Effects of shoe midsole cushioning on low back impact shock attenuation in recreational runners

By

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Abstract

Running is one of the most widely practiced and accessible forms of physical activity. When the foot contacts the ground during running, impacts on the order of two to three times body weight are generated. The impact force is attenuated by joints and propagated through each subsequent segment from the foot to the head over 600 times per kilometer and has been associated with the development of various pains and injuries across runners. Consequently, mechanisms of shock attenuation have been widely researched across the last few decades, where previous work has aimed to examine how joint positioning, eccentric muscle activation, passive soft tissue structures, and other strategies may help mitigate this impact on the musculoskeletal system. However, how these mechanisms are presented in the lumbar spine, which largely influence the delivery and experience of this impact shock in the upper body and head, is not well understood.

Further complicating this area is the role of running shoe cushioning, or the midsole. Softer and thicker midsoles have been shown to interact with lower limb kinematics and muscle activation, contributing to differences in leg stiffness and shock attenuation. Therefore, it was hypothesized that more compliant midsoles would produce similar results in the lumbar spine. Specifically, the purpose of this study was to investigate if increased shock transmission and shock attenuation occurred in the lumbar spine in response to softer midsoles. It was further hypothesized that differences would exist in sagittal knee and lumbar flexion angles and trunk muscle activation across midsole cushioning stiffness as well as between sex.

Twenty (10M, 10F) pain-free recreational runners who averaged a minimum weekly mileage of 16 km were recruited to participate in this study. Subjects were asked to run on a treadmill at 3.3 m/s for five minutes in each of three shoe conditions that ranged in their midsole cushioning stiffness, quantified prior to use in running via a mechanical testing system. Sagittal kinematics of the lumbar spine, pelvis, and right lower limb were collected using an active motion capture system, mean bilateral muscle activity, co-activation indices, and phase lags between co-activation of the lumbar erector spinae, rectus abdominus, and external obliques were measured via surface electromyography, and accelerometers were placed at the distal tibia, borders of the lumbar spine, and head to calculate peak resultant acceleration as well as shock attenuation in the frequency domain. All variables were calculated during stance phase and averaged across fifteen consecutive strides. Two-way mixed measures analyses of variances were used to assess differences across shoe conditions and between sexes.

Softer and more compliant midsoles resulted in increased ankle plantarflexion and knee extension leading to differences in low frequency shock attenuation, but the low back was not particularly responsive to midsole stiffness. Similar tibial and lumbar spine acceleration magnitudes were observed across all midsole stiffness conditions, and neither lumbar posture nor trunk muscle activation and co-activation changed with footwear. Minor differences were observed between sex, suggesting that females may employ slightly different shock attenuation mechanisms particularly at the hips and lower limbs, but future investigations are necessary to better understand the specific shock attenuation mechanisms involved. Overall, these results add to the evidence that midsole cushioning stiffness may influence the lower limb but suggest that such changes are accommodated by the time the shock reaches the lumbar spine.

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List of Abbreviations

ASTM	American Society for Testing and Materials
BW	Body Weight
BMI	Body Mass Index
CCI	Co-activation Coefficient
COM	Centre of Mass
EO	External Oblique
EVA	Ethylene Vinyl Acetate
FFS	Forefoot Strike
GRF	Ground Reaction Force
HTD	Heel-to-Toe Drop
IC	Initial Contact
IVD	Intervertebral Disc
LBP	Low Back Pain
LES	Lumbar Erector Spinae
MFS	Midfoot Strike
MVC	Maximal Voluntary Isometric Contraction
PGS	Nike Pegasus 38
RA	Rectus Abdominus
RCT	Nike React Infinity 2
RFS	Rearfoot Strike
ROM	Range of Motion
RRI	Running-Related Injury
SA	Shock Attenuation
TO	Toe Off
TPU	Thermoplastic Polyurethane
vGRF	Vertical Ground Reaction Force
ZMX	Nike ZoomX Invincible

1 Introduction

Running is defined as the movement strategy or aerobic activity with periods of double float at the start and end of the swing phase where neither foot is in contact with the ground (Novacheck, 1997). It is one of the most accessible and practiced sports worldwide (Hulteen et al., 2017), and associated with numerous health benefits (Rainville et al., 2004; Trompeter et al., 2017; Woolf & Glaser, 2004). Aerobic activity can increase blood flow and delivery of nutrients, promoting healing and reducing stiffness that can otherwise result in pain symptoms (Gordon & Bloxham, 2016; Nutter, 1988; Sculco et al., 2001). Thus, it is often recommended for individuals suffering with low back pain (Nutter, 1988; Rainville et al., 2004; Sculco et al., 2001).

However, runners are not excluded from such pain. Low back pain (LBP) is considered a global health issue that is estimated to affect nearly 80% of the population at some point in their life (Papageorgiou et al., 1995), with the annual cumulative incidence reaching nearly 20% (Cassidy et al., 2005). While incidence and prevalence of LBP in the running community are lower than the general population, up to one in ten recreational runners will experience LBP within their first year of running (Jacobs & Berson, 1986). Back pain can lead to high treatment costs, time off work and training, and an overall decreased quality of life (Mortazavi et al., 2015). Therefore, in both the general population and running community, understanding the etiology of injury and its prevention are continuous priorities. But before looking at the relationship between running and LBP, examining the role of the lumbar spine in a healthy population of runners is critical.

An area that has received particular attention is the repetitive exposure to impulsive forces and shock waves during running. When the foot hits the ground, ground reaction forces (GRF) on the order of 2 to 3 times body weight (BW) are generated, which compares with GRFs of 0.5 to 2 times BW during normal walking (Pratt, 1989). This force is experienced by the foot as it lands and decelerates, and then is transmitted to the next proximal segment, creating a transient *shock wave* that passes from the ground to the foot, ankle, tibia, and so on until it reaches the head (Whittle, 1999). The average recreational runner strikes the ground upwards of 600 times per kilometre (Lieberman et al., 2010), resulting in more than 12,000 impacts across both feet during a weekly mileage of 20 km (Shorten & Winslow, 1992). Given that early *in vitro* studies demonstrated that walking on concrete for extended periods led to the stiffening of the subchondral bone and subsequent degeneration of cartilage in the knees of rabbits (Radin et al., 1973), it is reasonable to suggest that low level, everyday loading could play a role in subsequent LBP or injury (Adams et al., 2000; M. L. Chu et al., 1986; Radin et al., 1973), and the even greater forces from running may have drastic implications on spine health.

Shock attenuation (SA) is the term used to describe these strategies of mitigating or dissipating the force experienced during impact and can occur via both active and passive mechanisms. For instance, lumbar lordosis and lumbar flexion during gait help dissipate the shock into bending and rotational deformations instead of axial compression (Adams, 2003; Castillo & Lieberman, 2018; Schache et al., 2002). This is further supported by adjusting leg stiffness through changing knee angles, as well as energy absorption through eccentric muscle contractions (Pratt, 1989). However, excessive trunk activation, as seen in individuals with LBP, may lead to compromised SA. Studies have shown that people with LBP have weaker trunk musculature with reduced endurance or increased co-activation of muscles to compensate for their impaired ability to stabilize the spine during dynamic movements (Ghamkhar & Kahlaee, 2015; Granata & Marras, 1995; Lamoth et al., 2006; van der Hulst, Vollenbroek-Hutten, Rietman, & Hermens, 2010; van der Hulst, Vollenbroek-Hutten, Rietman, Schaake, et al., 2010), contributing to reduced SA capacity (Nourbakhsh & Arab, 2002; Raabe & Chaudhari, 2018). Consequently, there is a need to better understand how the lumbar spine works in conjunction with muscle activation and the lower limb to execute shock attenuation.

