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Effects of Reduced Muscular Strength on Running Kinematics Relating to the Planus Foot

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A thesis submitted in partial fulfillment of the requirements for the Master of Science degree in Health and Rehabilitation Sciences

EFFECTS OF REDUCED MUSCULAR STRENGTH ON RUNNING KINEMATICS RELATING TO THE PLANUS FOOT

(Thesis format: Monograph)

by

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Abstract

The potential for injury is present in all forms of physical activity, particularly events involving repetitive motion. In attempt to identify kinematic changes following running induced reductions in muscular strength, 15 participants completed recorded treadmill runs before and after an outdoor run. Kinematics were recorded using a Vicon motion capture system and processed using 3D GAIT custom software. Individual factorial ANOVAs using side (dominant and non-dominant) and time (pre-run and post-run) as independent variables to assess ankle eversion and knee adduction revealed statistically significant effects for side suggesting that movements of the dominant side differ from the non-dominant. Using Pearson's product-moment correlation coefficients to compare successive strides, post-run reductions in stride-to-stride correlation were obtained for dominant and non-dominant knee adduction. Further, MANOVA analysis using standard deviation values suggested that dominant and non-dominant ankle movement variability may change following running induced reductions in muscular strength, primarily during the initial third of stance.

Keywords

Running, Pes Planus, Kinematics, Movement Variability

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Chapter 1: Introduction and Literature Review

1.1. Introduction

Physical fitness and exercise are quickly becoming a mainstay in the modern lifestyle of healthy living. One of the most popular and commonly understood forms of physical fitness is running (Dierks, Davis, & Hamill, 2010; Koblbauer, van Schooten, Verhagen, & van Dieen, 2013). It is well documented that injury has been linked to all levels of runners, ranging from beginner to the elite or Olympic level competitor (van Gent et al., 2007). Various epidemiological studies have cited that anywhere between 27% and 70% of competitive and recreational runners will experience at least one overuse injury in any one-year period (Ferber, Hreljac, & Kendall, 2009; Jacobs & Berson, 1986; Koplan, Powell, Sikes, Shirley, & Campbell, 1982; Lysholm & Wiklander, 1987; Marti, Vader, Minder, & Abelin, 1988; Walter, Hart, McIntosh, & Sutton, 1989). A review of literature performed by Johnson, Taunton, Lloyd-Smith, and McKenzie (2003) focused on running injury prevention, identified that "running injuries result from any combination of intrinsic and extrinsic factors that exceed a runner's capacity to withstand injury" (Johnston, Taunton, Lloyd-Smith, & McKenzie, 2003). Injury sources were attributed to improper training programs, poor footwear choices, inappropriate training surfaces, mal-alignment of the legs, muscular weakness and insufficient flexibility (Johnston et al., 2003).

1.2. Kinematic Effects of Fatigue

With a growing interest, and an increasing prevalence of injury in running, comes an increased demand for relevant research to understand what happens to the human body while it is engaging specifically in this activity (van Gent et al., 2007). Muscular fatigue is inherent to all forms of physical activity and has consistently been implicated as a mechanism of injury due to its potential to alter kinematics and kinetics in the lower extremity (Clansey, Hanlon, Wallace, & Lake, 2012; Cortes, Greska, Kollock, Ambegaonkar, & Onate, 2013; Dierks et al., 2010). Particularly, fatigue has been linked to reduced muscular strength, reduced neuromuscular control and increased peak rearfoot eversion (Clansey et al., 2012; Cortes et al., 2013; Dierks et al., 2010; Gerdle, Elert, &

Henriksson-Larsen, 1989; Iridiastadi & Nussbaum, 2006; Koblbauer et al., 2013).

One important study, performed by Clansey, Hanlon, Wallace, and Lake (2012) investigated the effects of fatigue in connection to impact-related injuries. Looking at 21 highly trained rearfoot striking male distance runners, the researchers analyzed data recorded using an accelerometer attached to the participant's left and right tibiae, located 0.10 meters above the ankle joint center, in addition to an accelerometer attached to the participant's forehead. Each participant was also out-fitted with a full-body six-degrees-of-freedom retroreflective marker set (Collins, Ghoussayni, Ewins, & Kent, 2009) in conjunction with a 12-camera motion capture system. Kinetic information was recorded using a force platform sampling at a rate of 1000Hz. Markers were tracked and labeled using the Qualisys Track Manger and processed in Visual 3D during each of the three gait analysis (Clansey et al., 2012).

Participants completed a pre-, mid- and post-gait analysis before, between and after two 20-minute fatigue inducing treadmill runs (Clansey et al., 2012). Each gait analyses consisted of six acceptable over-ground trials along a 15-meter runway at a speed of 4.5 meters per second (+/- 5%). The treadmill runs (1% gradient) were completed at each participant's lactate threshold, which was determined prior to testing.

Comparisons of post-run gait analysis to pre- and mid-run gait analyses focused on how fatigue affects the ability to cushion impact loading rates. Clansey et al. (2012) found that the average results of post-run testing showed statistically significant increases in peak rearfoot eversion, peak axial head acceleration, peak free moment, average vertical force loading rates and peak vertical loading rates in comparison to pre- and mid-run testing ($p < 0.05$) (Clansey et al., 2012).

The researchers claim that reduced ability to cushion impact stems from impairment of musculoskeletal functioning in controlling the motions of the lower limb (Clansey et al., 2012). Conclusions state that the measures used in their study could be used to identify people at risk of injury due to lower limb impact loading while running.

A recent study performed by Koblbauer, van Schooten, Verhagen, and van Dieen (2013) focused on the effects of fatigue in novice runners in relation to core endurance. Using 17 novice runners (10 male and 7 female), Koblbauer et al. (2013) assessed kinematic changes following a steady state running induced fatigue protocol. It was hypothesized that increases in rearfoot eversion and trunk flexion would be evident with the onset of fatigue. Additionally, it was anticipated that changes in kinematics would be more pronounced in participants with reduced core endurance (Koblbauer et al., 2013).

Changes from pre- to post-run were analyzed using Student's *T*-test at a significance level of $\alpha = 0.05$. The following 12 variables were examined: trunk flexion, extension, lateral flexion, rotation, hip flexion and extension, knee flexion and extension, and ankle inversion, eversion, dorsiflexion and plantar flexion were compared for dominant and non-dominant lower limbs. Statistically significant changes from pre- to post-run included increased peak trunk flexion and decreased trunk extension, as well as a 1.6° increased non-dominant rearfoot eversion ($p < 0.05$). Dominant rearfoot eversion trended toward significance but remained insignificant (1.0°, $p > 0.05$) (Koblbauer et al., 2013). Although the obtained statistically significant kinematic changes were subtle, Koblbauer et al. concluded that the cumulative effects of these changes may compound and potentially contribute to running injury (Koblbauer et al., 2013). Moreover, Koblbauer et al. (2013) state that increased ankle eversion when fatigued may benefit from increased pronation support and that the effects of fatigue may be an important determinant when selecting appropriate running shoes.

In a study performed by Mizrahi, Verbitsky, and Isakov (2000), the researchers identified shank loading imbalances resulting from fatigue. The sample consisted of 14 healthy male participants. Each participant completed a 30-minute run test at a speed 5% above their individual anaerobic threshold, resulting in muscular fatigue. Using Electromyography (EMG), myoelectric activity of both the gastrocnemius and the tibialis anterior were monitored throughout the run. Additionally, end-tidal carbon dioxide pressure was recorded to measure global or metabolic fatigue based on the development of metabolic acidosis (Mizrahi, Verbitsky, & Isakov, 2000).

Using accelerometers attached to the right leg of each participant, just above the tibial tuberosity, Mizrahi et al. (2000) found increased impact acceleration with induced fatigue ($p \leq 0.045$). Additionally, an imbalance between the antagonistic ankle flexor muscles was identified as tibialis anterior EMG activity significantly decreased from the

29th minute onwards ($p \leq 0.046$) whereas significant reductions were not noted for the gastrocnemius ($p > 0.05$). Working in conjunction to protect the tibia from harm by stabilizing the lower leg, the gastrocnemius and the tibialis anterior absorb shock and reduce tensile stress on the bone (Mizrahi et al., 2000). When an imbalance occurs, the muscles have a reduced ability to protect the bone from injury occurring from a reduction in any one of the above protective mechanisms (Mizrahi et al., 2000). Both EMG ($p \leq 0.056$) and mean power frequency ($p \leq 0.048$) recordings showed reductions in tibialis anterior activity in all of the runners, where as the gastrocnemius measures stayed the same. Mizrahi et al. concluded that the antagonistic muscle imbalance, in conjunction with the increased impact acceleration at heel strike as a result of global fatigue, was correlated with the development of excessive tibial bending stress and consequently, a higher risk of stress fracture.

Work performed by Larson et al. (2011) indicates that fatigue induced kinematic changes may have the potential to change running style. Focusing on foot strike pattern, Larson et al. filmed a total of 936 runners competing in the Manchester City Marathon in Manchester, New Hampshire at a frame rate of 30Hz. Two-dimensional video recordings were taken of each runner at the beginning, at a 10-kilometer point and at a 32-kilometer point in the race. Larson et al. found that a large percentage of participants who initiated the run with a mid-foot or forefoot strike pattern had switched to a rear foot strike pattern at both 10-kilometer and 32-kilometer points in the run. The researchers concluded that the switch in foot strike was due to fatigue in the triceps surae muscle group (Larson et al., 2011), although this was not directly recorded.

It is well documented that muscular fatigue contributes to reductions in muscular strength (Gerdle et al., 1989; Glace, McHugh, & Gleim, 1998; Iridiastadi & Nussbaum, 2006; Longpre, Potvin, & Maly, 2013; Murdock & Hubley-Kozey, 2012) and has consistently been linked to increases in peak rearfoot eversion (Clansey et al., 2012; Dierks et al., 2010; Koblbauer et al., 2013). There is, however, limited research focused on changes in frontal plane knee kinematics resulting from running induced fatigue.

Much of the current literature focused on fatigue-induced changes in frontal plane knee biomechanics (adduction/ abduction) places a greater emphasis on kinetic variables. Cortes, Greska, Kollock, Ambegaonkar and Onate (2013) assessed the effects of a short-term fatigue protocol on lower extremity biomechanics in 18 uninjured female soccer players during a sidestep-cutting task. Three-dimensional lower extremity joint kinematics and kinetics were recorded while participants completed two unanticipated tasks (stop jump and side cutting). A light beam was placed across the running area two meters in front of a force plate. When the light beam was interrupted as the participant ran through it, a randomly generated athletic task (stop jump or side cutting) was projected on to a screen in front of the runner indicating which task to perform. Participants were required to approach the force plate at a minimum speed of 3.5 meters per second. All trials were completed with participants wearing tight fighting clothing and Adidas Supernova running shoes in conjunction with 40 retroreflective markers on the feet and lower extremities.

Kinematic analysis was performed using an eight camera Vicon motion capture system at 300Hz in conjunction with self-designed kinematic model using Visual 3D software to quantify motion at the hip, knee and ankle joints. Kinetic analysis was performed using two force plates sampling at a rate of 1200Hz.

Trials were completed before (pre-fatigue), half way through (50% fatigue) and after (100% fatigue) a fatigue inducing protocol. The short-term fatigue protocol consisted of a series of three counter jump movement at 90% of maximal vertical jump, step up and downs on a 30 centimeter box for 20 seconds, three squats to 90% of knee flexion and a pro-agility shuttle run (Cortes et al., 2013)

Cortes et al. (2013) focused on the side step cutting task for statistical comparison. The researchers reported a reduction in internal knee-adduction moment from pre-fatigue to 50% fatigue to 100% fatigue ($F(2,34) = 5.712$, $p = 0.003$) at initial contact (Cortes et al., 2013). Statistically significant decreases were also reported for knee flexion at 100% fatigue in comparison to 50% and pre-fatigue ($F(2,34) = 5.112$, $p = 0.004$) (Cortes et al., 2013).

A noteworthy pattern was identified for a few dependent variables. Knee flexion at peak stance ($F(2,34) = 8.282$, $p = 0.001$), knee abduction at initial contact ($F(2,34) = 3.784$, $p = 0.03$), and knee adduction moment at peak stance ($F(2,34) = 3.755$, $p = 0.03$)

increased from pre-fatigue to 50% but decreased from 50% to 100% with values less than pre-fatigue levels (Cortes et al., 2013). The reduced knee abduction angle at 100% fatigue in comparison to pre-fatigue indicates that fatigue may, in some way, be responsible for a more adducted knee angle at initial contact. Knee abduction angle at peak stance was found to be similar among all three testing conditions (pre-, 50% and 100% fatigue).

A study performed by Longpré, Potvin and Maly (2013) focused on fatigue induced lower extremity kinetic and kinematic changes. A convenience sample of 20 healthy young women absent of knee pain, injury or surgery was investigated. The study consisted of two visits, one to orient and familiarize the participants with the equipment and procedure, and the second to perform the study. Walking gait analysis was performed prior to and following two separate bouts of fatigue. Gait analysis and baseline peak extensor and flexor torques, recorded using a dynamometer, were used as pre-fatigue reference points (Longpre et al., 2013).

Gait analysis was performed using an eight camera Vicon motion capture system sampling at 100Hz in conjunction with three force platforms sampling at a rate of 1000Hz. Each gait analysis required participants to walk barefoot at a self-selected pace until five successful analyses were complete. An analysis was considered successful when the participants' right foot alone fell in full contact with one of the three force platforms (Longpre et al., 2013). The Vicon system tracked 24 markers affixed to the pelvis, lower extremities and feet of each participant. The primary focus of kinetic and kinematic analyses was external knee joint moments and knee joint angles respectively (Longpre et al., 2013). Additionally, muscle activity in the rectus femoris, vastus lateralis and biceps femoris was monitored using surface mounted electrodes during both gait analyses and peak torque measurements to determine muscle contributions.

