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The Effect of Exertion on Intra-Limb Joint Coordination Variability During Running Using a Waveform Analysis Approach

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THE EFFECT OF EXERTION ON INTRA-LIMB JOINT COORDINATION VARIABILITY DURING RUNNING USING A WAVEFORM ANALYSIS APPROACH

by

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ABSTRACT

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About half of all runners sustain a running-related injury in a given year. Less variable joint coordination patterns may be detrimental as stress endured by the same tissue, encountered over many running cycles, could lead to overuse running injuries. The effects of fatigue may contribute to runners' risk of injury by altering joint coordination variability. Since fatigue is task-dependent, it is practical to consider a level of fatigue typically experienced by runners. The purpose of this study was to examine the influence of running in an exerted state on lower extremity joint coordination variability, using Principal Components Analysis (PCA) and traditional analysis methods. Sixteen healthy female runners were recruited. Data collection included three-dimensional motion analyses of the ankle, knee and hip before and after a run designed to mimic the subject's typical training experience. Joint coordination was defined using a vector coding technique for eight pairs of joints and planes of motion (e.g. ankle-frontal/knee-transverse) considered relevant to running injury risk. The within-subject variability for these eight coordination patterns was determined from the standard deviation of the coupling angle, averaged over each 25% of stance phase. A repeated measures MANOVA was used to determine differences in joint coordination variability before and

after the run. No significant differences were found for the eight coordination patterns. These results are limited by the analysis method, which requires a priori selection of time periods within stance phase as the dependent variables. PCA is an unbiased way to determine relevant differences in variability among full waveforms, and was used to determine fatigue-related changes in joint coordination variability for each of the eight coupling angle waveforms. A repeated measures MANOVA also did not reveal any differences in joint coordination variability for the eight coordination patterns before and after the run. These results suggest that healthy runners may not experience a change in joint coordination variability during their typical training run. This study established methods for using PCA to quantify changes in joint coordination variability. This can be used in injured populations to test the theory that overuse running injury is associated with low joint coordination variability.

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CHAPTER 1: INTRODUCTION

Running is a common mode of exercise, which is important to maintaining good health (Young & Dinan, 2005). However, about half of all runners will sustain a running-related injury in a given year (Taunton et al., 2002; Walter, Hart, McIntosh, & Sutton, 1989), with most of those injuries occurring at the knee (James & Jones, 1990; Taunton et al., 2002; Taunton et al., 2003; van Gent et al., 2007). Patellofemoral pain syndrome (PFP) is the most common running injury, affecting 1 in 4 people in the general population, with an even higher incidence among athletes (Barton, Munteanu, Menz, & Crossley, 2010; Taunton et al., 2002; Taunton et al., 2003; Thijs, De Clercq, Roosen, & Witvrouw, 2008). PFP affects more women than men (Almeida et al., 1999; Fulkerson & Arendt, 2000; Hutchinson & Ireland, 1995). Despite research conducted in this field, the injury rates have not dramatically changed (Taunton et al., 2002; Taunton et al., 2003; van Gent et al., 2007; Walter, Hart, McIntosh, & Sutton, 1989). Understanding the cause of running injuries is necessary for developing methods for prevention and better treatment options.

Exposure to multiple impact forces over the course of a run, or many runs, is suspected to play a role in most overuse running injuries. These impact forces may be especially harmful if combined with improper lower extremity mechanics during gait (Hreljac, 2004; James & Jones, 1990; Kannus, 1997; Nigg, 1986a). For example, large patellofemoral joint contact forces, which can be up to 7.6 times body weight during running, are thought to be a cause of PFP (Scott & Winter, 1990; Wilson & Davis, 2008). Knee mechanics that contribute to these forces, such as increased internal knee abduction moment and excessive knee valgus position, are considered risk factors for overuse

running injuries (Powers, 2010; Stefanyshyn, Stergiou, Lun, Meeuwisse, & Worobets, 2006). Both improper ankle and hip mechanics could result in the knee experiencing excessive transverse plane rotation, and could cause an increase in lateral patellofemoral joint stress (Barton, Bonanno, Levinger, & Menz, 2010; Barton, Munteanu, Menz, & Crossley, 2010; Boling, Padua, Marshall, Guskiewicz, & Pyne, 2009; Hreljac, 2004; James & Jones, 1990).

Abnormal frontal plane ankle motion and frontal and transverse plane hip motions have been shown to affect knee mechanics during gait. Optimal pronation is a necessary part of ankle movement during the early stance phase of running (Nigg, 1986b), but rearfoot eversion – which is the frontal plane component of pronation – is considered abnormal if the amount of motion is too low or high or if it occurs at the wrong time (Hreljac, 2004; Powers, 2003). It is suggested that excessive rearfoot eversion is associated with increased internal rotation of the tibia, which causes the femur to internally rotate instead of externally rotate during knee extension. Additionally, the hip abductors and external rotators control frontal and transverse plane motion of the femur. Weakness in these muscles could result in greater internal rotation of the femur (Boling, Padua, Marshall, Guskiewicz, & Pyne, 2009; Earl & Hoch, 2011; Snyder, Earl, O'Connor, & Ebersole, 2009). These abnormal joint motions at isolated joints are considered risk factors for injury due to the cumulative effects of exposing tissue to repeated impact forces. Furthermore, these joint motions are also thought to contribute to abnormal joint motion at adjacent joints. Therefore, the risk of overuse running injury may not be related to just one mechanism, but rather the coordination of ankle-knee and hip-knee motions during running and the variability of that coordination. The

coordination pattern considered to be most relevant to running injuries is the coupling of rearfoot eversion, tibial internal rotation and knee flexion (Tiberio, 1987). Asynchrony in this coupling pattern could result in excessive stress at the tibiofemoral joint or the patellofemoral joint.

When applied over many running cycles the abnormal stress could lead to development of a running overuse injury (Hamill, van Emmerik, Heiderscheit, & Li, 1999; Heiderscheit, 2000; N. Stergiou & Bates, 1997; N. Stergiou, Bates, & James, 1999; Tiberio, 1987). Additionally, dynamical systems theory suggests that lack of variability within a coupling pattern may also be an indication of an unhealthy state. Either runners are already injured and have low variability to replicate a pain-free pattern, or replication of a particular coordination pattern with low variability may stress the same tissue repeatedly, resulting in an eventual overuse injury (Hamill, van Emmerik, Heiderscheit, & Li, 1999; Heiderscheit, 2000). Investigations using discrete and continuous techniques for measuring joint coordination patterns have attempted to detect differences between healthy and pathological gait. The results of the studies, however, have been mixed. Some studies report reduced variability for injured runners with PFP (Hamill, van Emmerik, Heiderscheit, & Li, 1999), and ITBS (Miller, Meardon, Derrick, & Gillette, 2008), while others have found no differences in joint coordination variability between healthy and injured runners (Ferber, Davis, & Williams, 2005; Heiderscheit, Hamill, & van Emmerik, 1999). These inconsistencies may be due to limitations in analysis methods as well as the state of the runners.

As runners challenge themselves with increases in intensity, distance or both, they become fatigued or exhausted. Runners with PFP often do not have pain at the beginning

of a run, but complain of a gradual onset of pain as the run progresses. This may indicate that running in an exerted state could cause changes in joint coordination or variability that contribute to running injuries like PFP (Dierks, Manal, Hamill, & Davis, 2011). Studies investigating the effects of fatigue on running biomechanics typically utilize a protocol designed to bring runners to the point of exhaustion or maximum fatigue. This could allow the investigators to examine the greatest changes in biomechanics that occur as a result of fatigue. However, the best protocol for inducing fatigue-related changes in runners would be one that closely mimics a typical bout of exercise for a runner, while also providing an objective measure of exertion for all participants. Running in an exerted state may contribute to runners' risk of injury by altering mechanics such as rearfoot eversion, tibial internal rotation, knee flexion, knee adduction, knee internal rotation, and hip internal rotation. In addition to joint motion changes, joint coordination variability may be affected by running in an exerted state. Investigations in this area have presented results ranging from no changes in joint coordination and variability (Dierks, Davis, & Hamill, 2010; Miller, Meardon, Derrick, & Gillette, 2008), to decreased variability in an exerted or fatigued state (MacLean, van Emmerik, & Hamill, 2010; Trezise, Bartlett, & Bussey, 2011). The equivocal nature of these results on joint coordination variability may be due to limitations and differences in analysis methods.

Traditional investigations in biomechanics have focused on discrete variables, including peak forces, peak angles and excursions. Continuous methods of analysis have an advantage over discrete methods because they allow an investigator to examine the data over the entire stance phase rather than at discrete points. However, all of these approaches require an a priori decision about which dependent variables and events in a

stride cycle will be the most important to consider. By examining the full time series, or waveform, Principal Components Analysis (PCA) is an unbiased way to determine relevant differences in joint coordination variability (Daffertshofer, Lamoth, Meijer, & Beek, 2004; Deluzio, Wyss, Costigan, Sorbie, & Zee, 1999). PCA has been used to determine discriminating factors of gait between age groups (Chester & Wrigley, 2008), gender differences in cutting maneuvers as a risk factor of ACL injury (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007b; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007c; O'Connor & Bottum, 2009), gender differences in knee osteoarthritis patients (McKean et al., 2007), as well as differences between subjects with knee osteoarthritis and healthy controls (Deluzio, Wyss, Zee, Costigan, & Sorbie, 1997; Deluzio & Astephen, 2007; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007a). In addition, the portion of the gait cycle where the difference occurs is identified using PCA. The differences detected by PCA were not identified using traditional biomechanical methods of analysis. PCA can also be used to separate biological variability from random noise by extracting the variance common to all data and identifying the residual variance among the data (Daffertshofer, Lamoth, Meijer, & Beek, 2004; O'Connor & Bottum, 2009). This information could be used to test the dynamical systems theory that greater variability is a positive component of gait with regard to injury risk.

In conclusion, previous studies that have looked at the effect of fatigue on running mechanics have been limited by the design of the fatigue protocol, the a priori identification of discrete dependent variables, and analysis methods that cannot differentiate random noise from biological variability. Utilizing a running protocol that

mimics a person's typical running experience while objectively measuring their state of exertion will allow for better control and quantification of the effects of running in an exerted state on joint coordination variability. Additionally, analysis techniques that investigate continuous joint coordination variability and biological sources of that variability will be a valuable addition to the running injury literature.

Statement of Purpose

The purpose of this study is to examine the influence of running in an exerted state on lower extremity joint coordination variability using waveform analysis. In this investigation, a vector coding technique will be used to determine lower extremity joint coordination patterns, and differences in joint coordination and variability before and after a run in an exerted state will be analyzed using PCA. Results of the PCA will also be compared to results from a traditional method of analysis.

Hypotheses

Given the purpose of this research and the current literature on this topic, the following hypotheses were formed:

Primary hypothesis

Running in an exerted state will result in decreased variability in lower extremity joint coordination patterns.

Secondary hypotheses

Waveform analysis will identify joint coordination variability changes that occur during running in an exerted state that cannot be identified using traditional methods of analysis.

Delimitations of the Study

Results of this study may only be generalizable to the sample and conditions of the experiment.

1. Data will be collected on healthy, young runners; therefore any generalizations made will be limited to this population.
2. Though the run in an exerted state will occur on a treadmill, joint coordination and variability will be assessed overground in a laboratory setting, and so changes in joint coordination and variability may only be generalizable to this testing condition.

Assumptions of the Study

Some assumptions were made in conducting this study:

1. Participants will truthfully answer all questions in the background questionnaire.
2. Participants will accurately describe their perceived level of exertion during the run.
3. Participants will be at a similar level of exertion at the conclusion of the run.
4. The speed selected by the participants will accurately represent their typical pace for a difficult run.
5. Running to an exerted state on a treadmill will be similar to running overground for the data collection trials.
6. All lower-extremity segments are rigid bodies.
7. All lower-extremity joints are frictionless.

Significance of the Study

Investigating the effects of a run in an exerted state on lower extremity joint coordination variability may help determine if running in an exerted state contributes to

common mechanisms of overuse injuries. This information could be used to create better treatment options as well as products and protocols that could be implemented to prevent overuse running injuries. If the hypothesis that running in an exerted state results in lower joint coordination variability is true, treatment options that promote variability will be beneficial.

This study will test the relationship between decreased joint coordination variability and fatigue. Using a waveform approach to analyze joint coordination variability before and after a run in an exerted state will allow for a more accurate measure of variability. Additionally, the validity of a waveform approach to investigate changes in joint coordination variability will be explored.

CHAPTER 2: LITERATURE REVIEW

Running Injury

Background

It has been shown that regular physical activity is important for good health (Young & Dinan, 2005), and many people choose running as a convenient and inexpensive type of physical activity. Running has increased in popularity since the 1970s (Taunton et al., 2002; Walter, Hart, McIntosh, & Sutton, 1989), and therefore the number of people that get injured while running has increased. The percentage of runners who experience running-related injuries can be as high as 79.3% for lower extremity injuries, and up to 92.4% for all injuries (van Gent et al., 2007). About half of all runners will experience a running injury in a given year (Taunton et al., 2002; Walter et al., 1989).

Common injuries

The most common site of injury is at the knee, accounting for up to 50% of running injuries (James & Jones, 1990; Taunton et al., 2002; Taunton et al., 2003; van Gent et al., 2007). Patellofemoral pain (PFP) has consistently been the most common overuse running injury (Barton, Munteanu, Menz, & Crossley, 2010; Taunton et al., 2002; Taunton et al., 2003; Thijs, De Clercq, Roosen, & Witvrouw, 2008). The second-most common running injury, also occurring at the knee, is iliotibial band syndrome (ITBS) (Taunton et al., 2002). ITBS is characterized by sharp pain or burning at the lateral knee (Fredericson, Guillet, & Debenedictis, 2000; Fredericson & Wolf, 2005). It is the most common incidence of lateral knee pain in runners, and accounts for 1.6-12% of all running-related injuries (Fredericson et al., 2000; James & Jones, 1990; Lavine,

2010). Patellar-tendonitis (PT) is another knee injury that affects up to 5% of runners (Grau et al., 2008; Taunton et al., 2002). PT is commonly called "Jumper's Knee" due to its prevalence in athletes that have explosive extension or eccentric flexion of the knee (Johnson, Wakeley, & Watt, 1996). Like PFP, PT is characterized by anterior knee pain (Grau et al., 2008; James & Jones, 1990; Johnson et al., 1996).

As the most common running injury, the incidence of PFP is 1 in 4 in the general population, and even higher among athletes (Boling, Padua, Marshall, Guskiewicz, & Pyne, 2009; Thijs, Van Tiggelen, Roosen, De Clercq, & Witvrouw, 2007). Women are much more likely to have PFP than men (Almeida et al., 1999; Fulkerson & Arendt, 2000; Hutchinson & Ireland, 1995). Competitive and recreational runners as well as adolescents and young adults are at risk for developing PFP (Barton et al., 2010; Thijs et al., 2008). PFP is characterized by anterior or retropatellar knee pain (Crossley, Bennell, Green, & McConnell, 2001; Thijs et al., 2008), or pain and point tenderness in or around the patellofemoral joint that restricts physical activity (Barton et al., 2010; Boling et al., 2009).

PFP can be aggravated by walking, running, going up and down stairs or slopes, squatting, or prolonged sitting and kneeling (Barton et al., 2010). The source of the pain is unknown, but theories suggest it may be located in the cartilage, subchondral bone, synovium, lateral and medial retinaculum, or infrapatellar fat pad (Crossley et al., 2001). For about 25% of PFP patients, the symptoms are likely to persist up to 18 years after initial presentation (Barton, Bonanno, Levinger, & Menz, 2010; Boling et al., 2009), and PFP is associated with the development of patellofemoral osteoarthritis (Boling et al., 2009).

Mechanisms of injury

Running injuries are typically the result of overuse, or pain resulting from repetitive tissue (bone, cartilage, tendon, ligament or muscle) microtrauma. These repetitive stresses are necessary for positive remodeling of tissue, as long as the stresses are kept below critical limits. When this is the case, tissue deformation lasts as long as the mechanical stress is applied, and after a short amount of time the tissue returns to its original form, which is known as hysteresis (Nigg, 1986a). However, without sufficient time between the applications of stress, repeated exposure of tissue to low-magnitude forces creates microscopic injuries that eventually strain the tissue until the overuse injury occurs (Hreljac, 2004; James & Jones, 1990; Kannus, 1997; Nigg, 1986a). The response to stress depends on the type of tissue being stressed. Due to low nutrition flow, cartilage and tendons have little positive (remodeling) response to the stress stimulus. Conversely, bone and muscles have a much higher response (Denoth, 1986).

