	1.8274	9.6184	1.6094
RA EMG (%)	5.8908	0.6773	7.3987	0.8574
BF EMG (%)	64.4092	11.9851	82.8527	15.5249
Average Back Position (deg)	39.1539	1.3356	38.2226	0.6284
Back ROM (deg)	8.1222	1.4564	10.7387	1.5586
Max Knee Extension (deg)	-1.4385	1.4864	2.0813	1.0937
Max Knee Flexion (deg)	30.9791	1.2695	32.7796	1.3555
Knee ROM (deg)	32.4176	2.1457	30.6983	1.4074

Subject	8
Birthday	9/5/1968
Age	43
Gender	Male
Height (cm)	175.4
Mass (kg)	79.5
R Leg Length (cm)	95.8
L Leg Length (cm)	96.7
Preferred Speed (m/s)	2.41
Weekly Mileage (km)	64.4
Running Frequency (days/wk)	5
Running Experience (yrs)	3
Barefoot/Minimalist Running Experience	none
Footstrike C1	rearfoot
Footstrike C2	rearfoot

	C1		C2	
	Mean	Std	Mean	Std
ES EMG (%)	17.7597	1.9201	18.5123	3.0189
RA EMG (%)	7.2227	0.6191	5.5667	1.1873
BF EMG (%)	25.1682	4.3732	26.0456	3.0273
Average Back Position (deg)	20.7416	0.5093	19.3241	0.4737
Back ROM (deg)	8.2484	0.8487	11.3996	0.8835
Max Knee Extension (deg)	-3.2542	0.5403	-3.0805	0.7599
Max Knee Flexion (deg)	25.551	0.8919	26.8856	0.6439
Knee ROM (deg)	28.8052	0.9811	29.9661	0.7693

Subject	9
Birthday	3/2/1962
Age	50
Gender	Male
Height (cm)	168.5
Mass (kg)	80.8
R Leg Length (cm)	95.5
L Leg Length (cm)	95.6
Preferred Speed (m/s)	1.70
Weekly Mileage (km)	40.2
Running Frequency (days/wk)	4
Running Experience (yrs)	10
Barefoot/Minimalist Running Experience	barefoot strides on grass
Footstrike C1	forefoot
Footstrike C2	rearfoot

	C1		C2	
	Mean	Std	Mean	Std
ES EMG (%)	19.9874	3.3369	20.4214	1.8515
RA EMG (%)	6.4062	1.297	7.9192	0.6633
BF EMG (%)	6.7458	0.8289	7.1782	1.169
Average Back Position (deg)	3.5084	0.9896	3.8931	0.8279
Back ROM (deg)	6.6864	0.6352	7.11	0.8049
Max Knee Extension (deg)	18.7139	1.1566	18.212	0.6519
Max Knee Flexion (deg)	32.3735	0.5178	32.8647	0.8377
Knee ROM (deg)	13.6596	1.1635	14.6527	0.9893

Subject	10
Birthday	12/1/1991
Age	20
Gender	Female
Height (cm)	159.7
Mass (kg)	54.7
R Leg Length (cm)	91
L Leg Length (cm)	90.2
Preferred Speed (m/s)	2.28
Weekly Mileage (km)	12.9
Running Frequency (days/wk)	3
Running Experience (yrs)	3
Barefoot/Minimalist Running Experience	none
Footstrike C1	rearfoot
Footstrike C2	rearfoot

	C1		C2	
	Mean	Std	Mean	Std
ES EMG (%)	21.8021	5.192	20.963	7.0344
RA EMG (%)	7.8834	1.4187	7.6421	2.0894
BF EMG (%)	28.0797	2.5677	32.5114	3.5722
Average Back Position (deg)	26.4161	1.7584	25.6778	1.3386
Back ROM (deg)	9.4142	1.2373	8.9272	1.0299
Max Knee Extension (deg)	-0.1901	2.8808	1.0852	1.5035
Max Knee Flexion (deg)	32.5806	1.5242	34.5311	1.8887
Knee ROM (deg)	32.7707	3.8281	33.446	2.7313