The fatigue protocol was comprised of 50 isotonic knee flexions and extensions at 50% peak torque during a maximum voluntary isometric contraction (MVIC). Fatigue was considered a 25% reduction in MVIC (Longpre et al., 2013). Participants were allowed four sets of the fatigue protocol to reach "fatigue". If particpants were unable to reach 25% reduction in MVIC after four sets, that participant was removed from the study. Four participants were not able to reach a 25% reduction and their data was subsequently removed (Longpre et al., 2013).

Comparing pre-fatigue to post-fatigue gait analysis, Longpré et al. (2013) reported that fatigue reduced peak isometric torque from baseline ($p < 0.001$) but did not affect knee adduction moments, knee flexion angles, dynamic knee stiffness, or muscle co-activation (Longpre et al., 2013). Gait speed and stride length were also unaffected by fatigue. Fatigue did however reduce the peak knee extension moment from baseline ($p < 0.001$) (Longpre et al., 2013). Longpré et al. concluded that high intensity lower limb fatigue-inducing activity did not significantly alter knee joint mechanics while walking in young active women in a way that would increase their risk for injury (Longpre et al., 2013). Further investigation is required to determine if other forms of fatigue inducing activity, such as running, are able to significantly alter knee joint kinematics. Additionally, kinematic gait analysis performed with the participants running instead of walking and wearing running shoes instead of barefoot may reveal different findings.

Murdock and Hubley-Kozey (2012) assessed the effects of a high intensity quadriceps fatigue protocol on knee biomechanics. Assessing 20 healthy young adults (10 male and 10 female), participants performed gait profiles along a six-meter walkway before and after a quadriceps fatigue protocol. Muscle activity of the vastus lateralis, vastus medialis, rectus femoris, lateral and medial hamstrings and lateral and medial gastrocnemius was monitored using electromyography (EMG) via surface electrodes. Kinetic information was recorded using a single stationary force plate and kinematic data was captured at 100Hz using an Optotrak motion capture system synchronized with the force plate. Participants were outfitted with marker diode triads on their pelvis, thigh, lower leg and foot with 16 individual infrared emitting diodes on standard landmarks on the participant’s right side.

Gait analysis consisted of five successful trials. A trial was considered successful when the participant’s right foot came into full contact with the force plate, with no portion of their left foot touching the force plate (Murdock & Hubley-Kozey, 2012). Isokinetic strength measurements were performed using a dynamometer. Eight maximum voluntary isometric contractions were performed for knee extensors, flexors and foot plantar flexors to elicit maximum activation of each muscle group (Murdock & Hubley-Kozey, 2012). The quadriceps fatigue protocol consisted of 50 maximum effort knee extensions at 90° per second against an isokinetic dynamometer (Murdock & Hubley-Kozey, 2012).

Murdock et al. reported increased knee adduction angles ($p < 0.05$), primarily during swing, as well as increased net external adduction moment during stance (Murdock & Hubley-Kozey, 2012). To compound this, significant post-fatigue decreases were found in early stance leg external rotations ($p < 0.05$). Significant decreases were also reported for knee extensor torque (40%, $p < 0.05$) and bilateral quadriceps median power frequency (14-20%, $p < 0.05$). No pre- to post-fatigue muscle interactions ($p > 0.05$) or pre- to post-fatigue main effects ($p > 0.05$) were reported for any muscle activation characteristics.

1.3. Difference in Foot Posture

A study performed by Pierrynowski, Finstad, Kemecsey, and Simpson (2003) investigated subtalar joint inclination angle in an attempt to correlate joint orientation with different types of injuries. The researchers hypothesized that a higher subtalar joint axis would be found in participants with a history of knee pain and a lower subtalar joint axis would be found in participants with a history of foot pain (Pierrynowski, Finstad, Kemecsey, & Simpson, 2003). The hypothesis was based on the notion that the most accepted average subtalar joint inclination angle is 42° (+/- 9°) from horizontal and 16° medial to a line that extends from the center of the calcaneus to a point just between the first and second metatarsal heads (Close, Inman, Poor, & Todd, 1967; Inman, 1976). This joint orientation results in approximately equal amounts of frontal plane foot rotation (inversion/eversion) and lower leg transverse plane rotation (external/internal) (Pierrynowski et al., 2003). Deviations from this "average" subtalar joint inclination angle have the potential to alter the relative proportion of frontal plane foot rotation and lower leg transverse plane rotation (Pierrynowski et al., 2003).

Using questionnaires regarding the history of injury within the lower-extremity, 32 participants, mostly university students (25 women and 7 men ranging in age from 23 to 32 years old), were recruited (Pierrynowski et al., 2003). Sixteen participants were allocated to the knee injury group and 16 participants were allocated to the foot pain group. Using strategically placed markers and a kinematic data-acquisition system, participants performed both open and closed kinetic chain movements to determine the orientation of their subtalar joint axis. Data was analyzed using the van den Bogert's original code, simulated data, and data collected from known mechanical analogues of the foot (Pierrynowski et al., 2003). The findings of Pierrynowski et al. (2003) were consistent with the study's hypothesis. The subtalar inclination angle was higher in the knee injury group (43.9° +/- 5.5°) in comparison to foot injury group (36.8° +/- 7.3°) (Pierrynowski et al., 2003). The conclusions of Pierrynowski et al. stated that their study may provide evidence that different types of injuries may be correlated with different foot types (Pierrynowski et al., 2003). The researchers also declared that similar studies should be employed investigating different models of the rearfoot and lower leg coupling motions.

Focusing on foot type, Nawoczenski, Saltzman and Cook (1998) investigated how differences in foot structure affected three-dimensional kinematic behavior of the leg and rearfoot during running. This study consisted of 20 participants, 10 recreational runners in a low rearfoot group (pes planus) and 10 recreational runners in a high rearfoot group (pes cavus). Using anterior-posterior and lateral radiographic measurements to classify foot type, the researchers focused on the actions of a combined talocalcaneal and subtalar joint axis. The researchers found that this axis favored calcaneal inversion and eversion for their low rearfoot group and tibial medial and lateral rotation for their high rearfoot group (Nawoczenski, Saltzman, & Cook, 1998). This finding is consistent with the anatomy of the subtalar joint, as cited by Pierrynowski et al. (2003). Pierrynowski et al. found that a higher subtalar joint axis had increased amounts of transverse plane movement (tibial internal/ external rotation) in relation to transverse plane movement (rearfoot inversion/ eversion) resulting in knee pain and a lower subtalar joint axis had increased frontal plane movement in relation to transverse plan movement resulting in foot pain (Pierrynowski et al., 2003).

One limitation present in the work of Nawoczenski et al. is the use specially designed sandals for all testing procedures. The sandals were utilized to limit the confounding effects of footwear, unfortunately this also limits the ecological validity of the findings. A similar study performed with the participants donning their own footwear would have increased external validity, as it would be more relatable to the participants every day running conditions. Using specially designed sandals limited the applicability of the work of Nawoczenski et al. to real life running conditions.

A study performed by Lees, Lake and Klenerman (2005) investigated arch height and forefoot running. The researchers hypothesized that arch height affected the ability of the foot to absorb shock and would thus affect the choice of strike pattern. Looking at dynamic loading rate during forefoot running, Lees et al. (2005) focused on the first differential of vertical force recorded on a Kistler force platform at a sampling rate of 500Hz. This study involved 18 participants whose "clinically normal" feet were classified using the Arch Height Index (Butler, Hillstrom, Song, Richards, & Davis, 2008). The results of dynamic loading rates indicated three peaks and two intervening troughs (Lees, Lake, & Klenerman, 2005). Lees et al. were unable to correlate any of the force peaks or load rate peaks with the participant's foot type ($p > 0.05$). The researchers concluded that the height of a person's arch was not important in defining the functional capacity of the foot in action (Lees et al., 2005) and that other factors need to be assessed when investigating differences between arch heights.

Barnes, Wheat and Milner (2011) investigated forefoot and rearfoot kinematics, as well as tibial shock in 15 high arched (pes cavus) and 15 low arched (pes planus) participants. While wearing the same neutral midsole sandal, Barnes et al. (2011) had their participants run at a speed of 3.5 meter/second over a single force plate. Data was recorded for the right foot only. The researchers also had an eight-camera motion capture system set up to record kinematic data. Kinematic data was analyzed using a model designed by Carson et al., Digby et al. and Nester and Findlow, using 18 retroreflective markers (Barnes, Wheat, & Milner, 2011).

Using a significance level of $\alpha = 0.05$ for all statistical comparisons, multivariate analysis found a smaller forefoot abduction excursion and reduced forefoot abduction velocity in their low-arched group. The researchers suggest that this may result from a reduced available range of forefoot motion before reaching the end range of motion in the participants with low arches (Barnes et al., 2011). Barnes et al. Conclude by stating that because forefoot and rearfoot motion as a shock attenuation mechanism was not found ($p > 0.05$), foot kinematics associated with early stance is an area where further investigation is required to establish a link to injury risk (Barnes et al., 2011).

Focused on participants with a history of running-related lower extremity injuries, Williams, Hill and Nester (2001) investigated injury patterns in runners with different foot types. All eligible participants were screened with use of an arch ratio to determine whether they belonged to the high arch (HA) or the low arch (LA) group. The arch ratio was a measurement of the height to the dorsum of the foot from the floor (at 50% of the foot length), divided by the distance from the most posterior aspect of the calcaneus to the medial joint space of the first metatarsophalangeal joint (D. S. Williams, 3rd, McClay, & Hamill, 2001). Groups were determined based on normative values established in a previous study performed by Williams and McClay (2000), which measured a total of 102 feet (D. S. Williams & McClay, 2000). The mean arch ratio was determined to be 0.316 (SD ±0.027) (D. S. Williams & McClay, 2000). Williams et al. (2001) considered any measurement at or above 1.5 SD in the positive direction HA, and any measurement at or below 1.5 SD in the negative direction LA (D. S. Williams, 3rd et al., 2001).

Twenty HA (10 females and 10 males) and 20 LA (12 females and 8 males) participants who were not experiencing injury at the time and who ran at least six miles per week at a minimum eight minute per mile pace were included. Participants completed a questionnaire in which all running-related lower extremity injuries, as reported by a medical professional, were recorded. Participants were informed to be as specific about the location and nature of the injuries as possible (D. S. Williams, 3rd et al., 2001). To analyze the results, each injury was placed into one of three groups. Injury groups consisted of (1) medial or lateral; (2) bony or soft tissue; and (3) foot/ ankle or knee (D. S. Williams, 3rd et al., 2001).

Comparison of 70 injuries in the HA group to 64 injuries in the LA group using chi-square analysis, found statistically significantly differences in injury pattern ($p < 0.05$)

(D. S. Williams, 3rd et al., 2001). Lateral injuries were more common in the HA group, whereas medial injuries were more common in the LA group ($X^2 = 9.22$, $p = 0.002$) (D. S. Williams, 3rd et al., 2001). Additionally, boney injuries had an increased incidence in the HA group whereas soft tissues injuries were more commonly reported in the LA group ($X^2 = 3.42$, $p = 0.047$) (D. S. Williams, 3rd et al., 2001). Finally, knee injuries had an increased incidence in the LA runners and foot and ankle injuries were reported more commonly in the HA runners ($X^2 = 4.03$, $p = 0.045$) (D. S. Williams, 3rd et al., 2001).

Williams et al. (2001) report that a more pronated and flexible foot type, inherent to the LA foot type, may predispose runners to medial and soft tissue injuries, as well as knee injuries due to associated tibial internal rotation (D. S. Williams, 3rd et al., 2001). Conversely, the more rigid supinated position associated with the HA foot may predispose runners to lateral and boney injuries, as well as injuries of the foot and ankle (D. S. Williams, 3rd et al., 2001).

1.4. Stride-to-Stride Variability

Movement variability has been suggested as a potential contributing factor to injury and rehabilitation in both pathological and athletic populations (DeLeo, Dierks, Ferber, & Davis, 2004; Ferber, Kendall, & Farr, 2011; Hamill, van Emmerik, Heiderscheit, & Li, 1999; Miller, Meardon, Derrick, & Gillette, 2008). In the biomechanic literature, movement variability is defined as the variance from stride-to-stride during repeated motion, even when the goal of the motion remains constant (Ferber et al., 2011; Miller et al., 2008). In other viewpoints, such as engineering, variability is regarded as noise (Hamill et al., 1999). While noise is often considered an unwanted byproduct in the eyes of an engineer, the roll of pattern variability is gaining interest in a number of disciplines (Hamill et al., 1999). In a publication regarding self-organization of the brain and behavior, Kelso (1997) states that variability arises due to the non-linear interactions between the point of interest and its environment (Kelso, 1997). Further, Kelso claims that continuous fluctuations lend stability and coordination to the system (Kelso, 1997). This notion resonates well with movement variability. More specifically, stride-to-stride variability has been suggested to contribute to coordination change and a combination of stability and flexibility of movement during gait (Hamill et al., 1999).

Stride-to-stride variability has been identified as both beneficial and harmful to the individual depending on the type of variability measured (Ferber et al., 2011). Regarding gait, two types of stride-to-stride variability exist, global and local. Global variability focuses on factors that are not specific to one or paired segments such as stride length, stride width and stride time. With regards to global variability, increases in stride length and stride time have been linked to increased risk of falling in pathological and elderly populations (Hausdorff, Rios, & Edelberg, 2001).

Local variability addresses coupling or angles between two segments or joints, or within one joint or segment (Hamill et al., 1999). Focused on local variability, Hamill, van Emmerik and Heiderscheit (1999) performed two separate studies addressing Q-angle and patellofemoral pain syndrome. Using kinematic recordings (200Hz) of both overground and treadmill running, phase angles were calculated for thigh adduction/ abduction, thigh flexion/ extension, tibial rotation and foot inversion/ eversion. Of interest was continuous relative phase variability, which was defined as the difference between the normalized phase angles between two segments during the stance phase or during the entire stride (Hamill et al., 1999). Phase plots were constructed and used to illustrate the coupling of lower extremity segments during running.