Impact forces

The magnitude and rate of loading of impact forces are suspected to play a prominent role in the stresses that cause overuse running injuries. Impact forces are characterized as the high frequency forces associated with initial contact. For shod rearfoot-strike runners (the majority of runners), maximum impact force occurs about 10-35 ms after initial contact, and can be up to 2-5 times body weight (Hreljac, 2004; Nigg, 1986a).

Factors that influence the impact peak during running are: the hardness of the running shoe material, the geometry of the shoe sole, running velocity or a change in the positioning of the foot and leg at impact. There are many ways that runners can

manipulate these factors to better absorb impact forces, including hip and knee flexion, ankle dorsiflexion and pronation at the subtalar joint (Hintermann & Nigg, 1998; Nigg, Bahlsen, Denoth, Luethi, & Stacoff, 1986). For example, runners have been shown to run with greater foot and leg eversion with increasing running velocity. Impact forces are greater when running at a higher velocity, and the hardness of the contact is proportional to the area of contact with the ground. An increase in rearfoot angle decreases the initial contact area with the ground, and allows for a softer landing (Nigg et al., 1986).

Knee mechanics

The magnitude of knee joint forces and moments is thought to be linked to overuse running injuries. PFP is thought to be caused by excessive patellofemoral joint stress (Powers, Ward, Chen, Chan, & Terk, 2004; Wilson & Davis, 2008). The patella increases the moment arm of the quadriceps, allowing it to increase the knee extension torque. Mean patellofemoral joint contact forces during running can be up to 7.6 times body weight; with many repetitions, this may explain why the patellofemoral joint is commonly injured (Scott & Winter, 1990; Wilson & Davis, 2008). It is difficult to measure in vivo patellofemoral joint stress, so knee extensor and abduction moments are used to look at patellofemoral joint loading (Stefanyshyn, Stergiou, Lun, Meeuwisse, & Worobets, 2006; Wilson & Davis, 2008).

Frontal plane loading, characterized by increased internal knee abduction moment throughout stance phase, has been associated with PFP in both retrospective and prospective studies. During stance phase, the hip is adducted, and the ground reaction force results in a large external adduction moment at the knee, which is compensated for by an internal abduction moment. Increased knee abduction moments can be generated

by increased muscle or soft tissue forces. This could overpower vastus medialis, which stabilizes the patella medially, and lateral tracking of the patella can occur. The resulting lateral stress can lead to PFP (Stefanyshyn et al., 2006).

Excessive knee valgus has been shown to contribute to knee injuries, such as PFP (Powers, 2010). Additionally, transverse and frontal plane rotations of the hip and knee can change the Q-angle. An increased Q-angle causes greater retropatellar stress during knee flexion. Performing weight-bearing activities, such as running, with this alignment may cause inflammation of the tissues around the patella and lead to PFP (Boling et al., 2009; Wilson & Davis, 2008).

Risk factors

Overuse injuries can have a greater manifestation when coupled with extrinsic risk factors, such as training and environment, and/or intrinsic risk factors, such as muscle weakness or imbalance and biomechanical alignments. Additionally, factors including age and prior injuries can affect overuse injuries (Kannus, 1997). For example, elderly runners have a greater incidence of injury than young adult runners, and elderly runners take longer to recover from an injury (Fukuchi & Duarte, 2008; Fukuchi, Eskofier, Duarte, & Ferber, 2011). Having a prior injury has been shown to increase the risk of injury in runners, particularly if that injury has not been fully rehabilitated (Taunton et al., 2002; Taunton et al., 2003; van Gent et al., 2007; Walter et al., 1989).

Training

The predominant factor that puts runners at an increased risk of injury is training errors. Novice runners may be at a greater risk for injury (Taunton et al., 2002), though this trend is not always the case (Taunton et al., 2003; van Gent et al., 2007; Walter et al.,

1989). Inexperience with running may cause some runners to not recognize the early signs of an injury, and running through these signs may exacerbate the injury. However, an experienced runner may be more susceptible to injury due to a large volume of running. An excessive or competitive training load or rapidly increasing weekly mileage has been shown to put runners at a risk of injury. High mileage, high intensity workouts or a rapid change in training regimen does not give the body adequate time to adapt to the new forces (Fredericson & Misra, 2007; Hreljac, 2004; James & Jones, 1990; Taunton et al., 2002; van Gent et al., 2007; Walter et al., 1989). This has been the case for injuries such as PFP (Barton et al., 2010).

Strength

Muscle imbalances and various motions occurring at the knee could put runners at risk for injury. Vastus lateralis and vastus medialis oblique muscles are the primary dynamic stabilizers of the patella, so weakness or imbalance here could lead to patella maltracking, which could be a cause of PFP (Boling et al., 2009; James & Jones, 1990).

The hip joint relies on muscles to provide dynamic stability at the hip and distally along the lower extremity. Runners with weakness or impairments at the hip, particularly in the hip abductors and external rotators, could have greater internal rotation of the femur and increased Q angle. A large Q angle is thought to increase the lateral compressive forces at the patellofemoral joint that cause PFP (Boling et al., 2009).

In support of this theory, runners with weak hip abductors exhibited greater knee abduction during stance phase of running. Greater knee abduction could produce abnormal patellar pressures, which could have implications in PFP (Grau et al., 2008; Heinert, Kernozek, Greany, & Fater, 2008; Stefanyshyn et al., 2006). Also, strengthening

programs that improved the strength of the hip abductors and external rotators led to alterations in lower joint extremity loading. Specifically, rearfoot eversion and hip internal rotation ranges of motion as well as knee abduction and rearfoot inversion moments were reduced (Earl & Hoch, 2011; Snyder, Earl, O'Connor, & Ebersole, 2009). These results indicate that alterations at the hip can have an effect on the biomechanics of the distal portion of the lower extremity.

Footwear

Inappropriate footwear may also put runners at a risk of injury, and it has been shown to be a risk factor for PFP (Barton et al., 2010; Paluska, 2005). The most important considerations when constructing a running shoe is the influence on geometry and deformation of the foot and shoe (Stacoff & Luethi, 1986). The cushioning characteristics of the sole of a running shoe can influence the stress-deformation properties of the heel at initial contact, which could affect the mechanical trauma experienced by a runner (Denoth, 1986; Nigg, 1986a). Changes in the geometry or material of the midsole of a shoe may also influence the amount of pronation or cushioning. There are 15-20 individual parts of a running shoe, including midsole, heel counters, heel stabilizers, insoles, inserts, additional wedges and different lasts. Additionally, there are different strategies for combining the individual parts to make a running shoe. This means that running shoes can be constructed with specific running patterns in mind. It also means that unwanted side effects may occur when running shoes are constructed in certain ways (Stacoff & Luethi, 1986).

Pronation

Pronation is a motion consisting of simultaneous rearfoot eversion, abduction and dorsiflexion. The reverse movement, supination, is simultaneous rearfoot inversion, adduction and plantar flexion (Nigg, 1986b). Pronation occurs for approximately the first 70% of stance phase, with peak pronation at about 40%. Supination follows for the remaining 30% of stance phase (Hreljac, 2004; James & Jones, 1990).

An optimal amount of pronation is a necessary part of foot movement during the stance phase of running. Pronation is considered abnormal if the amount of motion is too low or high or if it occurs at the wrong time (Hreljac, 2004; Powers, 2003). Underpronation or a lack of pronation would make the foot too rigid and reduce its ability to absorb impact forces (Nigg, 1986b). Pronation is limited by passive structures in the foot – ligaments and bone – and overpronation may strain this connective tissue (Nigg, 1986b). Additionally, excessive or prolonged pronation may affect proximal aspects of the lower extremity, leading to common running injuries at the knee (Hreljac, 2004; James & Jones, 1990). For this reason, many running shoes are constructed with the aim of reducing pronation (Nigg, 1986b).

Overpronation is thought to be a risk factor for running injuries, and it may play a role in the mechanism for the development of PFP (Thijs et al., 2008; Thijs et al., 2007). It is suggested that excessive foot pronation is associated with excessive internal rotation of the tibia, which causes the femur to internally rotate during knee extension. This could result in the knee absorbing more transverse plane rotation, and could cause an increase in lateral patellofemoral joint stress (Tiberio, 1987). However, there is no consensus in the literature about the relationship between abnormal pronation of the foot and PFP

during static and dynamic evaluations (Barton et al., 2010; Barton et al., 2010; Thijs et al., 2008; Thijs et al., 2007). Runners who have overuse injuries may have abnormal pronation compared to healthy runners, although many runners with excessive pronation do not have overuse injuries (Hintermann & Nigg, 1998). Therefore, though it is frequently cited as a cause for injury, the relationship between abnormal pronation and running injuries is unclear (Hreljac, 2004).

Despite the lack of consensus that excessive pronation causes PFP, the use of orthotics has been an effective treatment for PFP, regardless of the type of material or how the orthotics are used (Bartold, 2001; Barton et al., 2010; Barton et al., 2010; Crossley et al., 2001; Eng & Pierrynowski, 1993; MacLean, Davis, & Hamill, 2008). The traditional hypothesis for the effectiveness of orthotics is that controlling excessive foot motion reduces abnormal lower limb internal rotation, which reduces the stress on the lateral patellofemoral joint. However, it has not been confirmed that this is the mechanism that is being corrected by the orthotics (Bartold, 2001; Heiderscheit, Hamill, & Tiberio, 2001; MacLean et al., 2008). Methodological differences, including the difficulty of measuring subtalar joint motion, and individual subjects' differing responses to orthotics may be the cause of equivocal results in the literature (Bartold, 2001; Heiderscheit et al., 2001; Neptune, Wright, & van den Bogert, 2000). Alternative hypotheses have been proposed that suggest orthotics improve the muscle activation patterns of the lower extremity, though this has not been observed in the short term (Barton et al., 2010; Boling et al., 2009; Rose, Shultz, Arnold, Gansneder, & Perrin, 2002).

Movement coordination and variability

There is significant coordination that occurs between the segments of the lower extremity. These coordinative structures, or muscle synergies, can allow the same goal to be reached by using different degrees of freedom, and they can use the same degrees of freedom to reach the same goals. The dynamical systems approach to studying movement coordination relies on the variability in the coordination patterns. Variability may provide information about the stability and flexibility of movement. A stable system can minimize metabolic cost, and variability is known to occur prior to phase transitions (Hamill, van Emmerik, Heiderscheit, & Li, 1999; Heiderscheit, 2000). Treatment of disordered movement is thought to be most successful when variability is high (Heiderscheit, 2000). Traditionally, variability has been thought to be a limitation to movement: increased variability of stride characteristics has been used to predict risk of falling, and is found in people with neuromuscular disease (Hamill, Haddad, Heiderscheit, Van Emmerick, & Li, 2006; Heiderscheit, 2000). However, variability in joint coordination can be considered functional, as it provides the system the ability to adapt (Hamill et al., 1999; Hamill et al., 2006; Heiderscheit, Hamill, & van Emmerik, 1999). For example, joint coordination variability has been found in sports events that require adaptation to changing environmental conditions, such as throwing a javelin and shooting a basketball (Bartletti, Wheat, & Robins, 2007). Additionally, expert marksmen have greater variability when stabilizing their arm than novice marksmen (Hamill et al., 1999). While low variability may be considered bad, too much variability may be bad as well. It is likely that there is an optimal level of variability for the system to work best (Miller, Meardon, Derrick, & Gillette, 2008). Applying this dynamical systems approach

to running injuries, it is thought that variability of joint coordination can be used to detect differences in healthy versus pathological gait.

The coordination pattern considered to be most relevant to running injuries is the following: pronation, tibial internal rotation and knee flexion are coupled while supination, tibial external rotation and knee extension are coupled for the reverse movement. It has been suggested that the maxima of these actions should occur at the same time. Asynchrony in these coupling patterns may occur, for example, if the subtalar joint pronates while the knee extends. In this case, the tibia could not externally rotate, affecting tibial-femoral transverse plane rotation and resulting in excessive stress at the tibiofemoral joint. A compensatory mechanism would be for the femur to internally rotate. This could lead to compression between the lateral patella and the lateral femoral condyle, and subsequent lateral tracking of the patella. This stress at the tibiofemoral joint or the patellofemoral joint, encountered over many running cycles, could lead to a running overuse injury (Hamill et al., 1999; Heiderscheit, 2000; Stergiou & Bates, 1997; Stergiou, Bates, & James, 1999; Tiberio, 1987).

Lack of variability with this particular coupling pattern may also be an indication of an unhealthy state. Completing a running task with a low amount of variability may allow injured runners to replicate a pain-free pattern. Thus, low variability may be indicative of an injury, though it does not necessarily mean it caused the injury. However, replication of a particular coordination pattern with low variability may stress the same tissue repeatedly, resulting in an overuse injury (Hamill et al., 1999; Heiderscheit, 2000). The problem with retrospective studies, however, is it is impossible

to know whether reduced variability caused the pain/injury, or if pain caused the runners to run with reduced variability.

Types of analysis.

Intra-limb coordination of joints or segments can be assessed by either discrete or continuous methods. Discrete methods are used to determine relative timing of joints or segments at one point in a movement cycle. An advantage to using discrete methods to evaluate movement coordination is that the data does not need to be manipulated beyond normal calculation of joint angles. The disadvantage of using discrete methods is that they evaluate coordination at only one point during the cycle (Hamill, Haddad, & McDermott, 2000).

Initial joint coordination studies examined the relative timing of pronation, tibial internal rotation and knee flexion with methods such as discrete relative phase (DRP) and joint excursion ratios (DeLeo, Dierks, Ferber, & Davis, 2004). In the time-series approach, a discrete relative phase (DRP) angle is determined for a particular point during the movement cycle. The DRP angle is calculated using the difference between times to the key event in the time-series of two joint or segment angles (Hamill et al., 2000).

The joint excursion ratio (commonly used for eversion/tibial internal rotation (EV/TIR) ratio) measures the relative excursion of each motion from heel-strike to its peak near midstance. The EV/TIR ratio is likely influenced by arch height, and the differences in EV/TIR ratio are most often attributed tibial internal rotation excursion more than eversion excursion. EV/TIR ratio has not been found to accurately indicate the

location of injuries. However, these measures are based only on discrete points of the gait cycle (DeLeo et al., 2004).

The lack of a difference between healthy and injured runners at discrete moments during the stride cycle does not mean differences do not exist. Continuous methods are used to determine coordination or coupling of movement over a period of time. This is different from discrete methods which only evaluate coordination at a single point in the cycle. Therefore, a continuous measure of coupling throughout the stride cycle is important (Hamill et al., 1999; Hamill et al., 2000). Traditionally, two types of continuous methods are used for determining coordination: continuous relative phase (CRP) and relative motion, also known as vector coding. While both methods are valid for measuring coordination and variability, they do not convey the same information at all times. The differences between the methods are most obvious when determining variability at specific instances or portions of a movement cycle (Miller, Chang, Baird, Van Emmerik, & Hamill, 2010). The decision of which method to use depends on the research question being asked (Hamill et al., 2000; Miller et al., 2010).

CRP is useful because it provides both continuous spatial and temporal information. CRP is calculated by creating a parametric phase plot – velocity plotted as a function of position – for each segment. Phase angles are then determined from the arctangent of this plot. After time-normalizing the phase angle, CRP is found by subtracting the phase angle of one segment from the other at every time point. When CRP is 0° the segments are in-phase, and when CRP is 180° the segments are anti-phase. CRP variability is the standard deviation of the CRP at each point in the cycle (Hamill et al., 1999; Hamill et al., 2000).

An additional normalization step must be taken for CRP before calculation of the phase angles. This will account for the frequency differences between waves. The goal of normalization should be to make the phase-plane more circular and center the phase plot about an origin. Different results will be obtained depending on the normalization procedures utilized (Hamill et al., 2000; Peters, Haddad, Heiderscheit, Van Emmerik, & Hamill, 2003; Wheat & Glazier, 2006).

CRP is used to compare the degree of in-phase or out-of-phase relationships for various coupling relationships. This has been done with mixed results. The use of angular velocity in the computation of phase angles provides temporal as well as spatial information, and may make CRP a more sensitive measurement of variability. However, the higher derivative of angular velocity may propagate errors in the displacement data. Additionally, it has been shown that normalization alters the data, and so some authors do not normalize, making comparisons between studies difficult (DeLeo et al., 2004; Wheat & Glazier, 2006). It is also difficult to generalize the in- or out-of-phase coupling for multiple joint segments or joint combinations throughout stance. Another limitation of CRP is that it is traditionally used for predominantly sinusoidal oscillators. However, most lower extremity joint movements – with the exception of the sagittal plane motion of the hip – are non-sinusoidal, which may affect the results of CRP (DeLeo et al., 2004; Heiderscheit, Hamill, & van Emmerik, 2002; Peters et al., 2003; Wheat & Glazier, 2006).

