THE EFFECT OF MIDSOLE CUSHIONING ON THE KINETICS, KINEMATICS, AND CONTROL OF RUNNING

by

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DISSERTATION

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ABSTRACT

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The incidence of musculoskeletal injuries associated with the stress of running, known as running-related injuries (RRIs), has risen in conjunction with the resurgence of running as a health-related physical activity approximately half a century ago, with some reports giving an annual prevalence as high as 79% of practitioners surveyed. Running shoes with increased amounts of midsole cushioning have been promoted to reduce the risk of sustaining RRIs by absorbing some of the high magnitude and frequency ground reaction forces (GRFs) imposed on the leg with every stride. However, individual variability, as well as the ability of the motor system to acutely adapt gait mechanics based on the haptic experience of ground contact, have led to disagreement in the literature and lack of consensus among experts as to the actual benefit of cushioned running shoes, and injury rate remains unaffected.

Study 1 (Chapter 2) takes a more comprehensive and ecological approach to defining the influence of midsole cushioning on limb loading dynamics than employed by previous

investigations by capturing a large (50) number of footstrikes (FS) from a large (>50) sample of subjects of both midfoot (MF) and rearfoot (RF) strike patterns performing unconstrained overground running in shoes with diametrically opposite levels of midsole cushioning. Advanced statistical techniques were then used to separate the contribution of the shoe from that of the motor response towards limb loading characteristics. Minor but significant adjustments to the runners' landing strategy were identified between cushioning conditions that were specific to the runners' habitual FS pattern. Further, although impact-period forces were attenuated with increased cushioning in RF runners, MF runners actually experienced a gain in these variables, and runners of both FS patterns experienced higher peak knee forces and moments. The added cushioning in running shoes seems to promote a more knee-dominant running strategy, with higher reliance on translational vs. rotational force attenuation in the limbs.

Additionally, overuse injury is a result of *accumulated* tissue stress. The amount of variability in relative limb segment articulations, particularly in time periods of high-magnitude loading such as initial ground impact, directly relates to the homogeneity of structural loading profile and therefore potential for injury. Study 2 (Chapter 3) used established continuous relative phase (CRP) methodology to quantify the amount of variability in critical joint couplings of the lower limbs during running, and then compared this quantity of coordinative variability between barefoot, minimalist, and maximalist footwear conditions. Footwear was found to have no effect on the amount of coordinative variability during treadmill running in our healthy subjects, although joint coupling patterns during the impact period of stance was altered between conditions.

The findings of these investigations provide vital insight into the objective influence of cushioning in the midsole of running shoes. It is necessary for practitioners to be cognizant of the potential effect running shoe cushioning may have on their gait mechanics to best realize potential benefits of cushioned footwear. Future investigations should further explore long-term musculoskeletal adaptations to specific footwear types, particularly in structures of the foot. Longitudinal investigations may also be able to discern a relationship between individual levels of coordinative variability and musculoskeletal health prognosis.

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DEDICATION

This work is dedicated to my mother and father, for never telling me not to dream, and for their unending love, understanding, and support.

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INTRODUCTION

Background

Humans have a long history with running and a short history with the running shoe. In addition to being a natural progression in motor development, the fundamental skill of distance running has been identified as a pivotal mechanism for the success of our genetic lineage (Bramble, 2004). Recent decades have seen the resurgence of distance running for recreation and exercise in the general population, and concurrent with this expansion in participation is the increased incidence of musculoskeletal injuries directly attributed to performance of the activity. In fact, running is “one of the most widespread activities during which overuse injuries occur” (Hreljac, 2004). Although precise estimates are difficult to obtain, it is has been reported that over a 1-year period, up to 80% of practitioners sustain an overuse injury (van Gent, 2007). In addition to relatively novel personal and environmental risk factors (i.e. sedentary lifestyle and artificial surfaces), the repetitive, high magnitude and loading rate of ground reaction forces (GRFs) transmitted through the limb during stance have been implicated as primary factors contributing to the development of chronic injuries (Radin, 1976; Stanish, 1984; Novacheck, 1998; Nigg, 2001; Burr, 2003; Hreljac, 2004; Milner, 2006; Zadpoor, 2011), which typically occur at or distal to the knee (Novacheck, 1998; van Gent, 2007; Pohl, 2009; Zadpoor, 2011). These stress-related injuries, such as stress fractures, plantar fasciitis, medial tibial stress syndrome (shin splints), achilles tendinopathies, and joint degradation, have been linked in a large body of literature to the demands imposed on the tissues via energy attenuation during the stance phase of running gait (Milner, 2006; Pohl, 2009; Lieberman, 2012; van Gent, 2007; Zadpoor, 2011). In light of the evolutionary basis for running towards the development of modern human physiology, this high occurrence of injury related to running stress elicits reasonable consternation.

Shoes for Protection

Shoes with increased midsole cushioning, particularly underneath the heel, have been advocated to reduce the likelihood of practitioners sustaining a "running-related injury" (RRI). This viewpoint is based on the logical assumption that cushioning acts as a buffer between the foot and the ground to attenuate potentially harmful GRFs transmitted through the foot and leg during stance. However, despite a relatively large body of literature investigating the effect of shoe properties on running mechanics, disagreement in research findings, not to mention their lack of translation to effective risk-reduction practices, have persisted. This lack of practical understanding of the functional benefit of cushioned running shoes towards injury prevention is reflected in the unchanged or even increased incidence of RRI over the decades (van Gent, 2007; Dubois, 2015; Taunton, 2002), as well as the extremely disparate array of shoes and professional opinions marketed today.

The "impact period", occurring roughly within the first 50 ms of stance, contains majority of high-frequency GRF signal power, as well as peak tibial acceleration magnitudes (Shorten, 1993). Additionally, forces experienced in this range occur too quickly for feedback-dependent adjustments to be made, and therefore must be accommodated through feedforward muscular pre-tuning. These high-frequency forces are also attenuated predominantly through passive (bone and soft tissue) rather than active (muscular) structures (Munro, 1978; Shorten, 1992). However, peak translational and rotational forces in the legs occur around mid-stance, and are also highlighted as possible injurious mechanisms (Winter, 1983; Nigg, 2003). The purpose of the investigations herein, however, is not to debate the specific force characteristics and dosages leading to injury development, but rather to provide valid and comprehensive information on how cushioning in running shoe midsoles affects running strategy and limb loading characteristics, in an attempt to more clearly define the practical contributions towards effective risk-reduction practice.

Motor response

Controlled material testing analyses do show that increasing midsole thickness and decreasing midsole stiffness reduces peak impact forces and slows loading rate by absorbing energy and spreading the impulse over a larger time frame (Nigg, 1988). However, determinants of *in vivo* force attenuation are multifactorial. In addition to the material properties of the running shoe and running surface, GRF characteristics and their resulting structural loads are determined by the specific state (velocity, orientation, compliance, etc.) of the interacting components (body segments, shoe, surface) of the kinetic system. Further, it is known that feed-forward control of these "kinetic determinants" is acutely modulated in response to the somatosensory information garnered during previous footstrikes (Wakeling, 2001; Bishop, 2006; Addison, 2015). This is to say that runners naturally adjust their stride based on GRF sensations. These adjustments are thought to be made to maintain efficiency of locomotion, and preserve head stability across changes in compliance of the foot-ground interface (i.e. surface or shoe) (Ferris, 1998, 1999; Kerdok, 2003). It is these gait adjustments that are indicated to result in the maintenance of GRF variables despite changes to midsole cushioning properties.

When comparing shod to barefoot (BF) running, clear kinetic and kinematic differences are observed, and runners almost unanimously change from a rearfoot (RF) to a midfoot (MF) pattern. Interestingly, then, many studies assessing gait changes between shoe conditions of varied cushioning levels report a lack of significant differences in the vast majority of joint kinematics assessed (Nigg, 1987; Dixon, 2000; Bonacci, 2013; Chambon, 2014; Sinclair, 2016) although a flatter foot placement and more vertical shank is somewhat consistently observed with increased midsole cushioning. However, evidence that joint compliance is significantly affected by shoe midsole or ground surface compliance supports previous hypotheses of gait adjustments between shod conditions of varied midsole cushioning levels (Ferris et al, 1998, 1999; Bishop et al., 2006; Hamill et al, 2014). Together these findings suggest that muscle activation strategies modulating joint stiffness during loading periods may be a more predominant response mechanism to increased impact experiences, versus changing segment orientation alone (Dixon, 2000; Nigg, 2001). Additionally, they indicate that wearing shoes, regardless of their midsole properties, sufficiently reduces the haptic feedback from ground contact to preclude substantial alterations to the running pattern when comparing shod conditions (Tenbroek, 2011; Bonacci, 2013). This consideration is critical in light of research showing that simple and short term “gait training” interventions, as well as cognizant running without shoes, are acutely effective in eliminating transient impact peaks and reducing initial loading rate (ILR) and GRF magnitude (Cheung, 2014; Samaan, 2014).

Many previous investigations examining the effect of varying levels of cushioning on limb loading dynamics have reported that a higher amount of midsole cushioning does *not* result in significantly different external vertical force magnitudes (Nigg, 1987, 1988a, 1988b;

Wakeling, 2001; Hamill, 2011; Paquette, 2013; Sinclair, 2013), although most simultaneously identified changes to running mechanics that would result in decreased force attenuation capabilities of the system (i.e. runners have increased joint stiffness and more forceful landings in softer shoes, and vice-versa) (Wakeling, 2001; Hamill, 2011; Paquette, 2013; Sinclair, 2013). Despite this, the analyses performed in these investigations do not account for adjustments to motor output when comparing kinetic variables between shoe conditions. In an attempt to better delineate the role of shoe cushioning from that of the motor response in regards to GRF attenuation, running speed is typically controlled through treadmill (Butler, 2003; Divert, 2005; Bishop, 2006; Squadrone, 2009; Willy, 2014) or over-ground (Dixon, 2000; Hamill, 2011; Bonacci, 2013; Paquette, 2013; Chambon, 2014; Sinclair, 2016) methods. However, many of these studies still report main effects for kinetic variables concurrently with kinematic adjustments that may influence loading, and it is of course possible to maintain forward velocity while altering gait strategy. Additionally, there is a potential that this added level of control imposes a constraint on the natural motor response to the experimental condition and might confound the external validity of findings. After inconclusive results, Chambon et al. (2014) suggested a more “ecological” protocol would decrease limitations in discerning the effect of midsole cushioning. It is being argued herein that an over-ground running protocol provides a more globally relevant context for the topic of “running injury risk”, and that the subtle but significant differences in kinetics and kinematics between treadmill and over-ground running (Nigg, 1995; Riley, 2007) may diminish the efficacy of research translation to real-world interventions.

To simplify the problem, the critical perspective is this: How does the footwear (or more specifically, the material properties of the midsole) affect the way one runs, and what does the running shoe itself contribute to differences in limb loading characteristics? To answer this question it is critical to evaluate all of the above-mentioned influential factors concurrently when performing analyses. The protocol and analytical methods employed in Study 1 (Chapter 2) are designed to address these issues to more clearly delineate the contribution of midsole cushioning to force attenuation, while large subject samples and trial numbers will allow for the detection of any subtle but potentially meaningful changes to gait kinematics that have been previously implied (Frederick, 2011).

Mechanically distinct footstrike patterns

More recently a paradigm shift has begun based on the concept that human physiology is adapted to running barefoot or with minimal material under the foot, rather than in cushioned shoes. This view is supported by investigations into the kinetic and kinematic differences between shod and barefoot (BF) running, which show a reduction of the GRF loading rate and lack of a discernible impact peak with the reorganization of the runners' gait to a MF pattern. Alas, for most practitioners today it is widely impractical to adopt a truly barefoot running practice, but this evidence and justification has led to the promotion of minimalist shoes, supposed to allow for a more natural MF running pattern. Paradoxically, perhaps, analyses of running mechanics in minimalist shoes show that running in 'barefoot-inspired' footwear is not the same as running barefoot, and may increase RRI risk by allowing for more forceful landings through protection of the sensitive plantar surface of the foot, without providing the shock-absorbing benefits of cushioned midsoles (Bonacci, 2013). This is highlighted in the literature through a frequent lack of kinematic differences between shod conditions of varied midsole properties, while shod vs. barefoot comparisons more often than not report runners switching from their habitual footstrike (FS) pattern.

Although a useful reference for how running mechanics can be adjusted based on the experience of ground contact, findings from BF vs. shod comparisons cannot be directly related to the effect of varied midsole cushioning levels on limb loading dynamics. This is a critical consideration since limb loading dynamics are distinct between RF and MF landings due to differences in landing kinematics (De Wit, 2000; Bishop, 2006; Lieberman, 2010; Ahn, 2014) and shock attenuation strategies (Gruber, 2014; Hamill, 2014). This would indicate a differential cushioning effect provided by the shoe based one's habitual running strategy (Dubois, 2015). Therefore, conclusions on the shoe effect must stay within a given FS pattern, and the difficulty in delineating the benefits of running shoe cushioning is compounded when considering that *both* FS patterns are commonly employed in shod distance running. However, the vast majority of investigations which examine the shoe effect do so exclusively in habitual RF runners, and little research exists on the differential effect of cushioning in the running shoe between these distinct running styles. It is hypothesized that variations in midsole cushioning will influence gait kinematics and limb loading characteristics differently between MF and RF runners. Therefore, both study protocols included in this work will analyze habitual runners of both MF and RF footstrike patterns. This footstrike (FS) by footwear (FW) design will allow for direct comparisons to be made between footstrike pattern and cushioning levels for all measured dependent variables.

Shoes as a feedback constraint, and implications for stress overload

Models created through the observations from Study 1 give a more detailed representation of lower-extremity loading across *one* footstrike. However, due to the repetitive nature of ground impact in running, it is easily accepted that development of chronic injuries attributed to running stress results from an *accumulation* of these repetitive loads on the specific tissues in question (i.e. critical dose) (Hreljac, 2004; Brandt, 2009; James, 2004). With an average male runner's stride length of approximately 2.4 meters (Cavanagh, 1989; Mercer, 2002) equating to over 400 loading cycles for each leg every kilometer, this highly repetitive application of stress suggests that the uniformity of loading patterns may also be a critical factor in determining total tissue loads across an activity bout, and therefore one's potential for crossing the "injury threshold". This reasoning was conceptually outlined by James et al. (2004) and was termed the *variability-overuse injury hypothesis*. The rationale essentially posits that decreased variability in repetitive movement results in decreased distribution of tissue loading pattern, volume, and timing, therefore increasing the overall volume and homogeneity of loads to specific tissue sites, and in turn the probability of developing tissue overload. This framework for the etiology of RRIs is exemplified in the therapeutic literature relating to occupational injuries caused by repetitive tasks (Srivisanan 2012).

Motor coordination involves reducing all possible degrees of freedom of the motor constituents into functional synergies, or coordinative structures, in order to achieve the movement goal. However, biological systems are inherently noisy, and a certain amount of spatiotemoral variability inevitably persists in repetitive movement coordination (Davids, 2003).

Traditionally this variability has been viewed as error with skillful performance considered to have high consistency of motion across repetitions with minimal deviation from central reference points (standard deviation from the mean), making the statistical assumption that variability in movement is both random and independent. More contemporary investigations, though, have shown that variability between task repetitions is neither random nor independent, but is distinguishable from noise (Dingwell, 2000; Stergiou, 2004, 2011; Delignieres, 2009) and has a deterministic origin (Miller, 2006; Harbourne, 2009). The 'dynamical systems' motor control theory takes a nonlinear approach to movement analysis, and views variability as an important component in the control of cyclical motion, imparting both stability and flexibility to the motor system. Furthermore, the dynamical systems model integrates somatosensory feedback into the feed-forward control of successive repetitions, highlighting the influential role of perception-action coupling in successful movement execution (Magill, 2007; Palmer, 2012).