The first study compared participants with a Q-angle greater than 15° to participants with a Q-angle less than 15° while running overground. There were no statistically significant findings for mean continuous relative phase or variability when comparing the high Q-angle to the low Q-angle groups ($p > 0.05$) (Hamill et al., 1999).

The second study compared participants who were symptomatic with patellofemoral pain syndrome to participants who were symptom free. Focusing on treadmill running, symptomatic patellofemoral pain participants were compared to pain free participants while running at three different velocities (2.5, 3.0 and 3.5 meters per second). Variability was reportedly less in the symptomatic group when compared to the asymptomatic group and was especially strong in the transition from stance to swing phase and from swing to stance phase ($p < 0.05$) (Hamill et al., 1999). Due to speculation that increased stride-to-stride variability may induce pain in symptomatic runners, Hamill et al. (1999) claim that the minimally deviated movement patterns, indicated by a reduction in variability, may enable these individuals to accomplish the task of pain free running by avoiding painful ranges of motion (Hamill et al., 1999). Further, the researchers state that the narrow continuous relative phase variability range may be indicative of the presence of injury but did not determine the cause of injury (Hamill et al., 1999).

Also assessing continuous relative phase, Drewes et al. (2009) compared rearfoot and shank coupling in participants with chronic ankle instability (n = 7) to healthy matched controls (n = 7). Three-dimensional treadmill analysis was recorded while participants performed treadmill walking and running. Results indicate the ankle instability group was more out of phase than the healthy controls. During walking trials, a statistically significant difference for continuous relative phase was obtained from 94% to 97% of the gait cycle ($p < 0.05$) in the ankle instability group indicating that the rearfoot was moving ahead of shank rotation in phase space (Drewes et al., 2009). Similarly, jogging trials indicate statistically significant differences were present from 47-58% and from 84-93% of the gait cycle ($p < 0.05$) in the ankle instability group. Referencing divisions of the gait cycle as described by Perry and Burnfield (2010), 47-58% of the gait cycle corresponds with late terminal stance (31-50%) and pre-swing (50-62%), 84-93% corresponds with late mid swing (75-87%) and terminal swing (87-100%), and 94-97% corresponds with terminal swing (Perry & Burnfield, 2010). Drewes et al. (2009) suggest that increased variability during terminal swing may influence foot position while preparing for heel strike, potentially influencing the risk of sustaining recurrent ankle sprains (Drewes et al., 2009). Thus, a less coordinated and more variable gait was thought to predispose individuals with chronic ankle instability to ankle inversion injuries.

Supporting the notion that increased stride-to-stride variability may be associated with injury, as reported by Drewes et al. (2009), McKeon (2009) reported a decrease in shank-rearfoot coupling variability following a four-week balance exercise program in a group of 29 individuals (12 males, 17 females) who presented with self-reported ankle instability. Comparisons were made for rearfoot inversion/ eversion, shank rotation, and coupling between the rearfoot and shank before and after a four week single-limb stance balance stabilization program. A statistically significant reduction in shank rearfoot coupling variability was noted during walking following the balance training (balance training post-test: 13.1° +/- 6.2°, balance training pre-test: 16.2° +/- 313°, $p = 0.03$). Pre- to post-test results did not statistically significantly change in the control group (post-test: 16.30° +/-4.4°, pre-test: 18.6° +/- 7.1, $p > 0.05$) (McKeon et al., 2009). Significant improvements were also reported in Foot and Ankle Disability Index and the Foot and Ankle Disability Index Sport scores in the balance-training group when comparing pre- to post-test measures (McKeon et al., 2009). Neither inversion/ eversion nor shank rotation kinematics statistically significantly differed.

McKeon et al. reported that the reduced post-test variability and improved self-reported Foot and Ankle Disability Index scores was indicative of increased ankle stability. Lending to the notion that increased variability may be linked to instability (Drewes et al., 2009; McKeon et al., 2009), these results indicate that greater local variability may be associated with injury.

Miller, Meardon, Derrick and Gillette (2008) compared lower extremity movement coordination variability in a group of eight runners who had been diagnosed with iliotibial band syndrome (ITBS) to a group of eight runners with no history of injury. Using an eight-camera 120Hz motion capture system and retroreflective markers, participants were analyzed while running on a treadmill at a self selected pace that would exhaust them within 20 minutes. Joint and segment angles were calculated for knee flexion/ extension, foot, knee and thigh adduction/ abduction and tibial internal/ external rotation. Phase plots were calculated for each movement. Movement couplings were tracked for five matched segments that were selected based on their likelihood of impacting strain on the iliotibial band (Miller et al., 2008). Continuous relative phase angles were calculated for thigh adduction/ abduction and tibial internal/ external rotation, thigh adduction/ abduction and foot inversion/ eversion, tibial internal/ external rotation and foot inversion/ eversion, knee flexion/ extension and foot adduction/ abduction and finally knee adduction/ abduction and foot internal/ external rotation. Continuous relative phase angles were calculated by subtracting the distal segment from the proximal segment. Between stride standard deviations for each subject at each time point were used to calculate continuous relative phase variability. Variability was then averaged across the full stride, stance, or swing to obtain a single value for each of these periods, and then averaged across subjects (Miller et al., 2008).

Statistical analysis using a factorial analysis of covariance (ANCOVA) with group (ITBS or control) and time (start of run or end of run) as factors and speed as a covariate indicated no significant main effects for time or for the interactions between group and time (Miller et al., 2008). Statistically significant findings for group indicate that runners with a history of ITBS were more variable for knee flexion/ extension and foot adduction/ abduction coupling at the start of the run (18.6° for ITBS vs. 15.3° for control, $p = 0.02$), less variable in thigh adduction/ abduction and foot inversion/ eversion coupling at the end of the run (30.5° for ITBS vs. 33.1° for control, $p = 0.03$) and also displayed a tendency for less variability in thigh adduction/ abduction and tibial internal/ external rotation coupling at the end of the run (31.4° for ITBS vs. 33.5° for control, $p = 0.09$) during the complete gait cycle (Miller et al., 2008). Increased variability was found in swing phase for runners with a history of ITBS in knee flexion/ extension and foot adduction/ abduction coupling (18.8° for ITBS vs. 15.4° for control, $p = 0.04$). Stance phase couplings demonstrate increased variability in knee flexion/ extension and foot adduction/ abduction coupling for the ITBS group at the start (18.6° for ITBS vs. 15.3° for control, $p = 0.02$) and end (19.0° for ITBS vs. 14.5° for control, $p = 0.003$) of the run (Miller et al., 2008). The only statistically significant group difference in continuous relative phase variability at heel strike was tibial internal/ external rotation and foot inversion/ eversion coupling at the start of the run (13.3° for ITBS vs. 24.2° for control, $p = 0.004$) indicating less variability in the ITBS group.

The findings of less variability in selected couplings were consistent with the results of Hamill et al. (1999), however, statistically significant increases in knee flexion/ extension and foot adduction/ abduction coupling variability, particularly during stance, were not. Miller et al. report this finding may indicate a lack of stability in this coupling. One of the conclusions reached based on a reduction in stability was that muscular strength may influence continuous relative phase variability (Miller et al., 2008). Due to the non-significant interaction effect between group and time, time did not have a differential effect on continuous relative phase variability, indicating that fatigue did not have a large, consistent effect on variability in the couplings that were analyzed (Miller et al., 2008).

Contrary to the findings of Hamill et al. (1999), the results of Drews et al. (2009), McKeon et al. (2009) and Miller et al. (2008) suggest that pathological runners may demonstrate increased movement variability (Miller et al., 2008). Supporting this notion, Ferber, Kendall and Farr (2011) report an increased stride-to-stride knee-joint variability in their assessment of participants with patellofemoral pain syndrome (PFPS) in comparison to a control group following a three-week hip-abductor muscle-strengthening protocol (Ferber et al., 2011).

In a method first employed and validated by Derrick, Bates and Dufek (1994), Ferber et al. (2011) used the Pearson product-moment correlation coefficient to investigate temporal similarity across the entire stance phase for individual variables (Derrick, Bates, & Dufek, 1994; Ferber et al., 2011). This method assesses curve correlation by using point-to-point Pearson product-moment correlation coefficients between paired data points of two comparative curves (Derrick et al., 1994). Derrick et al. (1994) report that the correlation coefficient may be used for this purpose when the two time series data sets begin and end with ordinal values equal to zero and demonstrate constant proportionality between all nonzero paired ordinal time points (Derrick et al., 1994).

Ferber et al. assessed two-dimensional frontal plane knee biomechanics in 25 participants, 15 of whom were clinically diagnosed with PFPS (5 men and 10 women) and 10 who were not (4 men and 6 women). Treadmill running at a speed of 2.55 meters per second for all recordings was performed while a 60Hz camera recorded kinematic data at the knee. Maximal isometric hip-abductor muscular strength was assessed at base line and after three weeks of the strengthening protocol with each participant lying on his or her side using a dynamometer to assess force output. Kinematic data collected from 10 consecutive footfalls was analyzed for stance and normalized to 101 data points corresponding to normalized timing of stance phase. Temporal patterns of knee genu valgum for each footfall were compared on a point-by-point basis to the subsequent footfall (i.e. footfall one was compared to footfall two, two compared to three, etc.) for all

101-time points and across all 10 consecutive footfalls. A 2 x 2 (group x time) repeated-measures analysis of variance was used to identify differences in peak isometric hip-abduction force, peak genu valgum angle and stride-to-stride knee joint variability.

Consistent the findings of Hamill et al. (1999) that reduced movement variability may be indicative of the presence of an injury due to protective movement patterns, Ferber et al. (2011) proposed a two part hypothesis. First, the researchers hypothesized that the PFPS group would exhibit a reduction in hip-abductor strength, greater peak genu valgum angle, and a decreased stride-to-stride knee-joint variability in comparison to the control group at baseline (Ferber et al., 2011). Further, the researchers hypothesized that in comparison to baseline values, the PFPS group would demonstrate an increased hip-abductor muscle strength, a reduction in pain, a decrease in peak genu valgum angle and an increase in stride-to-stride variability following their three-week rehabilitation protocol (Ferber et al., 2011).

Assessing baseline scores, the PFPS group demonstrated 28.71% less hip abductor strength ($p = 0.01$), no difference genu valgum angle ($p = 0.67$) and increased stride-to-stride knee joint variability ($p = 0.01$) in comparison to the control group (Ferber et al., 2011). Posttest results indicated a 32.69% abductor strength increase in the PFPS group and no strength differences in comparison to the control group ($p = 0.33$). No differences in genu valgum angle were reported in relation to baseline scores ($p = 0.55$) or to the control group ($p = 0.65$). Assessing post rehabilitation kinematic data, statistically significant reductions in stride-to-stride knee joint variability curves were reported for the PFPS in comparison to baseline values ($p = 0.01$), with no differences between the PFPS and control groups ($p = 0.36$) (Ferber et al., 2011). No differences were reported for the control group between testing sessions.

In contrast to the hypothesis, the PFPS group displayed increased stride-to-stride variability when compared to the control group at baseline and a reduction in variability following the strengthening protocol. This indicates that the PFPS group achieved a more consistent stride-to-stride kinematic pattern following rehabilitation. Although inconsistent with the findings of Hamill et al. (1999) this finding appears reasonable considering that restoring strength, in addition to a more consistent and predictable movement pattern would be expected following a rehabilitation protocol (Ferber et al., 2011).

The work of Ferber et al. (2011) is distinct from previous research investigating stride-to-stride variability as they focused on variability within one joint, not between two segments, as did Miller et al. (2009) and Hamill et al. (1999). Building on the work of Ferber et al. (2011), it seems reasonable to believe that fatigue induced strength reductions may also result to increase frontal plane kinematic variability, both at the level of the knee (adduction/ abduction) and the ankle (inversion/ eversion), independent from one another.

It is evident that some of the effects of muscular strength reduction and the kinematic differences between foot types have been researched separate from one another. It remains to be seen, however, if the effects of reduced muscular strength differ across foot type. Differences in foot type include but are not limited to; pes planus, or flat medial longitudinal arch, pes cavus, or high medial longitudinal arch and rectus foot type, or "normal" medial longitudinal arch. A statement put forth by Ounpuu (1990) in a study focused on the biomechanics of running, epitomizes the goals of this study. Ounpuu claimed that through a more in-depth understanding of the mechanics of locomotion, physicians would be able to more accurately diagnose and treat running injuries (Ounpuu, 1990). Although Ounpuu's research is more than 24 years old, its message is still applicable today.

It is of particular interest to investigate how, if at all, reductions in muscular strength affect lower limb kinematics within a pes planus population. These conditions were selected due to the fact that adverse changes in kinetic and kinematic measures have the potential to lead to running-related injury (Clansey et al., 2012). The ability to link running-related injuries to specific foot types may contribute to a reduction in running related medical care through the application of appropriate footwear, foot orthoses, and exercise prescription.

Given the lack of conclusive evidence focused on how the effects of muscular fatigue differ across foot type, the purpose of this study was to investigate how, if at all, reductions in muscular strength affect mean peak stance phase ankle eversion, mean peak stance knee adduction, stride-to-stride correlation and movement variability in runners with planus feet.

Consistent with the findings reported above, it was hypothesized that running would result in reductions in inversion and eversion strength at the ankle (Gerdle et al., 1989; Glace et al., 1998; Iridiastadi & Nussbaum, 2006; Longpre et al., 2013; Murdock & Hubley-Kozey, 2012), as well as increased peak rearfoot eversion (Clansey et al., 2012; Dierks et al., 2010; Koblbauer et al., 2013). Although the reported effects of fatigue on peak stance knee adduction are limited, it was hypothesized that reductions in muscular strength would lead to increases in peak stance knee adduction (Murdock & Hubley-Kozey, 2012). Additionally, despite the inconsistent findings surrounding stride-to-stride variability it was hypothesized that stride-to-stride variability would increase post-run.