Vector coding, therefore, is a way to determine continuous coordination for non-sinusoidal data. Using relative motion or a vector coding method to determine coordination is convenient because no normalization of data is required. It may be useful as a clinical tool because the original kinematic data are used in the analysis (Miller et al.,

2010). However, only spatial, and not temporal, information is presented. Relative motion measures coordination by using angle-angle plots. With the proximal segment or joint angle on the x-axis and the distal segment or joint angle on the y-axis, each point in the time-series is plotted. A vector is made between consecutive points, and the orientation of the vector relative to the right horizontal is called the coupling angle. The coupling angle describes the relative motion of the joints or segments. The coupling angle can be plotted as a function of the stride cycle. The variability of the coupling angle can be used to assess variability across multiple trials and/or between subjects (DeLeo et al., 2004; Hamill et al., 2000; Sparrow, Donovan, Vanemmerik, & Barry, 1987; Wheat & Glazier, 2006).

Coordination patterns.

Several studies have used discrete or continuous techniques of measuring joint coordination in an attempt to define healthy and pathological gait, yet differences in these analysis methods have made it difficult to compare the results of these studies. Various coordination patterns have been considered, depending on the topic of interest. Also, some studies have looked at the coordination patterns over the entire stance phase, while other studies have divided the stride cycle into a few phases to examine coordination patterns over a given period.

Joint coordination patterns of healthy runners that were determined using joint timing, excursion ratios, vector coding and CRP methods produced differing results. The joint timing relationships were relatively synchronous between rearfoot eversion, tibial internal rotation and knee flexion, but were asynchronous when knee internal rotation was involved. Excursion ratios revealed that the amount of eversion is twice the amount

of both tibial internal rotation and knee internal rotation. Vector coding methods showed that motion and variability was relatively similar across all coupling relationships and time periods of stance. CRP indicated that the coupling relationships were more out-of-phase and variable during the period of heel-strike to impact peak and in the last quarter of stance phase, while they were more in-phase and less variable during midstance after the acceptance of full body weight. The CRP results suggest that the less stable coordination patterns occur at the transitions between loaded and unloaded states. The coupling patterns that showed the most variability were rearfoot in/eversion-knee rotation and tibial rotation-knee rotation, while rearfoot in/eversion-knee flexion/extension was least variable (Dierks & Davis, 2007).

The joint coordination for abnormal or pathological gait has also been calculated using several methods, with inconsistent results. Subjects with excessive pronation had a smaller EV/TIR ratio due to a greater excursion of tibial internal rotation, and peak eversion occurring sooner. This would cause the rearfoot to invert while the tibia was still internally rotating and the knee was still flexing, and may put those runners at a greater risk of knee injury (McClay & Manal, 1997). In a similar study, it was found that the timing of joint angles is different for PT subjects than controls: maximum hip adduction is delayed and maximum tibial internal rotation is early, relative to maximum knee flexion (Grau et al., 2008). This early eversion and/or tibial internal rotation contradicts the pathological coordination pattern described by Tiberio (1987), which suggests that delayed maximum pronation, coupled with tibial internal rotation, leads to the development of knee injuries such as PFP and PT. While joint coordination may be a key component of running gait, it is important to be able to quantify it in a consistent and

meaningful way. These equivocal results could be due to inconsistencies or limitations in the methods of analysis.

Variability.

Support for the theory that reduced variability of joint coordination is indicative of pathological gait is found in studies comparing the gait of injured runners to healthy controls. Runners with ITBS exhibited less CRP variability in thigh ab/aduction-rearfoot in/eversion coordination over the complete stride cycle at the end of an exhausting run. Also, the ITBS runners had less CRP variability in rearfoot in/eversion-tibia in/external rotation coordination at heel-strike (Miller et al., 2008). Similarly, runners with PFP were shown to have less CRP variability in coordination patterns involving the knee than healthy runners (Hamill et al., 1999).

The connection between PFP and Q-angle was also explored in the context of joint coordination variability, but without the same results. A high Q-angle is thought to be a predictor of PFP, since a high Q-angle may cause excessive foot pronation, disrupting the coordination patterns described above. However, subjects with high and low Q-angles did not show a difference in CRP or CRP variability (Hamill et al., 1999; Heiderscheit et al., 1999).

Using a vector coding method, limited support for the theory that reduced variability of joint coordination is indicative of pathological gait has been shown. Increased variability in stride length was found for PFP patients, and across the whole stride cycle, unilateral PFP patients did not exhibit reduced variability in joint coordination in relation to their non-injured limb or a control group (Heiderscheit et al., 2002). Additionally, injured runners that improved their symptoms with the use of

custom orthotics did not exhibit differences in joint coordination or variability compared to healthy runners (Ferber, Davis, & Williams, 2005). However, reduced variability in thigh rotation-leg rotation coordination in a sub-phase of stance that included heel-strike for the PFP group does support the dynamical systems hypothesis that reduced variability in joint coordination may indicate pathological gait (Heiderscheit et al., 2002).

The time period during which coupling is examined may be important. Specifically, the beginning of stance phase has received a lot of attention. In three coupling relationships involving leg rotation – thigh flexion/extension, thigh ab/adduction and foot eversion/inversion – the greatest variability occurred in the initial period of stance from heel-strike to initial eversion for both high and low Q-angle subjects (Heiderscheit et al., 1999). This may indicate a flexible system at initial contact is necessary to prevent the body from repeatedly absorbing high impact forces in the same pattern, and/or allows the body to react to perturbations that may occur at initial contact, such as changes in terrain. However, this type of analysis requires a priori identification of the periods of stance phase to investigate, and not all studies define these periods in the same way. This makes comparisons between the studies difficult, and limits the ability to detect changes in variability which may not be occurring within these pre-defined periods.

Fatigue

Models of fatigue

Fatigue has many definitions, and there are several proposed models of how and where fatigue occurs. Simply, fatigue can be considered a decrease in force production, such that there is an increase in the perception of effort required, and eventually an

inability to produce the force (Enoka & Stuart, 1992). This type of definition suggests that fatigue suddenly occurs at task failure, however, muscles begin to have reduced force-generating capacity at the onset of exercise. Therefore, fatigue may be more aptly defined as an exercise-induced reduction in maximal voluntary muscle force, due to peripheral changes in the muscle and reduced drive from the central nervous system (Gandevia, 2001). There has been much debate over what causes fatigue and where it occurs.

Central and peripheral fatigue

Peripheral fatigue is typically defined as a decrease in the ability of skeletal muscle to produce force, and occurs distal to the neuromuscular junction. Central fatigue is defined as a reduction in the neural drive to muscle which results in the reduced force production proximal to the neuromuscular junction (Ament & Verkerke, 2009; Maclaren, Gibson, Parrybillings, & Edwards, 1989; St Clair Gibson & Noakes, 2004).

Rhythmic exercise, like running, ends when the target speed is not maintained. It is thought that running task failure occurs due to a lack of substrate supply, particularly carbohydrates, the accumulation of fatigue substances, or high muscle temperatures (Gandevia, 2001; St Clair Gibson & Noakes, 2004). Recently, it has been suggested that central fatigue can occur after prolonged running exercise (Millet & Lepers, 2004), though peripheral fatigue is likely the main reason for fatigue during running (Lattier, Millet, Martin, & Martin, 2004; Skof & Strojnik, 2006).

Catastrophic failure vs. Central governor model

The classic model of fatigue is the catastrophic failure model (Maclaren et al., 1989). Under this model, there is increased neural drive to maximize skeletal muscle

recruitment as metabolic changes lead to system failure of skeletal or cardiac muscle. Ultimately, at maximum skeletal muscle recruitment, the system failure results in skeletal muscle contractile failure, and the desired force catastrophically can no longer be produced. After a period of rest, the metabolic system returns to normal and full recovery is attained (St Clair Gibson & Noakes, 2004).

This catastrophic failure model of fatigue has been criticized for suggesting that the termination of exercise is always a result of a body system failure. In proposing the central governor model as an alternative, authors that criticize the catastrophic failure model point out that skeletal muscle recruitment is not maximized during voluntary exercise to exhaustion. They use this to suggest that the central nervous system regulates skeletal muscle contraction with the specific goal of preventing catastrophic failure. Therefore, the proposed central governor model holds that the development of physical exhaustion is not an absolute event, but rather the sensory representation of the neural processes that regulate exercise intensity so that homeostasis can be maintained. The interaction of physiological systems and environmental information are combined in a “governor” region of the brain that produces a pacing strategy for the athlete to regulate its exercise (Lambert, St Clair Gibson, & Noakes, 2005; Noakes & St Clair Gibson, 2004; Noakes, St Clair Gibson, & Lambert, 2004; Noakes, St Clair Gibson, & Lambert, 2005; St Clair Gibson & Noakes, 2004).

Though the central governor model may be most applicable to endurance exercise, it may not apply to all types of exercise, especially short bouts of maximal force. The central governor model has been challenged with examples of studies that have shown exercise to be terminated as a result of reduced skeletal muscle contraction,

and not only a reduction in motor unit recruitment (Weir, Beck, Cramer, & Housh, 2006). Therefore, it is possible that there is not one all-encompassing model of fatigue. The process by which a muscle becomes fatigued may have both central and peripheral factors, and is thought to be task-dependent (Enoka & Stuart, 1992; Gandevia, 2001; Weir et al., 2006).

Measuring fatigue

Fatigue in the context of exercise physiology has objectively measured physiological effects, but it also has a subjective psychological component. Another term related to a reduction in performance during physical exercise is exhaustion. Exhaustion can be defined as the moment in which the sense of effort required to maintain a desired force is greater than a person’s willpower to maintain that output (Ament & Verkerke, 2009). Physiologically, fatigue due to running can be measured using blood lactate tests or a rating of perceived exertion. Heart rate can be used as a measure of effort (Lucci, Cortes, Van Lunen, Ringleb, & Onate, 2011). Fatigue, being defined as a loss of force production, can be measured in specific skeletal muscles by observing a decrease in force produced during a maximum voluntary contraction following a fatigue protocol. It can also be observed as a decrease in speed during a maximal effort run (Nummela et al., 2008).

The cause of fatigue is multifactorial and task dependent, meaning the cause of fatigue is related only to the characteristics of the exercise or task inducing the fatigue. For exercise, such as running, the factors to consider are: how and when fatigue is measured, the subjects, and the fatigue protocol, which consists of the exercise mode, intensity and duration. With this approach, the fatigue protocol should be close to normal

exercise, making it practically relevant to study. However, fatigue may be difficult to control or measure in this setting, due to differences in subjects, the artificial environment of the laboratory, the methods used for determining fatigue and how closely the protocol matches real exercise (Cairns, Knicker, Thompson, & Sjogaard, 2005).

During the course of their training runs, runners rarely run to the point of exhaustion or maximum fatigue. Therefore, designing a study with a protocol that resembles a typical running session may give a more accurate picture of the biomechanical changes that occur during running. Called running in an exerted state, this was investigated when runners performed a prolonged run at training pace until their heart rate reached 85% of their maximum heart rate or the participants registered greater than 17 (very hard) on a rating of perceived exertion scale (Dierks, Davis, & Hamill, 2010; Dierks, Manal, Hamill, & Davis, 2011).

Some studies that investigated the effects of fatigue on running utilized a protocol designed to bring runners to the point of exhaustion or maximum fatigue. This could allow the investigators to examine the greatest changes in biomechanics that occur as a result of fatigue. However, many different exhaustion or maximum fatigue protocols have been used, including an exhaustive run at ventilatory threshold heart rate (Abt et al., 2011), a 10x400m interval workout (Collins et al., 2000), an exhaustive run at 3200 m maximum effort pace (Derrick, Dereu, & McLean, 2002), a maximal effort graded exercise test (Mercer, Bates, Dufek, & Hreljac, 2003), a 30-minute run above anaerobic threshold (Mizrahi, Verbitsky, Isakov, & Daily, 2000), a run at 4.5 m/s until volitional abandonment (VanGheluwe & Madsen, 1997), maximal effort during a 5,000 m run in competitive and noncompetitive settings as well as a treadmill run until volitional

abandonment at 5,000 m pace (Hanley, Smith, & Bissas, 2011; Nummela et al., 2008; Williams, Snow, & Agruss, 1991), a VO_2max test (Gerlach et al., 2005), and a marathon run (Chan-Roper, Hunter, Myrer, Eggett, & Seeley, 2012; Kyrolainen et al., 2000; Nicol, Komi, & Marconnet, 1991).

However, not all of these studies used objective, physiologically measured criteria of fatigue or exhaustion. The different running durations and intensities may affect runners differently (Abt et al., 2011). A study examining the effects of two fatigue protocols – one short and one long – on landing mechanics during a side-cutting maneuver found that the kinematics were altered similarly by both protocols. This suggests that a shorter protocol may be just as effective at eliciting changes due to fatigue as a longer protocol (Lucci et al., 2011). In the running literature, there are two common types of fatigue protocols: a marathon run and a short run at high intensity. The shorter method exhausts runners faster, and may result in kinematic changes that increase the risk of running injury. Yet exhausting runners faster may fatigue their cardiovascular system before compromising the neuromuscular system. The physiological and kinematic responses to this type of fatigue protocol have been reported to be different than a protocol similar to a marathon run (Abt et al., 2011). The best protocol for inducing fatigue in runners would be one that closely mimics a typical bout of exercise for a runner, while also providing an objective measure of fatigue for all participants.

Effects of fatigue on running biomechanics

Runners with PFP often do not have pain at the beginning of a run, but complain of a gradual onset of pain as the run progresses. This may indicate that prolonged running or exhaustion could cause mechanics that contribute to PFP. Despite this

hypothesis, there have been very few studies investigating the effects of fatigue on running mechanics for healthy and injured runners (Dierks et al., 2011). Among the studies that have been completed in this field, comparisons are difficult due to differences in fatigue protocols and 2D vs. 3D data collection.

In some studies, group differences pre- and post-fatigue were small, but some individual differences were large. This suggests that some biomechanical changes due to fatigue may be more important to some subjects than others (VanGheluwe & Madsen, 1997; Williams et al., 1991). A decrease in running economy has been observed following a marathon or exhaustive run, but the reduced economy was not found to be a result of changes in running biomechanics (Collins et al., 2000; Kyrolainen et al., 2000). However, the goal of kinematic adaptations may be to minimize metabolic cost, even at the expense of shock absorption (Hardin, Van den Bogert, & Hamill, 2004), which may put a runner at risk for injury.

Shock attenuation

Shock attenuation, or the absorption of impact forces, is vital for the prevention of overuse running injuries. It can be accomplished due to the shock absorbing properties of passive anatomical structures such as bone and the calcaneal fat pad, as well as external influences such as running shoes and the ground (Derrick, Hamill, & Caldwell, 1998; Valiant, 1990). Additionally, contraction of muscle plays a huge role in shock attenuation. It has been shown that muscle action at the joints, such as ankle, knee and hip flexion, help to reduce impact forces during running (Derrick et al., 1998). Running in an exerted state may increase a runner’s risk of overuse injury if the exhausted muscle

loses some shock absorbing ability or causes a change in movement pattern (Mercer et al., 2003).

Studies examining the effect of a run in an exerted state on shock attenuation have had mixed results. One study showed a decrease in impact forces and loading rate, suggesting a runner's attempt to reduce injury risk at the end of a VO_2max test (Gerlach et al., 2005). Meanwhile, other studies found increases in tibial accelerations at heel-strike which could increase a runner's risk of overuse injury (Mizrahi et al., 2000; Verbitsky, Mizrahi, Voloshin, Treiger, & Isakov, 1998). Another study showed that leg impact forces increased after an exhaustive run at 3200 m maximum effort pace, but changes in kinematics allowed for an increase in shock attenuation at the end of the run (Derrick et al., 2002). This was not confirmed in a later study; no differences in tibial acceleration were found, and shock attenuation was decreased at the end of a maximal effort graded exercise test (Mercer et al., 2003). A recent investigation indicates that there are no significant changes in impact accelerations or shock attenuation during an exhaustive run at ventilatory threshold heart rate (Abt et al., 2011). These results suggest that there may be individual differences in how runners absorb impact forces while running in an exerted state.