A further complication involves the interaction with footwear. The foot-ground interface can largely influence leg stiffness, muscle activation, and loading characteristics. Over the last few decades, vast research has been directed towards manipulating the midsole layer, which is a layer of foam between the lining of the insole and the outsole designed to provide cushioning. Studies have examined shoes with midsoles of varying thickness, softness, and stiffness, where there is a general agreement that softer and thicker cushioning can increase contact time and reduce loading rates in the lower extremities (Baltich et al., 2015; Chambon et al., 2014; Malisoux, Delattre, Meyer, et al., 2020; Nigg, 1986; Shorten & Mientjes, 2011; Sterzing et al., 2013), and even support for the use of footwear as a conservative treatment option for LBP (Wosk & Voloshin, 1985). However, limited evidence has been published regarding the effects of shoe cushioning on the spine. A noteworthy study investigated lumbar posture in response to shod running and found lumbar lordosis played an important role in dynamic lumbar SA (Castillo & Lieberman, 2018), but other mechanisms of SA such as lower limb kinematics and trunk activation were not examined, and only one shoe condition was examined. With the growing research on midsole cushioning, there is potential to understand how differences in midsole cushioning stiffness – a measure of the cushioning deformation under load, and thereby how the musculoskeletal system experiences the ground impact – may influence joint posture and muscle activation, and assist or impede shock attenuation in the lumbar spine. Studying this in healthy runners may inform the potential implications of these different SA factors on low back pain.

1.1 Research Questions & Hypotheses

The primary purpose of this thesis was to examine the effects of midsole cushioning on shock attenuation at the low back in pain free recreational runners. Findings from this project contribute to the advancement of knowledge in both the fields of spine biomechanics and the footwear industry in the context of running. This research aimed to answer the overarching question of how shoe cushioning interacts with loading – to provide some insight into the subsequent risk of injury as measured by shock attenuation – beyond the lower extremity. It was hypothesized that such relations exist, and they have implications on future work regarding injury risk and prevention, thus expanding on the limited work regarding LBP in running.

Specific questions that were examined and the corresponding hypotheses are:

1. How does midsole stiffness affect shock transmission and attenuation in the low back during running?

   Hypothesis 1: *Running in softer and more compliant midsoles will result in increased acceleration magnitudes at the low back.*

   Hypothesis 2: *Running in softer and more compliant midsoles will result in decreased shock attenuation in the low back.*

   Few studies have been conducted regarding segment accelerations in varying midsole stiffness, but lower tibial acceleration magnitudes have been observed when running in maximal shoes with thicker midsoles (TenBroek et al., 2014; Xiang et al., 2022). However, increased cushioning also results in increased leg stiffness and reduced knee angles during running, which could contribute to greater shock transmission to the low back leading to greater magnitudes recorded at the level of the low back (Hardin et al., 2004; Kulmala et al., 2018). No differences in head accelerations were observed across midsoles of differing stiffness (Bruce et al., 2019), suggesting that increased midsole stiffness must also result in greater shock attenuation at the low back. Therefore, softer and more compliant midsoles may lead to decreased shock attenuation in the low back, despite the increased shock transmission through the lower limbs resulting in higher acceleration magnitudes at the low back.

2. How does trunk and low back muscle activation change with midsole stiffness during running?

   Hypothesis 3: *Running in softer and more compliant midsoles will increase trunk and low back muscle activation.*

   Previous studies have found high inter-subject variability in lower limb muscle activation when running in shoes of different midsole hardness and stiffness (Apps et al., 2016; Roy & Stefanyshyn, 2006; Wakeling et al., 2002; X. Wang et al., 2017), but lumbar paraspinal muscle activity consistently decreased when running in minimal shoes over a 4-week training program (S. P. Lee et al., 2018), which are classified on the harder and stiffer end of the cushioning spectrum. Likewise, increased trunk muscle activation has been observed when activity or surface variables were more unstable (Behm et al., 2009; Buchecker et al., 2013; Lisón et al., 2016). Thus, decreasing midsole cushioning stiffness and hardness should lead to a change in trunk muscle activity that reflects the increased need to stabilize the spine when on unstable or cushioned surfaces, thereby increasing muscle activation.

3. How do low back and lower limb kinematics change with midsole stiffness during running?

   Hypothesis 4: *Running in softer and more compliant midsoles will result in less knee flexion and lumbar spine flexion at initial contact during running.*

Due to the greater shock experienced at contact with increased midsole stiffness, increased leg flexion angles and exaggerated lumbar lordosis are required to assist with shock attenuation (Delgado et al., 2013; Derrick, 2004; McMahon et al., 1987). Hence, in the opposite case, due to softer and more compliant midsoles producing less shock, running in softer and more compliant midsoles are expected to result in reduced knee and low back flexion angles.

4. Are there sex differences in shock attenuation when running in midsoles of different stiffness?

   Hypothesis 5: *Females will experience greater segment acceleration in the lower extremity compared to males during running.*

   Hypothesis 6: *Females will experience greater overall shock attenuation compared to males during running.*

   Although studies on sex differences in running and shock loading are inconclusive, preliminary research has shown that females experience greater tibial (Dufek et al., 2009; Giandolini et al., 2019) and sacral (Sinclair et al., 2012) acceleration compared to males. However, no difference in head acceleration magnitudes were observed, suggesting that females may have greater shock attenuation capacity between the sacrum and the head (Sinclair, 2016). Sex differences in lower extremity kinematics such as greater hip adduction and knee abduction in females may also influence resulting acceleration magnitudes (Baltich et al., 2015; Ferber et al., 2003; Sinclair et al., 2012), thus contributing to dissimilarities in attenuation mechanisms. Lastly, Dufek et al. (2009) reported significant differences in shock attenuation from the tibia to the head in females but not males when running on surfaces of varying stiffness, implying that the interface on which the running is performed is likely to present differently across sex.

1.2 Study Overview

To address the proposed research questions, a sample of healthy male and female recreational runners were recruited to undergo a running protocol while wearing shoes with different cushioning properties. Footwear midsole stiffness was characterized via mechanical testing and trunk and lower extremity shock attenuation, kinematics, and low back muscle activity were recorded during controlled running on a treadmill. Shock attenuation was calculated using accelerometers attached to the participants' lower limbs and trunk. Statistical analyses determined whether differences existed between sex and stiffness conditions. An outline of the experimental protocol and corresponding research questions is presented in Figure 1.1.