Using analytical tools developed by Kelso and colleagues (Schoner, 1988; Kelso, 1997), biomechanists are increasingly investigating the role of variability in the coordination of cyclical movement, including human gait. Evidence of this focus has demonstrated the functional role of variability in motor learning and skillful performance (Newell, 2001). What is more, there is a growing consensus supported by mathematical modeling and pathophysiological investigations (Heiderscheit, 2000; Stergiou, 2004; Hamill, 2012; van Emmerik) that a reduction in motor variability is indicative of disease or pathology. It is suggested that too much or too little variability in the coordination of the functional system leads to a "sub-optimal" state, resulting in decreased stability of the outcome parameter or, conversely, decreased flexibility of the system to respond to task demands. A broad but well-known example is that altered heart rate variability is associated with increased risk for future cardiac events (Denton, 1990; Goldberger, 1990; Tsuji, 1996; La Rovere, 1998). With respect to musculoskeletal coordination, disease populations such as parkinsonians demonstrate lower variability of relative pelvis-to-trunk motion during walking, and both parkinsonians and ACL-deficient individuals are shown to have lower variability of COM excursion magnitude and velocity during static postural control (Horak, 1992; Schieppati, 1994; van Emmerik, 1999; Davids, 2003). Other studies have shown similar 'rigidity' in the coordination of lower-limb segments during running in both ACL-deficient and symptomatic knee and back pain individuals (Heiderscheit, 2002; Moraiti, 2007; Seay, 2011). This is of particular interest due to the role of relative segmental motion in force attenuation during ground impact, as well as previous connections made between abnormal segment kinematics and injury risk (Bates, 1978; Stergiou, 1997, 1999; Stacoff, 2000; Powers, 2003; Kurtz, 2005).

Although it is difficult to identify a causal relationship between less variable states and injury, conclusions made by many leading investigators are suggesting that movement variability in running plays a functional role in the response of the functional system to the interactions of ground contact, and that a reduction of this coordinative variability may result in both a decreased ability of the system to respond to perturbations, as well as an increased consistency of structural loading, potentially resulting in an overuse situation (Hamill, 1999, 2012; Heiderscheit, 2000; Lipsitz, 2002; James, 2004). For example, degeneration of articular cartilage is associated with microscopic damage to the tissue accumulated through repetitive stressing (Hasler, 1999; Brandt, 2009), and more consistent loading patterns could exacerbate this reaction. In essence, higher variability results in *"a broader distribution of loads among different body tissues, distribution of* *loads among different locations within the same tissues, or a loading of the same tissues or locations at different times. Additionally, changes in the characteristics of the load due to variability might expose the affected area to a greater variety of force magnitudes, rates, and directions, thus potentially reducing or slowing the detrimental effects of repeated loading by permitting a longer adaptation time for tissues between loading events"* (James, 2004). Hamill et al. (2012) presented a mechanistic model of this effect, indicating an inverse relationship between the amount of coordinative variability and injury risk. However, although decreased variability is associated with injury and pathology, it is acknowledged that more evidence aimed at identifying causal factors in the reduction of coordinative variability is needed.

A relevant association within the previous investigations on functional variability in light of injury, pain, and other dyskinetic or morphological considerations is that proper coordinative control manifested as optimal system variability can be influenced by a disturbance of the afferent information integrated by the control system, potentially resulting in an "over-stable" state containing rigid, inflexible patterns (Wolpert, 1995; Hamill, 1999; van Emmerik, 2005; Moraiti, 2007; Stergiou, 2011). This relationship between somatosensory feedback and the execution of subsequent performance repetitions is not only implicit in the dynamical systems framework, but is also evidenced in biomechanical literature reporting alterations in gait mechanics with reductions to or alterations of proprioceptive information (Robbins, 1997; Galica, 2009; Palmer, 2012; Romer, 2013; Sterzing, 2013; Moore, 2014). Given that the material between the foot and the ground directly affects the nature and availability of the haptic information received during ground contact, it is hypothesized that increased midsole cushioning will result in an over-constraining of DOF within the functional structures, demonstrated as decreased stride-to-stride coordinative variability of lower-limb joint couplings. Despite the intended role of running shoes lowering stress-related injury risk, the effect of footwear on the variability of limb segment coordination through this action-perception framework has to date only been marginally investigated.

Ferber et al. (2005) have shown that orthotics can restore supposedly normal levels of variability in individuals with chronic knee pain, but this effect is only significant when pain is also ameliorated by the orthotic, making it difficult to determine the origin of the variability change. A between-subjects assessment of variability found habitually barefoot runners to be more variable than habitually shod runners (Altman, 2011), but effect of shoe cushioning cannot be directly assessed between these groups. Kurtz et al. (2003) have made initial steps to more directly elucidating the effect of midsole cushioning on coordinative variability, but (1) this analysis was only performed in RF runners, who do not rely as heavily on proprioception as do MF runners (Lieberman, 2012), (2) only single-joint motions were analyzed, as opposed to joint-coupling relationships more descriptive of segmental coordination, and (3) the 'spanning set' methodology that was used to quantify joint variability is controversial, whereas a continuous relative phase (CRP) approach is suggested to be a more appropriate means of analysis (Hanlon, 2016; Lamb, 2014). However, this study still reported reduced variability of knee and ankle motion in the shod running conditions when compared to barefoot. More recently, Tenbroek et al. (2011) manipulated shoe cushioning levels and used CRP analysis to quantify joint coupling variability, and in some comparisons found significant reductions in coordinative variability with increased cushioning. However, here again only RF runners were analyzed. Also, the shoe conditions employed exhibited different heel-toe wedge height, which is known to have an effect on running mechanics (Nigg, 1988; Lieberman, 2012).

Therefore, the purpose of Study 2 (Chapter 3) is to determine how coordinative variability during running, as measured by CRP analysis of joint coupling mechanics in a healthy, normally functioning system, is affected by the addition of different levels of midsole cushioning with similar wedge-heights. This effect will again be examined in both RF and MF runners to allow for more comprehensive conclusions. It is hypothesized that a reduction of coordinative variability initiated by the dampening of proprioceptive acuity may result in a state in which the biological structures are subject to a more repetitive, consistent stress and therefore may be at an increased risk of developing an overuse injury.

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Chapter 2

The influence of midsole cushioning on the kinetics and kinematics of running in midfoot and rearfoot runners

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Abstract

Despite the promotion of increased levels of cushioning in the running shoe midsole for reduction of running related injuries (RRIs), the clinical benefits of increased cushioning are tenuous, and confusion remains as to what this intervention contributes to mechanical load attenuation. Further, any effect is likely different between the mechanically distinct midfoot (MF) and rearfoot (RF). Therefore, in this investigation, 25 MF and 28 RF runners were recruited to perform 50 over-ground running trials in shoes with both minimalist and maximalist amounts of midsole cushioning, then stance-phase kinetics and kinematics were compared while statistically controlling for the runners' gait adjustments between conditions. MF and RF runners displayed group-specific adjustments to landing kinematics, with the most substantial being to knee joint compliance. High levels of cushioning primarily reduced impact loading rate ($p < 0.001$) and knee compression impulse ($p < 0.001$) in RF runners only. However, MF runners had a gain in these variables, and both groups experienced higher peak magnitude knee forces and moments when running in maximally cushioned shoes. Overall, midsole cushioning appears to promote gait mechanics more heavily reliant on passive translational versus active rotational energy attenuation, and a more knee-dominant running strategy.

Introduction

Within the last 4 or 5 decades an increase in the incidence of lower-extremity musculoskeletal injuries has been observed in conjunction with the resurgence of running for physical and mental well-being (Hreljac, 2004; van Gent et al., 2007). The high magnitude, rate and repetition of limb loading from ground reaction forces (GRFs) experienced during running have been implicated as primary factors contributing to the development of chronic injuries and joint deterioration (Burr & Radin, 2003; Hreljac, 2004; Milner, 2006; Nigg, 2001; Novacheck, 1998; Radin, 1976; Stanish, 1984; Zadpoor & Nikooyan, 2011), which typically occur at or distal to the knee (Novacheck, 1998; Pohl et al., 2009; van Gent et al., 2007; Zadpoor & Nikooyan, 2011). Shoes with increased midsole cushioning, particularly underneath the heel, have been advocated to reduce the risk of sustaining a "running-related injury" (RRI) (Nigg, 1988). However, despite a relatively large body of literature investigating the effect of shoe properties on running mechanics, disagreement in research findings, not to mention their lack of translation to effective risk-reduction practices, have persisted. This is evidenced by reports of unchanged or even increased incidences of RRI over the decades (Dubois, 2015; Lieberman, 2012; Taunton, 2002; van Gent et al., 2007).

Special attention in running research has been paid to the initial instances, or "impact" period, of the stance phase occurring within approximately the first 50 ms of ground contact. The high-frequency loads experienced in this time frame are faster than the quickest neuromuscular response, and therefore must be accommodated through muscular pre-activation prior to ground contact (Addison & Lieberman, 2015; Paul et al., 1978). They are also thought to be predominantly attenuated via passive (bone and cartilage) rather than active (musculotendinous units - MTU) structures (Chu et al., 1986; Gruber et al., 2014; Paul et al., 1978; Shorten & Winslow, 1992; Wosk & Voloshin, 1981). However, GRFs reach a maximum at mid-stance, and other researchers believe this time period may contribute more directly to injury risk (Dickinson et al., 1985; Winter, 1983). In addition to the attenuation of translational forces, moments generated through eccentric joint action have also received attention as a potential mechanism for MTU overload (Butler et al., 2003; McClay & Manal, 1999; Nigg et al., 2003; Scott & Winter, 1990). While cushioned running shoes (or surfaces) have been shown to reduce loading rate and tibial shock in some studies (De Wit et al., 2000; Dixon et al., 2000; Hamill et al., 2011; Lafortune et al., 1996; Willy & Davis, 2014), other investigations report no difference (Chambon et al., 2014; De Wit et al., 2000; Hamill et al., 2011; Laughton et al., 2003; Nigg et al., 1987; Paquette, 2013) or even an increase (Addison & Lieberman, 2015; Divert et al., 2005; Squadrone & Gallozzi, 2009) in forces experienced during the impact period. Further, peak forces and moments occurring at midstance have been observed to either increase (Bonacci et al., 2014; Divert et al., 2005) or not change (Bonacci et al., 2014; Braunstein et al., 2010; De Wit et al., 2000) when in cushioned running shoes. This lack of research consensus, along with the absence of clinical evidence for the benefit of midsole cushioning, highlights the confusion surrounding the capabilities of running footwear towards reducing stress-related injury risk.

Determinants of *in vivo* force attenuation are multifactorial. GRF characteristics and their resulting structural loads are determined by the specific state of the interacting components. It is known, however, that gait mechanics can be acutely adjusted by the motor control system in response to somatosensory information garnered during previous footstrikes (Addison & Lieberman, 2015; Bishop et al., 2006; Wakeling et al., 2001). Curiously, many studies assessing differences in landing kinematics between levels of midsole cushioning report a lack of significant effects in the majority of kinematic variables assessed (Bonacci et al., 2013; Chambon et al., 2014; Dixon et al., 2000; Sinclair et al., 2016). However, findings that joint compliance and peak joint moments are significantly different between midsole or surface conditions support previous hypotheses of gait adjustments in response to the material properties of the shoe midsole (Bishop et al., 2006; Ferris et al, 1998, 1999; Hamill et al, 2014). It has been proposed that a "muscular tuning" mechanism occurring prior to impact may explain how some kinetic variables may be maintained between shoe conditions (Dixon et al., 2000; Nigg & Liu, 1999) despite little or no difference in kinematics. When comparing shod to barefoot (BF) running, clear kinetic and kinematic differences are observed, and runners almost unanimously change their landing mechanics from a rearfoot (RF) to a midfoot (MF) pattern to accommodate for the heightened sensation of ground impact (Hamill et al., 2011; Lieberman et al., 2010; Lieberman, 2012). It is possible this response is only dampened when only transitioning between midsole cushioning levels.

In an attempt to control for the motor response, most investigations control running speed using treadmill (Bishop et al., 2006; De Wit et al., 2000; Divert et al., 2005; Squadrone & Gallozzi, 2009; Willy & Davis, 2014) or over-ground (Bonacci et al., 2013; Chambon et al., 2014; Dixon et al., 2000; Hamill et al., 2011; Paquette, 2013; Sinclair et al., 2016) methods. However, many of these studies still report main effects for kinetic variables concurrently with kinematic adjustments that may influence loading. We believe that an unconstrained over-ground running protocol provides a more globally relevant context for the topic of "running injury risk", and that the subtle but significant differences in kinetics and kinematics between treadmill and over-ground running (Nigg et al., 1995; Riley et al., 2007) may diminish clinical translatability of findings in these studies.

Additionally, the vast majority of between-shoe investigations do so exclusively in habitual RF runners. It has been shown that the distinctions in limb loading dynamics between RF and MF patterns are due to the differences in limb kinematics (Ahn et al., 2014; Bishop et al., 2006; De Wit et al., 2000; Lieberman et al., 2012) and shock attenuation strategies (Breine et al., 2017; Gruber et al., 2014; Hamill et al., 2014; Paul et al., 1978) between these running styles, strongly suggesting a differential cushioning effect provided by the shoe (Dubois et al., 2015). However, little research exists on the differential effect of midsole cushioning in the running shoe between these two categorically different running styles. A primary hypothesis for this investigation was that amount of midsole cushioning will affect RF and MF runners differently.

A final methodological consideration is with the natural variability within and between individuals performing repetitive tasks. Typically, sample sizes used in footwear biomechanics research are limited 10 or fewer, and capture as many footstrikes for analysis. Additionally, footstrikes are often averaged across trials and between subjects in a group. These procedural and analytical techniques have been recognized to limit statistical power and hinder the strength of conclusions (Bates et al., 1983; Bates, 1989; Maiwald et al., 2011; Oriwol & Maiwald, 2011). DeVita and Bates (1988) recommended at least 25 trials in order to “detect the subtle real differences that exist between shoe conditions” (Bates, 1989; DeVita & Bates, 1988), with the reminder that differences caused by experimental manipulations constitute only a portion of the total variance within the system(s) being evaluated. When attempting to determine the efficacy of cushioned shoes for the general running population, larger sample and trial numbers would greatly improve statistical confidence as well as the translatability of findings (Frederick, 2011).

Therefore, the purpose of this investigation is to examine the effect of cushioning in the running shoe on attenuation of GRFs, both in the initial instances of impact as well as at peak magnitudes, when controlling for adjustments to gait mechanics between the shoe conditions. This will be accomplished by first allowing for a natural motor response through unconstrained over-ground running, and then subsequently controlling for changes to influential covariates when making statistical comparisons. Also, this protocol will include habitual runners of both MF and RF footstrike patterns to elucidate any differential effect of cushioning. Finally, this study's large sample size and number of trials, in conjunction with linear mixed effects modeling techniques, will provide sufficient statistical power to detect any subtle but potentially meaningful changes to gait between conditions while accommodating for natural intra- and inter-subject variability.

Methods

Participants: 53 recreational runners (Table 1) were split into MF (n=25) and RF (n=28) groups based on footstrike (FS) pattern. Subjects were required to be habitual in their preferred FS and free of any injuries, chronic or otherwise, that might affect their gait. Initial visual determination of FS for grouping purposes was confirmed post-processing by the evaluation of strike index (SI) (Cavanagh & Lafortune, 1980) and the foot inclination angle at initial ground contact (FIC) (Altman & Davis, 2012).