Stride rate and stride length

The effect of fatigue on running is commonly studied with regard to stride parameters, but results are inconclusive. Several studies have shown no significant difference in stride length as a result of running in an exerted state (Collins et al., 2000; Derrick et al., 2002; Mercer et al., 2003; Nicol et al., 1991). An increase in stride length has been reported (Williams et al., 1991), as well as an increase in stride length with a

corresponding decrease in stride rate (Gerlach et al., 2005). Studies reporting a decrease in stride length have also reported a decrease in stride rate (Chan-Roper et al., 2012; Hanley et al., 2011), an increase in stride rate (Kyrolainen et al., 2000), or no significant difference in stride rate (Nummela et al., 2008). Another study reported just a decrease in stride rate (Mizrahi et al., 2000). These conflicting results could be the result of large individual differences in stride parameters, or differences in testing procedures.

Rearfoot mechanics

Several studies have looked at the effects of fatigue on rearfoot mechanics. Four studies have reported an increase in maximum rearfoot eversion during stance as well as an increase in maximum rearfoot velocity following a run in an exerted state (Derrick et al., 2002; Dierks et al., 2010; Dierks et al., 2011; VanGheluwe & Madsen, 1997). This suggests that runners may have greater pronation when running in an exerted state, putting them at risk for injuries induced by excessive pronation.

Knee mechanics

Knee flexion at heel-strike has been commonly studied after an exhausting run due to its role in shock attenuation. A few studies have reported no significant difference in knee flexion angle at heel-strike (Collins et al., 2000; Hanley et al., 2011; Williams et al., 1991), while others have reported an increase in knee flexion angle at heel-strike (Derrick et al., 2002; Mizrahi et al., 2000; Nicol et al., 1991). Though there seems to be an effect of the knee angle in the sagittal plane at heel-strike, this has not been seen throughout stance. Knee flexion during stance has been reported to decrease (Chan-Roper et al., 2012), and increase (Derrick et al., 2002), while some studies have reported no significant change (Abt et al., 2011; Dierks et al., 2010; Dierks et al., 2011). This

suggests that the increase in knee flexion due to exhaustion is most important at heel-strike to absorb the impact forces.

Other effects of a run in an exerted state on knee kinematics have been seen in the frontal and transverse planes. One study showed that runners had decreased maximum knee adduction during stance, meaning they had more of a valgus alignment, at the end of a typical training run (Dierks et al., 2011). However, there were no significant changes in knee adduction during stance in a similar study by the same group (Dierks et al., 2010). These investigations have also found an increase in knee internal rotation excursion, peak angle and peak velocity, as well as an increase in tibial internal rotation excursion and peak angle (Dierks et al., 2010; Dierks et al., 2011). All of these kinematic trends have been suggested to cause or exacerbate symptoms of PFP.

Hip mechanics

The same group that investigated knee mechanics in the frontal and transverse planes also looked at hip mechanics in those planes. The results showed no effect of a typical training run on maximum hip internal rotation or maximum hip adduction (Dierks et al., 2010; Dierks et al., 2011). However, there was a significant effect of the run on maximum hip adduction velocity (Dierks et al., 2010), and hip internal rotation excursion (Dierks et al., 2011). Greater hip internal rotation is part of an alignment that may be associated with a risk for PFP.

Runners with PFP

A study examining the effects of a typical training run on joint kinematics showed that a PFP group had, in general, lower peak angles and maximum velocities than a control group, even for variables thought to cause or exacerbate PFP. This includes

eversion, components of knee valgus, and internal rotation of the tibia, knee and hip. It is hypothesized that these kinematics, which are opposite from what was expected, are due to a pain reduction mechanism employed by the PFP group. The PFP subjects may have tried to reduce their motion in the direction of poor mechanics to avoid pain. This seemed to be successful at the start of the run when the subjects did not report feeling pain, however, by the end of the run, there was an increase in motion which coincided with an increase in pain. Similarly, the PFP subjects had less peak knee flexion than the controls. Reduced knee flexion is thought to reduce patellofemoral compressive forces and therefore reduce pain (Dierks et al., 2011).

Coordination and variability

Fatigue studies may give an indication of how certain muscles contribute to the coordination patterns during running. Tibialis posterior has a role in controlling rearfoot eversion, and selectively fatiguing this muscle has shown changes in joint coordination and an increase in joint coordination variability. This may be due to less control of joint movement because fewer muscles are being used to produce the movement. The reduced function of tibialis posterior may also lead to increased activation of other inverters, resulting in the observed increase in joint coordination variability (Ferber & Pohl, 2011). However, these results are for a walking study, and selective fatigue of tibialis posterior is not common under normal walking or running conditions. Thus, the results may not be generalizable to running injury mechanics. Other studies have examined the effects of running-induced fatigue on coordination, with conflicting results. Uninjured runners had no change in joint timing after a prolonged run (Dierks et al., 2010). Joint coordination patterns were unchanged following an exhausting run for ITBS subjects and healthy

controls, but the injured group exhibited a decrease in joint coordination variability (Miller et al., 2008). Runners with overuse knee injuries that had run with custom foot orthotics for six weeks showed a decrease in joint coordination variability during the course of a 30-minute run while shod but without the orthotics. There was no change in variability when these subjects ran for 30 minutes with their orthotics (MacLean, van Emmerik, & Hamill, 2010). Additionally, sprinters performing repeated sprint bouts with decreasing rest periods exhibited decreased coordination variability during stance phase (Trezise, Bartlett, & Bussey, 2011).

Waveform Analysis

Traditional biomechanics studies include analysis of kinematic, kinetic and electromyographic signals to help solve clinical problems. The complexity of human movement typically results in the use of biomechanical models that focus on specific aspects of the kinetic chain (Daffertshofer, Lamoth, Meijer, & Beek, 2004; Deluzio, Wyss, Costigan, Sorbie, & Zee, 1999). This requires identification of the relevant features of gait and statistical analysis prior to collecting the data. In order to reduce the vast amount of data that is collected when analyzing gait patterns, peak values or excursions are commonly reported. Coordination patterns have been established and investigated in an effort to describe the interactions of segments during movement. However, clinical relevance has mostly been found when comparing these patterns during distinct phases of the gait cycle, as mentioned above. Additionally, relative phase techniques are limited to determining the coordination between pairs of joints or segments, not complex multi-joint coordination (Forner-Cordero, Levin, Li, & Swinnen, 2005).

Principal Components Analysis (PCA) is a multivariate analysis technique used for pattern recognition among time series by extracting common sources of variation among data, and describes variability as either random or deterministic. It is an unbiased way to determine the relevant changes in coordination patterns as a result of various interventions or conditions, including injury, rehabilitation or environmental changes, which could give insight into the mechanisms of injury or the effectiveness of treatments. PCA provides these differences as a result of the analysis, rather than deciding a priori which dependent variables and events in a time series will be the most important (Daffertshofer et al., 2004).

Application to movement disorders

PCA has been used to determine discriminating factors of gait between various groups. When compared to a parameterization study using the same dataset, by investigating the magnitude and pattern of the waveforms, PCA found several temporal characteristics of gait that discriminate between age groups (Chester & Wrigley, 2008). Similarly, PCA was used to identify differences in lifting kinetics and kinematics in subjects that went on to develop lower back pain compared to subjects that remained healthy. These differences were not identified using traditional biomechanical methods of analysis (Wrigley, Albert, Deluzio, & Stevenson, 2005; Wrigley, Albert, Deluzio, & Stevenson, 2006). Studies investigating potential risk factors for ACL injury have used PCA to identify gender differences in movement patterns and muscle activation patterns during cutting maneuvers. These differences had not been identified using discrete variables, and may explain the gender bias in noncontact ACL injury (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007b; Landry, McKean, Hubley-Kozey, Stanish, &

Deluzio, 2007c; O'Connor & Bottum, 2009). PCA was also used to identify gender differences in the gait patterns of females and males with knee osteoarthritis (McKean et al., 2007). In addition, differences in the gait patterns of subjects with knee osteoarthritis and healthy controls, as well as the portion of the gait cycle where the difference occurs, were identified using PCA (Deluzio, Wyss, Zee, Costigan, & Sorbie, 1997; Deluzio & Astephen, 2007; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007a). The differences in gait patterns detected by PCA for knee osteoarthritis patients before and after unicompartmental arthroplasty were shown to be clinically relevant (Deluzio et al., 1999).

Variability

PCA may also be used to determine the relationship between pathological gait and variability (Daffertshofer et al., 2004). Dynamical systems theory predicts that greater movement variability may be protective of overuse running injuries by allowing a runner to adapt to changing environments and by preventing the same tissue from being stressed with every foot-strike (Hamill et al., 1999; Heiderscheit, 2000). Variability in joint coordination is typically obtained by calculating the standard deviation of the movement coordination pattern (Hamill et al., 1999). However, this describes all of the variability for a given subject or set of subjects. PCA extracts the main modes of variation in a data set and identifies the principles that cause variation in gait patterns. PCA is used to separate biological variability from random variability or noise. This information could be used to test the dynamical systems theory that greater variability is a positive component of gait with regard to injury risk (Daffertshofer et al., 2004; O'Connor & Bottum, 2009).

This method of evaluating the inter-trial variability indicated significant gender differences in the sagittal and frontal plane moments as subjects completed a cutting task. Using discrete kinematic and kinetic variables, there were no gender differences in variability. However, using a PCA approach, males exhibited greater variability than females. This may indicate why females are more susceptible to ACL injury than males (O'Connor & Bottum, 2009).

Limitations of PCA

A limitation of PCA is that the values obtained during PCA cannot be intuitively applied to clinical measurements or parameters. However, PCA may be helpful in determining what set of discrete variables adequately describe the differences identified during the waveform analysis (O'Connor & Bottum, 2009).

There are limitations to treating data as multivariate rather than functional. With multivariate analysis, each time point is treated as a separate variable, so measurements for each subject should be taken at the same time points. This implies that the time-ordering of the data is not accounted for in multivariate analyses, and so measurements at different time points could be exchanged without altering the results (Coffey, Harrison, Donoghue, & Hayes, 2011).

Summary

Running is a common mode of exercise, which is important to maintaining good health. However, about half of all runners will sustain a running-related injury in a given year. PFP is the most common running injury, though its cause is unclear and likely multifactorial. Exposure to many impact forces over the course of a run or many runs is suspected to play a role in most overuse running injuries. These impact forces may be

especially harmful if combined with improper mechanics. Large patellofemoral joint contact forces are thought to be a cause of PFP, and knee mechanics that contribute to these forces, such as increased internal knee abduction moment and excessive knee valgus, are considered faulty. Factors that may put a runner at risk for PFP include training errors, muscle imbalances in the quadriceps and at the hip, improper footwear and overpronation. PFP may also be caused by uncoordinated movement patterns within the lower extremity. Additionally, runners with less variable movement patterns may be at greater risk for stressing the same tissue repeatedly during running.

As runners challenge themselves with increases in intensity, distance or both, they become fatigued or exhausted. There is much debate about how and where fatigue occurs, and it is possible that there is not one all-encompassing model of fatigue. The process by which a muscle becomes fatigued may have both central and peripheral factors, and is thought to be task-dependent. The best protocol for inducing fatigue in runners would be one that closely mimics a typical bout of exercise for a runner, while also providing an objective measure of fatigue for all participants. However, many different exhaustion or maximum fatigue protocols have been used, and not all of these studies used objective, physiologically measured criteria of fatigue or exhaustion. The different running durations and intensities may affect runners differently. The effects of fatigue may contribute to runners' risk of injury by altering mechanics such as impact forces, stride parameters, rearfoot eversion, tibial internal rotation, knee flexion, knee adduction, knee internal rotation, and hip internal rotation. Joint coordination patterns may also be affected by running to exhaustion.

Traditional investigations in biomechanics have focused on discrete variables, including peak forces, peak angles and excursions. Joint coordination has been quantified using discrete and continuous methods. All of these approaches require an a priori decision about which dependent variables and events in a stride cycle will be the most important to consider. PCA is an unbiased way to determine relevant differences in coordination patterns and variability among time series. It can be used to detect changes in joint coordination patterns and variability as a result of an exhausting run. This could give insight into the mechanisms of running injuries, for instance PFP.

CHAPTER 3: METHODS

The purpose of this research was to determine the effects of running in an exerted state on lower extremity joint coordination patterns using waveform analysis. Participants ran at their typical training pace until they reached a state of exertion measured by 85% of age-calculated maximum heart rate and a score of 17 (very hard) on the rating of perceived exertion (RPE) scale, or until they asked to stop. Running mechanics were recorded for ten strides before and after the run. Using waveform analysis to identify differences in lower extremity joint coordination patterns during the course of a run in an exerted state may provide insight about the cause of running injuries, and lead to effective measures for prevention.

Participants

Sixteen female recreational runners were recruited for this study through fliers posted on the University of Wisconsin-Milwaukee campus and with local running organizations. Sample size estimations were based on a repeated measures MANOVA design with a medium effect size ($\eta_p^2 = 0.2$), 80% power and $\alpha = 0.05$ (Heiderscheit, Hamill, & van Emmerik, 2002). Participants were screened through a background questionnaire that assessed inclusion and exclusion criteria. Females between the ages 18 and 45 were required to have run a minimum of 15 miles per week for the past six months and be classified as low cardiac risk. Exclusion criteria included any current lower extremity pain or running-related injury that limited training within the past six months, any history of major surgery to the lower extremity, the use of orthotics, pregnancy, medical conditions or medications that could impair balance, or a forefoot-strike running pattern (Dierks, Davis, & Hamill, 2010; Dierks, Manal, Hamill, & Davis,

2011). Additionally, participants were asked to refrain from running in a race in the 48 hours prior to testing and all exercise in the 24 hours prior to testing. Information about the participants, including their typical running habits, is included in Table 1.

Table 1
Information about participants.

Participant Characteristics	**Mean**	**SD**
Height (m)	1.65	0.05
Mass (kg)	58.4	7.0
Age (yr)	25	7
Shoe size	8	1
Typical running time (min)	39	10
Typical running distance (miles)	4.5	1.3
Typical running pace (min/mile)	8.85	0.93
Typical weekly mileage (miles)	24	11

Instrumentation and Equipment

Data collection took place during one testing session in the Neuromechanics Laboratory at the University of Wisconsin-Milwaukee. Three-dimensional kinematic data were collected at 200 Hz with a ten-camera Eagle system (Motion Analysis, Inc., Santa Rosa, CA) and ground reaction forces were recorded at 1000 Hz using an AMTI force plate (OR6-5; Advanced Mechanical Technology Inc., Watertown, MA). The participants were fitted with a heart rate monitor (Polar Electro Inc., Woodbury, NY), and the warm-up and run to an exerted state took place on a treadmill (C964i; Precor, Woodinville, WA) with the participants wearing their own training shoes. During the data collection before and after running to an exerted state, participants ran in lab shoes (NBA-801; New Balance, Brighton, MA; mean size 8 ± 1) for standardization purposes. This is a heel-less shoe to allow for direct observation of rearfoot motion.

Experimental Protocol

Retroreflective markers were applied to the subjects' skin to track the motion of the pelvis, thigh, leg and foot. The tracking markers were placed on the left and right ASIS and PSIS, a four-marker plate on both the thigh and the leg, and a marker triad attached to the calcaneus. A standing calibration was recorded with additional calibration markers on the left and right iliac crests and greater trochanters, and lateral and medial femoral epicondyles, malleoli and first and fifth metatarsal heads of the right leg. The calibration markers were removed following a three-second standing calibration. The subjects had a five-minute warm up period on the treadmill which consisted of light jogging at 2.2 m/s. Each participant's pace for the data collection and the treadmill run was self-selected based on their typical pace for a training run (Dierks, Davis, & Hamill, 2010; Dierks, Manal, Hamill, & Davis, 2011).

Participants performed ten successful running trials at their self-selected pace, ± 5%, in the lab shoes across a 15-m runway containing an embedded force plate. A successful trial was defined as when right leg initial contact and toe-off occurred on the force plate. Kinematic and kinetic data were collected for each trial. Then the participants ran on the treadmill at their self-selected pace in their own training shoes and without the retroreflective markers attached. To mimic the participants' typical training run, they were permitted to listen to music via headphones, if they desired. Starting in the first minute of the run and at every five minutes during the run, the participants' heart rate and RPE were recording. When the participants reached a state of exertion measured by at least 85% of age-calculated maximum heart rate (ACSM, 2010) and a score of at least 17 (very hard) on the RPE scale (Borg, 1998; Dierks et al., 2010; Dierks, Manal,

Hamill, & Davis, 2011), they continued running for an additional two minutes before ending the run. Their final heart rate and RPE were recorded before they stopped the treadmill. Immediately at the end of the run, participants switched into the lab shoes and the tracking markers were reapplied. The participants performed ten successful running trials overground as kinematic and kinetic data were collected. After recording the running trials, the calibration markers were reapplied and a three-second standing calibration was recorded for the post-run markers. Participants were allowed to perform a cool-down run on the treadmill at the conclusion of the data collection.