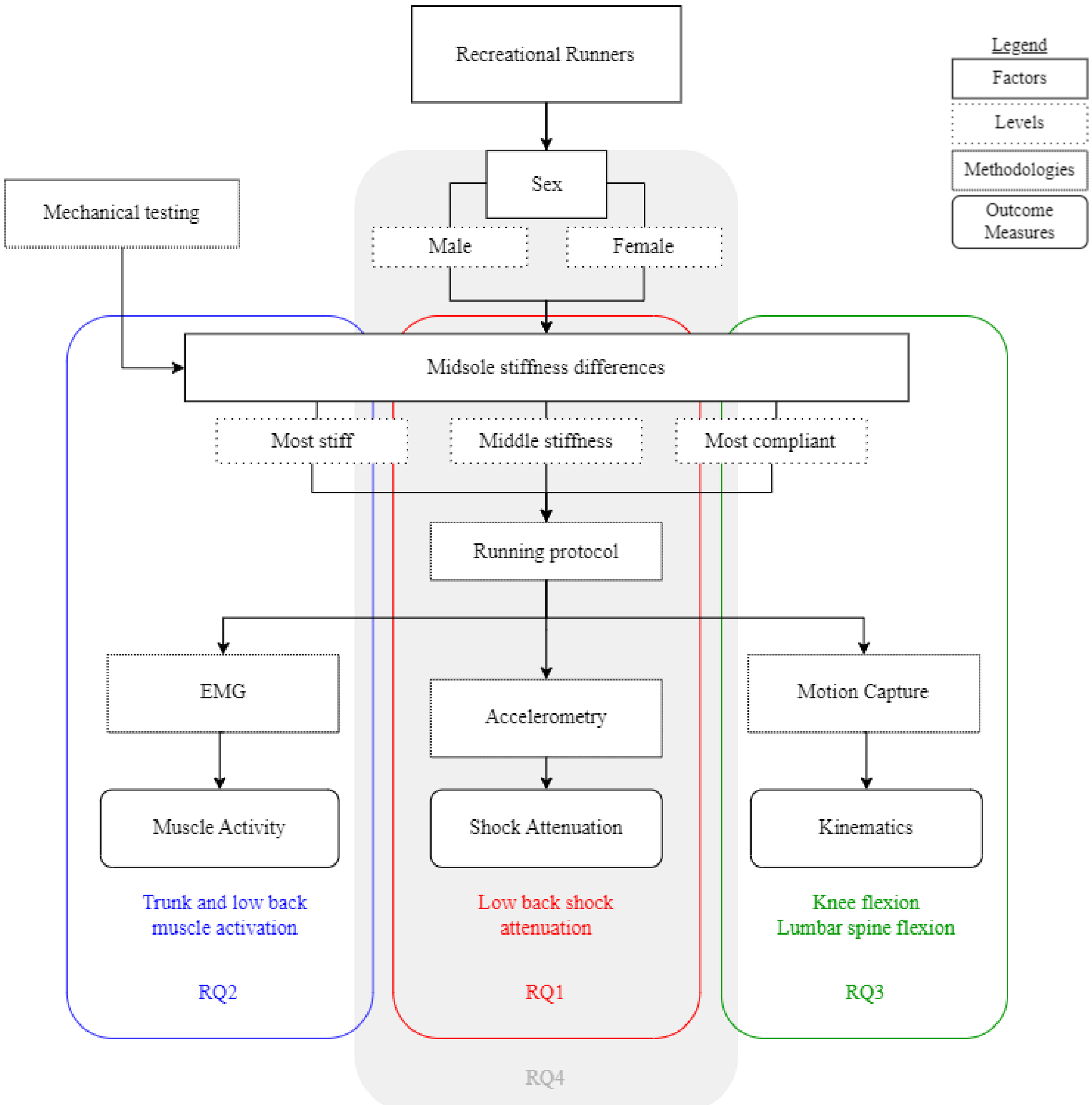

Figure 1.1: *Overview of anticipated study design and related research questions. RQ = research question, as noted in Section 1.1.*

2 Literature Review

This review is divided into five sections. First, an overview of running parameters and definitions will be provided (2.1). Second, the predictive factors that have been investigated with regards to low back pain in running are discussed (2.2). Third, the relationship between impact loading, the generated shock, and attenuation mechanisms (2.3). Fourth, the research on the role of footwear on injury prevention and shock attenuation, including how midsole cushioning is characterized is reviewed (2.4). Fifth, a summary of the key messages and current gaps in the literature are presented (2.5).

2.1 Running Parameters

2.1.1 Gait Cycle

The gait cycle begins when a foot contacts the ground, typically termed initial contact, and ends prior to the next initial contact by the same foot. The stance phase refers to when there is contact with the ground and is bookmarked by the initial contact and toe-off of the given foot, while the swing phase begins after toe-off (Figure 2.1). Unlike walking, running is characterized by the presence of double float or flight phase and absence of double stance, as only one foot is in contact with the ground at a time (Novacheck, 1997). Therefore, by definition, the stance phase is less than 50% of the gait cycle, with its duration decreasing as velocity increases (Thordarson, 1997). The phases are described in Figure 2.1, where stance is further divided into absorption and propulsion, and swing apportioned into initial and terminal periods.

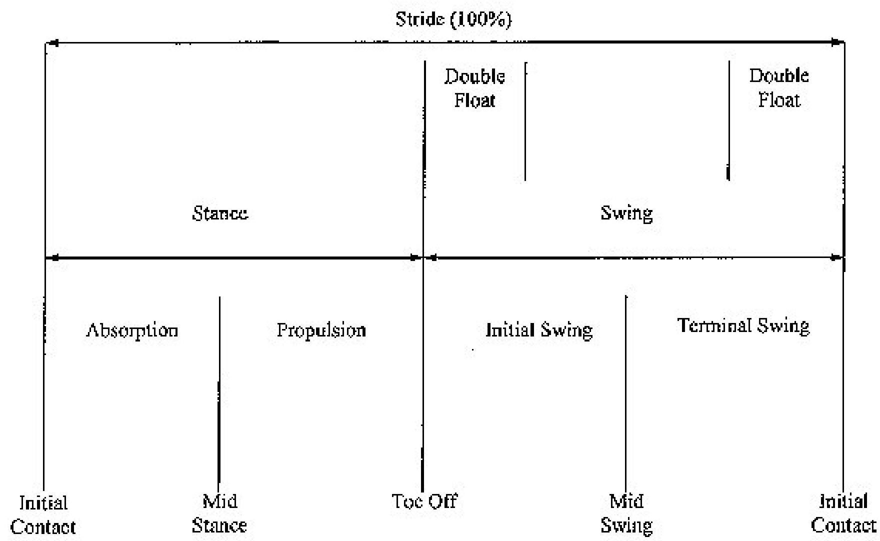

***Figure 2.1**: Events of the running cycle. Image adapted from* (Thordarson, 1997).

Events may be referred to according to their duration or position within the gait cycle phases. For instance, impact peak is suggested to be the first peak within the first 50 ms of the cycle (Cavanagh & Lafortune, 1980) or to coincide with 13% stance (Blackmore et al., 2016). Running biomechanics are particularly concerned with events during the stance period, termed midstance, as many parameters are relevant to the ground reaction forces experienced, periods of acceleration and deceleration, and the exchange of kinetic and potential energy (Novacheck, 1997).