Procedures: All study procedures were approved by the UTA IRB and consented to by study participants. Data collection took place during a single visit to the Applied Biomechanics Laboratory at the University of Texas at Arlington. Subjects were fit with a pair of highly cushioned (HI) (Hoka Stinson One One) and minimally cushioned (LW) (NewBalance Minimus) running shoes. These shoes were chosen because of their diametrically opposite amounts of midsole cushioning (LW = 8mm; HI = 34mm heel height), as well as their similarly minimal heel-toe drop (LW = 4mm; HI = 6mm) to ensure midsole architecture did not influence normal foot orientation at time of contact. A lower-body marker set on the Visual 3D 6DoF model was applied to the subject's pelvis, legs, and feet, using a single-segment foot and a Bell model pelvis. Shoe order was counterbalanced between subjects. Immediately prior to the running trials for either shoe condition, subjects performed two laps around a short indoor running track (total distance = 1/6 mile). Subjects then performed 50 successful over-ground running trials in both HI and LW shoes. A trial was considered successful if the right foot landed wholly on the in-ground force plate without the subject visibly targeting the plate or changing their speed over that stride.

Data Capture and Reduction: (See Table 2 for a full list of dependent variables, their abbreviations, and units) Marker data was captured at 500 Hz using 16 motion-capture cameras (T40s; Vicon Motion Systems Lt., UK) surrounding an in-ground force plate (Optima Human Performance System, Advanced Mechanical Technology, Inc., Watertown, MA) sampling GRF data at 1000 Hz. After marker identification, trials were transferred to Visual 3D v6 Professional software (C-Motion, Inc., Germantown, MD) for analysis. GRFs were normalized to subject mass and transformed into the shank coordinate system. Hip (HAC), knee (KAC), and ankle

(AAC) angles were calculated using Cardan sequence (medial/lateral, anterior/posterior, longitudinal axes), and sagittal joint angles were extracted at touchdown (TD). SI was measured as the distance of the GRF center of pressure (CoP) from the RHL marker proportional to the total foot segment length (Altman & Davis, 2012). FIC was found by normalizing the foot position in the model pose to the plane of the ground, then taking the normalized foot angle at the instance of ground contact (Altman & Davis, 2012). Peak frontal (Y) and sagittal (X) moments were extracted for the hip, knee, and ankle. Hip flexion/extension moment was excluded for lack of a clear pattern from which to extract a consistent peak value. Knee and ankle joint stiffness (KJS; AJS) were determined for sagittal plane rotations across the first 50ms of stance (eq. 1) (Hamill et al., 2014; Hamill et al., 2011).

(1) $K_{joint} = \Delta \text{ moment} / \Delta \text{ angle}$

As an indication of shock transmission, peak shank axial acceleration (pSA) was derived from segment velocity and normalized to gravity. Knee compression and sheer forces in the shank reference frame were integrated over the first 50 ms of ground contact (KCI_50; KSI_50) to give impact-period impulses. Impact loading rate (ILR) was determined as the average slope of the vertical GRF between the first 10-25 ms of ground contact. Calculating the difference between GRF and knee forces (GRFMag_Atten) allowed for a determination of the kinetic chain's force-attenuation capabilities between these two points.

Statistical Analysis: A linear mixed effects (LME) analysis was performed using IBM SPSS Statistic for Windows, version 23 (IBM Corp., Armond, NY) to model the influence of midsole cushioning amount on the individual's gait pattern, as well as on changes to limb loading characteristics. Kinetic variables were assessed using the following variables as covariates: braking impulse in the initial 50 ms of stance, horizontal velocity of the pelvis, shank axial velocity, initial knee angle, and knee and ankle joint stiffness. Covariates were excluded from analysis when not significant for a given dependent variable, as well as when one of these potential covariates was the dependent variable itself. Trials and subjects were set as random factors, with Shoe, FS, Shoe x FS, and significant covariates as fixed factors. Mixed effects comparisons of FS x Shoe and Shoe x FS were made to determine the specific differences between fixed factors of any significant main effects. Alpha was set at $p \leq 0.05$.

Results

Kinematic Response (Table 3): At a self-regulated habitual jogging speed there was no difference in forward running velocity between FS groups and MF runners did not change their forward velocity between shoes. However, RF runners ran slightly but significantly slower in HI shoes. There was also no between-group difference for shank axial velocity at TD, but both MF and RF runners decreased this variable when in HI shoes. Significant Shoe x FS interactions were found for both COMvel and Shankvel, with RF runners making greater adjustments to running speed and MF runners making greater adjustments to shank velocity, indicating different kinematic response strategies.

Both groups of runners decreased their SI when in HI shoes, denoting a ground contact point closer to the heel. Runners in both groups also changed their foot orientation at contact (FIC) when in HI shoes (Figure 1-E), such that MF runners landed with a flatter foot or less toe-down orientation and RF runners landed with a more inclined or toe-up foot. Ankle angles at the time of contact mirrored changes seen in FIC. MF runners did not change KAC, while RF runners decreased KAC in HI shoes. RF runners also landed with more knee extension than MF runners regardless of shoe. Although there were no differences in HAC between FS groups, MF runners **decreased** HAC when in HI shoes, while RF runners **increased** HAC in HI shoes, further highlighting their unique gait strategies.

In the first 50 ms of ground contact, AJS was not affected by shoes and was not significantly different between FS groups. However, both MF and RF runners significantly increased KJS in the HI compared to the LW shoe condition (Figure 1-B). MF runners had significantly lower KJS than RF runners in both shoes, but RF runners increased KJS more than MF runners between shoe conditions.

<u>Kinetics (Table 4):</u> Both MF and RF runners experienced a decrease in pAM_X with HI cushioning, with MF runners experiencing higher ankle plantarflexion moments than RF runners in both shoes. Similarly, a significant decrease in pAM_Z was seen going from LW to HI conditions in both FS groups, however there were no differences in pAM_Z between FS groups for either shoe. Both RF and MF runners experienced significantly higher pKM_X in HI vs LW shoes, but there were no differences between groups in either shoe. It is notable that MF runners experienced a significantly greater change in pAM_X, pAM_Z, and pKM_X than RF counterparts when going from LW to HI cushioning. There were no significant Shoe or FS main effects for pKM_Y. Finally, both MF and RF runners displayed similar reductions in pHM_Y between cushioning levels.

The differential contribution of cushioning between FS groups is especially apparent when comparing effects on translational forces. In HI shoes, ILR was significantly and substantially reduced in RF runners, but slightly **increased** for MF runners (Figure 1-A). Further, MF runners experienced a significantly higher KCI_50 in shoes, whereas RF had significantly lower KCI_50 with more cushioning (Figure 1-C). This effect was greater for RF runners, who also had significantly higher compression impulse in both HI and LW shoes than their MF counterparts. Both groups decreased KSI_50 in the HI vs LW shoes, and the RF group had lower shear impulse than MF in both HI and LW.

Both MF and RF runners experienced significantly **higher** pKCF in HI compared to LW shoes (Figure 1-D), but RF runners had lower pKCF than MF runners in both HI and LW shoes. Alternately, MF and RF runners had significantly **lower** pKSF in HI compared to LW. pKSF was also significantly lower for the RF group in both HI and LW shoes when compared to MF. A larger shoe effect on pKCF was seen in the MF group, but RF runners experienced a greater change in pKSF between shoes.

While rearfoot runners had greater GRFMag_Atten in HI shoes, indicating increased force attenuation capabilities, MF runners actually had lower GRFMag_Atten values in the HI shoes (Figure 1-F), which would indicate **decreased** attenuation of forces between the ground and knee. Further, MF runners experienced greater pSA in the HI shoe condition, while no shoe effect was observed in the RF group.

Discussion

<u>Kinematics:</u> Greater levels of cushioning resulted in a slightly less pronounced MF pattern in these runners. In contrast, RF runners displayed a more prominent RF pattern with

increased cushioning. What is notable is that both MF and RF runners adjusted their foot orientation at contact in similar anatomical directions (+X rotation) and with similar mechanical implications (more heel impact), suggesting increased reliance on the shoe for its shock attenuation capabilities. However, when considering the runners' habitual motor pattern, gait adjustments were directionally opposite between FS groups; increased cushioning resulted in a "less-expressive" MF pattern in this group of runners, while runners in the RF group "over-expressed" their RF pattern. This differential effect in kinematic response to increased midsole cushioning is a significant consideration for the practical translation of research findings, and is in support of the hypothesis that shoes contribute to structural loading characteristics differently between the FS patterns. It would seem that the intrinsic force attenuation capabilities available to MF runners are hindered or reduced when providing for shock attenuation through midsole cushioning, while RF runners are more dependent on the shoe for force attenuation and therefore may benefit more from high amounts of cushioning. This conclusion will be further explored when discussing changes to kinetic variables.

These kinematic findings show that, although not as large as changes reported between shod and BF conditions, increasing levels of cushioning in the running shoe dose in fact manifest in a subtle but significant kinematic response that can be observed immediately prior to the impact period of stance. However, it would also appear that even minimalist shoes provide enough tactile protection to allow for the maintenance of the FS pattern and only minor adjustments to gait patterns at TD, and muscular activation strategies may play a more predominant role in impact accommodation at this level. Further, although kinematic changes were distinct between the FS groups, both types of runners responded to increased cushioning in a way that would indicate an increased reliance on the shoe or a decreased regard for the experience of ground impact.

Joint Stiffness: Initial knee and ankle stiffness to serve as an indication of muscular pre-activation at the time of ground contact. A primary finding from this investigation is that runners in both groups increased KJS when in the maximalist shoes, and this adjustment was significantly larger in RF runners. These results are in disagreement with (Hamill 2014) who found both RF and MF runners did not significantly change KJS between shod conditions of varied midsole cushioning levels, and that RF runners displayed higher KJS than MF runners (Hamill et al., 2014). Bishop (2006) also found no difference in KJS between shoes. This discrepancy may be due to the different timeframes being analyzed, where the present investigation examined the first 50 ms of ground contact, compared to the entire first half of stance in the other reports. Taken together, this may suggest that motor strategies to prepare for impact (feedforward control) are distinct from online (feedback) control strategies that modulate stiffness during the active periods of stance. Ferris and colleagues have also found that when transitioning from harder to softer surfaces or shoes, runners acutely increase leg stiffness in an attempt to maintain CoM excursion (Ferris et al., 1998, 1999).

Further, the larger shoe effect on KJS compared to that on joint angles at TD supports the hypothesis that modulation of joint compliance via muscular activation is a more predominant strategy than changes to gait pattern when accommodating for increased impact sensations with reduced midsole cushioning. It would seem that adjustments are made to muscle activation levels, and therefore joint compliance, immediately prior to contact in order to maintain some characteristics of limb loading dynamics, so that the runner is able to maintain head stability

(Ferris et al., 1998; Gruber et al., 2014) and efficiency of locomotion (Butler et al., 2003; Kerdok et al., 2002; Latash & Zatsiorsky, 1993; McMahon, 1985) despite the softer foot-ground interface. That RF runners displayed a substantially larger increase in KJS with HI cushioning than their MF counterparts is another notable indication that strategies to accommodate impact forces are distinct between these FS patterns.

<u>Kinetics:</u> Figure 2 shows representative GRF curves in the positive axial direction of the shank's reference frame for MF **(a)** and RF **(b)** runners in both shoe conditions. Previous research more consistently reports decreased ILR with increased cushioning (De Wit et al., 2000; Dixon et al., 2000; Hamill et al., 2011; Paquette, 2013; Sinclair et al., 2016; Willy & Davis, 2014), even with impact peak magnitude sometimes being unaffected. These prior conclusions of the cushioning effect were substantiated through this investigation in the RF group, whose runners experienced a substantial (32%) decrease in ILR with higher cushioning. This is reasonable, as the heel pad itself has limited force attenuation capabilities and a highly cushioned midsole would greatly increase the time and area over which the force was distributed. MF runners, however, experienced a much less substantial but still significant **increase** in ILR. Changes to axial loading rates were reflected in changes to KCI_50, with RF runners experiencing significant decrease in axial loads in the impact period, and MF runners experiencing a significant increase. As with ILR, between-shoe differences in KCI_50 were substantially larger in the RF group. The direct relationship between loading rate and impulse is contradictory to the reports of Addison and Lieberman (2015) who showed an inverse relationship between these two variables. However, by integrating the force curve up to the impact peak, as was done in the Addison study, the time frame integrated is reduced as ILR is

increased, resulting in this inverse relationship. It might be argued that total impulse in the impact period is a more appropriate method of assessing time-dependent tissue stress. The differential shoe effect between FS patterns on kinetic variables in the impact period is well elucidated in the amount of force attenuated between the ground and the knee, and mirror the cushioning effect seen in axial loading rates and impulses for a respective footstrike. Between-shoe differences in GRF attenuation indicate increased attenuation capabilities in RF runners wearing highly cushioned shoes, while adding cushioning under the foot of a MF runner results in **reduced** impact force attenuation capabilities.

Despite RF runners experiencing reductions in impact-period forces, their peak shank acceleration was surprisingly unaffected. This is contradictory to findings by Sinclair et al. (2016), who measured lower peak shank accelerations and ILR in RF runners in cushioned vs. minimalist running shoes. In Sinclair's study, a tibia-mounted accelerometer was used to measure shank acceleration. Although Hennig showed that force plate measurements can provide a good indication of peak tibial acceleration (Hennig et al., 1993), it is possible that taking the double derivative or surface-mounted marker positions limited the sensitivity of acceleration calculations in this study. However, Laughton (2003) reported a reduction in ILR for both RF and MF runners with the use of a foot orthotic device, although pSA remained unchanged (Squadrone & Gallozzi, 2009), and it was posited that other variables such as peak and shear GRFs may also contribute to limb segment acceleration. It is also possible that even with thick midsoles, the cushioning bottoms out before peak accelerative forces are reached. MF runners actually experienced significantly higher peak shank axial acceleration in the maximally cushioned shoes. It is possible that the maximalist shoes constrained the foot articulations so as to reduce its deformation and therefore shock attenuation capabilities. The acceleration findings in both groups of runners suggest that cushioning itself may at best do little to reduce the shock transmitted through the lower leg during impact, which may be more readily modulated through cognizant alterations to gait kinematics (Samaan et al., 2014).

Between FS groups, directionally different changes were only observed in the kinetic variables associated with the impact period. However, significant Shoe x FS interactions in peak midstance variables highlight a continued distinction in mechanics between the two groups. Notably, pKCF was increased in both groups, and more so in MF runners. For rotational loads, reductions in peak ankle and hip moments with high cushioning were observed alongside increased peak knee extension moment. Together, these results suggest a more knee contribution across the active loading phase of stance when in cushioned shoes, as well as a reduced reliance on negative joint work at the expense of an increased reliance on passive translational force attenuation during load acceptance. Also, the significantly larger shoe effect on joint moments experienced by MF runners may be due to the fact that this group relies more heavily on negative rotational work, particularly at the ankle, for energy attenuation. Conversely, midsole cushioning had a greater effect on translational forces in RF runners.

In short, increased cushioning mitigated impact-related loading variables only in the RF group. In the MF group, however, maximal amounts of midsole cushioning was shown to **increase** these same variables, in addition to peak shank axial acceleration. Further, although peak hip and ankle moments were lower in the maximalist shoes, peak knee compression forces and extension moments were actually increased with high amounts of cushioning in *both* groups. This potentially results from an increased reliance on the knee joint during energy transfer of the stance phase, or more “knee-dominant” running strategy, with more midsole cushioning, and is reflected in the subtle changes to joint landing angles, as well as a more substantial increase to pre-tuned knee stiffness.

The findings from this investigation, both within and between FS groups, suggest that cushioned running shoe midsoles may not be as effective at mitigating the potentially harmful loads imposed on lower limb structures as is generally advertised, and may even increase structural loading depending on FS pattern or variable and time point of interest. Further, the response to midsole cushioning was distinct between MF and RF runners, which has significant implications towards proper footwear prescription. RF runners seem to rely more heavily on cushioning properties for impact attenuation, which is reasonable based on the decreased capabilities of intrinsic mechanisms for this function when considering RF landing mechanics. MF runners, on the other hand, are better able to use intrinsic mechanisms to accommodate impact forces and attenuate the initial loading response, which seem to be limited with substantial midsole material. The differential contribution of cushioned shoes between the two footstrike patterns is more significant in the initial instances of stance, which is a logical finding considering that dissimilarities between the running patterns are greatest in this time frame.