Data Reduction

The kinematic data were filtered using a 4^{th} order, zero-lag, recursive Butterworth filter with a cutoff at 12 Hz. Segment coordinate systems were anatomically-based and follow the right hand convention. During the standing calibration the pelvis, thigh, leg and foot coordinate systems were established. The x-axis pointed laterally, the y-axis pointed anteriorly and the z-axis pointed superiorly. Calculation of hip, knee and ankle joint angles was done using a joint coordinate system approach (Grood & Suntay, 1983). Ankle dorsiflexion/plantar flexion was defined as ankle motion in the sagittal plane (A_S), rearfoot in/eversion was ankle motion in the frontal plane (A_F), foot movement on the leg, or ankle in/external rotation, was ankle motion in the transverse plane (A_T), knee flexion/extension was knee motion in the sagittal plane (K_S), knee ab/adduction was knee motion in the frontal plane (K_F), leg movement on the femur, or knee in/external rotation, was knee motion in the transverse plane (K_T), hip ab/adduction was hip motion in the frontal plane (H_F), and hip in/external rotation was hip motion in the transverse plane (H_T). All kinematic data were time normalized to 100% of stance phase (101 data

points). The kinematic data processing was done using Visual3D software (v4.75.34; C-Motion, Inc., Rockville, MD).

The following calculations were performed with MATLAB (v8.0.0.783; Mathworks, Inc., Natick, MA). Vector coding is a technique that captures the relative motion of two joints for every point in the time series (Hamill, Haddad, & McDermott, 2000). With the proximal joint angle on the x-axis and the distal joint angle on the y-axis, each point in the time-series was plotted. A vector was made between consecutive points, and the orientation of the vector relative to the right horizontal was called the coupling angle. The range of coupling angles was -180° to 180°. The coupling angle can be interpreted as relative motion of the proximal and distal joints. To create a continuous waveform, the absolute value of the coupling angle at each time point in stance phase was used; therefore the range of coupling angles was 0° to 180°. This resulted in a loss of known directionality of each joint, but the joint with the greatest motion can be identified. A coupling angle of 0° or 180° indicates the distal joint is fixed and the proximal joint is in motion, while a coupling angle of 90° indicates that the proximal joint is fixed and the distal joint is in motion. A coupling angle of 45° or 135° indicates equal magnitude of motion of the proximal and distal joints. A coupling angle between 45° - 135° indicates more distal joint motion than proximal, while the opposite is true for coupling angles between 0° - 45° and between 135° - 180°.

The coupling angle was plotted as a function of time in the stride cycle. This was done for eight coupling patterns: ankle in/external rotation-rearfoot in/eversion (A_TA_F), rearfoot in/eversion-knee flexion/extension (A_FK_S), rearfoot in/eversion-knee in/external rotation (A_FK_T), rearfoot in/eversion-knee ab/adduction (A_FK_F), ankle

dorsiflexion/plantar flexion-knee flexion/extension (A_SK_S), rearfoot in/eversion-hip in/external rotation (A_FH_T), rearfoot in/eversion-hip ab/adduction (A_FH_F), and knee in/external rotation-hip in/external rotation (K_TH_T).

PCA was used to assess the variability of these coordination patterns over the waveform of stance phase before and after the run (O'Connor & Bottum, 2009; Wrigley, Albert, Deluzio, & Stevenson, 2006). Matrices for each coupling angle waveform were created. The individual trials populated *n* rows, and the 101 data points populated *p* columns in an X_{nxp} matrix. Eigenvector analysis of the covariance matrix $S_{101x101}$ determined the eigenvector matrix, $U_{101x101}$, by orthonormalizing $S_{101x101}$. The eigenvectors were the coefficients for the principal components (PCs) which represented the original data in new coordinates. The coefficients were the direction cosines that related the new axes to the old axes and are considered one mode of variation describing the variability within the entire original data set. The eigenvalues, L_{1x101}, were determined by $U'SU = L_{1x101}$. The eigenvalues represented the relative contribution or the rank of each PC to the total variation. A principal component score, Z_{nxp}, was calculated for each individual waveform by multiplying the individual trial's variation from the mean of all the trials, $\bar{x}_{1x101}$, by the transpose of the eigenvector matrix: $Z_{nx101} = (X_{nx101} - (1_{nx1} \text{ x } \bar{x}_{1x101})) \text{ x } U'_{101x101}$. The principal component scores represented the distance from each waveform to the mode of variability described by each principal component. The Z_{nx101} matrix was reduced to only the principal component scores that represent the primary modes of variation. The number of principal components retained was determined by using a SCREE plot. A SCREE analysis was performed on an equal-sized matrix of randomly generated numbers, and PCs that contributed modes of

variation in the original data that were greater than the modes of variation in the random numbers were retained. The variance not explained by the retained PCs represented random error. Within-subject variability was obtained by calculating the standard deviation of the principal component scores for the 10 trials of each subject.

The dependent variables for the PCA technique were the 10-stride standard deviation of the retained principal component scores for each coordination pattern before and after the run. For comparison, a traditional vector coding analysis of the joint coordination variability was done. The between trial mean and standard deviation of the coupling angle was calculated using circular statistics (RW.ERROR - Unable to find reference:577; MacLean, van Emmerik, & Hamill, 2010). Due to changing functional demands of the lower extremity during stance phase, rather than averaging the standard deviation of the coupling angle across the entire stance phase, a more sensitive analysis was performed by dividing stance phase into four periods (Period 1: 0-25%, Period 2: 25-50%, Period 3: 50-75%, Period 4: 75-100%), and averaging the standard deviation of the coupling angle over each period (Ferber & Pohl, 2011). The dependent variables for the traditional analysis were the 10-stride standard deviation of the coupling angle for each period of stance phase and each coordination pattern before and after the run.

Statistical Design and Analysis

For the PCA technique, each of the eight coordination patterns had several dependent variables, depending on the number of retained PCs for that waveform. Therefore, a repeated measures MANOVA for each waveform was done on the 10-stride standard deviation of the PC scores, for a total of eight MANOVAs. Pre and post exertion was the independent variable. If there was overall significance for a

coordination pattern, each individual PC that was used to describe the waveform was evaluated for differences pre and post exertion. In the traditional analysis, each of the eight coordination patterns had four dependent variables, one for each period of stance phase. A repeated measures MANOVA for each coordination pattern was done on the four dependent variables, for a total of eight MANOVAs. Pre and post exertion was the independent variable. If there was overall significance for a coordination pattern, each period of stance phase was evaluated for differences pre and post exertion. Statistical significance was determined at $\alpha = 0.05$ for all analyses. All statistical analyses were performed using SPSS (v19.0.0.1; SPSS, Inc., Chicago, IL).

CHAPTER 4: RESULTS

This research examined the effects of running in an exerted state on lower extremity joint coordination variability using waveform analysis. Heart rate and Rating of Perceived Exertion (RPE) were monitored during the run to ensure participants reached a high level of exertion. The run was designed to mimic each participant's typical training experience. Kinematic data were collected before and after the run. A vector coding technique was used to determine joint coupling angles, and Principle Components Analysis (PCA) was used to determine variability in joint coupling angles before and after the run. This technique was compared to a traditional method for determining variability in joint coupling.

Participants ran at a mean speed of 3.0 m/s (SD: 0.3) for a mean time of 39 minutes (SD: 19) until they reached the end criteria (Table 2). All participants reached one of the two stopping criteria (heart rate greater than 85% of age-calculated maximum, and RPE greater than 17) before ending the run. In seven cases, the participant asked to end the run when only one of the criteria was met. For the other nine participants, both criteria were met before ending the run (Table 3). There was no obvious effect of this difference in end criteria on the joint coordination variability observed before and after the run.

Table 2
Experiment information for individual participants and group means and standard deviations.

Participant code	Treadmill speed (m/s)	Time of run (min)	HR at start of run (% of max)	HR at end of run (% of max)	RPE at start of run (6-20)	RPE at end of run (6-20)	RPE after post run data collection (6-20)	Time of post run data collection (min)
P01	3.0	17	48.9	87.4	13	18	15	8.37
P02	3.0	27	54.3	88.4	9	17	12	8.50
P03	3.0	48	82.2	87.1	10	18	10	18.50
P04	3.4	32	74.0	91.5	8	18	--	7.80
P05	3.1	42	69.8	82.9	12	18	--	9.75
P06	3.7	77	68.2	82.1	7	17	12	15.77
P07	2.5	36	53.5	81.7	8	18	--	19.88
P08	2.8	45	52.4	83.2	7	17	13	11.50
P09	3.1	42	56.0	88.0	9	16	17	8.75
P10	2.7	17	62.7	98.3	7	18	12	14.50
P11	3.0	37	76.2	86.1	11	17	11	8.60
P12	2.9	37	61.2	84.2	9	19	18	9.50
P13	2.7	90	48.0	86.6	8	16	13	6.92
P14	3.1	17	80.7	91.4	11	18	18	7.30
P15	3.6	32	71.7	93.9	10	18	--	7.92
P16	3.3	27	59.2	91.1	9	18	15	10.72
Mean	3.0	39	63.7	87.7	9	18	14	10.89
SD	0.3	19	10.9	4.5	2	1	3	3.94

Table 3
Number of participants that reached each end criterion.

End Criteria	Number
HR $\geq$ 85% and RPE $\geq$ 17	9
HR $\geq$ 85%, asked to end	2
RPE $\geq$ 17, asked to end	5

The mean joint angles of all trials and all subjects for the ankle, knee and hip in the sagittal, frontal and transverse planes were plotted as a function of time (Figure 1). These kinematic data were collected and used to create a relative motion plot for each coordination pattern (Figure 2). The coupling angle for each coordination pattern was determined as described in the Methods section and plotted as a function of stance phase (Figure 3).

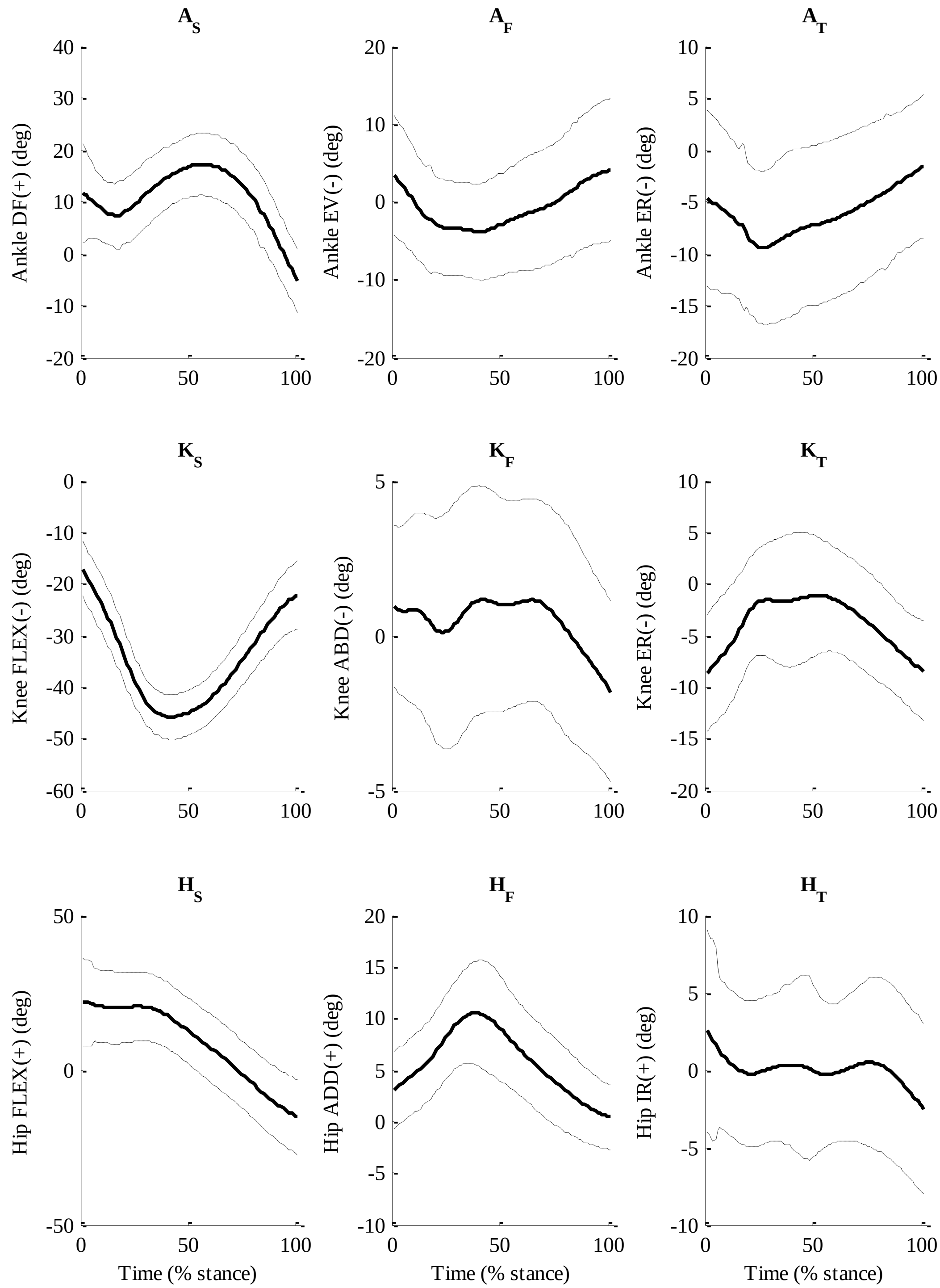

Figure 1. Mean (black line) individual joint angles, plus and minus one standard deviation (dashed lines) for the ankle, knee and hip in the sagittal, frontal and transverse planes.

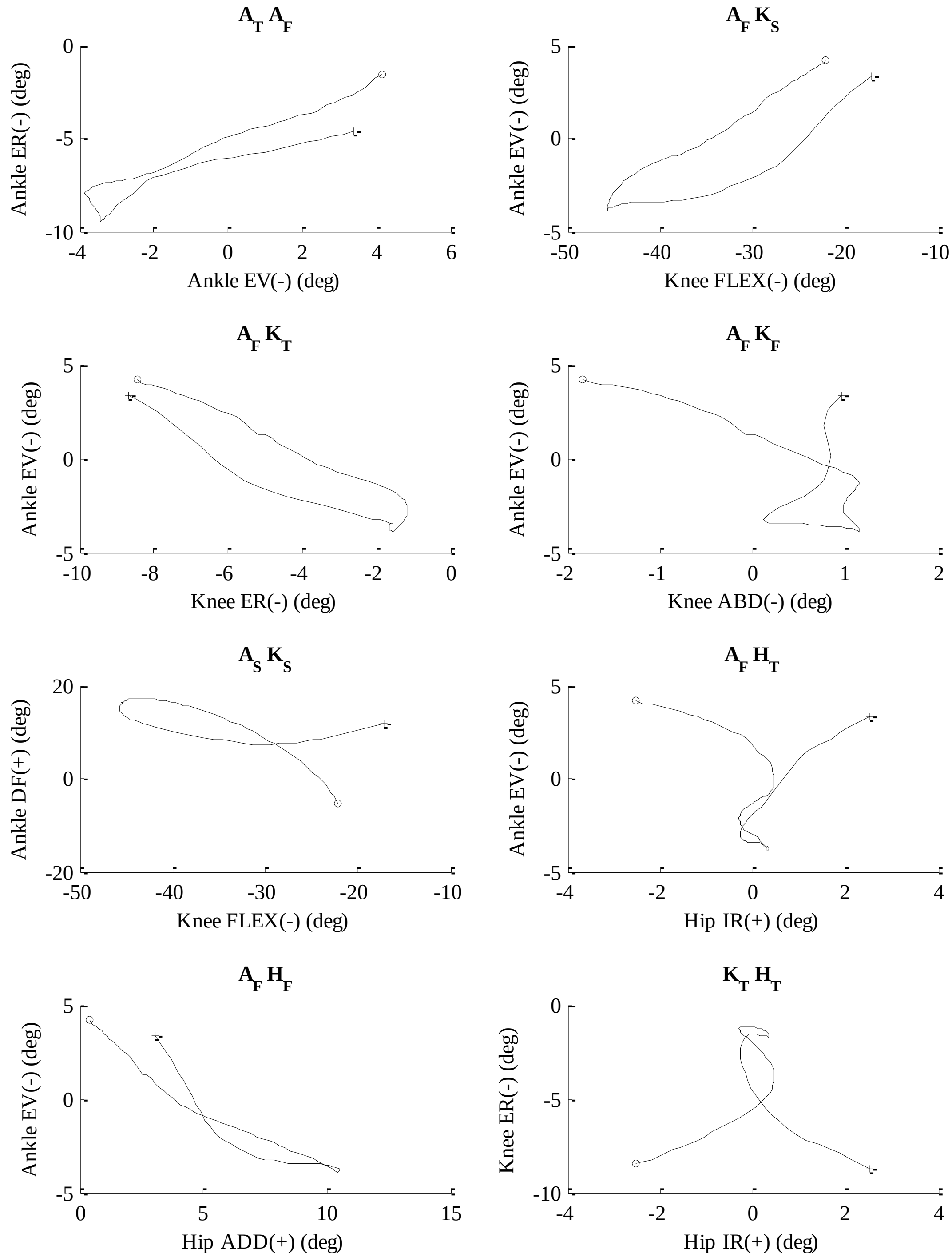

Figure 2. Relative motion plots for each coordination pattern during stance phase (heel-strike: +, toe-off: o). Data is the mean of all trials of all participants.