2.1.2 Foot Strike Pattern

Runners are classified by their preferred foot strike pattern, or strike index, during the initial contact. Majority of both elite (Hasegawa et al., 2007) and novice (Bertelsen et al., 2013) runners adopt a rearfoot strike (RFS) pattern, defined as initial ground contact with only the posterior third of the foot, and commonly confirmed by examining centre of pressure data (Cavanagh & Lafortune, 1980). Alternatively, RFS can be defined by a plantar angle between the earth horizontal and the plantar surface of the foot of greater than 0 degrees at the impact instance (Lieberman et al., 2010). Midfoot (MFS) and forefoot strike (FFS) runners make first contact with the middle and anterior thirds respectively, corresponding to plantar angles of 0 or less than 0 degrees (Cavanagh & Lafortune, 1980). These strike indices are typically favoured only by habitually barefoot endurance runners (Lieberman et al., 2010), and make up a very small proportion of runners. For instance, Hasegawa et al. (2007) found that the proportion of RFS, MFS, and FFS among 415 elite marathon runners to be 74.9%, 23.7%, and 1.4% respectively. The percentage of RFS among sub-elite and novice runners were even greater, at 88.9% (Larson et al., 2011) and 98.12% respectively (Bertelsen et al., 2013). Therefore, studies on recreational and long-distance running have often restricted their sample to RFS runners when foot strike pattern may be a confounding factor (Davis et al., 2016; Hamill et al., 2011; Kulmala et al., 2018).

This is the case when outcome measures such as responses to different landing geometries, speed, surfaces, and footwear conditions are assessed (Delgado et al., 2013; Hamill et al., 2011; Hardin et al., 2004; TenBroek et al., 2014). For instance, when comparing changes in lumbar range of motion (ROM) and shock attenuation between foot strike patterns on treadmill running, Delgado and colleagues (2013) reported that changing from RFS to FFS was associated with reduced absolute ROM and shock attenuation, but lower impact force values were observed in FFS runners. Multiple studies have also found correlations between increased speed and more anterior landing positions (i.e., MFS and FFS) (Giandolini, Horvais, et al., 2013; Hasegawa et al., 2007; Ruder et al., 2019), where speeds of 2.7 to 4.47 m/s have been used in experiments investigating RFS runners only. Lastly, studies have suggested that FFS patterns may be associated with reduced injury risk (Daoud et al., 2012; Lieberman et al., 2010), but modifying natural foot patterns can be accompanied with its own consequences and should be approached with caution (Almeida et al., 2015a; Giandolini, Arnal, et al., 2013; Giandolini, Horvais, et al., 2013).

2.1.3 Limb Preference

Limb preference is defined by the preferential use of one side of the body to perform a motor action (Carpes et al., 2010). Studies on lower limb preference, which is frequently determined by observing which leg is selected to kick a ball, in running have been founded on the rationale that asymmetry affects the risk of injury and performance due to anatomical and kinematic differences between the preferred and non-preferred legs (Carpes et al., 2010; Pappas et al., 2021). This theory has been partially verified through findings connecting greater asymmetries in plantar foot pressure with greater rates of tibial stress injuries on the side of the preferred limb (Bredeweg et al., 2013; Zifchock et al., 2006) or greater energy expenditure (Beck et al., 2016).

However, research that expressed asymmetry as a relative index did not find any difference between injured and non-injured runners (Zifchock et al., 2006), nor side to side kinematic and kinetic differences in female runners when running overground at 3.35 m/s (Brown et al., 2014). In support of the latter outcomes, Pappas et al. (2015, 2017) further found that significant asymmetry was only reported in flight time and maximal ground reaction force, but not in other kinematic variables such as leg and vertical stiffness in recreational males running on a treadmill at both moderate (4.44 m/s) and fast speeds (6.67 m/s). In this case, the reason why the asymmetry in flight time and ground reaction force were not expressed in other parameters could be explained by intra-limb compensations and neuromuscular activation factors where stiffness is maintained constant in both the preferred and supporting limbs regardless of force changes (Pappas et al., 2015). Hence, with respect to the potential implications of overlooking asymmetry when using unilateral data for bilateral analyses and interpretations, the authors concluded that it was acceptable to collect unilateral data and assume symmetry when mechanical variables presenting the lowest asymmetry index, such as leg stiffness, were the desired outcome measures (Pappas et al., 2015, 2017).

2.2 Low Back Pain in Running

To preface, epidemiologic and etiologic studies on pain and injuries among running or running-based sports have mostly focused on running-related injuries (RRI), which are defined as overuse injuries and/or musculoskeletal pain of the lower limbs or trunk, resulting in an interruption of running participation for a given duration (Hreljac, 2004; Maselli et al., 2020; Mortazavi et al., 2015; Yamato et al., 2015). Since back pain aligns with this definition, LBP is often referred to as an injury in running literature. On the contrary, in the spine biomechanics field, LBP is considered a symptom, not a diagnosis (Mortazavi et al., 2015). Researchers further agree that most LBP is classified as nonspecific or idiopathic, meaning the pain between the 12th thoracic vertebrae (T12) and gluteal folds has no definitive pathoanatomical diagnosis and is not directly attributed to any identifiable disc herniation, radicular pain, or neurological involvement (Balagué et al., 2012; Golob & Wipf, 2014; Hamill et al., 2009). Therefore, isolating risk and mechanical factors specific to LBP can be difficult.

In most cases, running is considered to have protective effects against LBP (Rainville et al., 2004; Trompeter et al., 2017; Woolf & Glaser, 2004). Aerobic exercise can have major benefits for cardiovascular disease and mortality risk (Maselli et al., 2020; van Gent et al., 2007). Research has also shown that jogging is not associated with an increased risk for lumbar disc disease or herniation (Mundt et al., 1993), and a recent study noted that long distance runners had better intervertebral disc hydration and hypertrophy, which are markers of spine health, compared to healthy individuals who had no spinal disease but were otherwise not physically active (Belavý et al., 2017).

However, it is evident that LBP exists among the running community, with annual incidence rates at 10% (Jacobs & Berson, 1986) and the trunk or back listed as one of the more likely body parts to be injured following the lower limbs (Baltich et al., 2017; Walter et al., 1989). While many intrinsic and extrinsic factors have been discussed regarding running-related injuries overall, little is known about predictive factors specific to LBP in running.

Studies on personal factors such as sex, age, height, weight, and genetics have been inconclusive. While females tend to be at lower overall risk of running injuries than males (Buist, Bredeweg, Bessem, et al., 2010; Buist, Bredeweg, Lemmink, et al., 2010; M. P. van der Worp et al., 2015), the opposite effect has been observed, albeit weakly, when examining predictive factors of LBP in the general population (Chenot et al., 2008; Hoy et al., 2010). However, in a study examining risk factors for LBP across 850 marathon runners, no correlation was found between sex and LBP (B. Wu et al., 2021). Some impact metrics that are linked to RRIs, such as vertical loading rates and tibial acceleration, have been observed to be greater in females (Davis et al., 2016; Giandolini et al., 2019; Milner et al., 2006), but other studies have reported no sex differences across peak forces (Baltich et al., 2015; Johnson et al., 2020), or only during single leg hops (Harrison et al., 2011). Sinclair (2016) noted that female runners experienced greater accelerations at the sacrum and lower overall shock attenuation compared to males, which could be owed to sex differences in muscle activation and running kinematics such as the greater hip adduction, internal rotation, and knee abduction in females (Baltich et al., 2015; Ferber et al., 2003), but how these mechanisms contribute to the perceived differences in LBP prevalence is still unknown.