Appendix V

SPSS Output

Running Trials

Paired Samples Statistics

		Mean	N	Std. Deviation	Std. Error Mean
Pair 1	ESEMG	15.3748	10	7.07735	2.23806
	ESEMG2	15.3866	10	5.62468	1.77868
Pair 2	RAEMG	9.7314	10	3.72676	1.17850
	RAEMG2	9.3200	10	2.87684	.90974
Pair 3	BFEMG	35.8902	9	19.03492	6.34497
	BFEMG2	40.9812	9	23.91993	7.97331
Pair 4	BackPosture	26.7379	10	9.90123	3.13104
	BackPostrure2	27.0120	10	10.36705	3.27835
Pair 5	BackROM	10.1799	10	3.80402	1.20294
	BackROM2	10.9589	10	3.37532	1.06737
Pair 6	MaxKneeExt	-2.6779	10	6.85328	2.16720
	MaxKneeExt2	-3.6091	10	6.72110	2.12540
Pair 7	MaxKneeFlex	-30.2869	10	5.47302	1.73072
	MaxKneeFlex2	-32.1881	10	4.66805	1.47617
Pair 8	KneeROM	27.6089	10	8.51815	2.69367
	KneeROM2	28.5790	10	7.37854	2.33330

Running Trials

Paired Samples Test

		Paired Differences					t	df	Sig. (2-tailed)
		Mean	Std. Deviation	Std. Error Mean	95% Confidence Interval of the Difference Lower	Upper			
Pair 1	ESEMG - ESEMG2	-.01178	2.14596	.67861	-1.54691	1.52335	-.017	9	.987
Pair 2	RAEMG - RAEMG2	.41140	1.59858	.50552	-.73215	1.55495	.814	9	.437
Pair 3	BFEMG - BFEMG2	-5.09101	7.59303	2.53101	-10.92753	.74551	-2.011	8	.079
Pair 4	BackPosture - BackPostrure2	-.27409	1.56337	.49438	-1.39246	.84428	-.554	9	.593
Pair 5	BackROM - BackROM2	-.77902	1.65433	.52315	-1.96246	.40442	-1.489	9	.171
Pair 6	MaxKneeExt - MaxKneeExt2	.93117	1.68978	.53435	-.27762	2.13996	1.743	9	.115
Pair 7	MaxKneeFlex - MaxKneeFlex2	1.90126	1.68873	.53402	.69321	3.10931	3.560	9	.006
Pair 8	KneeROM - KneeROM2	-.97012	1.91966	.60705	-2.34336	.40312	-1.598	9	.144

Impact Testing

Group Statistics

insolestatus		Impactlocation	N	Mean	Std. Deviation	Std. Error Mean
without insole	acceleration	Heel	8	23.8750	.05555	.01964
		Forefoot	8	23.8575	.07005	.02477
	pressure	Heel	8	1251.2375	2.94072	1.03970
		Forefoot	8	1250.3563	3.62011	1.27990
with insole	acceleration	Heel	8	11.5112	.41512	.14677
		Forefoot	8	20.2212	.54012	.19096
	pressure	Heel	8	603.3225	21.81241	7.71185
		Forefoot	8	1059.7800	28.33133	10.01664

Independent Samples Test

			Levene's Test for Equality of Variances		t-test for Equality of Means						
										99% Confidence Interval of the Difference	
insolestatus			F	Sig.	t	df	Sig. (2-tailed)	Mean Diff.	Std. Error Diff.	Lower	Upper
without insole	acceleration	Equal variances assumed	1.454	.248	.554	14	.589	.018	.032	-.077	.112
		Equal variances not assumed			.554	13.309	.589	.018	.032	-.077	.112
	pressure	Equal variances assumed	1.336	.267	.534	14	.601	.881	1.649	-4.028	5.790
		Equal variances not assumed			.534	13.436	.602	.881	1.649	-4.059	5.822
with insole	acceleration	Equal variances assumed	.428	.524	-36.164	14	.000	-8.710	.241	-9.427	-7.993
		Equal variances not assumed			-36.164	13.131	.000	-8.710	.241	-9.434	-7.986
	pressure	Equal variances assumed	.423	.526	-36.108	14	.000	-456.458	12.641	-494.089	-418.826
		Equal variances not assumed			-36.108	13.141	.000	-456.458	12.641	-494.469	-418.446

Independent Samples Test

insolestatus			Levene's Test for Equality of Variances F	Sig.	t-test for Equality of Means t	df	Sig. (2-tailed)	Mean Diff.	Std. Error Diff.	99% Confidence Interval of the Difference Lower	Upper
without insole	acceleration	Equal variances assumed	1.454	.248	.554	14	.589	.018	.032	-.077	.112
		Equal variances not assumed			.554	13.309	.589	.018	.032	-.077	.112
	pressure	Equal variances assumed	1.336	.267	.534	14	.601	.881	1.649	-4.028	5.790
		Equal variances not assumed			.534	13.436	.602	.881	1.649	-4.059	5.822
	acceleration	Equal variances assumed	.428	.524	-36.164	14	.000	-8.710	.241	-9.427	-7.993
		Equal variances not assumed			-36.164	13.131	.000	-8.710	.241	-9.434	-7.986
		Equal variances assumed	.423	.526	-36.108	14	.000	-456.458	12.641	-494.089	-418.826