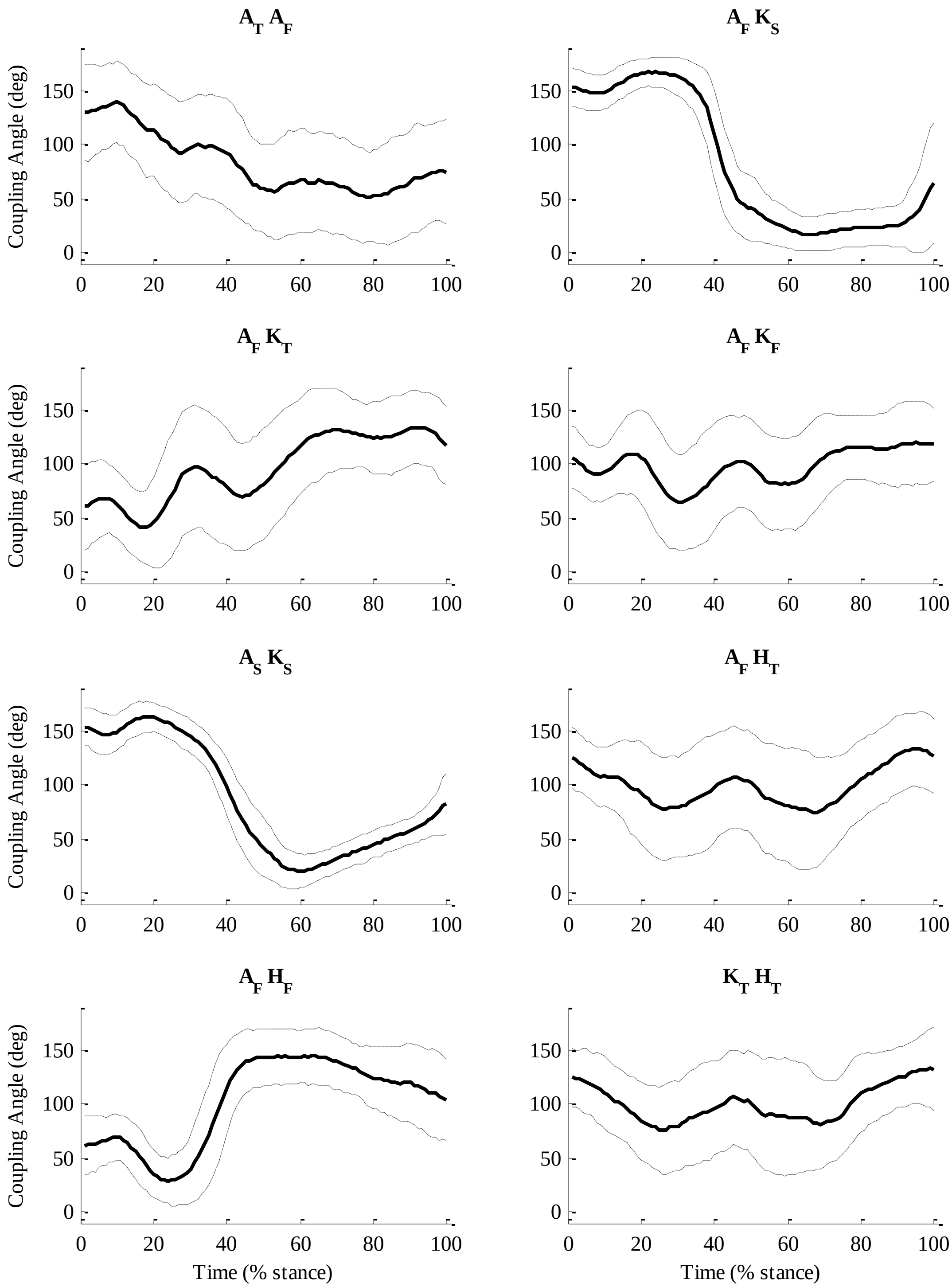

Figure 3. Mean coupling angle (black line) for all trials and participants during stance phase, plus and minus one standard deviation (dashed lines), for each coordination pattern.

PCA was done on the waveforms for all trials of each coordination pattern. The percent of the variance explained by the retained PCs ranged from 76% to 85%, (Table 4). For the $A_F K_T$ coordination pattern, the mean coupling angle was plotted along with ±1 standard deviation of the scores for the first three retained PCs (Figure 4). The first PC described the most variation (30%) in the relative knee transverse plane and ankle frontal plane motion, which occurred in midstance, between 30-60% of stance phase. PC2 detected variations in the relative ankle eversion and knee internal rotation in the first 10% of stance and from 20-50% of stance, which accounted for 19% of the total variation. Twelve percent of the total variation was represented by PC3 as a fluctuation in the relative ankle inversion and knee external rotation during late stance, from 60-100% of stance phase.

Table 4
Percent of variance explained for the retained PCs.

	PC1	**PC2**	**PC3**	**PC4**	**PC5**	**PC6**	**PC7**	**PC8**	**PC9**	**PC10**	**Total**
$\mathbf{A_T A_F}$	16	13	11	9	7	6	6	4	3		76
$\mathbf{A_F K_S}$	28	15	14	9	6	4	3	3			81
$\mathbf{A_F K_T}$	30	19	12	7	6	5	3	3			85
$\mathbf{A_F K_F}$	23	20	14	9	8	4	3	3			84
$\mathbf{A_S K_S}$	24	14	12	9	8	5	4	3	3		81
$\mathbf{A_F H_T}$	35	18	11	9	6	4	2				85
$\mathbf{A_F H_F}$	22	19	12	7	5	4	4	4	3	2	82
$\mathbf{K_T H_T}$	35	17	11	8	6	5	3				84

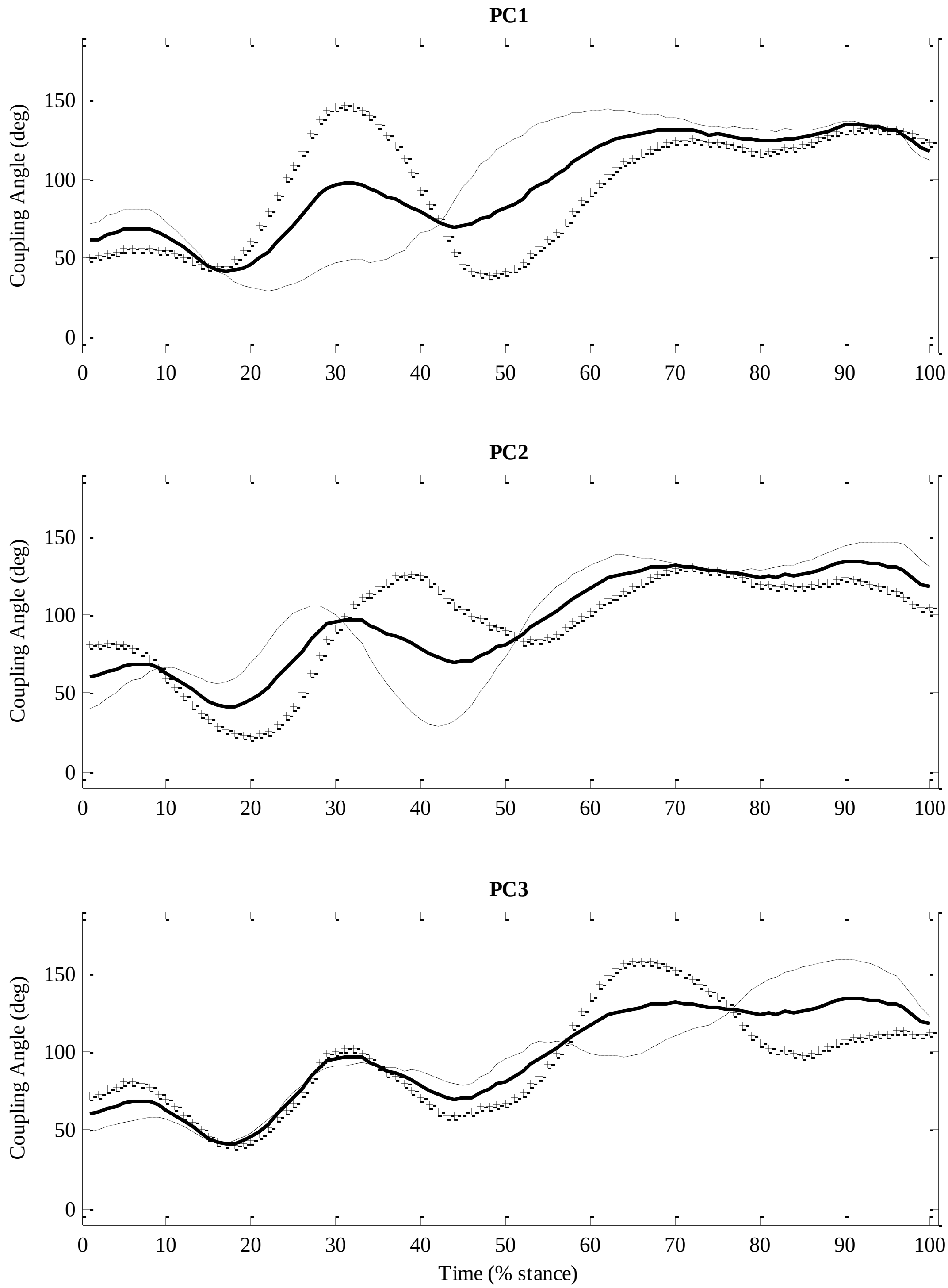

Figure 4. The plus one standard deviation (+) and minus one standard deviation (-) contributions of the scores for the first three retained PCs of the mean $A_F K_T$ coupling angle (black line).

The standard deviation of the PC scores for each retained PC was calculated before and after the run (Table 5). A repeated measures multiple analysis of variance (MANOVA) was performed on the standard deviation of the PC scores for each coordination pattern (Table 6). No significant differences between before and after the run were present for any of the eight coordination patterns, therefore no post-hoc analyses were done.

Table 5
Mean (SD) of the standard deviation for the retained PC scores before (Pre) and after (Post) the run.

		PC1	**PC2**	**PC3**	**PC4**	**PC5**	**PC6**	**PC7**	**PC8**	**PC9**	**PC10**
A_TA_F	Pre	94.2 (33.1)	98.7 (31.2)	92.3 (22.4)	108.6 (45.3)	95.6 (24.4)	75.0 (24.9)	86.4 (22.6)	70.6 (25.6)	69.7 (18.9)	
	Post	82.9 (30.7)	91.6 (34.9)	79.9 (35.7)	88.5 (38.6)	85.6 (25.0)	79.7 (39.4)	80.3 (34.4)	67.7 (26.5)	61.4 (27.3)	
A_FK_S	Pre	40.2 (24.2)	45.7 (14.9)	44.9 (17.2)	34.6 (11.3)	34.5 (10.6)	31.8 (8.7)	31.4 (10.5)	29.1 (7.1)		
	Post	42.4 (23.8)	54.4 (17.6)	38.0 (17.6)	42.9 (15.5)	32.7 (17.6)	29.2 (9.2)	35.2 (10.8)	27.2 (9.2)		
A_FK_T	Pre	95.4 (30.4)	85.4 (30.5)	67.9 (19.5)	62.9 (30.5)	64.7 (20.2)	52.6 (24.9)	51.0 (20.1)	40.8 (17.2)		
	Post	94.2 (41.5)	87.1 (30.7)	59.2 (23.0)	54.2 (20.6)	58.7 (23.2)	48.9 (18.1)	50.2 (17.5)	51.0 (23.3)		
A_FK_F	Pre	105.8 (51.2)	75.7 (28.7)	54.8 (21.7)	60.5 (24.3)	64.2 (21.4)	47.6 (12.0)	50.4 (17.2)	46.4 (15.8)		
	Post	90.6 (40.2)	76.0 (37.6)	48.9 (16.6)	57.8 (21.7)	64.0 (31.3)	44.8 (18.6)	43.4 (14.8)	46.7 (24.7)		
A_SK_S	Pre	41.7 (12.3)	29.4 (9.6)	33.8 (12.6)	28.5 (10.6)	28.2 (8.9)	25.4 (9.1)	26.3 (8.9)	19.6 (4.9)	23.0 (6.3)	
	Post	40.7 (15.4)	29.3 (10.5)	29.7 (9.1)	29.9 (11.6)	28.5 (12.3)	23.2 (10.7)	25.1 (8.9)	23.4 (11.4)	26.6 (10.4)	
A_FH_T	Pre	96.7 (44.5)	76.4 (28.3)	90.4 (36.7)	77.6 (37.3)	64.0 (18.9)	52.9 (21.5)	53.3 (25.7)			
	Post	88.3 (42.6)	75.6 (27.5)	73.0 (34.8)	70.3 (27.1)	54.0 (23.2)	48.1 (12.9)	48.9 (18.9)			
A_FH_F	Pre	57.6 (27.2)	60.3 (26.7)	62.0 (21.2)	53.8 (17.5)	47.4 (19.0)	38.6 (16.1)	51.9 (23.5)	44.4 (15.2)	35.4 (13.2)	37.8 (14.4)
	Post	56.5 (47.6)	51.6 (23.0)	53.6 (29.4)	52.0 (20.9)	45.5 (21.4)	45.1 (20.4)	41.3 (14.8)	42.8 (15.1)	39.4 (10.1)	38.5 (15.2)
K_TH_T	Pre	77.2 (36.1)	64.8 (32.2)	77.9 (32.7)	65.3 (16.2)	54.9 (26.8)	46.4 (16.5)	48.8 (21.2)			
	Post	66.3 (45.4)	70.5 (29.0)	62.8 (26.7)	58.3 (25.5)	55.9 (23.4)	43.4 (18.5)	40.3 (10.5)			

Table 6
Results for the eight MANOVAs performed for the variability of the scores for each waveform.

	Wilk's Lambda	**F**	**p**	η_p^2
A_TA_F	0.811	3.489	0.081	0.189
A_FK_S	0.985	0.228	0.640	0.015
A_FK_T	0.987	0.201	0.660	0.013
A_FK_F	0.948	0.825	0.378	0.052
A_SK_S	1.000	0.001	0.976	0.000
A_FH_T	0.895	1.761	0.204	0.105
A_FH_F	0.983	0.257	0.619	0.017
K_TH_T	0.915	1.388	0.257	0.085

For comparison purposes, differences in joint coordination variability before and after the run were investigated using traditional methods. The standard deviation of the coupling angles for each period of stance phase was calculated before and after the run (Table 7). Each period corresponds to 25% of stance phase. A repeated measures MANOVA was performed on the standard deviation of the coupling angles for each quartile of stance phase of each coordination pattern (Table 8). No significant differences between before and after the run were present for any of the eight coordination patterns, therefore no post-hoc analyses were done.

Table 7
Mean (SD) of the standard deviation for the coupling angle for each quartile of stance phase before (Pre) and after (Post) the run.