Higher physical height and a body mass index (BMI) greater than 24 kg/m^2 have been suggested as possible risk factors for LBP among runners, but these results have yet to be replicated (Woolf et al., 2002; B. Wu et al., 2021), nor are these factors strongly linked to LBP in the general population (Croft et al., 1999). Greater BMI is associated with increased running injury risk overall, but studies investigating injury risk are often limited to the lower extremity, and a range of thresholds have been suggested (Buist, Bredeweg, Lemmink, et al., 2010; Malisoux, Delattre, Urhausen, et al., 2020; Nielsen et al., 2013). It is well known that age is related to natural degenerative changes in the spine that can be associated with LBP (Adams, 2003; Hoy et al., 2010), but runners between the ages of 31 and 40 have been demonstrated to have the greatest prevalence of LBP (B. Wu et al., 2021).

Extrinsic factors are also variable. One of the most agreed upon findings is that previous injury or LBP is highly associated with subsequent injury (Buist, Bredeweg, Bessem, et al., 2010; Buist, Bredeweg, Lemmink, et al., 2010; Nielsen et al., 2013; Papageorgiou et al., 1996; M. P. van der Worp et al., 2015). Irregular physical activity and warm-up (Woolf et al., 2002; G. Wu et al., 2002), participating at a higher competitive level (Malliaropoulos et al., 2015), and wearing specific footwear (Woolf et al., 2002) have been weakly associated with increased likelihood of experiencing LBP in elite marathon runners.

However, these studies had several limitations; one notable attribute being that they were cross-sectional designs which are not the best method for investigating risk factors (Maselli et al., 2020). Based on the study samples, the generalizability of these factors to recreational runners is also questionable, with a range of thresholds used to define a recreational runner. Some studies used a weekly minimum of 30 minutes of running participation in their inclusion criteria (Baltich et al., 2015; Nigg et al., 2012), others recruited participants who ran at least 16 to 20 km per week in the past 6 months (Davis et al., 2016; Hamill et al., 2009; J. F. Seay et al., 2011), and Videbaek et al. (2015) defined recreational runners as those who have three or more months of running experience. Furthermore, there is limited evidence that training volume is a predictor of RRI (M. P. van der Worp et al., 2015), but no maximum quantity has been established, and no correlation was found between mileage or duration with LBP (B. Wu et al., 2021). Strength of the risk factors also differed across sex, such as greater weekly running distances associated with RRI development in females compared to males (M. P. van der Worp et al., 2015). Lastly, fatigue and running posture have been linked to LBP during running, but it is unknown whether these factors are predictive or a consequence of injury (B. Wu et al., 2021).

Overall, these findings highlight some of the attributes that may be of concern regarding exclusion criteria in a study, but also point to the heterogeneity and complexity of assessing LBP in running. Perhaps attention would be better directed at analyses of the mechanical factors that contribute to LBP or an understanding of factors that alter the mechanical exposure in the low back.

2.3 Impact and Shock Attenuation in Running

Across running literature, the greatest concern for injury has been related to the repeated impact between the foot and the ground. During a 5-km run, the body may be subject to approximately 3000 impacts with the ground (Shorten & Winslow, 1992).

According to the law of the conservation of momentum, when the foot contacts the ground during running, the sudden halt in its movement dictates an exchange in energy and momentum between the two objects. The foot exerts a downward force, and the ground produces a reactionary upwards force equal to the rate of change of momentum (Addison & Lieberman, 2015; Whittle, 1999). The upward ground reaction force (GRF) acts to decelerate the moving limbs and it results in the force transmitted from the foot to the ankle joint, to the tibia, knee joint, femur, and so on until it reaches the head. Each joint and segment is therefore subject to a wave of force, or acceleration, which is referred to as a transient shock wave (Whittle, 1999). Attenuation of this shock wave is critical to prevent the disruption of vestibular and visual systems (Pozzo et al., 1991), as well as to minimize the potential for tissues becoming overloaded and injured. The capacity of various mechanisms to attenuate impact is discussed in the following sections, as well as research on impact forces, measured by the GRF, and impact shock, represented by segmental acceleration, and their relationship with lower extremity running-related injuries and disorders. Evidence will also be provided regarding the lack of examination on how these variables are experienced at the low back.

2.3.1 Impact Forces

Traditionally, the gold standard for kinetic analysis has been via strain-gauge force platforms (Cavanagh & Lafortune, 1980). Force plates can be used to define the magnitude and direction of the GRF, as well as resolve the resultant GRF into vertical, mediolateral, and anteroposterior components (Collins & Whittle, 1989). Applying a minimum threshold such as 10 N of force allows for the detection of gait events such as initial contact and toe-off (O'Connor et al., 2007), and can also be used to measure the frequency components of the impact (Gruber et al., 2014; Whittle, 1999).

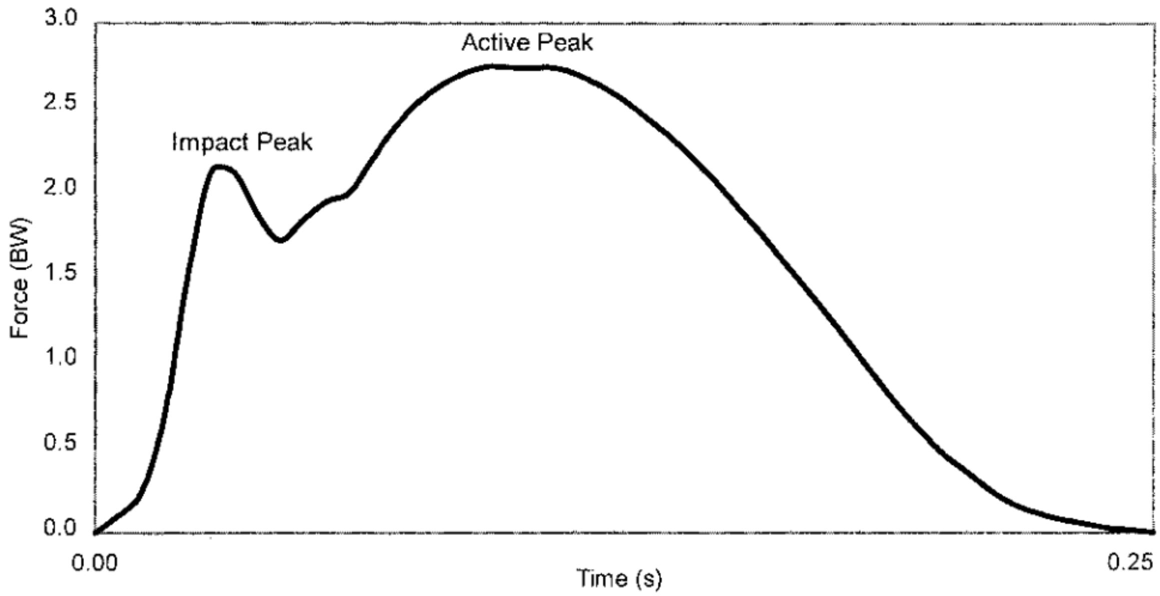

Figure 2.2: Representative vertical ground reaction force versus time curve during stance phase for running. Image adapted from (Hreljac, 2004).