Figure Legends

Figure 1: Graphical representation of changes to select dependent variables between shoe condition (x-axis) by footstrike group (y-axis) for ILR (A), KJS (B), KCI_50 (C), pKCF (D), SI (E), and GRFMag_Atten (F).

Figure 2: Force-time curves of GRFs acting along the longitudinal axis of the shank for midfoot (a) and rearfoot (b) runners in both shoe conditions.

Table 1: Subject Demographics

	Age		Height (m)		Weight (kg)	
Footstrike	$\bar{x}$	± sd	$\bar{x}$	± sd	$\bar{x}$	± sd
Rearfoot (RF)	24.61	6.25	1.73	0.11	69.02	11.76
Midfoot (MF)	24.40	5.35	1.76	0.07	73.59	10.63
Total	24.51	5.84	1.74	0.10	71.18	11.47

Table 2: List of dependent variables, their abbreviations, and units.

Dependent Variables	Abbreviations	Units
Ankle Angle at Contact	AAC	°
Ankle Joint Stiffness	AJS	N·m/kg/rad
Attenuation of GRF Magnitude Impulse - 50 ms	GRFMag_Atten	%GRF
Foot Inclination at Contact	FIC	°
Hip Angle at Contact	HAC	°
Impact Loading Rate	ILR	N/kg/s
Knee Angle at Contact	KAC	°
Knee Joint Stiffness	KJS	N·m/kg/rad
Knee Shear Impulse - 50 ms	KSI_50	N/kg·ms
Knee Compression Impulse - 50 ms	KCI_50	N/kg·ms
Peak Knee Compression Force	pKCF	N/kg
Peak Knee Shear Force	pKSF	N/kg
Peak Shank Axial Acceleration	pSA	G's
Pelvis Horizontal Velocity 10 ms Before Contact	COMvel	m/s
Peak Right Ankle Plantar/Dorsiflexion Moment	pAM_X	N·m/kg
Peak Right Ankle In/Eversion Moment	pAM_Z	N·m/kg
Peak Right Hip Add/Abduction Moment	pHM_Y	N·m/kg
Peak Right Knee Flexion/Extension Moment	pKM_X	N·m/kg
Peak Right Knee Add/Abduction Moment	pKM_Y	N·m/kg
Shank Axial Velocity 10 ms Before Contact	Shankvel	m/s
Stance Time	ST	s
Strike Index	SI	%Foot length from heel

Table 3: Kinematic variable pairwise comparison of the shoe effect separated by FS group.

Variable Units	FS	Mean Difference (LW – HI)	Std. Error	Sig.	95% Confidence Interval for Difference	
					Lower Bound	Upper Bound
COMvel [b] m/s	MF	-0.0076	0.0049	0.120	-0.0172	0.0020
	RF	0.0300	0.0046	0.000	0.0281	0.0463
Shankvel [a,b] m/s	MF	0.0780	0.0062	0.000	0.0659	0.0904
	RF	0.0380	0.0059	0.000	0.0259	0.0492
SI [a,b] % FtLength	MF	0.0530	0.0040	0.000	0.0452	0.0609
	RF	0.0420	0.0038	0.000	0.0341	0.0490
FIC [a,b] degrees	MF	-0.5200	0.1400	0.000	-0.7942	-0.2452
	RF	-1.2290	0.1332	0.000	-1.4902	-0.9679
AAC [a,b] degrees	MF	-0.4550	0.1207	0.000	-0.6920	-0.2189
	RF	-1.3260	0.1148	0.000	-1.5510	-1.1009
KAC [b] degrees	MF	0.0564	0.1135	0.620	-0.2789	0.1662
	RF	-0.4830	0.1080	0.000	0.2710	0.6945
HAC [a,b] degrees	MF	0.3910	0.1071	0.000	0.1808	0.6006
	RF	-0.3350	0.1018	0.001	-0.5347	-0.1354
KJS [a,b] N·m/kg/rad	MF	-0.2730	0.0417	0.000	-0.3548	-0.1914
	RF	-0.5590	0.0402	0.000	-0.6381	-0.4805
AJS N·m/kg/rad	MF	-0.1646	0.9295	0.859	-1.9869	1.6577
	RF	0.4263	0.8939	0.633	-1.3260	2.1787

[a] Significantly different between shoe conditions in MF runners
[b] Significantly different between shoe conditions in RF runners.

Note: Values for pairwise comparisons represent the measurement for the HI condition subtracted from the measurement for the LW condition. For example, an FIC with a greater plantarflexion (negative) angle in LW shoes vs. HI shoes equates to more toe-down landing in LW shoes (-0.520 ± 0.140 degrees, $p < 0.001$). Tables for estimated means by dependent variable can be found in the attached data sources.

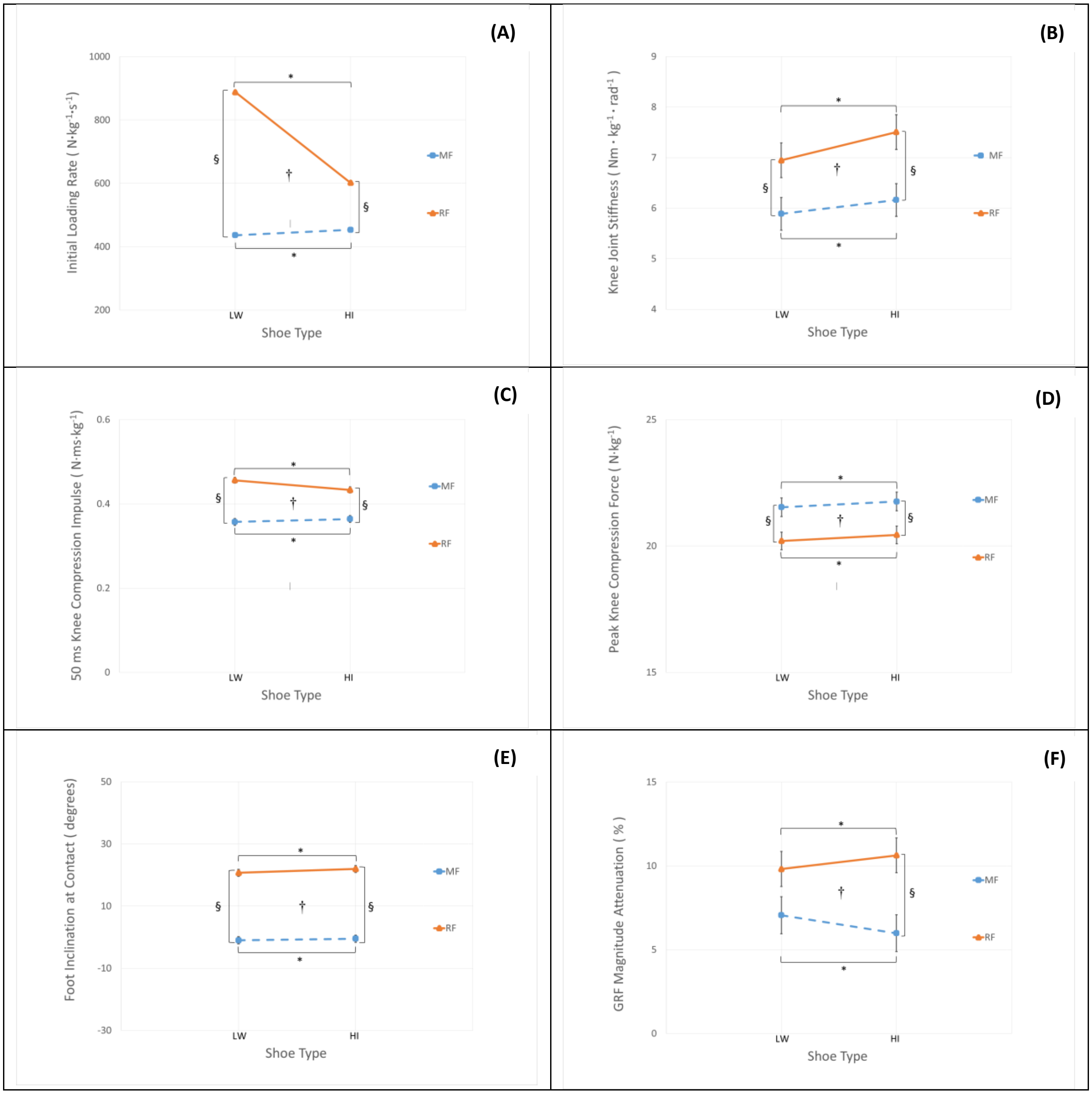
Table 4: Kinetic variable pairwise comparisons of the shoe effect separated by FS group.

Variable Units	FS	Mean Difference (LW – HI)	Std. Error	Sig.	95% Confidence Interval for Difference	
					Lower Bound	Upper Bound
pAM_X [a,b]	MF	0.2710	0.0049	0.000	0.2619	0.2811
$N \cdot m \cdot kg^{-1}$	RF	0.2300	0.0048	0.000	0.2208	0.2396
pAM_Z [a,b]	MF	0.0340	0.0019	0.000	0.0298	0.0374
$N \cdot m \cdot kg^{-1}$	RF	0.0070	0.0018	0.000	0.0035	0.0106
pKM_X [a,b]	MF	-0.2510	0.0108	0.000	-0.2724	-0.2299
$N \cdot m \cdot kg^{-1}$	RF	-0.1010	0.0105	0.000	-0.1215	-0.0804
pKM_Y	MF	0.0305	0.0156	0.051	-0.0001	0.0610
$N \cdot m \cdot kg^{-1}$	RF	0.0179	0.0151	0.237	-0.0118	0.0475
pHM_Y [a,b]	MF	0.2390	0.0874	0.006	0.0671	0.4099
$N \cdot m \cdot kg^{-1}$	RF	0.3230	0.0823	0.000	0.1619	0.4846
ILR [a,b]	MF	-16.9970	4.3064	0.000	-25.4394	-8.5546
$N \cdot kg^{-1} \cdot s^{-1}$	RF	285.9650	4.0958	0.000	277.9353	293.9943
KCI_50 [a,b]	MF	-0.0080	0.0012	0.000	-0.0099	-0.0053
N/kg	RF	0.0230	0.0012	0.000	0.0208	0.0253
pKCF [a,b]	MF	-0.4040	0.0299	0.000	-0.4622	-0.3451
N/kg	RF	-0.3200	0.0291	0.000	-0.3769	-0.2628
KSI_50 [a,b]	MF	0.0060	0.0006	0.000	0.0050	0.0072
N/kg	RF	0.0060	0.0006	0.000	0.0050	0.0072
pKSF [a,b]	MF	0.0460	0.0174	0.009	0.0115	0.0797
N/kg	RF	0.1730	0.0169	0.000	0.1402	0.2065
GRFMag_Atten[a,b]	MF	1.0730	0.0888	0.000	0.8984	1.2466
%	RF	-0.8140	0.0869	0.000	-0.9843	-0.6437
pSA [a]	MF	-0.0730	0.0144	0.000	-0.1007	-0.0443
G's	RF	-0.0035	0.0141	0.805	-0.0311	0.0241

[a] Significantly different between shoe conditions in MF runners

[b] Significantly different between shoe conditions in RF runners.

Note: Values for pairwise comparisons represent the measurement for the HI condition subtracted from the measurement for the LW condition. For example, greater GRFMag_Atten in HI shoes results in a negative pairwise difference for RF runners (-0.8140 ± 0.0869 %, $p < 0.001$).



(A)
Initial Loading Rate (N·kg⁻¹·s⁻¹)
Shoe Type
LW
HI
MF
RF
(B)
Knee Joint Stiffness (Nm · kg⁻¹ · rad⁻¹)
Shoe Type
MF
RF
(C)
50 ms Knee Compression Impulse (N·ms·kg⁻¹)
Shoe Type
MF
RF
(D)
Peak Knee Compression Force (N·kg⁻¹)
Shoe Type
MF
RF
(E)
Foot Inclination at Contact (degrees)
Shoe Type
MF
RF
(F)
GRF Magnitude Attenuation (%)
Shoe Type
MF
RF

Figure 1.

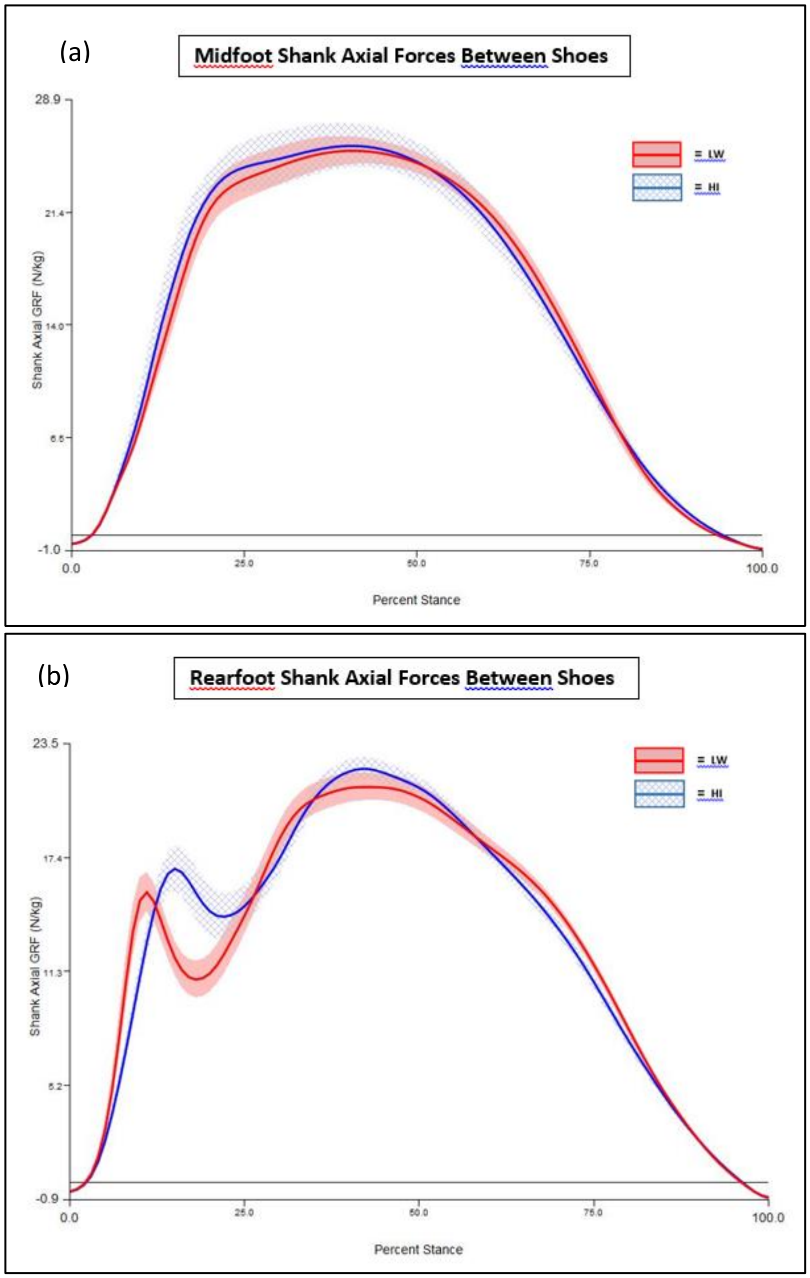

(a)
Midfoot Shank Axial Forces Between Shoes
= LW
= HI
28.9
21.4
14.0
6.5
-1.0
Shank Axial GRF (N/kg)
0.0
25.0
50.0
75.0
100.0
Percent Stance
(b)
Rearfoot Shank Axial Forces Between Shoes
= LW
= HI
23.5
17.4
11.3
5.2
-0.9
Shank Axial GRF (N/kg)
0.0
25.0
50.0
75.0
100.0
Percent Stance

Figure 2.