Chapter 2: Methods

2.1 Sample

Participants were recruited from Runner's Choice and Running Room, two community based running groups located in London, Ontario. Individuals that expressed interest were provided with a presentation wherein the study purpose was described and the study procedures were outlined. Interested parties were issued a letter of information detailing the study and were invited to have their arch height assessed using the Arch Height Index following the presentation as a pre-screening measure. Additionally, both Runner's Choice and Running Room were given a participant recruitment poster to advertise in their respective stores. The research protocol, recruitment method, and mechanism for obtaining informed consent were approved by the Health Sciences Research Ethics Board, at the University of Western Ontario; approval #103376 (Appendix A).

Inclusion was limited with respect to several factors. First and foremost, eligible participants were required to have a pes planus foot type as determined using the Arch Height Index. In a study performed by Butler et al. (2008) using 11 participants, the Arch Height Index displayed high intrarater (ICC's between 0.96 and 0.99) and interrater reliability (ICC's between 0.98 and 0.99) for each of the three measurements involved (Butler et al., 2008). Using the Arch Height Index to measure a group of 100 male and female recreational runners, Butler et al. determined that the mean arch height was 0.34 with a standard deviation of 0.03. For the purpose of this study, any arch height greater than one standard deviation (0.03) away from the mean (0.34) in the "planus" direction (< 0.31) was considered "pes planus". Additionally, only healthy adults between the ages of 18 and 60 years of age with the absence of leg length discrepancy were considered. Eligible participants were required to demonstrate a history of distance running and/or training having completed at least one organized running event greater than 5 kilometers in the 12-month period prior to completion of the study. Beginner and novice level runners may have an increased risk for running related injury (Koblbauer et al., 2013) and were therefore excluded. Furthermore, eligible participants were required to commute to the Fowler Kennedy Sport Medicine Clinic at Western University on the day of their testing session.

Individuals were excluded from this study if they: i) did not have a pes planus foot structure defined as any measurement greater than 0.31 using the Arch Height Index; ii) presented with a leg length discrepancy greater than three millimeters; iii) had a recent history of surgery, injury, or break or sprain that may impact their running; iv) had not completed an organized run greater than 5 kilometers in the 12 months prior to the date of completion of the study; v) required the use of custom foot orthoses to complete the run.

The initial sample consisted of 30 participants, 14 of whom were excluded due to ineligible arch height and one who was excluded from analysis due to a data processing error. Each of the 15 participants included in the final sample had competed in at least one organized run in the 12 months prior to the date of testing. Competition distances ranged from 5 kilometers to full marathon (42.2 kilometers). See Table 2.1 for participant demographics.

Due to the fact there are no published studies to parse out the differences between foot type as it relates to the variables tested, an adequate estimate of effect size upon which to justify sample size is difficult. Assuming a medium effect size (delta= 0.75), an alpha of 0.05, and using a 2 x 2 repeated measures multivariate analysis of covariance (MANCOVA) with two levels of the independent variable (time x side) and two covariates (arch height and change in strength), we would expect to detect a significant difference 80% of the time with a sample of 16 participants [http://euclid.psych.yorku.ca/cgi/power.pl].

2.2 Scheduling

After the initial arch height measurement, each eligible participant was scheduled to attend a testing session at the SoleScience Pedorthic Clinic in the Fowler Kennedy Sports Medicine Clinic. Participants were booked to begin their trial on an individual basis, with no more than three participants assessed per test day. The length of testing sessions was dependent upon the distance and pace of the run. The majority of testing sessions spanned 90 to 120 minutes.

2.3 Screening

Upon arrival to the clinic, participants were asked to complete a short screening form that was approved by the Western University Research and Ethics Board (Appendix B). The form included unique participant identification number, age, sex, height, weight, length of time they had been distance running, distance selected for the run (any distance from 5 to 8 kilometers), fastest time completed at the distance selected, which was their dominant foot, defined as the foot that the participant would use to kick a soccer ball (Koblbauer et al., 2013), as well as any previous injuries to their feet or lower limbs.

Upon completion of the screening form, an official arch height measurement was taken to verify eligibility. Next, leg length was assessed using the average of two measurements using the tape measure method (Beattie, Isaacson, Riddle, & Rothstein, 1990) to determine if a leg length discrepancy (LLD) was present, and if so, if it was within "normal" limits. Subotnick (1981) suggest that LLDs as little as three millimeters in the running population may require intervention to facilitate injury reduction (Subotnick, 1981). Other studies claim that five millimeters may serve as the operational definition of a leg length discrepancy, further yet, six millimeters of discrepancy has been deemed the point at which a discrepancy reaches clinical significance (Friberg, Nurminen, Korhonen, Soininen, & Manttari, 1988; Holmes, Pruitt, & Whalen, 1993). Due to inconsistencies within the current literature, "normal" limits were considered any LLD measuring less than three millimeters, as measured using the tape measure method.

2.4 Pre-Run Testing

All testing was performed with participants wearing their own footwear to best approximate everyday running conditions, and because unfamiliar footwear may have the potential to influence lower extremity kinematics and possibly increased the risk of injury.

While seated on the plinth with knees extended to 180° and a 90° bend at the hips, the participant's isometric inversion strength was measured, followed by isometric eversion strength using a hand held dynamometer (Nicholas Manual Muscle Tester). Each leg was tested for both inversion and eversion strength twice and an average of the two measures for each leg was recorded. A study performed by Aydog et al. (2004) demonstrated high intra-tester and inter-tester reliability when using a hand held dynamometer to measure isokinetic inversion and eversion strength with the ankle in a neutral position (Aydog, Aydog, Cakci, & Doral, 2004). Additional support is evident in the work of Kelln et al. (2008) (Kelln, McKeon, Gontkof, & Hertel, 2008). All dynamometer testing was done by the same examiner (A.F.).

Pre- and post-run kinematic information was recorded using a 7-camera Vicon motion capture system (Vicon, Oxford, UK) with the participant running on a KISTLER Gaitway II S H/P/COSMOS treadmill. The same treadmill was used for all gait recordings and was found to be accurate to 0.11 meters per second using the run speeds reported by the Vicon 3D Gait system. Comparing treadmill running to over-ground running, Fellin et al. (2010) report that kinematic curve analysis was similar, on average, for the hip, knee and rearfoot joints (Fellin, Manal, & Davis, 2010). The one movement that significantly differed between over-ground and treadmill running was ankle dorsiflexion. Fellin et al. speculated that the difference was a result of decreased stride length required while running on a treadmill (Fellin et al., 2010). Kinematics assessed in this study took place at the level knee and rearfoot permitting the validity of treadmill running for use in this study. Maintaining a constant surface type (treadmill at +1 inclination) and speed between the pre- and post-run analysis for each individual, permitted changes in lower limb kinematics to be identified as fatigue related. One participant requested that their gait speed be slowed from a pre-run testing speed of 2.2 meters per second to 2.0 meters per second during post-run testing due to feelings of fatigue induced unease. The minimal reduction in speed was deemed acceptable, as it was unlikely to affect kinematics.

The Vicon motion capture system consisted of seven infrared cameras used to track a grouping of markers attached to the participant. Marker placement was based on a model designed by Milner et al. (2006). Molded thermoplastic shells, each with four noncollinear markers, were attached bilaterally to the posteriolateral proximal thigh and posteriolateral distal shank (Milner, Ferber, Pollard, Hamill, & Davis, 2006). One additional rigid shell with three markers was attached to a belt around the participant's waist using Velcro to approximate the sacroiliac joint. Sixteen individual retroreflective markers were used to define the anatomic coordinate system and inertial parameters of each of the segments in the lower extremity (Milner et al., 2006). Individual markers were attached bilaterally to the participant's lateral femoral epicondyle, medial and lateral knee at the level of the lateral femoral epicondyle, as well as the medial and lateral malleolus to approximate the ankle joint center (Milner et al., 2006). Additionally, three markers were attached bilaterally to the posterior heel of the participant's running shoes to approximate rearfoot motion: two markers marking the vertical bisection of the heel and a third on the lateral portion of the shoe to differentiate left from right foot (Milner, Ferber, Pollard, Hamill, & Davis, 2006). All participants were asked to wear tight fitting running clothing to comply with the markers.

All kinematic data were sampled at 200Hz which is a well cited as an acceptable sampling rate (Gehring, Mornieux, Fleischmann, & Gollhofer, 2013; Segal et al., 2006). Ankle, knee and hip joint kinematic angles were calculated in frontal, transverse, and sagittal planes of motion using 3D GAIT custom software (Gait Analysis Systems Inc., Calgary, Alberta, Canada). The resulting kinematic data was presented in an excel spreadsheet separated by joint type (hip, knee or ankle) and plane of motion (frontal, sagittal or transverse) for both the left right side.

Prior to gait analyses, each participant performed one static calibration trial to establish marker orientation. The participant was instructed to stand in a relaxed neutral position with their feet positioned 30 centimeters apart pointing straight ahead and their arms crossed in front of them (Leigh, Pohl, & Ferber, 2013). Following successful completion of the static trial, a recorded walking trial, lasting approximately 30 seconds, was performed at a speed of 1.3 meters per second (+/- 0.11 meters per second). Finally, an approximately thirty-second running trial was completed, at a speed of 2.4 meters per second (+/- 0.639 meters per second). The run speed of 2.4 meters per second served as a starting point allowing the runner to increase or decrease the speed to match their typical running speeds. For the purposes of this study, only the data recorded during the running trial was used for kinematic analysis. The kinematic variables assessed included mean peak stance phase ankle eversion, mean peak stance knee adduction, and stride-to-stride variability for both mean peak stance ankle eversion and mean peak stance knee adduction.

2.5 Run

Following the completion of the pre-run testing, each participant was escorted to the TD Stadium track, a standard 400-meter synthetic rubber track at Western University, where all running trials were completed. The use of a flat 400-meter rubber track allowed for a controlled running environment free from obstacles, elevation changes and running surface changes. A controlled running environment helped to limit the potential for injury during the study, and also standardized the run for each participant. Runners were permitted to perform a self-selected warm up prior to commencing the run if so desired. Upon completion of a warm up, each participant then completed a run of a selected distance (any from five to eight kilometers). Each run was completed on the 400-meter track permitting the distance to be monitored through number of laps completed. The run was supervised by one of the investigators of the study at all times to count the number of laps and provide support for the participant in case of injury. Each participant was encouraged to run at a pace that would match or better his or her fastest completed run at the distance selected. See table 2.2 for all run times. No injuries were reported at any point during or following the study.

2.6 Post Run Testing

Immediately upon completion of the run, prior to any cool-down exercises or stretches, each participant was escorted back to the Fowler Kennedy Sports Medicine Clinic. Within two minutes of completing the run, exactly the same procedure was performed for isometric strength, followed by kinematic analysis. All kinematic analyses were performed on the same treadmill at the same walking and running speeds as were used during the pre-run testing. For safety reasons, one participant was permitted to slow the speed of their pre-run testing speed of 2.2 meters per second to 2.0 meters per second during post-run testing due to feelings of fatigue induced unease. Marker placement from pre-run kinematic analyses was marked directly on the participants skin and shoes using tape and a pen to ensure that marker placement was as close as possible for both pre- and post-run kinematic testing. Following the post-run testing, the study was complete. Participants were asked if they had any questions regarding any of the measures that were taken, confidentiality procedures, or any other relevant information. Each participant was then permitted to return outside and complete any self-selected cool down exercises if desired.

2.7 Statistical Analysis

Changes in lower extremity kinematics were analyzed in SPSS (SPSS version 21.0: SPSS Inc., Chicago, IL) using 2 x 2 factorial analysis of variance (ANOVA). Individual factorial ANOVAs were performed for mean peak stance phase ankle eversion and mean peak stance phase knee adduction with a Bonferroni correction to the alpha value in order to account for multiple comparisons. In each analysis, side (dominant versus non-dominant) and time (pre-run versus post-run) were used as independent variables.

Kinematic data was assessed using a total of 10 stance phases with each stance phase consisting of 101 normalized time points. Temporal similarity between subsequent stance phases was assessed in a protocol employed by Ferber et al. (2011). Frontal plane ankle and knee kinematics for each stance phase were compared on a point-by-point basis to the subsequent stance phase using the Pearson product-moment correlation coefficient for all 101 normalized time points across the first 10 accepted strides. The resulting nine correlation coefficients for each participant were averaged across all 15 participants yielding a mean score for each of the nine stride comparisons. To assess differences between pre- and post-run analyses, confidence intervals across the nine pre-run average scores were compared to the nine post-run average scores.

Looking further into stance phase variability, standard deviations were calculated across the mean peak values (ankle eversion and knee adduction) of the first 10 strides individually for each participant. Standard deviations were calculated across the 10 strides at each time point, for each participant. The standard deviations were then averaged across all participants for each independent variable resulting in one standard deviation for each of the 101 time points for dominant and non-dominant sides at both pre- and post-run times of ankle eversion and knee adduction. In a method similar to that employed by Miller, Meardon, Derrick and Gilette (2008), stance phase was assessed in segments. Values at 10%, 50% and 90% of stance phase were used in a 2 x 3 x 2 multivariate analysis of variance (MANOVA). Side (dominant and non-dominant), portion of percent stance (10%, 50% and 90%) and time (pre-run and post-run) were used as independent variables to assess ankle eversion and knee adduction variability. Significant interaction effects were assessed by performing individual factorial analysis of variance (ANOVA) for each dependent variable, with pairwise comparisons of estimated marginal means to compare each pre-run portion of stance phase to its respective post-run portion as post-hoc testing.