		Period 1	**Period 2**	**Period 3**	**Period 4**
A_TA_F	Pre	27.7 (8.7)	32.4 (8.2)	31.9 (10.6)	25.7 (10.7)
	Post	23.5 (7.6)	30.1 (8.7)	27.8 (12.7)	26.0 (11.8)
A_FK_S	Pre	9.1 (4.2)	16.1 (2.5)	11.3 (4.4)	11.1 (5.5)
	Post	10.1 (4.3)	16.6 (4.3)	12.0 (5.0)	12.7 (6.2)
A_FK_T	Pre	18.7 (5.2)	26.4 (4.7)	23.5 (5.8)	17.5 (5.0)
	Post	18.5 (6.4)	27.0 (7.3)	19.0 (6.2)	16.8 (5.2)
A_FK_F	Pre	17.5 (4.1)	25.0 (5.0)	25.8 (7.3)	16.6 (4.8)
	Post	17.1 (5.0)	25.3 (6.7)	20.2 (7.3)	16.1 (5.3)
A_SK_S	Pre	9.2 (3.3)	13.4 (3.7)	10.1 (3.9)	5.7 (3.9)
	Post	10.2 (3.4)	11.7 (4.6)	10.8 (4.5)	6.5 (4.5)
A_FH_T	Pre	18.7 (5.2)	27.5 (8.1)	23.9 (7.5)	19.6 (7.7)
	Post	18.0 (4.5)	26.8 (6.6)	19.5 (5.8)	17.2 (6.9)
A_FH_F	Pre	14.4 (4.8)	21.6 (6.9)	17.8 (4.3)	18.5 (7.1)
	Post	16.1 (4.8)	20.0 (7.3)	16.0 (6.2)	17.4 (6.9)
K_TH_T	Pre	16.2 (6.1)	24.2 (7.5)	20.8 (10.0)	16.8 (4.9)
	Post	15.8 (4.9)	23.0 (6.1)	14.9 (7.0)	15.0 (7.8)

Period 1 = 0-25%, Period 2 = 25-50%, Period 3 = 50-75%, Period 4 = 75-100% stance

Table 8
Results for the eight MANOVAs performed for the inter-trial variability for each quartile of stance phase.

	Wilk's Lambda	**F**	**p**	**η_p^2**
A_TA_F	0.843	2.800	0.115	0.157
A_FK_S	0.914	1.419	0.252	0.086
A_FK_T	0.938	0.990	0.335	0.062
A_FK_F	0.913	1.431	0.250	0.087
A_SK_S	0.991	0.138	0.716	0.009
A_FH_T	0.892	1.816	0.198	0.108
A_FH_F	0.986	0.208	0.655	0.014
K_TH_T	0.828	3.111	0.098	0.172

CHAPTER 5: DISCUSSION

The purpose of this study was to examine the influence of running in an exerted state on lower extremity joint coordination variability using waveform analysis. The primary hypothesis of this study was that running in an exerted state would lead to a decrease in joint coordination variability, determined by Principal Components Analysis (PCA). This hypothesis was not supported, as no significant differences in joint coordination variability before and after the run were found for any of the eight coordination patterns. The secondary hypothesis that PCA would reveal different results than a traditional analysis of changes in joint coordination variability was also not supported since neither method was able to detect significant differences pre- and post-run. These results suggest that there is a weak or no relationship between joint coordination variability and the level of exertion experienced by healthy runners during their typical run.

Dynamical systems theory suggests overuse running injuries may be associated with a decrease in movement variability, which would prevent a runner from adapting to changing environments and would expose the same tissue to stress with every foot-strike (Hamill, van Emmerik, Heiderscheit, & Li, 1999; Heiderscheit, 2000). While the current study cannot directly establish this link since only healthy subjects were examined, previous authors have suggested that neuromuscular fatigue may be linked to injury by reducing movement variability (MacLean, van Emmerik, & Hamill, 2010; Miller, Meardon, Derrick, & Gillette, 2008; Trezise, Bartlett, & Bussey, 2011). The results of this body of literature have been mixed, with some evidence supporting changes in

movement variability with fatigue, using both vector coding and CRP methods of analysis.

The results of the eight MANOVAs, one for each coordination pattern, performed for the inter-trial variability for each quartile of stance phase did not reveal a difference in joint coordination variability after a run in an exerted state. This result is somewhat consistent with studies that have investigated changes in joint coordination and variability after exertion. MacLean et al. (2010) showed no changes in joint coordination variability of a control group from the beginning to the end of a 30-minute run. However, an injured group that had been running with orthotics for at least six months exhibited a decrease in joint coordination variability at the end of a 30-minute run when they were shod but not wearing their orthotics. Trezise et al. (2011) did show a decrease in joint coordination variability after running in an exerted state, but these results may not be generalizable to typical training runs: only two subjects were used in the study, and they performed repeated sprint bouts with decreasing rest periods, rather than a prolonged run at their typical training pace. Another study examining walking kinematics after selective tibialis posterior fatigue actually showed an increase in joint coordination variability, due to less control of joint movement or increased activation of other inverters as a result of the tibialis posterior fatigue (Ferber & Pohl, 2011). However, walking kinematics and selective fatigue of tibialis posterior is also not generalizable to typical training runs. Overall, these results suggest that the effects of a typical training run may not be enough to cause a decrease in joint coordination variability in healthy runners that is detectable by traditional measures.

It was hypothesized that by extracting common sources of variation among data and avoiding a priori identification of dependent variables, PCA would reveal differences in joint coordination variability as the result of a typical training run that could not be observed using traditional analyses. However, this hypothesis was not supported. For each coordination pattern, 7-10 PCs were retained and only 76-85% of the total variance was explained by the retained PCs. The variance explained by the first PC, which is the greatest mode of variation in each coordination pattern, was only 16-35% of the total variance. These numbers are very different from the PCA results of running kinematics for studies investigating gender differences in cutting tasks, where the first three or four retained PCs accounted for 94-100% of the total variance of the kinematic data (Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007a; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007b; O'Connor & Bottum, 2009). Additionally, O’Connor & Bottom (2009) showed that the first PC of the kinematic data accounted for 73-81% of the total variance in the data. The low percentage of variance explained in the current study suggests that there was no major factor influencing the variance in joint coordination. As a result, PCA did not reveal a decrease in joint coordination variability after a run in an exerted state, as was hypothesized

It is possible that running in an exerted state that is similar to a typical training run does not evoke a significant change in joint coordination variability for healthy runners. If the theory that decreased variability represents a risk for injury is valid, the results of this study indicate that healthy runners will not increase their risk for injury by participating in their typical training run. While fatigue effects on individual joint kinematics were observed in this study (Appendix E), these changes do not appear to

influence the joint coordination variability after a typical run. This suggests that healthy runners are able to maintain the same range coordination patterns throughout their typical running experience.

The fact that neither the traditional or PCA method of quantifying joint coordination variability demonstrated significant changes after a run in an exerted state indicates that a link between fatigue and movement variability in a healthy population may be weak or non-existent. However, choices in conducting the study may have also affected the ability to detect changes in joint coordination variability due to exertion. Factors such as the exercise protocol, the choice to conduct overground data collection trials, and analysis techniques could have affected the outcome of the study.

The exercise protocol, slightly modified from the procedure used by Dierks et al. (2010; 2011), was designed to mimic the participants' typical training run. Participants ran for about the same time and pace as their self-reported information, which suggests that the results should indicate what occurs ecologically during running. The time of the run and treadmill speed in the current study was similar to these measures in similar studies (Dierks et al., 2010; Dierks et al., 2011; MacLean et al., 2010). Though the speed was similar to the self-selected pace used by participants in a study by Miller et al. (2008), the mean time of the run was longer in the current study. These similarities suggest that the level of exertion reached by runners in the current study is similar to the level of exertion reached in experiments with procedures also designed to mimic a runner's typical training run. All participants also reached a level of exertion that was at least 85% of their age-calculated maximum heart rate or a rating of at least 17 on the RPE scale, or both, which suggests all participants reached a similar level of exertion.

However, there was no metabolic measure of fatigue during this study, and participants may have varied in the level of exertion they reached. If the level of exertion was not high enough, differences before and after the run would be difficult to detect.

Another factor affecting the level exertion reached by the runners could be differences between treadmill and overground running, since overground running was analyzed but the participants ran in an exerted state on a treadmill. It has been shown that the kinematics and kinetics are similar for overground and treadmill running at the same speed (Riley et al., 2008). However, the perception of speed during treadmill running is faster than overground running (Kong, Koh, Tan, & Wang, 2012). Therefore, participants may have perceived that they were running faster on the treadmill, and rated their RPE higher than they might have if they were performing the run in an exerted state overground. Anecdotally, several participants commented that they felt the treadmill run was more difficult than their typical overground training runs.

After the run, participants changed shoes and tracking markers needed to be reapplied. Due to some markers falling off during the post-run data collection, this took longer for some participants than others, and the mean time from the end of the run to the end of the post-run data collection was 10.89 minutes (SD: 3.94). During this time, participants may have recovered from their run as they were not continuously running in an exerted state prior to the final data collection. While great care was taken to minimize this data collection time window, this may account for a lack of differences between the pre- and post-run joint coordination variability.

To assess whether the current study’s baseline kinematic results are representative of typical running behavior, the joint angle time series were compared to previous

running studies. The plots of the mean joint angles resemble the shape of the frontal and transverse plane angles for the ankle, knee and hip, and the sagittal plane knee angle reported by Dierks et al. (2010) in a study comparing kinematics at the beginning and end of a run in an exerted state. The sagittal plane ankle and hip angles appear similar to those reported by Ounpuu (1994). The only differences in these plots are related to differences in marker placement between the studies, which would not influence the joint coordination variability. This suggests that joint kinematics for the participants in this study are similar to those of previous studies that investigated running gait. Therefore, the joint coordination and variability measures that are based on these kinematics should be considered representative of typical running gait.

Coordination was quantified through the use of vector coding. Vector coding is a technique used to determine joint coordination and variability between joints. It is a method that provides a measure of coordination across the time series, which allows for a more robust assessment than methods that simply compare the timing of discrete gait events. The joint coupling angle and variability are dependent on how this approach is employed. It is recommended that the distal joint angle be plotted on the y-axis and the proximal joint angle on the x-axis, and the mean and standard deviation of the relative angle for multiple trials should be calculated using circular statistics (Hamill, Haddad, & McDermott, 2000). Using an arctangent function, it is possible to determine the coupling angle with a range from -180° to 180° (or 0° to 360°) at every point for each individual trial, and plot that as a function of time (Heiderscheit, Hamill, & van Emmerik, 2002). Due to the circular nature of the coupling angle, this does not produce a continuous waveform, as -180° is the same angle as 180°, but they are not continuous in the time

function. For the current study, the absolute value of the coupling angle (range 0° to 180°) had to be used to create a continuous waveform for the PCA, which influenced the joint coordination and joint coordination variability.

The reduced range of angles eliminated the ability to account for directionality within the coupling angle. For example, while both -45° and 45° represent equal relative of motion of the proximal and distal joints, -45° represents proximal motion in the positive direction and distal motion in the negative direction, and 45° represents both proximal and distal motion in the positive direction. That directionality was lost when the absolute value of the coupling angle was used.

Nevertheless, the coupling angle results from the current study can be compared with those from studies that have reported the coupling angle as a function of time. For example, the relative angle plot and the plot of coupling angle over stance phase for A_FK_S can be compared to the stance phase portion of the same plots reported by Heiderscheit et al. (2002). That study used a range of 0° to 360° for the coupling angle, but if considered on a 0° to 180° scale, those plots appear similar to the plots in the current study for A_FK_S. Similarly, the plot of the coupling angle over stance phase for A_TA_F can be compared with the same plot produced by Ferber et al. (2005), where the range of the coupling angle was further reduced to 0° to 90°. When considered on this scale, the plot reported by Ferber et al. (2005) appears similar to the one in the current study for A_TA_F.

Several studies, including this one, are most interested in the inter-trial variability of the coupling angle. As such, they may publish just the standard deviation of the coupling angle, rather than plots or information about the coupling angle itself. Differences in how the coupling angle and variability are represented make it difficult to

compare results across studies. The variability of the coupling angle is affected by the range of the coupling angle. Reducing the range of the coupling angle to 0 to 180°, from -180° to 180° (or 0° to 360°), will reduce the coupling angle variability. For example, coupling angles of -6° and 5° would be represented as 6° and 5° once the absolute value was applied, thus reducing the variability between them. Additionally, the variability of the coupling angle is traditionally averaged across sub-periods of stance or swing phase to make the data more manageable and to give functional meaning to the data. However, there is not a standard set of periods used in every study, making comparisons between studies difficult.

In this study, the joint coordination variability for the traditional (non-PCA) method was determined as the standard deviation of the coupling angle over four periods, each representing 25%, of stance phase. These results can be compared with other studies that have reported similar measures, though a direct comparison is not possible if different periods of stance phase were defined. The variability of A_TA_F for each period of stance was greater than that reported in studies by Ferber et al. (2005; 2011) and MacLean et al. (2010). Compared to the variability over the entire stride cycle reported by Heiderscheit et al. (2002), the variability of A_FK_S and A_SK_S for each period of stance was greater, while the variability of A_FK_T and K_TH_T for each period of stance was about the same. One possible reason for the reduced variability reported by Ferber et al. (2005; 2011) is the 0° to 90° range of coupling angle, compared to the range of 0° to 180° used in the current study. Based on this argument, the variability in the current study should be less than that reported by Heiderscheit et al. (2002) and MacLean et al. (2010), as they used a coupling angle range of 0° to 360°. However, Heiderscheit et al. (2002) reported

variability over the entire stride cycle and not individual periods of stance, and MacLean et al. (2010) used periods of stance that were different from the quartiles used in the current study. Additionally, both of those studies reported the variability of consecutive footfalls while running on a treadmill, while the current study reported the inter-trial variability of one stance phase during ten overground running trials. Other studies, using non-vector coding methods, have also examined joint coordination and variability between consecutive footfalls as participants ran on a treadmill (Dierks et al., 2010; Miller et al., 2008). An additional study has looked at joint coordination variability of consecutive footfalls for sprinters during overground running (Trezise et al., 2011). It is possible that the inter-trial variability is greater than the variability of consecutive strides. This may have washed out any changes in variability that resulted from the run in an exerted state.

A post-hoc analysis of the kinematic data was performed to compare the results to other studies that assessed kinematic differences in gait before and after a run in an exerted state. Discrete variables (mean and standard deviation of the angle at heel-strike, the peak angle, and the excursion from heel-strike to the peak angle) were also determined for each joint and plane of motion (Appendix E).

Previous studies have reported an increase in maximum rearfoot eversion and eversion excursion during stance (Derrick, Dereu, & McLean, 2002; Dierks et al., 2010; Dierks et al., 2011; VanGheluwe & Madsen, 1997). While no differences in peak eversion angle were detected in the current study, there was a significant increase in eversion excursion after the run. The increase in eversion excursion in this study is due

to a significantly greater inversion angle at heel-strike after the run, which is consistent with the results presented by Derrick et al. (2002).

Knee motion in the sagittal plane is thought to be related to absorbing impact forces during running. While some studies have reported an increase in knee flexion angle at heel-strike after an exhausting run (Derrick et al., 2002; Mizrahi, Verbitsky, Isakov, & Daily, 2000; Nicol, Komi, & Marconnet, 1991), the results of this study are consistent with those that report no changes in knee flexion at heel-strike (Collins et al., 2000; Hanley, Smith, & Bissas, 2011; Williams, Snow, & Agruss, 1991). There was, however, a decrease in peak knee flexion and knee flexion excursion after the run. One study has shown a decrease in knee flexion during stance, and that was during the course of a marathon (Chan-Roper, Hunter, Myrer, Eggett, & Seeley, 2012). Yet, another study has shown an increase (Derrick et al., 2002), while some studies have reported no significant change (Abt et al., 2011; Dierks et al., 2010; Dierks et al., 2011) in knee flexion during stance after shorter runs. The runs in the studies by Derrick et al. (2002) and Abt et al. (2011) were designed to be shorter runs at a fast pace. Failing to match the patterns of the discrete variables in these studies suggests that the participants in the current study did not reach the same level of exertion as those who ran at a faster pace. Rather, they matched the pattern of runners in a marathon.

There was a decrease in hip adduction excursion after the run in this study. This was not found in other studies that examined hip mechanics in the frontal and transverse planes (Dierks et al., 2010; Dierks et al., 2011). A previously-observed increase in hip internal rotation excursion after the run was not seen in this study (Dierks et al., 2010; Dierks et al., 2011). The runners in the current study completed a run that was modeled

after the procedure employed by Dierks et al. (2010; 2011). However, inconsistencies between these discrete joint measures suggest that the runners in this study did not reach the same level of exertion as those in the Dierks et al. (2010; 2011) studies.