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Chapter 3

A quantification of lower-limb coordinative variability during running with different levels of midsole cushioning

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Abstract

Cushioned material between the foot and the ground during the stance phase of running dampens haptic feedback information used by higher control centers to inform subsequent movement executions. The purpose of this investigation was to examine whether this constraint on the motor system would manifest in reduced coordinative variability in lower-limb joint couplings and a more ‘rigid’ gait pattern, implying increased consistency of tissue stress during time periods of limb loading. 10 midfoot and 10 rearfoot experienced distance runners, all of whom were habitual in their categorized footstrike pattern and free of any pain or injury, ran on a treadmill while barefoot and in shoes with minimalist (New Balance Minimus) and maximalist (Hoka Stinson One One) amounts of midsole cushioning.

No differences in leg segment coordinative variability were observed between footwear conditions in either group of runners. Peaks in variability were often observed around periods of gait transition, primarily in foot-shank couplings. Further, a novel observation was the altered spatiotemporal properties of the segments’ phase plots between footwear conditions, manifesting in altered joint coupling patterns. The effect of increased midsole cushioning material on proprioceptive acuity during stance was not a sufficient disturbance to the motor system to manifest in altered levels of coordinative variability. Healthy, experienced runners likely display learned coordination patterns containing ‘normal’ levels of variability. Subjective observations of altered joint coupling patterns between footwear conditions may have significant implications towards healthy mechanical function. Prospective investigations should be employed to assess for any association between injury risk and individual-specific levels of coordinative variability.

Introduction

Mechanical deterioration of tissue results from an excessive accumulation of stress without adequate recovery (Stanish, 1948; Novacheck, 1998). Although concrete loading parameters (rate, magnitude, volume, etc.) causal to injury are difficult to define and are individual-specific, from a dose-response perspective there exists a theoretical 'critical dose threshold' beyond which tissues deteriorate into an injured state due to mechanical overload (Hreljac, 2000; James, 2004; Brandt, 2009). During running, ground reaction forces (GRFs) proportional to approximately 3x an individual's bodyweight (BW) are transmitted through the feet and legs with every stride. Previous research has predominantly focused on defining differences in GRF (loading rate, impact magnitude, etc.) and limb loading (joint moment, shank axial force and acceleration, etc.) characteristics between footwear and gait pattern conditions. However, a typical male endurance runner with a stride length of 2.4 meters (Cavanagh, 1989; Mercer, 2002) contacts the ground over 400 times per kilometer *per leg*. The highly repetitive application of stress suggests that the uniformity (or conversely the variability) of loading patterns to specific tissue sites may also be a critical factor in determining total tissue stress across an activity bout, and therefore one's potential for crossing the injury threshold. This reasoning was conceptually outlined by James et al. (2004) and was termed the '*variability-overuse injury hypothesis'*. This model states that decreased variability in repetitive movement results in decreased distribution of tissue loading pattern, volume, and timing, therefore increasing the overall volume and homogeneity of loads to specific tissue sites, and in turn the probability of developing tissue overload.

Variability is inherent in nature, including in human movement. Although classical paradigms on the regulation of human physiological systems (i.e. motor control, cardiovascular control, brain activity, hormonal secretion) have viewed variability as being either noise in or a disturbance to proper system functioning, scientists across this broad spectrum of disciplines are now recognizing that variability in cyclical patterns is distinguishable from noise, has a deterministic origin, and plays an integral role in proper system function (Goldberger, 1990; La Rovere, 1998; Edwards, 1999; Glass, 2001; Lipsitz, 2002). By examining the structure of variability in cyclical or repetitive motor tasks, Kelso and others have shown it to be an important component in the stability of the movement pattern (Haken, 1985; Kelso, 1996; van Emmerik, 2000; Latash, 2006), and regions of increased variability are indicative of transition between two stable states or between distinct phases of segment motion (Heiderscheit, 2000; Clark and Phillips, 1993; Kelso, 1996; Dierks and David, 2011). Variability in the coordination of body segments for task performance, as in segment-to-segment couplings in the trunk and lower limbs during running gait, is theorized to provide a source of adaptation to dynamic task demands as well as flexibility within the system to accommodate perturbations or unexpected events (Clark and Phillips, 1993; Turvey, 1990; van Emmerik et al., 1999; Heiderscheidt, 2000).

Suggestions have already been widely made that, based on the ‘inverted-U’ model (Figure 1), healthy and stable movement systems exhibit an optimal level of variability across movement cycles (VanEmmerik, 2005; Latash, 2006; Stergiou, 2011; Hamill, 2012). Alteration to ‘normal’ variability in kinetic and kinematic parameters during static postural control and locomotion have been found in a number of neurological diseases (Hausdorff, 1998; Edwards, 1999; Webster, 2006; Delignieres, 2009), in individuals with ACL deficiency (Tzagarakis, 2010) or patellofemoral pain (Heiderscheit, 2002), and in fall-risk elderly (Hausdorff, 1997a). These studies all report increased variability in some *outcome* parameter of the motor task. Therefore, increased variability in the outcomes measured was associated with reduced stability in the system. However, healthy motor patterns and more optimal performance is characterized by *lower* levels of variability in the movement outcome – or end-point variability – but *higher* levels of variability in relative segmental motion during task execution, referred to as coordinative variability. In fact, other studies have shown that these same clinical populations which display unsteady gait and increased variability in stride characteristics (higher end-point variability) also display *decreased* levels of coordinative variability quantified via segmental couplings (van Emmerik et al., 1999; Heiderscheit, 2002; Stergiou, 2004, 2011; Moraiti, 2006). Lipzits (2002) and Hamill (2012) have synthesized the available evidence into the 'loss of complexity hypothesis in relation to injury or pathology'. This model suggests that injury or disease may occur when the reduction of movement complexity within functional synergies, and therefore the overall functional capacity of the system, decreases across a critical threshold and the individual becomes clinically afflicted. However, although strong associations have been made between altered motor variability and injury or pathology, a causal relationship remains unclear. A primary question is whether any alteration to motor variability exist prior to the affliction, which would suggest and increased risk for disease or injury, if not highlighting a causal mechanism.

From a perspective based on Dynamical Systems theory of motor control, afferent information is integrated by the central nervous system to inform control of subsequent iterations of a repetitive or cyclical movement (Gandevia, 1992; Wolpert, 1995). Runners are known to acutely adjust their gait pattern based on sensory information from ground contact (Hennig, 1996; Nigg, 2001; Moore, 2014). Based on this model, removal or alteration of feedback information can influence the control of repetitive movement and, therefore, potentially affects the nature/quantity of variability within successive repetitions. Compromise of peripheral sensory pathways is associated with altered gait variability and stability in neuropathic patients (Dingwell, 1999; Hausdorff, 1997b), and the introduction of subsensory vibratory noise has been shown to increase tactile perception (Collins, 1996; Khaodhiar, 2003) and reduce gait variability (Galica 2009) in clinical populations. Increased midsole cushioning inherently alters the structure and quantity of somatosensory feedback acquired from the foot-ground interaction during the stance phase of gait. Therefore, through this mechanism it is possible that protective cushioning material in running shoe midsoles impose a constraint on the motor system such that the dampening of proprioceptive acuity manifests in alterations to the variability of segmental coordination in running.

The use of foot orthotics has in fact been reported to restore supposedly normal levels of coordinative variability in PFP patients (Ferber, 2005), but this effect could be due to the pain relief garnered from the intervention, and so it is likely not appropriate to extrapolate conclusions from this evidence to the proposed mechanism of the current investigation. More directly, Kurtz et al. (2003) report reduced variability of knee and ankle joint motions when runners were shod compared to barefoot, although no differences were found between shod conditions. Further, Tenbroek (2012) also found that, at least in some of the joint couplings analyzed, runners displayed greater joint coupling variability when barefoot. Again, however, no differences or consistent trends were seen between the shod conditions of varied midsole properties. The possibility was presented in both of these investigations that differences between midsole cushioning properties were insufficient to elicit detectable differences in coordinative patterns. Another restriction of these studies is the inclusion of only rearfoot runners. Due to the distinct differences in mechanics between the rearfoot (RF) and midfoot (MF) running patterns, as well as the consideration that MF runners are thought to rely more heavily on proprioception (Lieberman, 2012), it is possible that MF runners will respond differently to changes in midsole cushioning properties.

Therefore, the present study attempts to more clearly and comprehensively elucidate whether midsole cushioning influences the variability of segment-to-segment coordination of the lower limbs during running. The quantity of variability in relative segmental motion, or joint couplings, may give insight into control dynamics during running gait, as well as the consistency of stress patterns accumulated throughout the exercise bout. This assessment will be accomplished by using continuous relative phase (CRP) techniques (Hamill, 1999) to quantify the variability in lower-limb joint couplings within timeframes surrounding the periods of gait transition from swing to stance and from stance to swing, and then comparing these quantities between footwear conditions of extremely disparate midsole properties. Both habitual RF and MF runners will be included in this assessment to examine for any differential response. It was hypothesized that increased midsole cushioning, by dampening proprioceptive feedback from the foot-ground interaction, will result in decreased joint coupling variability.

Methods

Participants: 10 MF and 10 RF distance runners (Table 1) were recruited from the university's campus as well as the surrounding community. To qualify, runners must have been completing at least 10 miles/week for 4 or more weeks without experiencing any pain, discomfort, or injury. Considerations for subject criteria aimed to include experienced distance runners with proficient mechanics whose cyclical motor patterns should reflect healthy or normal levels of variability.

Procedures: Participants reported to the biomechanics laboratory for a single testing session. After giving informed consent approved by the university's IRB they donned form-fitting spandex shorts and t-shirt, and 14mm diameter Reflective surface markers were applied to the pelvis, legs and feet following segment definitions of the Visual 3D 6DoF model. Static poses were captured in each of the three footwear conditions – hi cushioning (HI; Hoka Stinson One One), low cushioning (LW; New Balance Minimus), and no cushioning or barefoot (NO). Reflective markers remained in place attached to the shoe upper when changing between footwear conditions. Maintaining marker placement in the NO condition required subjects to perform the barefoot dynamic trials first, then shoe order was counterbalanced between subjects. Subjects ran at 5 and 7 mph (2.24 and 3.13 m/s) in each footwear condition, with at least 1 minute rest between running trials to avoid any fatigue component. Each running trial lasted between 1.5 and 2 minutes, with the first minute being a 'washout' for familiarization, after which 50 consecutive strides of the right leg were captured for analysis.

Data Analysis and Reduction: CRP was quantified for 6 joint couplings which describe the primary components of lower-limb articulations responsible for energy transfer during stance. (Hamill, 1999; Tenbroek, 2012; Frank, 2013; Seay et al., 2006) (Table 2). The CRP methodology allows for the examination of relative motion patterns between interacting segments, and contains information about both the angular position and the angular velocity of the segments across time. Coordinative variability is then quantified as the standard deviation of the CRP ensemble curve of all movement cycles.

Marker trajectory data were captured with 16 Vicon MX T40S cameras sampling at 500 Hz across 50 consecutive strides of the right leg (T40s; Vicon Motion Systems Lt., UK). A single Vicon Bonita video camera synced at 100 Hz was placed just above belt level approximately 3 ft to the right of the treadmill to capture time events of footstrike (FS) and toe-off (TO). Data were then transferred to Visual 3D v6 Professional software (C-Motion, Inc., Germantown, MD) for processing and extraction of joint angles and angular velocities. Joint angles were calculated using and X-Y-Z Cardan sequence. Angular velocities were derived from joint angle changes. To quantify variability of limb segment coordination, the continuous relative phase approach outlined by Hamill (1999) was followed. Transitioning to a custom C# program built in Visual Studio 2010 (Microsoft Corporation, Redmond, WA) for the remainder of the data reduction, angles were normalized by the maximum range of the signal (equation 1)

$$\theta'_i = \frac{2*[\theta_i - \min(\theta_i)]}{\max(\theta_i) - \min(\theta_i)} - 1 \tag{1}$$

and angular velocities were normalized by the absolute maximum (equation 2) (Robertson, 2013).

$$\omega'_i = \frac{\omega_i}{\max[\max(\omega_i), \max(-\omega_i)]} \tag{2}$$

Phase plots containing all 50 strides in a given footwear/speed condition were created for each segment rotation and phase angles were extracted for each data point (Figure 2) (equation 3).

$$\phi(i) = tan^{-1}\left(\frac{\omega'(\mathrm{i})}{\theta'(i)}\right) \tag{3}$$

CRP was then calculated by subtracting the distal segment phase angle from that of the proximal (equation 4)

$$CRP(i) = \phi_A(i) - \phi_B(i) \tag{4}$$

with the final relative phase angle expressed over a range of ± 180 degrees. Composite CRP curves for each coupling and condition were created from the mean ± sd of the 50 consecutive strides. Variability was then quantified as the integral of the ± sd curve over the 100 ms time frame surrounding touchdown (TD) (50 ms before and after the event instance) as well as 100 ms surrounding toe-off (TO). Trials were interpolated to 100 points over each of the specified intervals, and changes to coordinative variability were assessed across footwear conditions within a FS group.

A Priori Statistical Power Analysis: G*Power version 3.1.9.2 was used to determine the sample size required to detect an effect size of 0.6 for ANOVA with one between subjects' factor with two levels and one within subjects' factor with 3 repeated measures using a correlation among repeated measures of 0.5 and an alpha level of 0.05. To obtain an estimated power of 80%, 9 subjects per group would be required.

Statistical Analysis: A 2x3 (FSxFW) ANOVA with 1 between-subjects factors (RF, MF) and 1 within-subjects factors (HI, LW, NO) was performed for both speeds to assess for a shoe effect on coordinative variability, as well as any between-FS interaction. The distributions for each coupling variable were analyzed for normality and the existence of outliers using histogram plots, boxplots and Quantile-Quantile plots. RF runners who switched their footstrike to a MF pattern when running barefoot were excluded from analysis in this condition.

Results

Results of statistical analyses were similar between running speeds, therefore the remainder of the report will describe results from the 5 mph (2.24 m/s) trials. Five of the rearfoot runners switched their footstrike to a MF pattern when running barefoot, and so were excluded from analysis in this condition. CRP variability was not found to be different between footwear conditions ($p > 0.05$) (Figures 3a and b). Further, no differences in CRP variability were found between FS patterns ($p > 0.05$). For both FS groups, mean estimates for variability integrated over the 100 ms time frames ranged between (2.58 – 6.48 Deg·s) at TD (Table 3) and (2.64 – 7.46 Deg·s) at TO between all shoe conditions (Table 4). Peaks in variability were more distinct in the $Th_{Ad/Ab}$ –$Sh_{I/E}$ and the distal shank-foot couplings, $Tib_{F/E}$ – $Ft_{F/E}$ and $Tib_{I/E}$-$Ft_{I/E}$. (Figure 4), with peak variability often exceeding 50 degrees and sometimes, but not always, occurring around periods of stance-swing transition. Although no different their amount of variability, the trajectory of the CRP plots over the time periods analyzed are visibly altered between the footwear conditions in both groups (Figure 5).

Discussion

The purpose of this investigation was to quantify the amount of lower-limb joint coupling variability in time periods surrounding heel-strike and toe-off to determine whether different levels of midsole cushioning constrained natural motor variability in these regions of gait transition by dampening proprioceptive acuity during ground contact. Reductions in coordinative variability from some ‘normal’ or ‘optimal’ level have been associated with injury and pathology, and are thought to arise from a constraining of the DoF within the functional synergies involved in the motor task. However, it is not known whether a less-variable system can be a causal mechanism for the injury or pathology, although logical arguments to this effect have been made (Lipsitz, 2002; Hamill, 2012). Based on the ‘variability-overuse injury’ hypothesis (James, 2004), decreased coordinative variability in lower-limb joint couplings during loading phases in running may predispose specific tissue sites to an excessive accumulation of stress. It was hypothesized that increasing underfoot material would sufficiently influence feed-back control pathways to result in a constraining of functional DoF and lower levels of coordinative variability in leg joint couplings. However, the consistency of variability in all couplings across footwear conditions leads these investigators to reject this hypothesis in favor of the null. Several possible explanations for these observations exist and will be explored, although the evidence of this investigation/protocol indicates that dampening haptic feedback through increased levels of midsole cushioning is not a sufficient disturbance to the system’s control pathways to manifest as reduced coordinative variability in experienced distance runners.