Table 2.1 Participant demographics

Participant	Sex	Age (Years)	Foot Dominance	Run Distance (Km)	AHI	
					Non-Dominant Foot	Dominant Foot
1	F	48	Rt	8	0.2933	0.2873
2	F	52	Rt	5	0.2857	0.2892
3	F	40	Rt	5	0.3155	0.3118
4	F	48	Rt	5	0.3112	0.3095
5	F	49	Rt	5	0.3092	0.2991
6	M	32	Lt	5	0.2954	0.3145
7	F	30	Rt	5	0.3001	0.2752
8	M	57	Rt	5	0.3113	0.3113
9	M	52	Rt	5	0.3119	0.3111
10	F	31	Rt	5	0.291	0.2920
11	F	24	Rt	5	0.3189	0.3138
12	F	23	Rt	5	0.3079	0.3021
13	F	50	Rt	5	0.3135	0.3115
14	M	60	Rt	5	0.2824	0.2804
15	M	39	Rt	5	0.3148	0.2896
Where AHI = Arch Height Index; Lt = Left foot; Rt = Right foot						

Table 2.2 Participant run times and speeds

Participant ID	Run Time (minutes:seconds)		Recorded Treadmill Run Speed (meters/second) Table	
	Personal Best	During Study	Pre-Run Analysis	Post-Run Analysis
1	50:30*	48:00:00*	2.0934	2.1133
2	30:40:00	29:21:00	2.3298	2.0477
3	25:40:00	24:58:00	2.533	2.4999
4	28:00:00	27:41:00	2.2617	2.184
5	24:00:00	24:11:00	2.2302	2.2298
6	16:45	20:10	2.253	2.2418
7	25:00:00	22:30	3.0395	2.9873
8	27:00:00	26:30:00	2.4292	2.4355
9	30:00:00	29:08:00	2.3207	2.4379
10	24:00:00	23:13	2.4262	2.4149
11	28:30:00	31:29:00	2.3974	2.4379
12	20:00	22:30	2.8238	2.9382
13	27:58:00	27:00:00	2.4783	2.4994
14	29:33:00	31:10:00	2.3203	2.2954
15	22:00	22:20	2.3834	2.3676
* Run times for participant 1 are presented for the 8-kilometer distance that was completed during the study. All other run times are presented for a 5-kilometer distance				

Chapter 3: Results

3.1 Results

The biomechanical effects of reduced muscular strength on distance runners with planus feet were investigated. Using a significance level of $\alpha = 0.05$, $\beta = 0.20$ for all statistical comparisons, individual factorial ANOVAs were performed for mean peak stance phase ankle eversion and mean peak stance phase knee adduction with a Bonferroni correction to the alpha values (SPSS version 21.0: SPSS Inc., Chicago, IL). The independent variables in each analysis were side (dominant and non-dominant) and time (pre-run and post-run). Pearson's product-moment correlation coefficient was used to analyze pre- versus post-run stride-to-stride variability and a 2 x 3 x 2 MANOVA was used to assess movement variability. Variability analysis focused on frontal plane movements at the ankle and knee with side (dominant and non-dominant), portion of stance phase (initial, middle and finals thirds) and time (pre-run and post-run) as independent variables. The null hypothesis that both pre- and post-run group mean scores and pre- and post-run correlation coefficients and movement variability were equal was tested. The alternative hypothesis was that post-run kinematics, stride-to-stride variability and movement variability would statistically significantly differ from pre-run measures.

Descriptive statistics for each dependent variable are listed in Table 3.1. Post-run testing revealed an average reduction in dominant leg inversion strength of 33.8% and an average reduction of non-dominant leg inversion of 28.8% in comparison to pre-run testing. See Table 3.2 for participant strength recordings. Having met the assumptions of an ANOVA, no statistically significant interaction effect was obtained between side and time ($p > 0.05$) for ankle eversion. There was however a significant effect for side [F (1)= 9.996, p = 0.003]. Similarly, factorial ANOVA analysis revealed no statistically significant interaction effect between side and time ($p > 0.05$) for knee adduction, but did show a significant effect for side [F (1)= 15.492, p = 0.000].

Individual stride-to-stride correlation coefficients were calculated and averaged across participants for dominant and non-dominant legs at the pre- and post-run testing periods for stance phase ankle eversion and stance phase knee adduction across all participants. The dominant leg pre-run Pearson product-moment correlation coefficient for stance phase knee adduction was $r = 0.91$, 95% CI [0.90, 0.92] compared to a post-run value of $r = 0.81$, 95% CI [0.78, 0.86]. The non-dominant leg stride correlation coefficient for stance phase knee adduction was $r = 0.87$, 95% CI [0.83, 0.91] for pre-run testing and $r = 0.80$, 95% CI [0.76, 0.84] for post-run. Focusing on ankle eversion, the dominant leg pre-run correlation coefficient was $r = 0.95$, 95% CI [0.94, 0.97] in comparison to a post-run $r = 0.95$, 95% CI [0.93, 0.96], where as non-dominant ankle eversion values revealed correlation coefficients of $r= 0.94$, 95% CI [0.93, 0.95] and $r = 0.92$, 95% CI [0.90, 0.94] for pre- and post-run tests, respectively. Box and whisker plots can be found in Figures 3.1 through 3.4.

Frontal plane stance phase ankle and knee variability was studied using a 2 x 3 x 2 MANOVA with side (dominant and non-dominant), percent stance (10%, 50% and 90%) and time (pre-run and post-run) as independent variables. Using the Wilk's Lambda distribution, the multivariate interaction effect for side*percent stance*time approached significance [$F(4,54) = 2.427$, $p = 0.059$, $\eta^2= 0.152$]. Although the interaction effect did not reach statistical significance ($\alpha = 0.05$), "surely, God loves .06 nearly as much as the .05" (Rosnow & Rosenthal, 1990) justifying the interpretation of the side*percent stance*time univariate analysis. Univariate analysis using the Greenhouse-Geisser epsilon adjustment revealed a statistically significant effect for eversion [$F(1.397) = 4.562$, $p = 0.035$, $\eta^2= 0.246$].

To assess which pre-run percent of stance phase statistically significantly differed from its respective post-run percent of stance phase, individual 3 x 2 (percent stance by time) factorial ANOVAs were performed for dominant and non-dominant ankle eversion. A statistically significant percent stance*time interaction was obtained [$F(2) = 3.511$, $p = 0.034$, $\eta^2= 0.077$]. Using pairwise comparisons of estimated marginal means for post-hoc analysis, 10% stance statistically significantly differed between pre-run and post-run times for dominant ankle eversion ($p = 0.030$, $\eta^2 = 0.055$), but did not for 50% and 90% ($p > 0.05$). No statistically significant percent stance*time interactions were present for non-dominant ankle eversion ($p > 0.05$).

Table 3.1 Means (and standard deviations), separated by time and side (run as separate factorial ANCOVAs)

	Peak Stance Ankle Eversion (degrees)		Peak Stance Knee Adduction (degrees)	
	Dominant	**Non-Dominant**	**Dominant**	**Non-Dominant**
Pre-Run	8.3 (2.8)	6.4 (2.6)	1.7 (2.7)	-2.2 (5.1)
Post-Run	8.0 (3.6)	5.2 (2.6)	1.8 (3.6)	-1.9 (3.8)

Table 3.2 Individual participant inversion strength, separated by side and time

Participant	Ankle Inversion Strength (lbs)			
	Dominant Leg		**Non-Dominant Leg**	
	Pre-Run	**Post-Run**	**Pre-Run**	**Post-Run**
1	12.9	8.7	11.4	8
2	12	9.7	11.3	9.1
3	14.9	10.4	13.8	9.9
4	11.1	10.1	11	10
5	9.6	5.3	8.9	6.6
6	13.4	11.7	12.7	9
7	7.7	7.7	7.1	6.5
8	14.1	10.9	16.9	13.2
9	8	6.2	8.4	6.9
10	23.8	10.2	26.1	13.3
11	12.5	9.9	11	8.7
12	18.8	13.8	19.4	14.7
13	12.9	6.3	12.1	7.4
14	20.75	16.7	20	18
15	24.5	16.7	19.9	15.8
Mean	14.46	10.29	14	10.47
% Reduction	33.75		28.82	

Table 3.3 Portion by time movement variability for dominant ankle eversion

Dependent Variable: Dominant Ankle Eversion

Portion	Time	Mean	Std. Deviation	N
10%	Pre	1.520	.8055	15
	Post	1.147	.4518	15
	Total	1.333	.6692	30
50%	Pre	.693	.1163	15
	Post	.953	.1885	15
	Total	.823	.2029	30
90%	Pre	1.193	.4788	15
	Post	1.140	.3924	15
	Total	1.167	.4310	30
Total	Pre	1.136	.6339	45
	Post	1.080	.3653	45
	Total	1.108	.5152	90

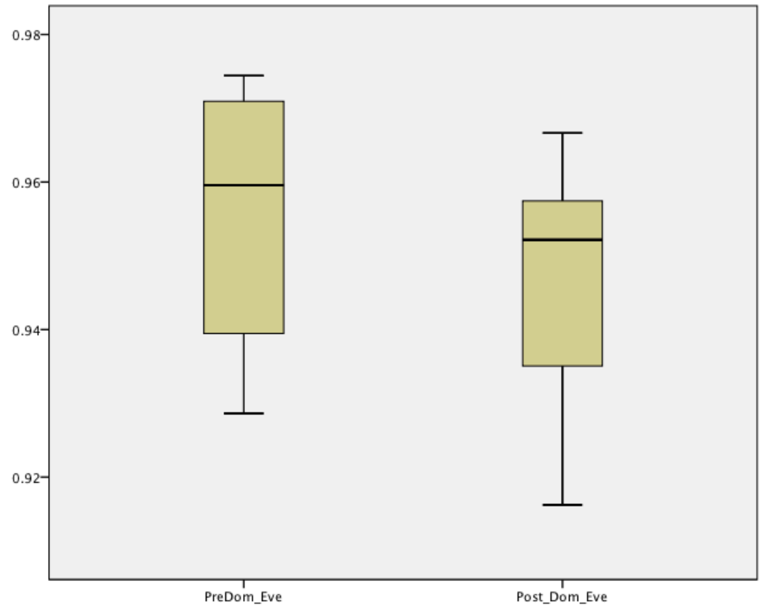

Figure 3.1 Box and whisker plot for dominant ankle eversion stride-to-stride correlation

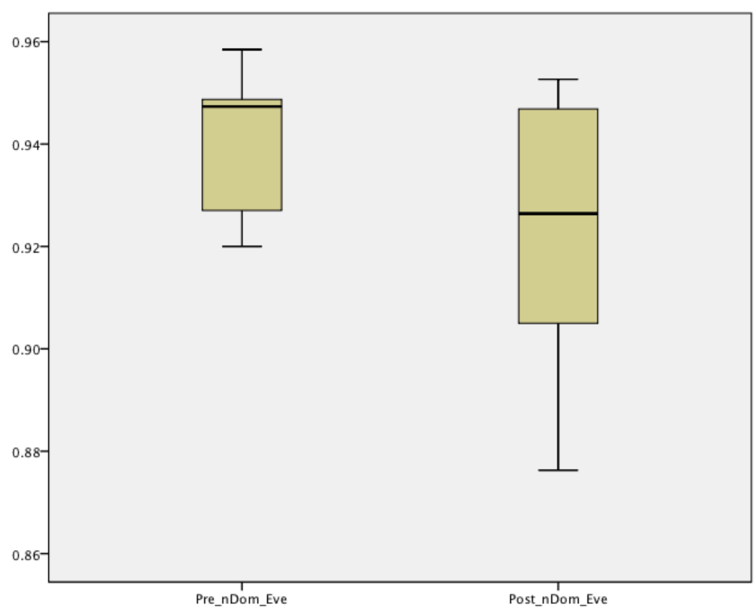

Figure 3.2 Box and whisker plot for non-dominant ankle eversion stride-to-stride correlation

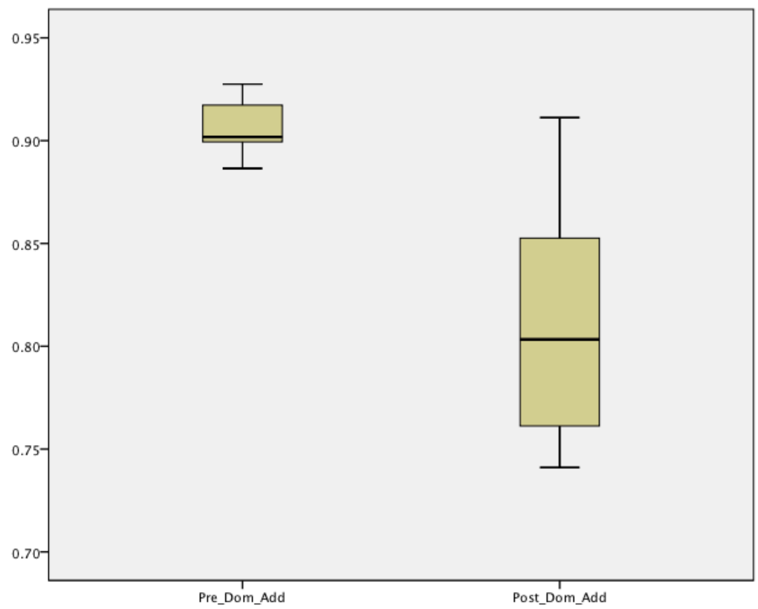

Figure 3.3 Box and whisker plot for dominant knee adduction stride-to-stride correlation

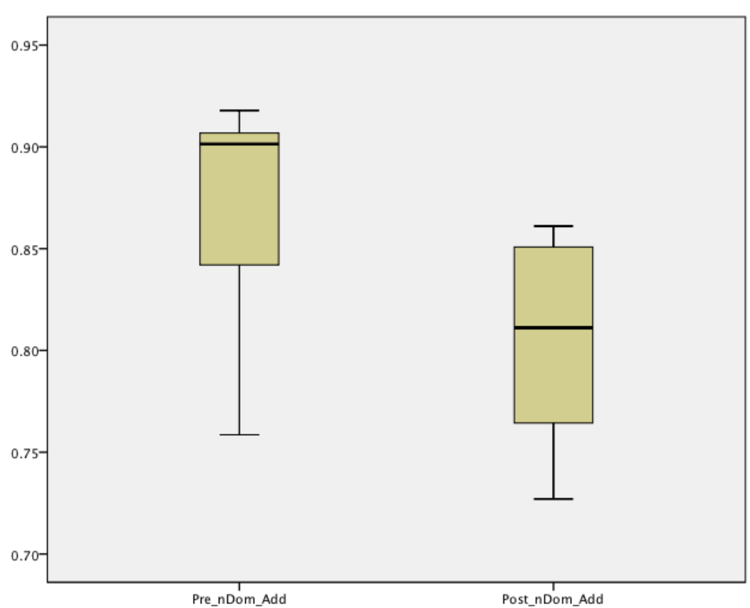

Figure 3.4 Box and whisker plot for non-dominant knee adduction stride-to-stride correlation

Chapter 4: Discussion

4.1 Kinematics

Despite the high prevalence of repetitive stress injuries in the running population, conclusive evidence is lacking regarding the identification of risk factors specific to foot posture in relation to reduced muscular strength. The goals of this study were to investigate the effects of reduced muscular strength on peak knee and peak ankle frontal plane kinematics and variability. To study the aforementioned, a convenience sample of experienced distance runners completed two treadmill runs while lower extremity kinematics were recorded. One recorded treadmill run was completed before, and another immediately after an outdoor run on a 400-meter synthetic, rubber composite track at a self-selected pace. Any distance between 5 and 8 kilometers was permitted at the participant's discretion. Participants were encouraged to complete the run at a pace to match or better their fastest run time at the individual distance selected.