In rear-foot strikers (RFS), the impact creates a characteristic double peak in the vertical ground reaction force (vGRF) profile during stance (Cavanagh & Lafortune, 1980), which are described as the impact and active peaks respectively (Hreljac, 2004; Shorten & Mientjes, 2011) (Figure 2.2), or as a high frequency force component of 10-20 Hz that is superimposed on the lower frequency motion of the centre of mass (COM) at 4-8 Hz (Gruber et al., 2014). The magnitude of the impact peak is therefore determined by the velocity of the moving foot and the mass that is decelerated, over the time during which this exchange occurs – usually 5-25 ms within the first 50 ms of stance (Pratt, 1989; Whittle, 1999). Meanwhile, the active peak reflects the vGRF during midstance (Collins & Whittle, 1989), and while it is greater in magnitude than the impact peak, it is also observed over a longer duration (Gruber et al., 2014). For this reason, it has been suggested that force during toe-off may be relevant for overuse injuries such as stress fractures, while impact peaks may be important for chronic disorders such as LBP (Dickinson et al., 1985).

Average impact and active peak magnitudes of 2.2 body weight (BW) and 2.8 BW respectively have been recorded in (Cavanagh & Lafortune, 1980)ng (Cavanagh & Lafortune, 1980), but running pattern (Farahpour et al., 2016; Lieberman et al., 2010; J. A. Mercer & Horsch, 2015), speed (Hamill et al., 1983; Munro et al., 1987), stride length (Derrick et al., 1998; J. Seay et al., 2008), shoe condition (Baltich et al., 2015; Clarke et al., 1983; de Wit et al., 1995; Hamill et al., 2011; Kulmala et al., 2018; Lieberman et al., 2010; Malisoux, Delattre, Meyer, et al., 2020; Nigg et al., 2012; Sterzing et al., 2013), and surface properties (Dixon et al., 2000) may all influence the resulting magnitude.

Likewise, other force metrics are also influenced by various intrinsic and extrinsic factors. Vertical loading rate, typically defined as the maximum slope before impact peak magnitude, has been noted to be greater in runners with a history of stress fractures and other similar injuries, albeit inconsistently across sex (Bredeweg et al., 2013; Davis et al., 2016), running pattern (Breine et al., 2017), and injury types (Johnson et al., 2020; H. van der Worp et al., 2016). Contact time has also been observed across different subject groups and conditions (Bredeweg et al., 2013; Hamill et al., 1983; McCallion et al., 2014). Loading rates of 77.2 BW/s and 113 BW/s were reported when running at 3 m/s and 5 m/s respectively, with stance times of 270 ms and 199 ms, demonstrating that these metrics are speed-dependent (Munro et al., 1987). Finally, impulse, which can be calculated as the integral of the impact peak over contact time (Addison & Lieberman, 2015), may be used as a gauge of cumulative load (Matijevich et al., 2019) or to assess the relative contributions of impact peak and contact time.

However, one of the major drawbacks with kinetic analyses is that the number of consecutive gait cycles that can be collected is limited by the number of available force plates. Although instrumented treadmills may be a solution, impact forces cannot be used to inform the actual magnitude of impact shock (Shorten & Mientjes, 2011). The GRF reflects the net force acting on the average acceleration of the whole body, rather than the force experienced by individual limbs during impact (Shorten & Winslow, 1992).

Furthermore, recent literature has called to attention the lack of correlation between certain force metrics and internal bone loading (Matijevich et al., 2019) or overall injury risk (Johnson et al., 2020; Malisoux, Delattre, Meyer, et al., 2020). When comparing impact peaks with tibial bone loads computed via a lower limb model during running, Matijevich and colleagues (2019) observed mostly weak or negligible correlations. Tibial bone loads were much greater due to the contribution from muscle contraction that is not detected by force analyses, and the metrics did not coincide temporally as peak bone loads occur closer to midstance (Matijevich et al., 2019).

Likewise, in a prospective study examining peak and average forces across runners who developed lower extremity injuries, the authors found no correlation between magnitudes and injury rate (Malisoux, Delattre, Meyer, et al., 2020). Although higher loading rates are associated with stress fractures (Davis et al., 2016; Zadpoor & Nikooyan, 2011), similar relations have also been reported in other metrics, such as tibial acceleration (Milner et al., 2006; J. H. Zhang et al., 2016), which is also highly correlated with loading grate ($r = 0.87$) (Hennig & Lafortune, 1991). Force metrics can be also used to predict shock characteristics via frequency transfer functions (Lafortune et al., 1995). In other words, acceleration may be a viable surrogate for measuring impact loading and injury risk.

Lastly, studies using impact forces to measure the effects of different interventions on impact attenuation have produced results that conflict with attenuation theory. Specifically, Shorten and Mientjes' (2011) work showed that shoe cushioning attenuates a high frequency component of impact peaks, but force outputs do not reflect the decomposition of high and low frequency components. Therefore, more compliant shoes may produce greater impact magnitudes rather than the expected attenuation – termed the impact peak anomaly (Shorten & Mientjes, 2011). Hence, other metrics, such as the propagation of impact shock, may be better suited for inferring injury risk.

2.3.2 Impact Shock

Measuring the acceleration of specific segments allows for an understanding of the shock at impact and its attenuation throughout the body. Typical accelerometers can be sensitive up to ±200 *g* or higher, if necessary, where the device is calibrated to 1 $g = 9.81$ m/s^2 to align with earth gravitation (Kavanagh & Menz, 2008). Given that voluntary motion during gait is typically within the ±2-5 *g* range, and the acceleration of shock less than ±20 *g*, these sensors are widely used among gait analyses, and a validated method of measuring shock attenuation (Kavanagh & Menz, 2008; Lafortune et al., 1996; J. A. Mercer et al., 2003; Shorten & Winslow, 1992).

Light et al. (1979) performed much of the early ground-breaking work on the transmission of transient shock throughout the skeleton. Using two bone-mounted accelerometers, they observed acceleration magnitudes of up to 80 m/s^2 (8 *g*) in the tibia and approximately 0.5 *g* in the skull, demonstrating that the transient is attenuated as it passes through successive joints and that it contributes to compressive loading (Light et al., 1979). Their use of accelerometers provided an alternative to optoelectronic or force plate analysis methods that were not restricted to laboratory settings, had minimal encumbrance, and offered direct acceleration data rather than the need to differentiate kinematic data, but were largely invasive, as the accelerometers were rigidly attached to the bone via stiff Kirschner wires.