Figure 4 displays the pattern of absolute variability across the entire gait cycle. Visible peaks in the CRP variability occur around periods of gait transition from stance to swing and swing to stance, as expected based on Kelso's and other's work on Dynamical Systems theory (Heiderscheit, 2000). These patterns are similar to those observed in Hamill's (1999) healthy runners. Interestingly, the visible peaks are greater in the thigh-shank axial, as well as both of the distal shank-foot rotations examined. Basketball players have been reported to have greater amplitudes in distal segment range of motion in free-throw shooting, and it was hypothesized that increased movement trajectories in distal segments allow for more effective last-minute adjustments to ensure movement outcome accuracy (Robbins, 2003; Button, 2003; Bartlett, 2007). Miller (2002) reported an increasing trend in absolute variability of segment end-point speed along the segment chain, and it is logical that distal segments in a kinetic chain would contain the sum of the variability from more proximal segments within the chain.

Dierks and Davids (2007) described lower limb joint coupling patterns and variability in healthy runners performing a series of overground trials; although the range of CRP values used in their investigation terminated at 90˚ the range of reported within-subject variability (from 6.6° to 23.6°) is comparable to values seen Hamill (99) and this investigation (Figure 4) when scaled to a cutoff of 180˚. Additionally, they found the least amount of variability in the phase of gait roughly between midstance and half the remaining distance to TO, which generally matches patterns seen in our and Hamill's findings. These investigations together may contribute valuable reference for 'healthy normal' coordinative variability in this motor task.

A notable subjective observation is the influence shoe cushioning has on the coupling pattern, where within-subject differences can be seen in segment phase plot patterns and resulting continuous relative phase of the coupling (Figure 5). Faulty joint coupling patterns have been associated with various musculoskeletal injury risks (DeLeo, 2007). Although outside the scope of the current topic, this effect is likely meaningful and should be explored in detail in future investigations.

In general the findings of this study agree with previous investigations into midsole cushioning effects by Kurtz and Stergiou (2003), Tenbroek (2011), and Frank (2013) in that no differences in variability were found between shod conditions with varied midsole cushioning properties. Although the authors have suggested a potentially too-little of a difference in midsole characteristics between the employed shoe conditions may explain their lack of significant findings, the use of extremely opposite midsole types in the present investigation seems to substantiate their between-shoe findings. However, our results disagree with the findings of these investigations in the barefoot condition, where they found RF runners to be more variable in joint displacements across stance (Kurtz 2003) and in some of the total joint couplings assessed across the gait cycle (Tenbroek 2011).

Some differences in methodology exist which will be put forth for consideration. First, the exclusion of five of our rearfoot subjects from analysis in the barefoot condition because of their altered footstrike pattern limits statistical power and may preclude comparisons in this specific condition. However, the findings within our RF runners concurred with those of the MF group, all of whom were able to maintain their habitual footstrike pattern between the three conditions. To our knowledge, this is the first investigation to quantify lower-limb coordinative variability between shoe conditions in a group of healthy MF runners. It might be considered, even, that the change of FS pattern when barefoot is itself a representation of higher levels of motor variability, as the runner explores a broader solution manifold, or spectrum of possible solutions, to accommodate for the greater impact experience of running barefoot (Latash, 2012).

The spanning set methodology used in Kurtz's study analyzes single-segment motion, and it may be possible that the more variable motion for the individual segment is reflected in the adjacent segment, and therefore the amount of variability in the relative phase of the joint coupling is maintained. In Tenbroek's investigation, which used similar CRP methodology as the current, it seems the average variability over the entire gait cycle was used for comparisons, which may explain the slight (3 of 6 couplings showed significant differences in this condition) disagreement between our results. Therefore, taking these findings together we believe it is likely the drastic difference in sensation between true barefoot and even minimally shod rearfoot running may still result in a large enough high impact perturbation for the runners to adopt a more variable coordination strategy. However, in the protocol employed in our study, it seems that the influence of increased midsole cushioning between shoes on somatosensory information received during footstrike was not substantial enough of a disturbance e wasto normal control mechanisms to effect the variability in coordination patterns. Running on the flat, smoot surface of the treadmill belt provided for very consistent sensory input and there was little to no chance of uncertainty in this environment. On the other hand, outdoor trail running on a natural substrate would likely provide higher potential for the influence of cushioned shoes to dampen somatosensory information, not to mention increase the likelihood for perturbations.

Additionally, it is known that higher motor control centers integrating sensory feedback to inform subsequent motor execution rely more heavily on visual rather than somatosensory input (Magill 2007). In the protocol employed, the subjects' visual field was surrounded proximally (under one meter) by the treadmill display panel and frame arms, and slightly more distally (a few meters) by the perimeter beam for the cameras and the walls and ceiling of the lab space. This entirely static and relatively proximal visual references could have minimized the influence of alterations to proprioception imposed by the experimental conditions insufficient to disturb control pathways, and allowed for the maintenance of coordination parameters. An immersive dynamic visual field may make the system more sensitive to alterations to tactile perceptive acuity.

Finally, the subjects recruited for this protocol were healthy, experienced distance runners, considered representative of proficient and stable mechanics (i.e. not leading to injury). Therefore it is possible they exhibit well-learned and resilient gait control parameters having optimal or normal levels of variability, which are not acutely influenced by the experimental intervention employed. By quantifying local dynamic stability, Frank (2013b) has shown that experienced runners display more stable (non-divergent) joint patterns during treadmill running than do novice runners. Future work manipulating footwear conditions in novel populations or in different performance environments will give further insight into the influence of running shoes on motor control mechanisms during the high-stress task of running.

Summary and Conclusions

From a dynamical systems perspective of motor control, variability is an important component of stability in movement system, and strong associations have been made across human physiology disciplines between variability structure (i.e. pattern and quantity) and system performance and health. Clinical evidence from therapeutic disciplines outline the risk for musculoskeletal injury or deterioration from high-stress repetitive tasks (Srinivasan 2012). Due to the highly repetitive nature of limb loading in running, it has been suggested that the variability limb segment articulations, particularly those concerned with energy transmission or attenuation, could influence one's risk of developing overuse injury (Hamill 1999, 2012; Heiderscheit 2002; James 2004). However, causal mechanisms between reductions to coordinative variability remain unclear.

In this investigation, the effect of adding cushioned material to the plantar surface of the foot, thereby imposing a constraint on feedforward control mechanisms by altering proprioceptive information gained via foot-ground interaction, was found to have no effect on the amount of variability in lower-limb segment coordination patterns. Caution must be emphasized when drawing relationships between variability quantity and injury risk, and carefully controlled prospective investigations would shed more light on the underlying mechanisms of tissue overload, as well as the consequences of altered levels of coordinative variability. Considerations such as the nature of available visual information, a population's typical training (i.e. practice) context, individual-specific levels of motor variability should be explored in future work examining the influence of variability towards RRI risk, and the role of the running shoe in movement coordination.

Figure Legends

Figure 1: A representation of the 'inverted U' model of variability within functional synergies, where healthy cyclical patterns display high levels of complexity and contain some optimal amount of variability. Too much variability (randomness) and too little variability (periodicity) would reduce the functional capacity of the system below a hypothetical threshold (dashed line) into disease or injury, where signals displaying highly complex patterns are considered most stable and are associated with good health and function.

Figure 2: Phase plot of knee flexion/extension over multiple strides, with phase angle (Φ) calculated from the origin to the respective data point (i). This cyclical phase plot contains both visible consistency and variability in the pattern, representing typical healthy motor coordination which is neither completely periodic nor completely random. Note: number of consecutive strides plotted were reduced from 50 to 10 for clarity.

Figure 3a: FW x FS estimated means and 95% confidence intervals for each joint coupling during the 100 ms interval surrounding touch-down (TD). No significant differences or trends are seen in CRP variability integrated over this time frame between FW conditions ($p > 0.05$) in either MF or RF runners.

Figure 3b: FW x FS estimated means and 95% confidence intervals for each joint coupling during the 100 ms interval surrounding toe-off. No significant differences or trends are seen in CRP variability integrated over this time frame between FW conditions ($p > 0.05$) in either MF or RF runners.

Figure 4: Plots showing the absolute variability in degrees over the entire gait cycle for the joint couplings analyzed. Vertical dashed lines represent touch-down. Columns are separated by footwear (NO; LW; HI).

Figure 5: Visual comparison of joint coupling patterns during the 100 ms surrounding touch-down between footwear conditions (NO = barefoot; LW = minimalist; HI = maximalist) in a representative midfoot (a) and rearfoot (b) runner.

Table 1: Subject Demographics.

	Age		Height (m)		Weight (kg)		Running Volume (km/wk)	
	$\bar{x}$	± sd	$\bar{x}$	± sd	$\bar{x}$	± sd	$\bar{x}$	± sd
Midfoot (MF)	25.00	3.13	1.80	0.03	79.97	13.25	31.51	14.25
Rearfoot (RF)	26.40	6.62	1.63	0.11	64.56	12.22	31.46	14.47
Total	25.70	5.23	1.71	0.12	72.27	14.90	31.99	14.37

Table 2: Joint couplings assessed, and their respective notations.

Segment Rotations	Notation
Thigh internal/external rotation vs. Tibia internal/external rotation	$Th_{I/E} – Tib_{I/E}$
Thigh flexion/extension vs. Tibia internal/external rotation	$Th_{F/E} – Tib_{I/E}$
Thigh flexion/extension vs. Tibia flexion/extension	$Th_{F/E} – Tib_{F/E}$
Thigh adduction/abduction vs. Tibia internal/external rotation	$Th_{Ad/Ab} – Tib_{I/E}$
Tibia flexion/extension vs. Foot plantar/dorsiflexion	$Tib_{F/E} – Ft_{P/D}$
Tibia internal/external rotation vs. Foot eversion/inversion	$Tib_{I/E} – Ft_{I/E}$

Table 3: CRP variability over the 100 ms surrounding touch-down.

	Midfoot			Rearfoot			Significance
	NO	LW	HI	NO	LW	HI	p
$Th_{I/E}$ – $Tib_{I/E}$	5.94 ± 0.78	6.52 ± 0.71	5.63 ± 0.55	6.41 ± 0.78	6.13 ± 0.71	5.76 ± 0.55	0.56
$Th_{F/E}$– $Tib_{I/E}$	4.37 ± 0.56	3.9 ± 0.60	3.41 ± 0.46	4.28 ± 0.56	4.14 ± 0.60	3.97 ± 0.46	0.51
$Th_{F/E}$ – $Tib_{F/E}$	3.51 ± 0.38	2.44 ± 0.34	2.69 ± 0.48	2.93 ± 0.38	2.82 ± 0.34	2.59 ± 0.48	0.19
$Th_{Ad/Ab}$ – $Tib_{I/E}$	6.00 ± 0.67	4.76 ± 0.59	4.21 ± 0.61	5.15 ± 0.63	6.36 ± 0.59	5.86 ± 0.70	0.61
$Tib_{F/E}$ – $Ft_{F/E}$	3.17 ± 0.52	2.58 ± 0.17	3.19 ± 0.52	3.57 ± 0.52	2.50 ± 0.17	2.85 ± 0.52	0.19
$Tib_{I/E}$ – $Ft_{I/E}$	5.60 ± 0.97	6.28 ± 0.95	6.09 ± 0.78	6.48 ± .097	5.57 ± 0.95	5.92 ± 0.78	0.99

Values are reported as estimated means ± standard error of estimate. Tests of significance are for footwear (FW) main effects. NO = barefoot, LW = minimalist shoe; HI = maximalist shoe.

Table 4: CRP variability for the 100 ms surrounding toe-off.

	Midfoot			Rearfoot			Significance
	NO	LW	HI	NO	LW	HI	p
$Th_{I/E} – Tib_{I/E}$	7.15 ± 0.72	7.04 ± 0.63	6.59 ± 0.74	6.70 ± 0.72	5.73 ± 0.63	7.46 ± 0.74	0.47
$Th_{F/E} – Tib_{I/E}$	3.54 ± 0.63	3.90 ± 0.44	4.31 ± 0.56	4.75 ± 0.63	4.98 ± 0.44	5.08 ± 0.56	0.44
$Th_{F/E} – Tib_{F/E}$	3.21 ± 0.64	2.87 ± 0.48	3.05 ± 0.32	3.89 ± 0.64	3.06 ± 0.48	3.03 ± 0.32	0.43
$Th_{Ad/Ab} – Tib_{I/E}$	5.43 ± 0.81	6.41 ± 0.56	7.29 ± 0.66	6.03 ± 0.81	5.47 ± 0.56	6.39 ± 0.66	0.12
$Tib_{F/E} – Ft_{F/E}$	3.57 ± 0.42	3.30 ± 0.53	4.23 ± 0.57	3.38 ± 0.42	2.64 ± 0.53	3.31 ± 0.57	0.23
$Tib_{I/E} – Ft_{I/E}$	5.68 ± 0.74	6.79 ± 0.68	6.43 ± 0.85	6.07 ± 0.74	6.63 ± 0.68	6.87 ± 0.85	0.31

Values are reported as estimated means ± standard error of estimate. Tests of significance are for footwear (FW) main effects. NO = barefoot, LW = minimalist shoe; HI = maximalist shoe.

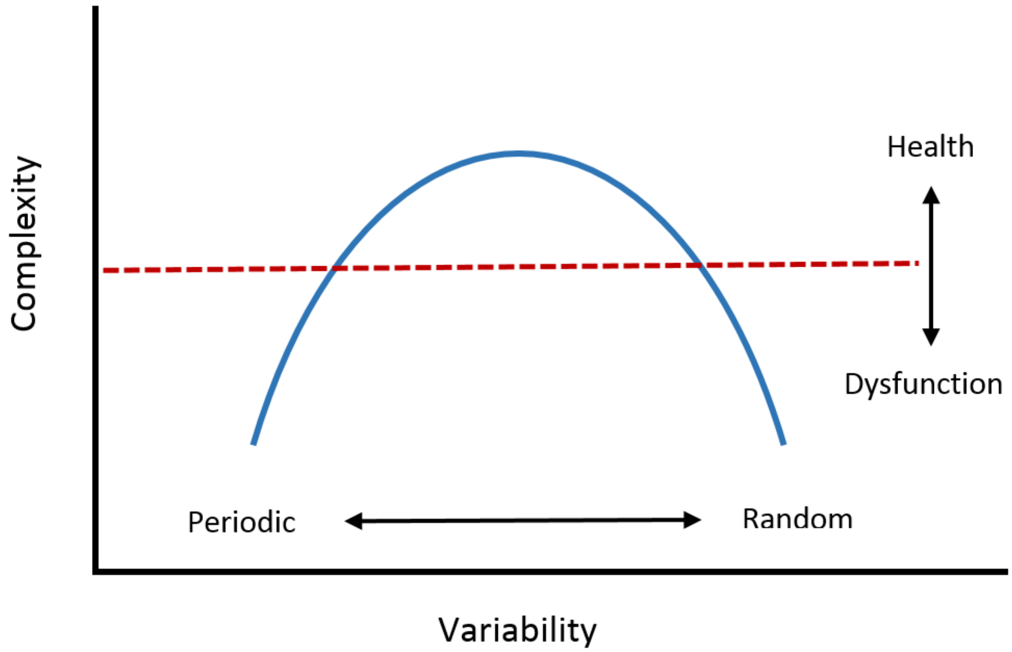
Complexity
Health
Dysfunction
Periodic
Random
Variability

Figure 1.