Focusing on movements at the ankle, one of the primary muscles contributing to ankle inversion is the tibialis posterior (Magee, 2008). During gait, the tibialis posterior muscle functions to control the magnitude and rate of pronation (Perry & Burnfield, 2010). Due to the fact that rearfoot foot eversion is one of the primary components of pronation and is also one of the key characteristics of the planus foot (D. S. Williams, 3rd et al., 2001), inversion strength was of primary interest in the population under investigation. Inversion muscular strength was recorded using a hand-held dynamometer before and after the out-door run to determine the degree of muscular fatigue.

Inversion strength recordings suggest that the procedure induced fatigue through a reduction in muscular strength. A 33.8% average reduction in dominant leg inversion muscular strength and a 28.82% average reduction in non-dominant leg inversion muscular strength were obtained between pre- and post-run testing sessions. Previous authors suggest that muscular weakness may play a primary role in the development of overuse injuries (Ferber et al., 2009; Ferber et al., 2011; Ireland, Willson, Ballantyne, & Davis, 2003; Robinson & Nee, 2007). In their assessment of patellofemoral pain syndrome, Ferber et al. (2011) identified that participants positive for patellofemoral pain had a 28.7% reduction in hip-abductor muscular strength at baseline in comparison to the pain free control group. The authors suggested such a reduction in muscular strength may play a role in the development of patellofemoral pain syndrome (Ferber et al., 2011). Further, in their investigation of fatigue-induced changes in knee kinematics, kinetics and stiffness, Longpre et al. (2013) utilized a 25% reduction in peak isometric torque to classify fatigue. Modeling the significance of the muscular strength reductions found by Ferber et al., reductions in inversion muscular strength of 33.8% and 28.82% were deemed large enough to potentially influence lower extremity kinematics.

In the present study, individual 2 x 2 factorial ANCOVAs were used to assess differences in knee adduction and ankle eversion with time (pre- and post-run) and side (dominant and non-dominant legs) as independent variables in each analysis. In contrast with the hypotheses, no significant multivariate effects were obtained for the interaction between time and side ($p > 0.05$).or for time ($p > 0.05$). There was however a significant multivariate effect for side when looking at both ankle eversion [F (1)= 9.996, p = 0.003] and knee adduction [F (1)= 15.492, p = 0.000] suggesting that ankle and knee kinematics of the dominant and non-dominant legs did differ. These findings are not associated with time and can therefore not be related to running induced reductions in muscular strength.

Despite the presence of current literature to support the original hypotheses of this study (Clansey et al., 2012; Dierks et al., 2010; Koblbauer et al., 2013), there is a body of research supporting insignificant kinematic changes in relation to reductions in muscular strength. In a study performed by Pohl, Rabbito and Ferber (2010), the researchers reported no statistically significant differences in rearfoot eversion following a tibialis posterior fatigue protocol. The authors suggest alterations to the force output of the tibialis posterior muscle did not alter rearfoot motions during gait. Similarly Abt et al. (2011) report no changes in peak magnitude or time to peak magnitude for rearfoot eversion or knee flexion following an exhaustive run (Abt et al., 2011). Abt et al. state that further research is required to investigate fatigue induced kinematic changes to determine the threshold at which kinematic changes may occur (Abt et al., 2011).

Pohl et al. (2010) did however report a statistically significant increase in rearfoot eversion excursion (0.7°, $p < 0.05$), but the authors question the clinical significance of this degree of reduction as it was smaller than the precision error of a within-day gait analysis (0.9°) (Pohl, Rabbito, & Ferber, 2010).

Similar to the present study, irrespective of foot type, Kobelbauer et al. (2013) investigated running induced fatigue in relation to rearfoot eversion. Interestingly, the researchers reported a statistically significant increase in non-dominant rearfoot eversion (1.6°, $p < 0.05$) but no change for dominant rearfoot eversion (1.0°, $p > 0.05$) (Koblbauer et al., 2013). Despite statistical significance, the clinical significance of the degree of movement reported by Koblbauer et al. is in question. With 13° being cited as the point at which rearfoot eversion becomes excessive (Michaud, 1997), the total degrees of motion that the rearfoot is subject to is an important factor. As such, a change from 6° of rearfoot eversion to 7.6° of rearfoot eversion would still be classified within a "normal" range of rearfoot eversion. Koblbauer et al. do however report a change from 11.5° (standard deviation of 3.9°) of rearfoot eversion pre-fatigue to 13.1° (standard deviation of 4.6°) of rearfoot eversion. While the mean value does cross the 13° threshold reported by Michaud (1997), the large standard deviation suggests that a portion of their 17 participant sample was everting less than 13.1°. Additionally, Koblbauer et al. did not record or report the precision error found within their kinematic analysis. Reflecting on the 0.9° root-mean-square error reported by Pohl et al. (2010), the 1.6° increase in non-dominant rearfoot eversion may warrant little clinical significance.

Building on the suggestion of Abt et al. (2011), further research is required to determine if there is a level of muscular and/ or cardiovascular fatigue that may serve as the threshold upon which kinematic changes may occur. Further, the differences need to parsed by foot type in attempt to identify injury risk factors inherent to each foot posture. Clansey et al. (2012) indicate that the inconsistent kinematic findings reported in the current literature may be attributed to the type of activity and level of fatigue used in each of the studies (Clansey et al., 2012). For example, the procedure in this study focused specifically on running induced muscular fatigue in a sample of experienced distance runners. Other studies have used different fatigue protocols and kinematic recording apparatus and procedure (Clansey et al., 2012; Cortes et al., 2013; Koblbauer et al., 2013; Longpre et al., 2013.; Mizrahi et al., 2000). In absence of true gold standard to record joint and segment kinematics, procedural variations are abundant.

4.2 Variability

The presence of stride-to-stride variability has been identified as an important factor in the stability, flexibility and coordination of movement (Hamill et al., 1999; Kelso, 1997). It remains to be seen, however, how much or little movement variability is beneficial, and at which point variability may become harmful. If each stride were identical, noted by the absence of variability, responding to environmental changes (uneven terrain) would be difficult. Contrarily, if each stride were completely different than the previous, noted by high variability, the coordination of movement would be highly unpredictable. With this in mind, an acceptable range of variability must exist, albeit allusive at this point. Of particular interest in this report are the differences in stride-to-stride stance phase correlation and the differences between pre- and post-run stance phase variability.

Assessing the correlation of each stride with the subsequent stride, across 10 gait trials, stride-to-stride variability increased for frontal plane knee mechanics. Pre-run assessment indicated a stride-to-stride Pearson's product-moment correlation coefficient of $r = 0.91$, 95% CI [0.90, 0.91] for the dominant side and $r = 0.87$, 95% CI [0.84, 0.91] for the non-dominant side, whereas post-run correlations decreased to $r = 0.81$, 95% CI [0.77, 0.85] for the dominant knee and $r = 0.80$ for the non-dominant knee, 95% CI [0.77, 0.84]. While $r = 0.81$ and $r = 0.80$ can all still be considered strong correlations (Cohen, 1988), the reductions are notable, particularly for the dominant knee which is evident in the differences in confidence intervals between pre- and post-run analysis. For this reason, variability was further investigated to determine which portions of stance phase were becoming more variable.

Changes in frontal plane stance phase ankle and knee variability were assessed using a 2 x 3 x 2 MANOVA. Although the three-way interaction was not statistically significance ($p = 0.059$) univariate analysis was justified in that "surely, God loves .06 nearly as much as the .05" (Rosnow & Rosenthal, 1990). Univariate analysis suggested that differences were present for ankle eversion.

Assessing dominant ankle eversion, it is evident that the largest differences in pre- and post-run standard deviations took place at 10% stance ($p = 0.030$, $\eta^2 = 0.055$)

suggesting a statistically significant difference in variability during this portion of stance. Assessing mean scores (Table 3.3), it is apparent dominant ankle frontal plane variability decreases at the post run testing period

The observed differences in movement variability during the initial portion of stance phase is consistent with that observed by Hamill et al. (1999). Hamill et al. report that both their pain-free and patellofemoral pain groups experienced the largest stance phase differences in segment coupling variability from the time the heel made contact with the ground until pronation occurred (Hamill et al., 1999). During this portion of stance phase the foot is making initial contact with the ground, followed by the mitigation of impact forces during the loading response (Perry & Burnfield, 2010). Increasing variability during either of these events, initial contact or loading response, may increase the risk of injury. Lieberman et al. (2010) report that the moment the foot hits the ground may be the most injurious portion of the gait cycle while running. As Lieberman et als. findings suggest, the impact forces associated with the abrupt collision of the heel with the ground (the first 50 milliseconds of stance phase) are approximately 1.5-3 times body weight (Lieberman et al., 2010).

With the increased impact forces, in addition to the increased stride rate experienced during running, it is evident that the initial third of stance phase is a critical portion of the gait cycle. Changes in movement variability during this portion of stance may have important clinical implications. Foot care specialists, physiotherapists and running coaches often prescribe footwear and training programs for the running population based on clinically oriented gait analyses. These brief encounters are often accomplished while the subject is walking or running in a non-fatigued state and are often centered on the degree and rate of movement that takes place at and immediately following initial contact with the ground. If movement variability at the ankle is impacted by reductions in muscular strength, the training program or prescribed footwear may not be appropriate while running in a fatigued state. For this reason, it may be beneficial for the practitioner to perform clinical gait analyses while the runner is in a fatigued state prior to making any evidence based decisions. This may also prove beneficial to the physical therapy world, through the early identification of potential injury risk factors. More specifically, the application of pre-habilitation training programs targeted at reducing changes in movement variability may have a significant influence on injury rates in distance runners.

Building on the clinical relevance of the current findings, comparison must be made to the work of Hamill et al. (1999). Hamill et al. reported decreases in lower extremity segment coupling variability in their injured sample. While the symptomatic group did demonstrate a reduction in movement variability, albeit using a different method than the current study, the researchers claim that this did not indicate a reason for the injury. Hamill et al. state that the observed reductions in segment coupling variability may be an indication of a non-healthy state but that this information does not provide any indication to the cause of the injury (Hamill et al., 1999). Hamill et al. claim that the task of running with an injury may be performed by using segment actions that are repeatable within a very narrow range, allowing the accomplishment of the task with minimal pain (Hamill et al., 1999). Further, the researchers suggest that while reduced variability may permit pain free gait, the resulting movement patterns may result in constant soft-tissue stress potentially exposing the injured runner to long-term degenerative changes (Hamill et al., 1999).

Focusing on movement variability and the occurrence of acute injury, Drewes et al. (2009) suggest that altered ankle kinematics and joint coupling during late swing phase may be predispose to inversion ankle sprains at initial contact (Drewes et al., 2009). As frontal plane ankle movement variability increases, the foot is making initial contact with the ground in a progressively altered position with each stride. The larger the increases, the less predictable foot placement at initial contact becomes. Without proper accommodations, the prevalence of injuries, such as inversion ankle sprains, may increase. Further, if the muscles in the shank are spending increased time adjusting to differences in foot placement, less time may be allotted to the loading response, in turn, potentially impairing the bodies ability to respond to the impact of initial heel contact.

In the eyes of a sports medicine clinician, someone who consistently assesses and treats injured athletes, the differences in movement variability between injured and uninjured patients must be considered. If repetitive movements, such as running, are accomplished with too little variability, as may be the case with an inured athlete, this may have long-standing degenerative implications. Contrarily, if repetitive movement becomes too variable, as may be the case with advanced states of running-induced fatigue, the potential for acute injury, such as inversion ankle sprains, may increase. For these reasons, movement variability is an area that warrants further investigation.

Much of the previous literature investigating movement variability has focused on segment coupling (Drewes et al., 2009; Hamill et al., 1999; McKeon et al., 2009; Miller et al., 2008). Building on the work of Ferber et al. (2011), this study is one of few that focuses attention on the movement variability within a single joint and aside from Miller et al. (2008), is one of the first to investigate which portions of the gait cycle are subject to fatigue induced changes in variability. With the presence of literature suggesting that variability may be associated with injury, it is strongly recommended that further research be conducted investigating the effects that reduced muscular strength may have on movement variability.

4.3 Limitations

Inherent to all forms of research, the current report has its limitations. To start, the sample was limited to healthy participants. It has been suggested that injured runners have altered kinematic patterns (Hamill et al., 1999) limiting the applicability of this research to that population. Further, distance running requires a consistent repetitive motion. It is possible that other types of athletes or non-runners may have altered kinematic patterns also limiting the applicability of this research beyond the bounds of the distance running population. Specific to the participants in this study, it may also have been possible that reductions in muscular strength were not large enough to alter kinematics. Further investigation into the effects of advanced states of fatigue is recommended.