There was a decrease in the variability of knee internal rotation excursion after the run. This was the only significant change in variability observed for an individual joint in this study. The lack of further differences in variability of these kinematic variables could be a reason why there were no changes in joint coordination variability from before the run to after the run.

Future Research

Future research on this topic should use a metabolic measure of fatigue to ensure all participants reach the same level of exertion at the end of the run. Additionally, completing the post-run data collection immediately after the run, or during the run, would be ideal to avoid recovery of the participants before the data is collected. A comparison of inter-trial variability and the variability of consecutive footfalls would be useful to determine if both methods are adequate for quantifying joint coordination variability.

Investigating the muscles that cause the kinematics changes during running may be useful in determining the variability of running patterns before and after a run. EMG could be used to record the muscle activity of the lower extremity, and PCA could be applied to the EMG signals to detect changes in muscle activation variability.

Since using PCA did not identify changes in joint coordination variability, it is unclear if the method is faulty or if there were no changes to detect from before the run to after the run. Changes in joint coordination variability have been limited, but were

shown to exist when examining an injured population (Heiderscheit et al., 2002; MacLean et al., 2010). This PCA technique could be used to look for changes in joint coordination variability in an injured population. Additionally, a prospective study could be done to see if PCA can predict if runners with lower joint coordination variability will become injured.

Summary

The experimental protocol caused participants to run to a level of exertion similar to that of their typical training run, therefore these results reflect ecological patterns of joint coordination variability in runners. No changes in joint coordination variability were observed for any of the eight coordination patterns using traditional analyses. PCA also did not identify differences in joint coordination variability. The way the kinematic data was processed to produce the coupling angle may have limited the observed variability of the coupling angle. Additionally, inter-trial joint coordination variability, measured in this study, may be different from the variability of consecutive footfalls. The low amount of variance explained by the retained PCs suggests that there is not one major factor that mediates the variance in the joint coordination data. It is possible that healthy runners do not experience a change in joint coordination variability during their typical training run.

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APPENDIX A: CONSENT FORM

Informed Consent | IRB Protocol Number: 13.257
Version: 1 | IRB Approval Date: 2/15/2013

UNIVERSITY OF WISCONSIN – MILWAUKEE
CONSENT TO PARTICIPATE IN RESEARCH

THIS CONSENT FORM HAS BEEN APPROVED BY THE IRB FOR A ONE YEAR PERIOD

1. General Information

Study title: The Effect of Fatigue on Intra-Limb Coordination Variability during Running Using a Waveform Analysis Approach

Person in Charge of Study (Principal Investigator):
The Principal Investigator (PI) for this study is Kristian O'Connor, PhD. Dr. O'Connor is a faculty member in the Department of Kinesiology. The co-PI on this study is Lauren Benson. Lauren is a Master's student in the Department of Kinesiology.

2. Study Description

You are being asked to participate in a research study. Your participation is completely voluntary. You do not have to participate if you do not want to.

Study description:

- The purpose of this study is to examine the influence of running in an exerted state on movement patterns of the ankle, knee and hip.
- Investigating these effects may help determine if running in an exerted state contributes to common overuse injuries. This information could be used to create better treatment options for, or ways to prevent, overuse running injuries.
- The study is being done at UW Milwaukee, where there will be 16 participants.
- Participants will be tested during one 1.5-hour session.

3. Study Procedures

What will I be asked to do if I participate in the study?
If you agree to participate you will be asked to go to the Neuromechanics Laboratory at UW Milwaukee (Enderis Hall, Room 132) for one testing session.

You will be asked to wear clothing appropriate for physical activity; however, clean, tight-fitting shorts and shoes will be provided for you during all testing sessions. The tasks you perform will include:

1. You will be given a questionnaire containing questions pertaining to previous lower extremity injuries and your ability to run to a high level of exertion on a treadmill. (10 minutes)
2. You will warm up at a low intensity on a treadmill. (5 minutes)
3. Markers will be applied to your pelvis and right leg at specific landmarks, and you will put on laboratory shoes. (2 minutes)

4. Baseline movement data will be collected during 10 running trials on a 15m raised runway with an imbedded force plate. (10 minutes)
5. The markers will be removed and you will put on your own shoes. (2 minutes)
6. You will run at your typical training pace on a treadmill until you reach a predetermined level of exertion. Your heart rate will be monitored continuously during the run, and you will be asked how your rate your perceived level of exertion at five-minute intervals. (45 minutes)
7. Markers will be applied again and you will put on laboratory shoes. (2 minutes)
8. Movement data will be collected during 10 running trials on a 15m raised runway with an imbedded force plate. (10 minutes)
9. You will be allowed to cool down as necessary. (5 minutes)

4. Risks and Minimizing Risks

What risks will I face by participating in this study?
Physical risks

- Muscle soreness as a result of the testing. (Unlikely)
- Injuries such as muscle strain or muscle tightness as a result of the testing session. (Unlikely)
- Minor skin irritation due to the spray tape adhesive or tape. (Unlikely)

Psychological, social, economic risks

- None

Protection of Physical Risks:
To reduce the above risks, practice trials will be performed prior to data collection to allow you to become familiar with each procedure prior to performing a maximal effort trial. If you feel any soreness or irritation while participating in this study, please tell the investigators as soon as possible. If you are injured while participating in this research study, you will initially be provided care by the investigator(s), who are all trained in first aid and CPR. Students will then be referred to the Norris Health Center for follow-up care. Non-students will be referred to their primary care physician and will be responsible for all expenses incurred.

Risks to Privacy and Confidentiality:
Since your private information will be collected for this study, there is always a risk of breach of confidentiality (less than 1%).

Protection of Risks to Privacy and Confidentiality:
All data will be stored in a locked filing cabinet in a locked room. All data will be given a letter and number that is uniquely associated with you. This code will not contain any partial identifiers (i.e. last four digits of your SSN) and will be stored in a separate locked office in a locked filing cabinet. No identifiers will be stored with the research data. Only those individuals with an active role in this study will have access to the research data and only the PI and Co-PIs will have access to identifying information. When all participants' have completed active participation in the study and data collection is completed, the code will be destroyed. All appropriate measures to protect your private information will be taken.

5. Benefits

Will I receive any benefit from my participation in this study?
The information which is obtained may be useful scientifically and possibly helpful to others.

6. Study Costs and Compensation

Will I be charged anything for participating in this study?
You will not be responsible for any of the costs from taking part in this research study. You are responsible for your own transportation to and from UWM and for any parking costs for the testing session.

Are subjects paid or given anything for being in the study?
You will be given $20 in cash for participating in this study.

7. Confidentiality

What happens to the information collected?
All information collected about you during the course of this study will be kept confidential to the extent permitted by law. We may decide to present what we find to others, or publish our results in scientific journals or at scientific conferences. Only the PI and co-PI will have access to the information. However, the Institutional Review Board at UW-Milwaukee or appropriate federal agencies like the Office for Human Research Protections may review this study's records.

The confidentiality of your data and information will be safeguarded as outlined in "Risks & Minimizing Risks" section under the "Protection of Risks to Privacy and Confidentiality" header.

8. Alternatives

Are there alternatives to participating in the study?
There are no known alternatives available to you other than not taking part in this study.

9. Voluntary Participation and Withdrawal

What happens if I decide not to be in this study?
Your participation in this study is entirely voluntary. You may choose not to take part in this study. If you decide to take part, you can change your mind later and withdraw from the study. You are free to not answer any questions or withdraw at any time. Your decision will not change any present or future relationships with the University of Wisconsin Milwaukee. If you choose to withdraw, we will use the information collected about you to that point. If you are a student, your refusal to take part in the study will not affect your grade or class standing.

10. Questions

Who do I contact for questions about this study?
For more information about the study or the study procedures or treatments, or to withdraw from the study, contact:

Kristian O'Connor, PhD
Department of Kinesiology
Enderis 471
P.O. Box 413
Milwaukee, WI 53201
414-229-2680

Who do I contact for questions about my rights or complaints towards my treatment as a research subject?
The Institutional Review Board may ask your name, but all complaints are kept in confidence.

Institutional Review Board
Human Research Protection Program
Department of University Safety and Assurances
University of Wisconsin – Milwaukee
P.O. Box 413
Milwaukee, WI 53201
(414) 229-3173

11. Signatures

Research Subject's Consent to Participate in Research:
To voluntarily agree to take part in this study, you must sign on the line below. If you choose to take part in this study, you may withdraw at any time. You are not giving up any of your legal rights by signing this form. Your signature below indicates that you have read or had read to you this entire consent form, including the risks and benefits, and have had all of your questions answered, and that you are 18 years of age or older.

____
Printed Name of Subject/ Legally Authorized Representative

____ ____________________
Signature of Subject/Legally Authorized Representative Date

Principal Investigator (or Designee)
I have given this research subject information on the study that is accurate and sufficient for the subject to fully understand the nature, risks and benefits of the study.

____ ____________________
Printed Name of Person Obtaining Consent Study Role

____ ____________________
Signature of Person Obtaining Consent Date

APPENDIX B: MEDICAL HISTORY AND PHYSICAL ACTIVITY QUESTIONNAIRE

IRB#: 13.257
Approval Date: 2/15/2013

Participant Code: _____
Date: _____________

Medical History & Physical Activity Questionnaire

Screening Criteria
Please answer the following questions to the best of your ability. Eligible participants will answer "yes" to these questions.

☐ Yes ☐ No Are you between the ages of 18 and 45 years old?

☐ Yes ☐ No Do you wear shoes between women's sizes 7 and 10?

☐ Yes ☐ No Are you currently a recreational runner (at least 15 miles a week for the past 6 months)?

☐ Yes ☐ No Do you run without orthotics?

☐ Yes ☐ No Have you refrained from running in a race in the 48 hours prior to testing and all exercise in the 24 hours prior to testing?

Medical History Questionnaire
For your safety, a list of conditions that would make you unable to participate in this study has been prepared. Please read this list carefully and consider whether any of the conditions apply to you. If any of these conditions are true for you, you will not be able to participate in this study. For each condition, please indicate "yes" or "no" if this is true or not for you.

☐ Yes ☐ No Do you have a medical condition that may impair your balance performance (i.e. concussion, neurological impairments, etc)?

☐ Yes ☐ No Are you taking medications/drugs that may make you dizzy or make you tired (i.e. cold medications, sleeping medications, muscle relaxants)?

☐ Yes ☐ No Have you had, in the last 6 months, a lower extremity injury that caused you to decrease the amount of physical activity you undertake?

☐ Yes ☐ No Do you currently have any lower extremity pain or injury(ies)?

☐ Yes ☐ No Have you ever had major orthopedic surgery on your lower extremities?

☐ Yes ☐ No Do you have a physical condition that would prevent you from being able to run at a high level of exertion on a treadmill for a time equal to your typical training run?

☐ Yes ☐ No Are you pregnant or do you have reason to believe that you may be pregnant?

Comments/Notes: ____

____

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APPENDIX C: RECRUITMENT FLYER

RUNNERS
NEEDED FOR RESEARCH STUDY

Title: The effect of fatigue on intra-limb coordination variability during running using a waveform analysis approach.

Purpose: To examine the influence of running in an exerted state on movement patterns of the ankle, knee and hip.

Who Can Participate?

- Female runners, ages 18 to 45
- Must not have any current injuries or pain in the low back, hips, or legs
- Must not have had an injury to the low back, hips or legs in the last 6 months
- Must have never had major surgery to the lower extremity
- Must currently be a recreational runner (at least 15 miles a week for 6 months)

What Would I Have To Do?

- One Testing Session (~1.5 hours)
    - Fill out questionnaires
    - Light warm-up
    - Baseline data collection
    - Run in an exerted state on a treadmill
    - Post-run data collection
    - Cool down

Compensation

- $20 Cash

In case of any questions or to volunteer, please contact:

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This research project has been approved by the University of Wisconsin-Milwaukee Institutional Review Board for the Protection of Human Subjects (IRB Protocol Number 13.257 approved on 2/15/2013)

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APPENDIX D: DATA COLLECTION WORKSHEET

Date				
Subject #				
Height		inches		m
Weight		Newtons		kg
Age				
Shoe size				
Running time		minutes		
Running distance		miles		
Running pace		min/mile		
Weekly milage		miles		
Music				

Heart rate		max
		85%

Treadmill speed		miles/h

Timing gate	fastest	slowest	
			s

Time of run (min)	HR	RPE
0		
5		
10		
15		
20		
25		
30		
35		
40		
45		
50		
55		
60		
65		
70		

End time		min
End criteria		
Post test time		min
Final RPE		

APPENDIX E: POST-HOC ANALYSIS

Table E1
Mean (SD) for discrete kinematic variables of the ankle, knee and hip in the sagittal, frontal and transverse planes.

	A_S			**A_F**			**A_T**		
	Pre	**Post**		**Pre**	**Post**		**Pre**	**Post**	
HS(°) mean	9.8 (8.2)	13.5 (7.6)		0.5 (5.7)	6.1 (7.5)	*	-4.1 (7.3)	-5.1 (9.5)	
HS(°) sd	4.2 (7.0)	2.2 (0.9)		3.3 (2.4)	2.5 (1.3)		1.8 (1.9)	1.6 (1.1)	
Peak(°) mean	16.8 (6.9)	20.0 (5.9)		-7.8 (5.6)	-4.0 (7.3)		-10.3 (6.4)	-11.2 (8.7)	
Peak(°) sd	3.7 (6.3)	1.6 (0.9)		2.6 (2.0)	2.6 (2.6)		1.4 (0.8)	1.4 (0.9)	
ROM(°) mean	6.9 (3.9)	6.5 (4.0)		8.3 (2.8)	10.1 (4.3)	*	6.2 (2.7)	6.0 (3.1)	
ROM(°) sd	3.1 (2.4)	2.2 (0.8)		2.4 (1.1)	3.5 (2.7)		1.9 (1.4)	2.1 (1.2)	
	K_S			**K_F**			**K_T**		
	Pre	**Post**		**Pre**	**Post**		**Pre**	**Post**	
HS(°) mean	-17.5 (4.6)	-16.9 (5.5)		0.5 (3.1)	1.4 (1.9)		-8.5 (6.1)	-9.1 (5.1)	
HS(°) sd	1.8 (0.5)	2.0 (0.7)		0.6 (0.2)	0.7 (0.3)		1.6 (0.6)	1.7 (0.6)	
Peak(°) mean	-47.0 (4.1)	-44.9 (4.3)	*	2.5 (3.6)	3.2 (2.3)		0.4 (4.8)	0.6 (5.8)	
Peak(°) sd	1.3 (0.5)	1.3 (0.6)		0.9 (0.5)	0.8 (0.3)		1.0 (0.3)	0.8 (0.2)	
ROM(°) mean	29.5 (4.3)	28.1 (4.6)	*	2.0 (1.7)	1.8 (1.4)		8.9 (3.2)	9.8 (2.5)	
ROM(°) sd	2.0 (0.6)	2.3 (0.7)		0.8 (0.6)	0.7 (0.4)		1.5 (0.6)	1.8 (0.6)	*
	H_S			**H_F**			**H_T**		
	Pre	**Post**		**Pre**	**Post**		**Pre**	**Post**	
HS(°) mean	23.2 (13.1)	20.7 (10.4)		2.7 (3.2)	3.3 (4.0)		1.7 (5.3)	3.5 (4.3)	
HS(°) sd	1.2 (0.4)	4.3 (11.7)		1.1 (0.4)	1.5 (1.1)		1.1 (0.5)	3.0 (6.2)	
Peak(°) mean	-13.5 (13.7)	-16.8 (10.6)		11.1 (4.9)	10.7 (5.3)		4.1 (6.4)	5.1 (4.1)	
Peak(°) sd	1.4 (1.0)	1.1 (0.3)		1.4 (1.9)	1.1 (0.7)		1.9 (3.3)	3.1 (6.2)	
ROM(°) mean	36.7 (6.0)	37.5 (6.5)		8.4 (3.7)	7.4 (3.0)	*	2.5 (4.3)	1.6 (1.4)	
ROM(°) sd	1.7 (1.1)	4.6 (11.6)		1.5 (1.9)	1.6 (0.9)		1.6 (3.4)	1.1 (1.1)	

*HS = heel-strike angle, Peak = peak angle, ROM = excursion from HS to Peak; * Significant Pre/Post effect, $P < 0.05$*