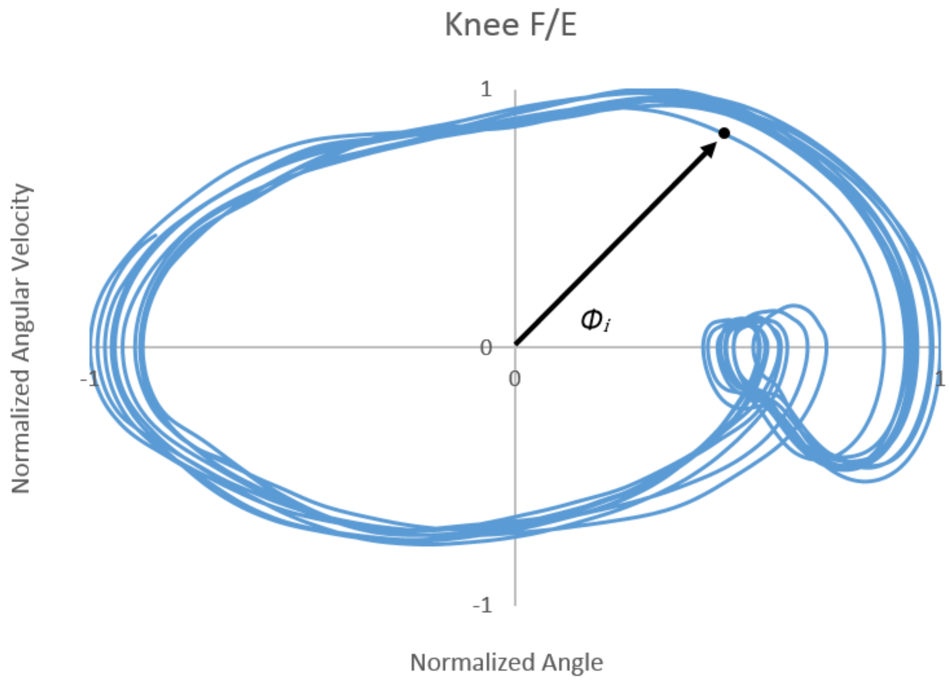
Knee F/E
1
0
-1
-1
0
1
Φ_i
Normalized Angular Velocity
Normalized Angle

Figure 2.

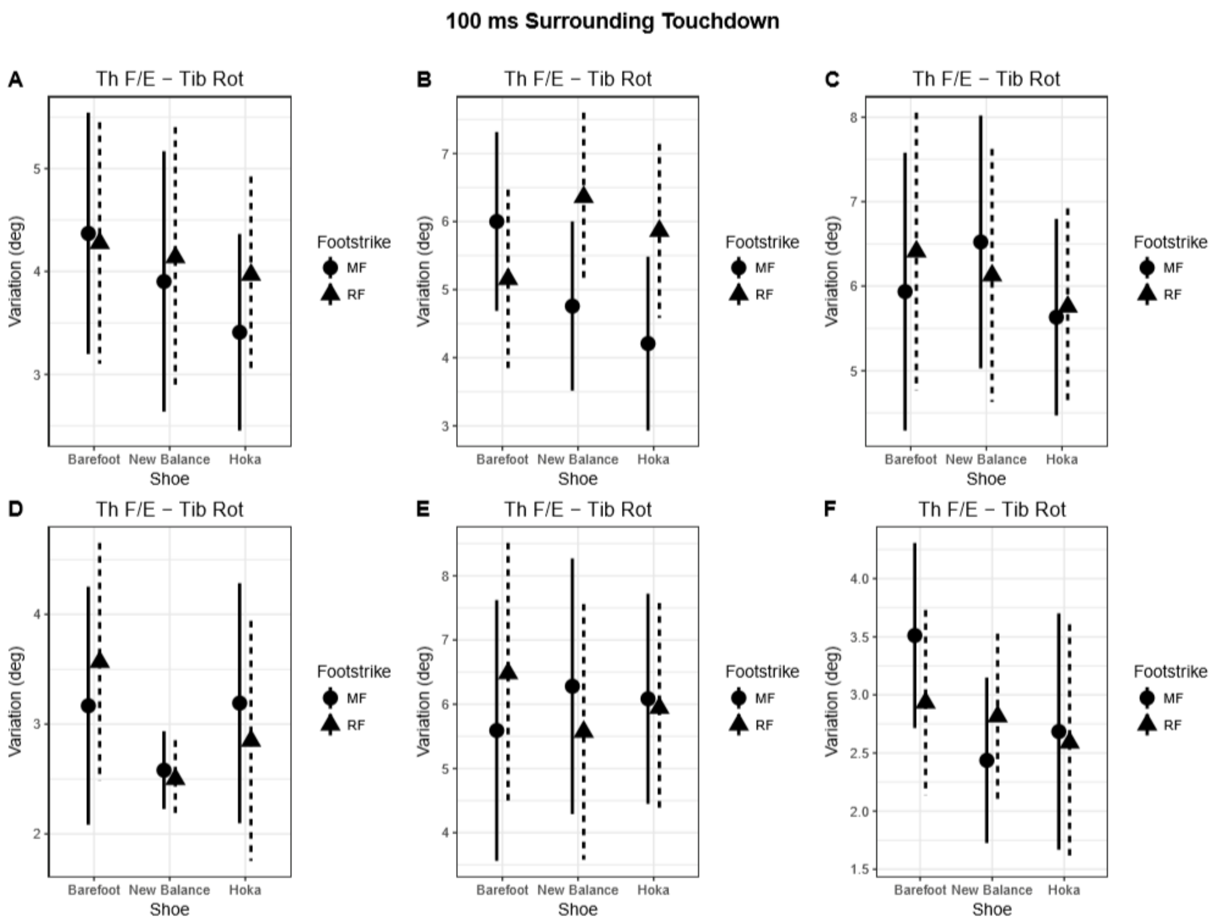
100 ms Surrounding Touchdown
A
B
C
D
E
F
Th F/E − Tib Rot
Variation (deg)
Shoe
Barefoot
New Balance
Hoka
Footstrike
MF
RF

Figure 3a.

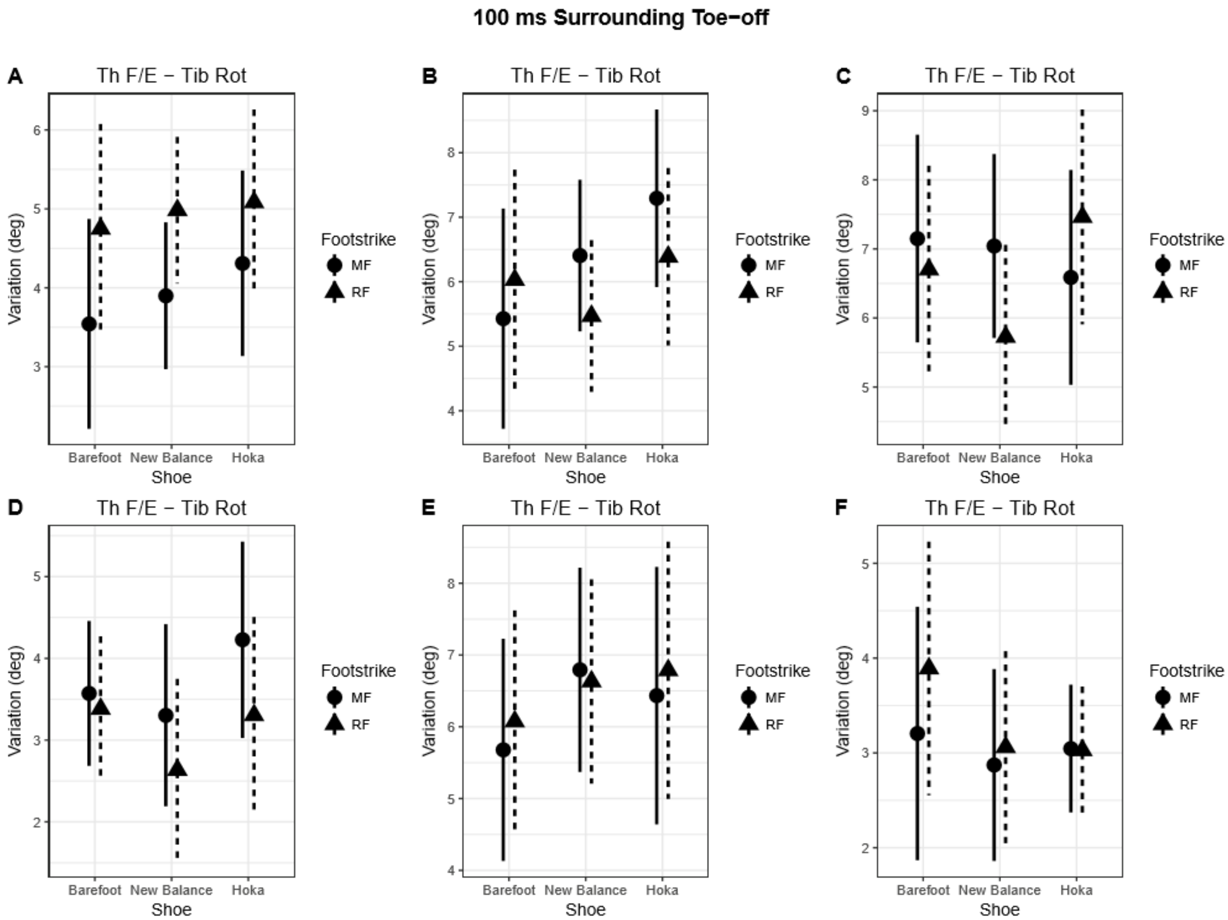
100 ms Surrounding Toe−off
A
Th F/E − Tib Rot
B
Th F/E − Tib Rot
C
Th F/E − Tib Rot
D
Th F/E − Tib Rot
E
Th F/E − Tib Rot
F
Th F/E − Tib Rot
Variation (deg)
Shoe
Barefoot
New Balance
Hoka
Footstrike
MF
RF

Figure 3b.

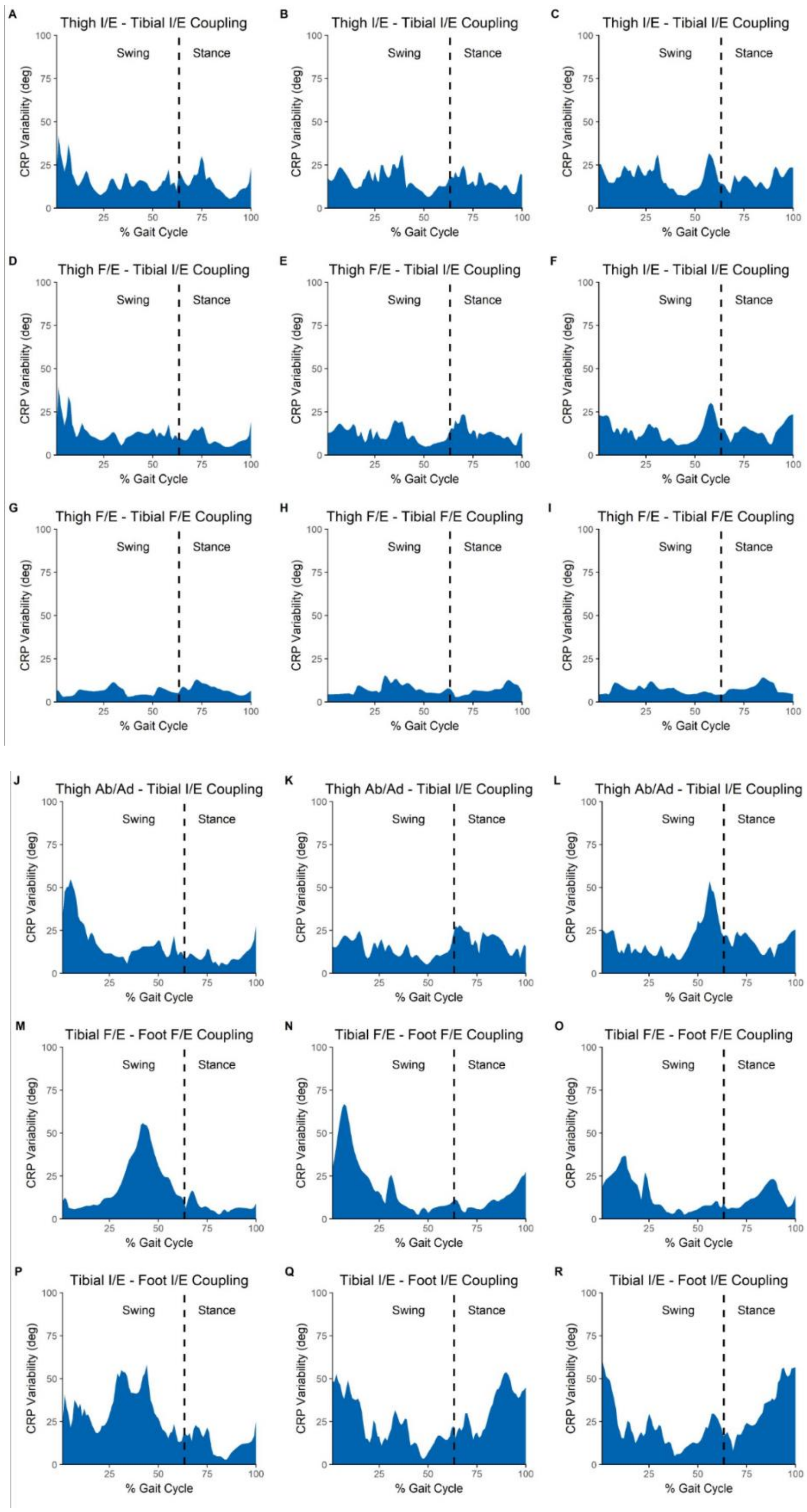
A
Thigh I/E - Tibial I/E Coupling
B
Thigh I/E - Tibial I/E Coupling
C
Thigh I/E - Tibial I/E Coupling
D
Thigh F/E - Tibial I/E Coupling
E
Thigh F/E - Tibial I/E Coupling
F
Thigh I/E - Tibial I/E Coupling
G
Thigh F/E - Tibial F/E Coupling
H
Thigh F/E - Tibial F/E Coupling
I
Thigh F/E - Tibial F/E Coupling
J
Thigh Ab/Ad - Tibial I/E Coupling
K
Thigh Ab/Ad - Tibial I/E Coupling
L
Thigh Ab/Ad - Tibial I/E Coupling
M
Tibial F/E - Foot F/E Coupling
N
Tibial F/E - Foot F/E Coupling
O
Tibial F/E - Foot F/E Coupling
P
Tibial I/E - Foot I/E Coupling
Q
Tibial I/E - Foot I/E Coupling
R
Tibial I/E - Foot I/E Coupling
Swing
Stance
CRP Variability (deg)
% Gait Cycle

Figure 4.