In addition to the limitations within the sample under investigation, the gait analysis procedure used to record knee and ankle kinematics also has its short falls. The markers used to identify rearfoot motion consisted of two markers set parallel to the running surface. A neutral rearfoot position was identified as a perpendicular bisection of the markers and the running surface. Frontal plane motion was likened to movement in either direction away from neutral with eversion being movement toward the mid-line of the body and inversion being movement away from the mid-line of the body. Eversion occurs about the subtalar joint, which is a tri-planer joint meaning the movements at this joint are far more complex than the single plane side-to-side movements of the two rearfoot markers. Kinematic analyses using a multi-segmental foot models, such as the Oxford Foot Model, are showing that foot mechanics are far more complex than we currently understand. The use of multi-segmental foot model in a study similar to this would permit further and more accurate movements within the foot to be assessed.

Building on the markers used to record lower extremity kinematics, it is also possible that error may have been present due to marker placement. The procedure used in this study required that the markers be removed between the pre- and post-run analysis while the participant completed the outdoor run portion of the study. Although all marker placements were performed by the same investigator (AF), who is also a trained in anatomy of the lower extremity as a Canadian Certified Pedorthist and a Registered Kinesiologist, in addition to using athletic tape and pen markings to identify pre-run marker placements in attempt replicate identical placement for each recording, it is possible that markers may have been in slightly altered positions.

A final limitation took place with one of the participant's kinematic recording trials. The participant requested that the run speed be slowed from a pre-run testing speed of 2.2 meters per second to 2.0 meters per second during post-run testing due to feelings of fatigue induced unease. Although the minimal reduction in speed was deemed acceptable, it is possible that the change 0.2 meters per second may have resulted in altered knee or ankle kinematics.

4.4 Future Directives

Due to the inconsistency reported in the literature regarding fatigue induced kinematic changes to gait, future research should endeavor to further investigate this matter. More specifically, attempting to identify the degree of muscular fatigue at which kinematic changes take place. Future research should also consider alternate foot postures. It is highly possible that populations with pes cavus or pes rectus foot types may demonstrate different results than reported in this study. Additionally, other considerations should be taken into account when classifying foot posture. Further investigation should examine resting standing rearfoot frontal plane angle, in addition to the available range of frontal plane rearfoot motion. Due to the fact that the pes planus population has an everted resting rearfoot posture, it is possible that the participants in this study were already reaching their maximum range of rearfoot eversion, even while running in a non-fatigued state.

This study was successful in identifying fatigue-induced changes in movement variability. Due to the fact that movement variability is a relative young method used to investigate gait, further research is warranted. Future studies should assess differences in alternate activities, as well as further investigation into pathological and diseased population. The identification of an amount, or range of movement variability that is beneficial to gait would be applicable to both athletic and pathological populations.

4.5 Conclusions

Reduced muscular strength and its implications on gait kinematics may in fact differ from some of what the current literature suggests. What can be gleaned from some of the current research is that there are many factors that may implicate gait kinematics and that muscular strength may just be a small piece of a larger puzzle. What is readily evident from this report is that reduced muscular strength does affect movement variability. Whether the observed changes in movement variability are harmful to the individual remains to be seen. Future research is required to fully understand the implications that increased movement variability may have.

Bibliography

Abt, J. P., Sell, T. C., Chu, Y., Lovalekar, M., Burdett, R. G., & Lephart, S. M. (2011). Running kinematics and shock absorption do not change after brief exhaustive running. *J Strength Cond Res, 25*(6), 1479-1485. doi: 10.1519/JSC.0b013e3181ddfcf8

Barnes, A., Wheat, J., & Milner, C. E. (2011). Fore- and rearfoot kinematics in high- and low-arched individuals during running. *Foot Ankle Int, 32*(7), 710-716.

Beattie, P., Isaacson, K., Riddle, D. L., & Rothstein, J. M. (1990). Validity of derived measurements of leg-length differences obtained by use of a tape measure. *Phys Ther, 70*(3), 150-157.

Butler, R. J., Hillstrom, H., Song, J., Richards, C. J., & Davis, I. S. (2008). Arch height index measurement system: establishment of reliability and normative values. *J Am Podiatr Med Assoc, 98*(2), 102-106. doi: 98/2/102 [pii]

Clansey, A. C., Hanlon, M., Wallace, E. S., & Lake, M. J. (2012). Effects of fatigue on running mechanics associated with tibial stress fracture risk. *Med Sci Sports Exerc, 44*(10), 1917-1923. doi: 10.1249/MSS.0b013e318259480d

Close, J. R., Inman, V. T., Poor, P. M., & Todd, F. N. (1967). The function of the subtalar joint. *Clin Orthop Relat Res, 50*, 159-179.

Cohen, J. (1988). *Statistical Power Analysis for the Behavioral Sciences* (Second ed.). Hillsdale, New Jersey: Lawrence Erlbaum Associates.

Collins, T. D., Ghoussayni, S. N., Ewins, D. J., & Kent, J. A. (2009). A six degrees-of-freedom marker set for gait analysis: repeatability and comparison with a modified Helen Hayes set. *Gait Posture, 30*(2), 173-180. doi: 10.1016/j.gaitpost.2009.04.004

S0966-6362(09)00110-6 [pii]

Cortes, N., Greska, E., Kollock, R., Ambegaonkar, J., & Onate, J. A. (2013). Changes in lower extremity biomechanics due to a short-term fatigue protocol. *J Athl Train, 48*(3), 306-313. doi: 10.4085/1062-6050-48.2.03

DeLeo, A. T., Dierks, T. A., Ferber, R., & Davis, I. S. (2004). Lower extremity joint coupling during running: a current update. *Clin Biomech (Bristol, Avon), 19*(10), 983-991. doi: 10.1016/j.clinbiomech.2004.07.005

Derrick, T. R., Bates, B. T., & Dufek, J. S. (1994). Evaluation of time-series data sets using the Pearson product-moment correlation coefficient. *Med Sci Sports Exerc, 26*(7), 919-928.

Dierks, T. A., Davis, I. S., & Hamill, J. (2010). The effects of running in an exerted state on lower extremity kinematics and joint timing. *J Biomech, 43*(15), 2993-2998. doi: 10.1016/j.jbiomech.2010.07.001

S0021-9290(10)00339-8 [pii]

Drewes, L. K., McKeon, P. O., Paolini, G., Riley, P., Kerrigan, D. C., Ingersoll, C. D., & Hertel, J. (2009). Altered ankle kinematics and shank-rear-foot coupling in those with chronic ankle instability. *J Sport Rehabil, 18*(3), 375-388.

Ferber, R., Hreljac, A., & Kendall, K. D. (2009). Suspected mechanisms in the cause of overuse running injuries: a clinical review. *Sports Health, 1*(3), 242-246. doi: 10.1177/1941738109334272

10.1177_1941738109334272 [pii]

Ferber, R., Kendall, K. D., & Farr, L. (2011). Changes in knee biomechanics after a hip-abductor strengthening protocol for runners with patellofemoral pain syndrome. *J Athl Train, 46*(2), 142-149. doi: 10.4085/1062-6050-46.2.142

Friberg, O., Nurminen, M., Korhonen, K., Soininen, E., & Manttari, T. (1988). Accuracy and precision of clinical estimation of leg length inequality and lumbar scoliosis: comparison of clinical and radiological measurements. *Int Disabil Stud, 10*(2), 49-53.

Gehring, D., Mornieux, G., Fleischmann, J., & Gollhofer, A. (2013). Knee and Hip Joint Biomechanics are Gender-specific in Runners with High Running Mileage. *Int J Sports Med*. doi: 10.1055/s-0033-1343406

Gerdle, B., Elert, J., & Henriksson-Larsen, K. (1989). Muscular fatigue during repeated isokinetic shoulder forward flexions in young females. *Eur J Appl Physiol Occup Physiol, 58*(6), 666-673.

Glace, B. W., McHugh, M. P., & Gleim, G. W. (1998). Effects of a 2-hour run on metabolic economy and lower extremity strength in men and women. *J Orthop Sports Phys Ther, 27*(3), 189-196. doi: 10.2519/jospt.1998.27.3.189

Hamill, J., van Emmerik, R. E., Heiderscheit, B. C., & Li, L. (1999). A dynamical systems approach to lower extremity running injuries. *Clin Biomech (Bristol, Avon), 14*(5), 297-308.

Hausdorff, J. M., Rios, D. A., & Edelberg, H. K. (2001). Gait variability and fall risk in community-living older adults: a 1-year prospective study. *Arch Phys Med Rehabil, 82*(8), 1050-1056. doi: 10.1053/apmr.2001.24893

Holmes, J. C., Pruitt, A. L., & Whalen, N. J. (1993). Iliotibial band syndrome in cyclists. *Am J Sports Med, 21*(3), 419-424.

Inman, V. T. (1976). *The Joints of the Ankle*. Baltimore: Williams & Wilkins.

Ireland, M. L., Willson, J. D., Ballantyne, B. T., & Davis, I. M. (2003). Hip strength in females with and without patellofemoral pain. *J Orthop Sports Phys Ther, 33*(11), 671-676. doi: 10.2519/jospt.2003.33.11.671

Iridiastadi, H., & Nussbaum, M. A. (2006). Muscular fatigue and endurance during intermittent static efforts: effects of contraction level, duty cycle, and cycle time. *Hum Factors, 48*(4), 710-720.

Jacobs, S. J., & Berson, B. L. (1986). Injuries to runners: a study of entrants to a 10,000 meter race. *Am J Sports Med, 14*(2), 151-155.

Johnston, C. A., Taunton, J. E., Lloyd-Smith, D. R., & McKenzie, D. C. (2003). Preventing running injuries. Practical approach for family doctors. *Can Fam Physician, 49*, 1101-1109.

Kelln, B. M., McKeon, P. O., Gontkof, L. M., & Hertel, J. (2008). Hand-held dynamometry: reliability of lower extremity muscle testing in healthy, physically active,young adults. *J Sport Rehabil, 17*(2), 160-170.

Kelso, S. (1997). *Dynamic Patterns: The Self-Organization of Brain and Behavior*. Cambridge, Massachusetts: The MIT Press.

Koblbauer, I. F., van Schooten, K. S., Verhagen, E. A., & van Dieen, J. H. (2013). Kinematic changes during running-induced fatigue and relations with core endurance in novice runners. *J Sci Med Sport*. doi: S1440-2440(13)00139-4 [pii]

10.1016/j.jsams.2013.05.013

Koplan, J. P., Powell, K. E., Sikes, R. K., Shirley, R. W., & Campbell, C. C. (1982). An epidemiologic study of the benefits and risks of running. *JAMA, 248*(23), 3118-3121.

Larson, P., Higgins, E., Kaminski, J., Decker, T., Preble, J., Lyons, D., . . . Normile, A. (2011). Foot strike patterns of recreational and sub-elite runners in a long-distance road race. *J Sports Sci, 29*(15), 1665-1673. doi: 10.1080/02640414.2011.610347

Lees, A., Lake, M., & Klenerman, L. (2005). Shock absorption during forefoot running and its relationship to medial longitudinal arch height. *Foot Ankle Int, 26*(12), 1081-1088. doi: 909331 [pii]

Leigh, R. J., Pohl, M. B., & Ferber, R. (2013). Does tester experience influence the reliability with which 3D gait kinematics are collected in healthy adults? *Phys Ther Sport*. doi: S1466-853X(13)00039-4 [pii]

10.1016/j.ptsp.2013.04.003

Lieberman, D. E., Venkadesan, M., Werbel, W. A., Daoud, A. I., D'Andrea, S., Davis, I. S., . . . Pitsiladis, Y. (2010). Foot strike patterns and collision forces in habitually barefoot versus shod runners. *Nature, 463*(7280), 531-535. doi: 10.1038/nature08723

nature08723 [pii]

Longpre, H. S., Potvin, J. R., & Maly, M. R. (2013). Biomechanical changes at the knee after lower limb fatigue in healthy young women. *Clin Biomech (Bristol, Avon), 28*(4), 441-447. doi: 10.1016/j.clinbiomech.2013.02.010

S0268-0033(13)00026-0 [pii]

Lysholm, J., & Wiklander, J. (1987). Injuries in runners. *Am J Sports Med, 15*(2), 168-171.

Magee, D. (2008). *Orthopedic Physical Assessment* (5 ed.). St. Louis, Missouri: Elsevier Inc.

Marti, B., Vader, J. P., Minder, C. E., & Abelin, T. (1988). On the epidemiology of running injuries. The 1984 Bern Grand-Prix study. *Am J Sports Med, 16*(3), 285-294.

McKeon, P. O., Paolini, G., Ingersoll, C. D., Kerrigan, D. C., Saliba, E. N., Bennett, B. C., & Hertel, J. (2009). Effects of balance training on gait parameters in patients with chronic ankle instability: a randomized controlled trial. *Clin Rehabil, 23*(7), 609-621. doi: 10.1177/0269215509102954

Michaud, T. (1997). *Foot Orthoses and Other Forms of Conservative Foot Care*. Newton, Massachusetts: Thomas C. Michaud.

Miller, R. H., Meardon, S. A., Derrick, T. R., & Gillette, J. C. (2008). Continuous relative phase variability during an exhaustive run in runners with a history of iliotibial band syndrome. *J Appl Biomech, 24*(3), 262-270.

Milner, C. E., Ferber, R., Pollard, C. D., Hamill, J., & Davis, I. S. (2006). Biomechanical factors associated with tibial stress fracture in female runners. *Med Sci Sports Exerc, 38*(2), 323-328. doi: 10.1249/01.mss.0000183477.75808.92

00005768-200602000-00019 [pii]

Mizrahi, J., Verbitsky, O., & Isakov, E. (2000). Fatigue-related loading imbalance on the shank in running: a possible factor in stress fractures. *Ann Biomed Eng, 28*(4), 463-469.