Impact Testing

Group Statistics

impactlocation		Insolestatus	N	Mean	Std. Deviation	Std. Error Mean
heel	acceleration	without insole	8	23.8750	.05555	.01964
		with insole	8	11.5112	.41512	.14677
	pressure	without insole	8	1251.2375	2.94072	1.03970
		with insole	8	603.3225	21.81241	7.71185
forefoot	acceleration	without insole	8	23.8575	.07005	.02477
		with insole	8	20.2212	.54012	.19096
	pressure	without insole	8	1250.3563	3.62011	1.27990
		with insole	8	1059.7800	28.33133	10.01664

Independent Samples Test

impactlocation			Levene's Test for Equality of Variances F	Sig.	t-test for Equality of Means t	df	Sig. (2-tailed)	Mean Diff.	Std. Error Diff.	99% Confidence Interval of the Difference Lower	Upper
heel	acceleration	Equal variances assumed	14.826	.002	83.496	14	.000	12.364	.148	11.923	12.805
		Equal variances not assumed			83.496	7.251	.000	12.364	.148	11.852	12.876
	pressure	Equal variances assumed	14.842	.002	83.262	14	.000	647.915	7.782	624.750	671.080
		Equal variances not assumed			83.262	7.254	.000	647.915	7.782	621.006	674.824
forefoot	acceleration	Equal variances assumed	11.233	.005	18.884	14	.000	3.636	.193	3.063	4.209
		Equal variances not assumed			18.884	7.235	.000	3.636	.193	2.970	4.303
	pressure	Equal variances assumed	11.322	.005	18.873	14	.000	190.576	10.098	160.516	220.637
		Equal variances not assumed			18.873	7.229	.000	190.576	10.098	155.616	225.537

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Vitae

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Grants Received

McClellan, J.R., Dufek, J.S. The Effect of Footwear on Lower Back Mechanics During Treadmill Running. Altra Footwear, equipment grant $800, 2011.

McClellan, J.R. Dual-Task Performance Among Children with Cerebral Palsy Following a Novel Physical Therapy Intervention. Graduate and Professional Student Association, UNLV, travel grant $100, 2011.

Abstracts and Presentations

Hickman, R., Dufek, J.S., Blahovec, A., Kuiken, A., **McClellan, J.R.**, Mears, J., Riggins, H. Feasibility and effects on gait of large amplitude movement training in children with CP. American Physical Therapy Association Annual Conference, Tampa, FL, June 2012.

McClellan, J.R., Dufek, J.S., Hickman, R. Effectiveness of a Novel Therapeutic Technique on Improving Gait Characteristics among Children with Cerebral Palsy. American College of Sports Medicine Annual Meeting, San Francisco, CA, May 2012.

Agnelli, C., **McClellan, J.R.**, Tarno, J., Nielson, J., Mercer, J.A. Insight to muscle activity during the lacrosse shot. American College of Sports Medicine Annual Meeting, San Francisco, CA, May 2012.

McClellan, J.R., Dufek, J.S. Are Shock Characteristics at Impact Influenced by Midsole Hardness During Running? Rocky Mountain Regional American Society of Biomechanics Meeting, Boise, ID, April 2012.

McClellan, J.R., Dufek, J.S., Hickman, R., Blahovec, A., Kuiken, A., Mears, J., Riggins, H. Dual-Task Performance Among Children with Cerebral Palsy Following a Novel Physical Therapy Intervention. Southwest Regional American College of Sports Medicine Annual Meeting, Reno, NV, October 2011.

Agnelli, C., **McClellan, J.R.**, Tarno, J., Nielsen, J., Mercer, J.A. Preliminary inspection of muscle activity during a lacrosse shot. Southwest Regional American College of Sports Medicine Annual Meeting, Reno, NV, October 2011.

McClellan, J.R., Dufek, J.S. The Effect of Shoe Choice on Variability of Walking. Intermountain Graduate Research Symposium, Logan, UT, March 2011.

McClellan, J.R., Forrest, D.M., Neumann, E.S. and Dufek, J.S. Effects of a custom orthopedic brace on balance characteristics for Charcot-Marie-Tooth patients. Southwest Regional American College of Sports Medicine Annual Meeting, San Diego, CA, October 2010.

Professional Organizations

American College of Sports Medicine, 2011-2012
American Society of Biomechanics, 2011-2012
Southwest Chapter of ACSM, 2011-2012
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