Yet, their findings aligned with studies demonstrating the relation between impulsive forces and cartilage degeneration (M. L. Chu et al., 1986; Radin et al., 1973), and added to the evidence that these transient shock waves may serve as an explanation for osteoarthritic degeneration or LBP (Light et al., 1979; Voloshin & Wosk, 1982). Specifically, these studies observed that repetitive exposure to this impact shock led to stiffening of the subchondral bone and subsequent breakdown of the joints in animals and cadavers, which both act as shock absorbing structures (M. L. Chu et al., 1986; Radin et al., 1973).

Studies comparing bone and skin-mounted options have since demonstrated that recordings from the skin were sufficient for measuring the magnitude and acceleration of the transient (Collins & Whittle, 1989; Light et al., 1979). Thus, lightweight (<4 grams), skin-mounted sensors are often applied. Saha and Lakes (1977) further established that pre-loading the skin under and adjacent to the location which the accelerometer is attached, such as via adhesives or elastic straps, should be performed to minimize soft tissue oscillations and any consequential motion artifact.

Using these methods, peak axial acceleration has been measured at the distal tibia, as a surrogate of GRF impact loading (J. J. Chu & Caldwell, 2004; Delgado et al., 2013; Gruber et al., 2014; Hamill et al., 1995; Lafortune et al., 1996; Mo et al., 2021; TenBroek et al., 2014); the femoral condyle (Voloshin & Wosk, 1982); various levels of the lumbar spine (Castillo & Lieberman, 2018; Delgado et al., 2013; Ogon et al., 2001), such as the L3 spinous process given that it best resembles lower trunk acceleration during walking due to minimal transverse plane rotation at this level (Kavanagh & Menz, 2008); and at the forehead, as a measure of the remaining non-attenuated shock (J. J. Chu & Caldwell, 2004; Gruber et al., 2014; Hamill et al., 1995; Lafortune et al., 1996; Light et al., 1979; J. A. Mercer et al., 2003; TenBroek et al., 2014; Voloshin & Wosk, 1982). Examples of typical acceleration magnitudes from these sensor locations and methods are presented in Table 2.1. In general, greater acceleration has been observed in runners with a history of stress fractures (Milner et al., 2006), with increased speed and stride length (J. A. Mercer et al., 2003; Ruder et al., 2019), and during steeper downhill grades (J. J. Chu & Caldwell, 2004).

Shock attenuation (SA) can then be calculated as a ratio of the incoming acceleration or inferior sensor and outgoing acceleration or superior sensor to provide information about the magnitude of shock that is mitigated or delivered to the structures in between. Early studies used unitless ratios between the amplitudes of shock measured at the tibial tuberosity and forehead (Voloshin et al., 1981; Voloshin & Wosk, 1982), but more recent work has provided full body SA values from -9.44 ± 2.67 dB (TenBroek et al., 2014) to -17.9 ± 16.2 dB (Gruber et al., 2014) during running, using frequency-domain analyses to demonstrate the attenuation of specific signal power. Specifically, the frequency content of the signal power of the impact shock can be examined by calculating the power spectra of the signal, where ranges of 3-8 Hz, 9-20 Hz and > 30 Hz have been identified as representative of the voluntary motion of the lower extremity, impact of the foot during initial contact, and resonant frequency of the sensor or vibration of soft tissues respectively (Edwards et al., 2012; Gruber et al., 2014; Shorten & Winslow, 1992). This approach allows for further analysis and interpretation of the SA mechanisms at work, where eccentric muscle activation and active mechanisms (Section 2.3.3) are more responsive to low frequency shock, and passive structures (discussed in Section 2.3.4) may better attenuate higher frequency impacts (Pratt, 1989).

***Table 2.1**: Examples of studies investigating shock transmission and attenuation across various interventions and their corresponding values.*

Author	Intervention	Testing Protocol	Accelerometer Location(s)	Acceleration	Shock Attenuation
(Light et al., 1979)	Footwear with hard and complaint heels	Overground walking	Tibial tuberosity, head via bite bar	Tibia 5 *g* Head 0.5 *g*	N/D

(Voloshin et al., 1981)	Healthy group vs. Various joint disorders group	Overground walking	Tibial tuberosity, forehead	N/D	Healthy group 3.68 ± 1.33 Joint disorders group 2.83 ± 1.15 (expressed as a ratio of accelerometer amplitude)
(Shorten & Winslow, 1992)	Treadmill speed	Treadmill running	Distal anteromedial tibia, head via bite bar	(5 m/s) Tibia 0.707 ± 0.29 g^2/Hz Head 0.265 ± 0.11 g^2/Hz	(5 m/s) -15 dB *
(Hamill et al., 1995)	Stride frequency	Treadmill running	Distal anteromedial tibia, forehead	(preferred stride frequency) Tibia 0.086 ± 0.04 g^2/Hz Head 0.019 ± 0.01 g^2/Hz	(preferred stride frequency) -25 dB *
(J. A. Mercer et al., 2003)	Stride length, stride frequency	Treadmill running	Distal anteromedial tibia, Forehead	(exemplary subject) Tibia 6 *g* * Head 2 *g* *	(preferred stride length and frequency) -15 dB *
(J. J. Chu & Caldwell, 2004)	Treadmill grade	Treadmill running	Distal anteromedial tibia, forehead	(0% grade) Tibia 7.86 ± 2.25 *g* Head 1.77 ± 0.30 *g*	(0% grade) 76.1 ± 4.8%, -14.8 dB
(Milner et al., 2006)	Healthy group vs. History of tibial stress fracture group	Overground running	Distal anteromedial tibia	Healthy group 5.81 ± 1.55 *g* Stress fracture group 7.70 ± 3.21 *g*	N/A
(Delgado et al., 2013)	Foot strike pattern (barefoot)	Treadmill running	Distal anteromedial tibia, forehead via helmet	(RFS) Tibia 6.1 ± 2.2 *g* Head 1.5 *g* *	(RFS) 73.4 ± 10.9%
(Gruber et al., 2014)	Foot strike pattern (shod)	Treadmill running	Distal anteromedial tibia, head	(RFS) Tibia 5.07 ± 1.49 *g* Head 0.51 ± 0.28 *g*	(RFS) 3-8 Hz, -17.9 ± 16.2 dB 9-20 Hz, -165.1 ± 43.3 dB
(Castillo & Lieberman, 2018)	Lumbar lordosis	Treadmill walking and running	L5/S1, T12/L1	(running) L5/S1 0.065 ± 0.06 g^2/Hz T12/L1 0. 044 ± 0.04 g^2/Hz	(running) -0.77 ± 3.07 dB
(Xiang et al., 2022)	Footwear cushioning level	Treadmill running	Distal anteromedial tibia, Proximal anteromedial tibia	(control shoe) Distal tibia 7.13 ± 1.37 *g* Proximal tibia 5.32 ± 1.10 *g*	(control shoe) 3-8 Hz, -28.12 ± 23.12 dB 9-20 Hz, -23.53 ± 42.64 dB

* peak values inferred from graphical data; N/A = not applicable; N/D = no data

Shock propagation has also been compared in patients with LBP, other degenerative diseases, and healthy controls. Voloshin and Wosk (1982) found that LBP was correlated with a reduced capacity of the musculoskeletal system to attenuate shock waves, while other pathologies were not. This diminished capacity was observed between the acceleration collected at the femoral condyle and the head, suggesting that attenuation is accomplished to prevent overloading of the head since excessive head acceleration can result in disruption to vestibular and visual systems (Lafortune et al., 1996; Voloshin & Wosk, 1982). Hence, the trunk and lower extremities are vital in attenuating the impact shock, and the low back may be implicated when the shock is not adequately attenuated in the knee and distal segments, such as in the case of pathological or meniscectomized joints (Voloshin & Wosk, 1982, 1983).