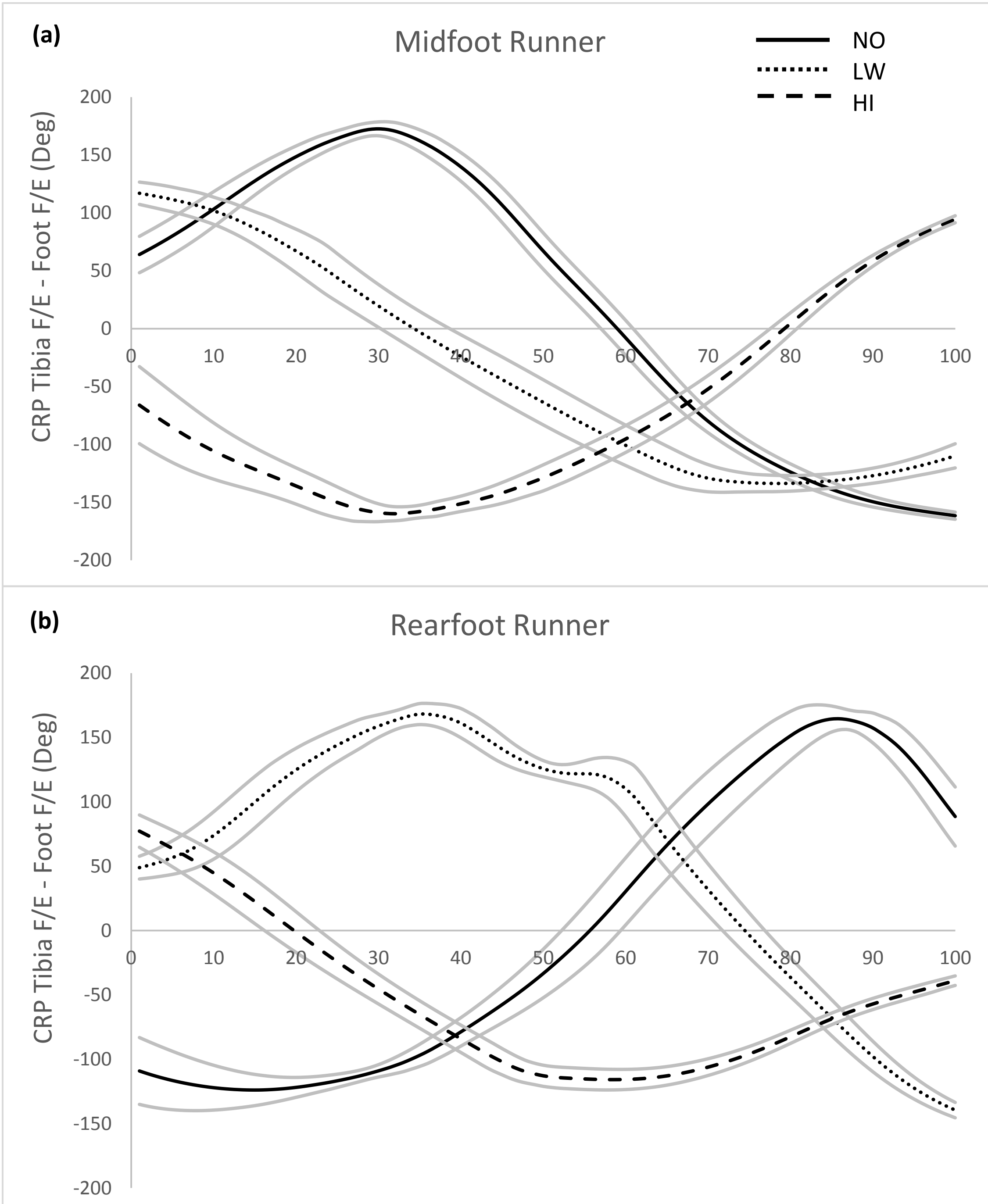
(a)
Midfoot Runner
NO
LW
HI
CRP Tibia F/E - Foot F/E (Deg)
200
150
100
50
0
-50
-100
-150
-200
0
10
20
30
40
50
60
70
80
90
100
(b)
Rearfoot Runner
CRP Tibia F/E - Foot F/E (Deg)
200
150
100
50
0
-50
-100
-150
-200
0
10
20
30
40
50
60
70
80
90
100

Figure 5.

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Chapter 4

Summary

and

Future Directions

The development of running-related injuries (RRI), which ultimately hinder further participation in this most basic of health-related activities, is a persistent issue in today's world-wide running community. This problem is especially stimulating and personal when considering the evolutionary influence running is thought to have had on human musculoskeletal structure and function. Although a number of novel factors likely contribute to the occurrence of RRI, the biomechanical perspective states that injury results from tissue overload, or too much of the wrong kind of stress without adequate recovery. Despite advancements in our understanding of the mechanical factors involved in running performance, as well as in footwear material and design, little consensus has been reached on proper application of footwear and performance techniques to minimize the risk of meeting with pain, injury, and ultimately reduced musculoskeletal integrity and exercise adherence.

Methodological considerations in the available literature make it difficult to separate the contribution of midsole cushioning towards GRF attenuation from that of motor adjustment. What is more, MF running is biomechanically distinct from the RF pattern, characterized primarily to differences in landing kinematics and force attenuation strategies between the two strategies. This would strongly suggest that increasing the compliance of the foot-ground interface through midsole cushioning would have different effects between runners of the different FS patterns. However, the majority of the literature to date have focused exclusively on RF runners. The investigations presented in this work took novel approaches to evaluate how cushioning material under the foot contributes to limb loading forces in the lower leg.

The experimental design in Study 1 provided a comprehensive and ecological context from which to examine the effect of midsole cushioning on force attenuation by allowing both

MF and RF runners to self-pace during overground running in market-available footwear with maximally different amounts of midsole cushioning properties. Although cushioning material objectively provides padding and attenuation, the ability of the human motor system to acutely adjust one's gait based on the somatosensory information provided to the CNS, primarily from receptors in the foot and lower leg, confounds attempts to specifically define the mechanical contribution of the cushioning to force attenuation separate from that of adjustments to gait. Therefore, the statistical methods employed in in the first investigation aimed to control for these adjustments by using linear mixed effects analyses and inputting as covariates those dependent variables shown to significantly influence the kinetic variables being assessed. Further, statistical limitations of previous literature predicated by natural inter- and intra-subject variability were accommodated through this statistical technique, as well as the use of large number of subjects (25+ per group) and trials (50).

A primary observation from this study is the unique response to, and contribution of, midsole cushioning between the respective RF and MF running patterns. Minor but statistically significant changes to sagittal joint angles at TD between FW conditions were detected in both groups, and were specific to running style. In maximally cushioned shoes, RF runners landed with a more flexed hip, extended knee, and toe-up foot, which can be viewed as a heightened expression of their RF pattern. Conversely MF runners tended towards a less expressive MF pattern, with less plantarflexion and a flatter foot orientation at TD. These observations in both groups suggest that midsole cushioning allows for a decreased reliance on intrinsic force attenuation capabilities and an increased reliance on the shoe. Further, runners in both groups displayed increased knee joint stiffness (KJS) in the impact period in the maximally cushioned shoes, and this increase was greater in RF runners.

The assessment of kinematic strategy to prepare for ground impact shows that, although changes to segment orientations at TD were small, increasing the level of midsole cushioning does in fact manifest in a subtle but significant adjustment to landing strategies. These findings are consistent with the expected adjustments based on findings from barefoot (BF) vs. shod comparisons, which suggest that runners will change their gait in a way to reduce the impact loading when the proprioceptive experience of ground contact is enhanced with decreased midsole cushioning. However, when compared to BF running studies, the general maintenance of limb segment orientation at TD indicates that even minimalist shoes provide enough tactile protection to the bottom of the foot to allow the individual to run ‘at will’ with less regard for the sensation of impact and a heavier reliance on the shoe, which may paradoxically result in an increase in GRF characteristics and higher magnitude loads transmitted to the limb than if running completely barefoot with full tactile consequence. Larger adjustments to landing strategy were seen in changes to knee joint stiffness (KJS) during the impact period of stance, which is modulated through muscular pre-tuning prior to touch-down (TD).These results together substantiate suggestions that between differences in midsole cushioning properties, muscular activation strategies may play a more predominant role in impact accommodation between shod conditions of differing midsole cushioning properties than by changing segment orientation alone (Nigg, 2001; Hamill, 2014).

Additionally, when statistically isolating the effect of midsole cushioning on kinetic dependent variables, both groups of runners experienced larger peak knee compression forces and peak knee moments when running in highly cushioned shoes. The kinetic and kinematic results together suggest that when in highly cushioned shoes, both types of runners use a more ‘knee-dominant’ strategy that is more heavily reliant on passive translational rather than active rotational attenuation mechanisms.

Current evidence indicates that midsole cushioning primarily contributes to attenuation of GRF transmission through decreased LR and tibial shock (Sinclair 2013, 2016). This conclusion is substantiated through our findings, where the largest kinetic differences between shoe conditions were reductions in initial loading rate (ILR) and a gain in impact-period GRF attenuation between the ground and the knee in the RF group. However we found this was not the case in MF runners, who experienced an increase in ILR and reduced impact-related GRF attenuation. High-frequency forces have received large attention as a likely injurious mechanism, and a mismatch between tissue resonance frequency and that of input forces may be a harmful stimulus and lead to tissue degradation (Wakeling, 2003; Boyer, 2007). Reducing the high-frequency shock transmitted through the heel would likely provide benefits to bone and knee injury or deterioration (Radin, 1976; Burr, 2003), and may be a critical benefit of padding provided by the midsole cushioning, at least in those runners landing heel-first.

Gruber (2014) showed that MF runners have lower spectral power in the high-frequency ranges than RF counterparts, thought to be due to greater active shock attenuation mechanisms, such as eccentric joint work and deformation of the foot during the loading phase of stance in MF runners. To our knowledge, GRF spectral density power attenuation comparisons have not been performed between footwear conditions in both running patterns. An assessment of shock attenuation in both high and low frequency spectrums in both FS patterns would provide more detail on the cushioning contribution towards reduction of likely injurious mechanisms, respective to one's running style.

Effects of the shoes as measured by both kinematic and kinetic variables were notably amplified in the RF runners, reiterating the differential effect of midsole cushioning on running mechanics between these two categories of gait patterns. Although some more recent studies have incorporated habitual runners of both FS (Paquette, 2013; Hamill, 2014), to our knowledge evidence highlighting the significantly different contribution of midsole cushioning to reductions in limb loading variables has not before been presented, and is a relevant consideration in footwear prescription. Segment kinematics, particularly of the foot and ankle during time periods surrounding TD, are the most distinct aspects between the FS patterns. These between-group differences were reflected in the directionally opposite effect of increased midsole cushioning on impact-period kinetic variables. One limitation of the present study and others examining the effect of footwear on impact characteristics is the use of a single-segment foot in the inverse dynamic calculations performed for kinetic variables. Articulations of the foot and ankle, such as the flattening of the medial longitudinal arch or the tri-planar pronation motion, play a large role in shock attenuation and energy storage and return, (Nigg, 1993; Leardini, 2007), and especially in MF runners. More detailed foot modeling would provide further insight into the mechanisms underlying the differential contribution of midsole cushioning to dynamic limb loading characteristics between the respective FS patterns.

Simple gait-training interventions have been proven effective in reducing impact characteristics (Samaan, 2014). It is possible that prospective intervention studies of this nature may prove more successful in reducing injuries regardless of footwear. Another stimulating question generated by this evidence is whether reductions proprioceptive feedback available when wearing more highly cushioned shoes deprives the individual of information beneficial to learning a less harmful gait pattern (i.e. the sensation of ground impact may be a necessary feedback stimulus to consciously regulate one's gait in attempt to reduce these sensations, where a lack of this sensory information may reduce efficacy of learning). Indeed, proponents of barefoot running practice argue that without the protection provided by shoes, the runner would be required to ambulate in such a way that would not lead to eventual tissue degradation, injury, and dysfunction, but would in fact stimulate running mechanics more in harmony with our evolved physiological capacity.

The cushioning effect on motor output and tissue overload risk is further explored in Study 2, which uses concepts based in dynamical systems theory to investigate whether midsole cushioning affects coordinative variability in the leg segment articulations or joint couplings. The runner is often modeled as a 'mass-spring' system to examine the transfer of energy during the stance phase of gait (Farley, 1996, 1998; Kerdok, 2002). The compression and return of the leg spring during stance is primarily accomplished through joint articulations or couplings. Coupling patterns that are too consistent or invariable would increase the homogeneity of stress profiles to specific tissue sites, correlating to increased local stress accumulation and thus RRI risk (James, 2004). Additionally, rigid coordination mechanics are thought to represent a functional system that is less capable in maintaining a stable movement outcome in the face of perturbations or unexpected environmental alterations, and therefore represents overall decreased functional capacity of the system (Hamill, 2012).

The objective of this investigation was accomplished by quantifying the amount of variability in lower-limb segment articulations, or joint couplings, between footwear conditions using continuous relative phase (CRP) techniques. Despite the differences in tactile experience afforded by the footwear conditions employed – from maximal padding to thin rubber protection to completely barefoot – no difference in coordinative variability was observed. Certain limitations must be taken into consideration, however the result of this and previous investigations strongly suggest that, at least in a controlled laboratory setting, alterations to somatosensory information are not a sufficient disturbance to overall sensory feedback to influence the control of leg segment coordination in healthy, pain-free runners. Nonetheless, much work has left to be done in understanding the complete function of coordinative variability in running, and the consequences of different subject-specific amounts of variability when performing high-stress tasks like running. The experimental design employed in Study 2 provides a foundation for future work examining the effect of proprioceptive acuity on intrinsic levels of coordinative variability within the leg segments during running.

A potential limitation of Study 2's protocol was the entirely static and relatively proximal visual environment. Similar protocols to the one employed that incorporate an immersive dynamic visual environment, such as running down a forest trail, may increase the motor system's sensitivity to the available somatosensory feedback, and thus the potential for cushioning to effect. This experimental setup would also have the capability to incorporate perturbation protocols, which would allow for questions to be asked about how variability contributes to running pattern stability. One's ability to recover from perturbation, or running over uneven surface, might be affected by the structure of the variability in their movement coordination patterns. Habitual barefoot runners are thought to have higher levels of coordinative variability than habitually shod runners (Altman, 2011), but this suggestion stimulates intriguing questions on where this differences arise. It is possible that group-specific training context has instilled in these two types of runners a different level of variability in their learned motor coordination. One accustomed to running through uneven and sometimes unpredictable natural landscape would likely employ different coordination strategies than one accustomed to the monotony of treadmill running.

Results from Study 2 and previous investigations which have quantified coordinative variability during running in healthy and experienced runners (Dierks, 2007; Hamill, 1999) could prove a useful reference for future investigations focusing on clinical and at-risk populations, as well as a platform for future interventional studies. For example, gait speed has been found to have a significant effect on the variability of pelvic-trunk coordination during walking (van Emmerik, 1996), and is an important consideration for investigations of the present focus. Future studies may manipulate gait speed based on a percentage of the subject-specific maximal range to highlight whether this variable is a relevant consideration in defining the effects of footwear interventions. Further, separating runners into groups or a continuum based on their individual levels of variability may allow for prospective investigations to more clearly define the true implications of motor variability towards one's risk for sustaining tissue overload.

A novel and notable finding from these data is the within-subject difference in CRP coupling patterns between all three footwear conditions. Faulty or excessive joint kinematics are hypothesized to increase RRI risk by generating abnormal tissue stress patterns (Stergiou, 1999; DeLeo, 2004). It is likely that a more compliant foot-ground interface provided by increased amounts of midsole cushioning would alter spatiotemporal properties of the foot and subsequent in-series segment motion, not to mention time- and length-dependent muscle and soft tissue stiffness properties contributing to force attenuation. Although the use of foot orthotic devices have been shown to significantly influence joint excursions at both the ankle and knee (Laughton, 2003) little is known about the effect of midsole cushioning alone on joint coupling patterns. This effect of midsole cushioning properties on CRP patterns in lower-limb joint couplings will therefore need to be quantified for comparison in future analyses.

In short, health and performance professional must exercise caution when attempting to delineate causal mechanisms between coordinative variability and risk for tissue overload. Carefully controlled prospective investigations would allow for stronger associations to be made between performance variability and overuse injury risk, as well as underlying mechanisms contributing to alterations in variability. Manipulation of dependent variables such as the nature of available visual information, a population's typical training (i.e. practice) context, and amount of individual-specific motor variability should be explored in future work examining the influence of variability towards RRI risk, as well as the effect of the running shoe on movement coordination.

The debate surrounding the efficacy of running shoes on injury prevention was best and most notably articulated by Daniel Lieberman, who offered the conclusion that "how one runs is probably more important than what is on one's feet, although what is on one's feet probably affects how one runs." Evidence provided in this work lends support to this argument, and highlights the fact that while cushioning and protection provided by the running shoe may impart some benefit to the wearer, it is critical for the practitioner to be cognizant of how they run, and what they are running in (and on) when developing individual-specific health intervention or performance training programs.

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