Murdock, G. H., & Hubley-Kozey, C. L. (2012). Effect of a high intensity quadriceps fatigue protocol on knee joint mechanics and muscle activation during gait in young adults. *Eur J Appl Physiol, 112*(2), 439-449. doi: 10.1007/s00421-011-1990-4

Nawoczenski, D. A., Saltzman, C. L., & Cook, T. M. (1998). The effect of foot structure on the three-dimensional kinematic coupling behavior of the leg and rear foot. *Phys Ther, 78*(4), 404-416.

Ounpuu, S. (1990). The biomechanics of running: a kinematic and kinetic analysis. *Instr Course Lect, 39*, 305-318.

Perry, G., & Burnfield, J. M. (2010). *Gait Analysis: Normal and Pathological Function* (2nd ed.). Thorofare, NJ: SLACK Incorporated

Pierrynowski, M. R., Finstad, E., Kemecsey, M., & Simpson, J. (2003). Relationship between the subtalar joint inclination angle and the location of lower-extremity injuries. *J Am Podiatr Med Assoc, 93*(6), 481-484.

Pohl, M. B., Rabbito, M., & Ferber, R. (2010). The role of tibialis posterior fatigue on foot kinematics during walking. *J Foot Ankle Res, 3*, 6. doi: 10.1186/1757-1146-3-6

Robinson, R. L., & Nee, R. J. (2007). Analysis of hip strength in females seeking physical therapy treatment for unilateral patellofemoral pain syndrome. *J Orthop Sports Phys Ther, 37*(5), 232-238. doi: 10.2519/jospt.2007.2439

Rosnow, R., & Rosenthal, R. (1990). Statistical procedures and the justification of knowledge in physchological science. *American Psychologist, 44*, 1276-1284.

Segal, A. D., Orendurff, M. S., Klute, G. K., McDowell, M. L., Pecoraro, J. A., Shofer, J., & Czerniecki, J. M. (2006). Kinematic and kinetic comparisons of transfemoral amputee gait using C-Leg and Mauch SNS prosthetic knees. *J Rehabil Res Dev, 43*(7), 857-870.

van Gent, R. N., Siem, D., van Middelkoop, M., van Os, A. G., Bierma-Zeinstra, S. M., & Koes, B. W. (2007). Incidence and determinants of lower extremity running injuries in long distance runners: a systematic review. *Br J Sports Med, 41*(8), 469-480; discussion 480. doi: bjsm.2006.033548 [pii]

10.1136/bjsm.2006.033548

Walter, S. D., Hart, L. E., McIntosh, J. M., & Sutton, J. R. (1989). The Ontario cohort study of running-related injuries. *Arch Intern Med, 149*(11), 2561-2564.

Williams, D. S., 3rd, McClay, I. S., & Hamill, J. (2001). Arch structure and injury patterns in runners. *Clin Biomech (Bristol, Avon), 16*(4), 341-347. doi: S0268003301000055 [pii]

Williams, D. S., & McClay, I. S. (2000). Measurements used to characterize the foot and the medial longitudinal arch: reliability and validity. *Phys Ther, 80*(9), 864-871.

Appendix A: Ethics Approval Forms

Western Research | Research Ethics

Use of Human Participants - Ethics Approval Notice

Principal Investigator: Dr. Colin Dombroski
File Number:103376
Review Level: Delegated
Approved Local Adult Participants:20
Approved Local Minor Participants:0
Protocol Title: Effects of Muscular Fatigue Across Running Styles Relating to the Pes Planus Foot
Department & Institution: Health Sciences\Physical Therapy,Western University
Sponsor:
Ethics Approval Date: March 13, 2013 **Expiry Date:**January 31, 2014
Documents Reviewed & Approved & Documents Received for Information:

Document Name	Comments	Version Date
Western University Protocol		2013/01/02
Other	Revised Screening/ Data Collection Form	2013/02/25
Revised Letter of Information & Consent	Revised LOI and Consent	2013/02/25
Revised Western University Protocol	Additional Changes	2013/02/13
Recruitment Items	'REVISED' Participant Recruitment Poster	2013/02/13

This is to notify you that The University of Western Ontario Research Ethics Board for Health Sciences Research Involving Human Subjects (HSREB) which is organized and operates according to the Tri-Council Policy Statement: Ethical Conduct of Research Involving Humans and the Health Canada/ICH Good Clinical Practice Practices: Consolidated Guidelines; and the applicable laws and regulations of Ontario has reviewed and granted approval to the above referenced revision(s) or amendment(s) on the approval date noted above. The membership of this REB also complies with the membership requirements for REB's as defined in Division 5 of the Food and Drug Regulations.

The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable responses to the HSREB's periodic requests for surveillance and monitoring information. If you require an updated approval notice prior to that time you must request it using the University of Western Ontario Updated Approval Request Form.

Members of the HSREB who are named as investigators in research studies, or declare a conflict of interest, do not participate in discussion related to, nor vote on, such studies when they are presented to the HSREB.

The Chair of the HSREB is Dr. Joseph Gilbert. The HSREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000940.

[redacted]

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London, ON, Canada N6A 3K7 t. 519.661.3036 f. 519.850.2466 www.uwo.ca/research/services/ethics

Western Research

Research Ethics

Use of Human Participants - Revision Ethics Approval Notice

Principal Investigator: Dr. Colin Dombroski
File Number: 103376
Review Level: Delegated
Protocol Title: Effects of Muscular Fatigue Across Running Styles Relating to the Pes Planus Foot
Department & Institution: Health Sciences\Physical Therapy,Western University
Sponsor:
Ethics Approval Date: January 16, 2014 **Expiry Date:**
Documents Reviewed & Approved & Documents Received for Information:

Document Name	Comments	Version Date
Letter of Information	LOI	
Advertisement	Poster	
Instruments	screen form	
Revised Western University Protocol	2.4 Inclusion/ 2.5 Exclusion Revised Submitted- {2013/09/25}	
Revised Western University Protocol	2.3 Methodology Revised Submitted- {2013/09/25}	
Approval Notice		
Revised Study End Date		

This is to notify you that The University of Western Ontario Research Ethics Board for Health Sciences Research Involving Human Subjects (HSREB) which is organized and operates according to the Tri-Council Policy Statement: Ethical Conduct of Research Involving Humans and the Health Canada/ICH Good Clinical Practice Practices: Consolidated Guidelines; and the applicable laws and regulations of Ontario has reviewed and granted approval to the above referenced revision(s) or amendment(s) on the approval date noted above. The membership of this REB also complies with the membership requirements for REB's as defined in Division 5 of the Food and Drug Regulations.

The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable responses to the HSREB's periodic requests for surveillance and monitoring information. If you require an updated approval notice prior to that time you must request it using the University of Western Ontario Updated Approval Request Form.

Members of the HSREB who are named as investigators in research studies, or declare a conflict of interest, do not participate in discussion related to, nor vote on, such studies when they are presented to the HSREB.

The Chair of the HSREB is Dr. Joseph Gilbert. The HSREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000940.

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Ethics Officer to Contact for Further Information

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Western Research

Research Ethics

Use of Human Participants - Ethics Approval Notice

Principal Investigator: Dr. Colin Dombroski
File Number:103376
Review Level: Delegated
Approved Local Adult Participants:20
Approved Local Minor Participants:0
Protocol Title: Effects of Muscular Fatigue Across Running Styles Relating to the Pes Planus Foot
Department & Institution: Health Sciences\Physical Therapy,Western University
Sponsor:
Ethics Approval Date: June 18, 2013 **Expiry Date:**January 31, 2014
Documents Reviewed & Approved & Documents Received for Information:

Document Name	Comments	Version Date
Advertisement	Revised Recruitment Poster	
Revised Letter of Information & Consent	Letter of Information	
Revised Western University Protocol		

This is to notify you that The University of Western Ontario Research Ethics Board for Health Sciences Research Involving Human Subjects (HSREB) which is organized and operates according to the Tri-Council Policy Statement: Ethical Conduct of Research Involving Humans and the Health Canada/ICH Good Clinical Practice Practices: Consolidated Guidelines; and the applicable laws and regulations of Ontario has reviewed and granted approval to the above referenced revision(s) or amendment(s) on the approval date noted above. The membership of this REB also complies with the membership requirements for REB's as defined in Division 5 of the Food and Drug Regulations.

The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable responses to the HSREB's periodic requests for surveillance and monitoring information. If you require an updated approval notice prior to that time you must request it using the University of Western Ontario Updated Approval Request Form.

Members of the HSREB who are named as investigators in research studies, or declare a conflict of interest, do not participate in discussion related to, nor vote on, such studies when they are presented to the HSREB.

The Chair of the HSREB is Dr. Joseph Gilbert. The HSREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000940.

[redacted]

Ethics Officer to Contact for Further Information

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Western Research

Research Ethics

Use of Human Participants - Revision Ethics Approval Notice

Principal Investigator: Dr. Colin Dombroski
File Number:103376
Review Level: Delegated
Protocol Title: Effects of Muscular Fatigue Across Running Styles Relating to the Pes Planus Foot
Department & Institution: Health Sciences\Physical Therapy,Western University
Sponsor:
Ethics Approval Date: October 29, 2013 **Expiry Date:**January 31, 2014
Documents Reviewed & Approved & Documents Received for Information:

Document Name	Comments	Version Date
Letter of Information	LOI	
Advertisement	Poster	
Instruments	screen form	
Revised Western University Protocol	2.4 Inclusion/ 2.5 Exclusion Revised Submitted- {2013/09/25}	
Revised Western University Protocol	2.3 Methodology Revised Submitted- {2013/09/25}	

This is to notify you that The University of Western Ontario Research Ethics Board for Health Sciences Research Involving Human Subjects (HSREB) which is organized and operates according to the Tri-Council Policy Statement: Ethical Conduct of Research Involving Humans and the Health Canada/ICH Good Clinical Practice Practices: Consolidated Guidelines; and the applicable laws and regulations of Ontario has reviewed and granted approval to the above referenced revision(s) or amendment(s) on the approval date noted above. The membership of this REB also complies with the membership requirements for REB's as defined in Division 5 of the Food and Drug Regulations.

The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable responses to the HSREB's periodic requests for surveillance and monitoring information. If you require an updated approval notice prior to that time you must request it using the University of Western Ontario Updated Approval Request Form.

Members of the HSREB who are named as investigators in research studies, or declare a conflict of interest, do not participate in discussion related to, nor vote on, such studies when they are presented to the HSREB.

The Chair of the HSREB is Dr. Joseph Gilbert. The HSREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000940.

[redacted]

Ethics Officer to Contact for Further Information

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Western Research

Research Ethics

Western University Health Science Research Ethics Board
NMREB Amendment Approval Notice

Principal Investigator: Dr. Colin Dombroski
Department & Institution: Health Sciences\Physical Therapy,Western University

NMREB File Number: 103376
Study Title: Effects of Muscular Fatigue Across Running Styles Relating to the Pes Planus Foot
Sponsor:

NMREB Revision Approval Date: August 07, 2014
NMREB Expiry Date: December 31, 2014

Documents Approved and/or Received for Information:

Document Name	Comments	Version Date
Revised Study End Date		

The Western University Non-Medical Science Research Ethics Board (NMREB) has reviewed and approved the amendment to the above named study, as of the NMREB Amendment Approval Date noted above.

NMREB approval for this study remains valid until the NMREB Expiry Date noted above, conditional to timely submission and acceptance of HSREB Continuing Ethics Review.

The Western University NMREB operates in compliance with the Tri-Council Policy Statement Ethical Conduct for Research Involving Humans (TCPS2), the Ontario Personal Health Information Protection Act (PHIPA, 2004), and the applicable laws and regulations of Ontario.

Members of the NMREB who are named as Investigators in research studies do not participate in discussions related to, nor vote on such studies when they are presented to the REB.

The NMREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000941.

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Appendix B: Participant Screening Form

Effects of Muscular Fatigue Across Running Styles Relating to the Pes Planus Foot

Screening Form

Participant ID: ______________________________

Age (*years*): ______________________________

Sex: ______________________________

Height (*ft/inches*): ______________________________

Weight (*lbs*.): ______________________________

Dominant Foot (*left or right*): ______________________________

Length of Time You Have Been Distance Running (*years/months*): ______________________________

Distance of Run: ______________________________

Fastest Completed Run at this Distance: ______________________________

Any Previous Injuries to Your Feet or Lower Limbs: ______________________________

Arch Height Information
(*recorded by investigators*):

Seated Heel to Toe (SHT): **Left:** ____________ **Right:** ____________

Standing Heel to Ball (SHB): **Left:** ____________ **Right:** ____________

1/2 of Heel to Toe (1/2 HT): **Left:** ____________ **Right:** ____________

Dorsal Height (DH): **Left:** ____________ **Right:** ____________

Arch Height Index (AHI): **Left:** ____________ **Right:** ____________

Curriculum Vitae

Name:	Adam David Froats
Post-secondary Education and Degrees:	Wilfrid Laurier University Waterloo, Ontario, Canada 2006-2011, Hons. BA(kin)
	The University of Western Ontario London, Ontario, Canada 2011-2012, DipPed
	The University of Western Ontario London, Ontario, Canada 2012-current, Msc(HRS)
Honours and Awards:	Dean's Honor Roll Wilfrid Laurier University 2010-2011
Related Work Experience	Teaching Assistant The University of Western Ontario 2012-2014.
	Registered Kinesiologist/ Canadian Certified Pedorthist SoleScience Inc., London, Ontario, Canada. 2012-2014.
Symposia Lectures	2013- Pedorthic Association of Canada *"Patient Outcome Measures"*
Publications:	Dombroski CE, Balsdon MER and Froats A. (2014). *The use of a low cost 3D scanning and printing tool in the manufacture of custom-made foot orthoses: a preliminary study*. BioMed Central 7: 443.