This leads to the different mechanisms of SA, which have been researched over the last few decades, but primarily in the lower extremities. Mechanisms of SA can be classified as active or passive, where active mechanisms include eccentric muscle activation, proprioception, and joint positioning, and passive mechanisms refer to the viscoelastic elements of musculoskeletal structures such as bone, cartilage, and the heel pad, as well as extrinsic interventions such as footwear (Pratt, 1989).

2.3.3 Active Mechanisms

Firstly, eccentric contraction allows for increased shock absorption; as the knee extensors and ankle plantarflexors lengthen during deceleration, energy from the impact is absorbed and can be stored for later use (Derrick et al., 1998; Pratt, 1989). Spinal musculature also activates in response to sudden loading to stabilize the spine and prevent soft tissues from being loaded (Ghamkhar & Kahlaee, 2015; Mannion et al., 2000; Ogon et al., 2001). Erector spinae activation has been shown to peak right after initial contact during gait (Thorstensson et al., 1982), while the abdominals and obliques activated just prior (Cappozzo, 1983; Cromwell et al., 1989). These activation patterns are related to the control of the trunk movements in the sagittal and lateral planes respectively but given that erector spinae activity contracts to oppose the forward motion of the torso during deceleration, it could be argued that these contractions may also assist with shock attenuation (Cromwell et al., 1989). However, this has been shown to largely increase spinal compression forces due to overreaction of the back muscles (Mannion et al., 2000). There is also evidence that muscle reaction time to gait stimuli is too slow, with a 30 ms latency, to serve as the leading mechanism of SA and adequately respond to the high frequency components of this shock wave (Ogon et al., 2001; Pratt, 1989; Wakeling et al., 2003). This mechanism is further hindered in those with LBP, who may have weaker trunk musculature than healthy individuals (Ghamkhar & Kahlaee, 2015).

Therefore, anticipatory muscle activation may play a larger role in active SA. Muscle tuning, which is the mechanism by which muscle pre-activates prior to impact depending on properties of the impact signal gathered from previous steps (Nigg & Wakeling, 2001), has been observed in the lower extremity (Wakeling et al., 2003; X. Wang et al., 2017) and trunk musculature (Ghamkhar & Kahlaee, 2015; Lamoth et al., 2006). This is particularly evident when variability exists in the impact conditions, but the signal remains the same over multiple continuous strides, such as in the case of fatigue or surface irregularities (Derrick, 2004), and when the input signal frequency is close to the natural frequency of the soft tissue compartment of interest (Boyer & Nigg, 2007; Wakeling et al., 2003). These changes occur to minimize soft tissue vibration and acceleration caused by the transient shock (Boyer & Nigg, 2007; Wakeling et al., 2003).

This activation pattern has also been linked to increased co-activation of the erector spinae and rectus abdominus among those with LBP, owing to the guarding hypothesis, which can contribute to greater loads experienced at the low back (Ghamkhar & Kahlaee, 2015; van der Hulst, Vollenbroek-Hutten, Rietman, & Hermens, 2010). On the other hand, Cai and Kong (2015) demonstrated that runners with LBP had weaker knee extensor and multifidus strength compared to healthy runners, which they suggested may have resulted in greater force transmission to the spine and LBP. Thus, the role of muscle activation with regards to joint loading and SA is unclear.

Rather, the primary role of musculature in SA may be to control knee joint positioning, which has been repeatedly established as a primary locus of SA (Derrick, 2004; Derrick et al., 1998; Edwards et al., 2012; Lafortune et al., 1996). When running, studies have shown that leg stiffness changes as a function of surface properties to maintain a constant bouncing movement of the COM (Ferris et al., 1998). Leg stiffness has been defined as the ratio of peak GRF to peak leg compression, often measured by taking the vertical displacement of the greater trochanter (Addison & Lieberman, 2015). This relation can be modeled by a spring-mass system, with the elastic behaviour of leg compression and recoil during stance represented as a spring, and the body COM as a point mass (Blickhan, 1989; McMahon & Cheng, 1990). Therefore, leg stiffness is dictated by knee flexion angles, and running with increased knee flexion increases leg compliance, which decreases the amplitude of the shock wave transmitted to the head (Derrick et al., 1998; Lafortune et al., 1996).

Certain movement strategies such as "Groucho running" which involves exaggerated knee flexion (McMahon et al., 1987) or decreasing heel strike velocity while increasing ankle dorsiflexion (J. J. Chu & Caldwell, 2004) may also be effective SA mechanisms due to changes to joint and limb stiffness. This last theory has contributed to the rise in FFS running patterns, where SA was observed to be greater in RFS, but the magnitude of shock is less among FFS runners (Delgado et al., 2013; J. A. Mercer et al., 2003).

Based on these relationships, studies have examined the influence of different footwear and surface properties on joint stiffness in search of potential interventions for impact loading. For instance, increased shoe cushioning resulted in greater leg stiffness (Bishop et al., 2006; Kulmala et al., 2018), which may null the effect on impact shock attenuation provided by greater cushioning. Shoe hardness also altered lower extremity muscle activity, producing changes in joint loading (Wakeling et al., 2002). Temporal data has shown that proprioception at the foot-ground interface when barefoot can trigger a musculature response within the timeframe of a monosynaptic stretch reflex, suggesting that afferent proprioception may assist with SA by reducing muscle activation latency (Ogon et al., 2001). However, use of insoles or footwear cushioning in tandem may confound these effects. Therefore, more research is needed on the specific role of spinal muscles in SA, and how footwear may interact with these mechanisms.

2.3.4 Passive Mechanisms

The literature on passive mechanisms is more unanimous. For viscoelastic structures such as cartilage, menisci, and bone, its viscoelastic nature implies that the response of theses tissues is load-rate dependent, therefore its mechanism of impact attenuation and dissipation of energy may vary based on the frequency which the structures are loaded (Pratt, 1989). For instance, Gruber et al. (2014) suggested that passive SA assists with the attenuation of high frequency shock, while active mechanisms are better able to respond to signals less than 10 Hz. These tissues also reduce loading by increasing the duration over which the perturbation is experienced by providing additional thickness for deformity, such as in the case of the calcaneal heel pad (Pratt, 1989; Whittle, 1999). However, heavy reliance on specific structures, such as the menisci, could lead to overload and damage, thereby reducing the overall SA capacity of the musculoskeletal system (Gruber et al., 2014; Voloshin & Wosk, 1982).

Other intrinsic mechanisms may include lumbar lordosis, the sagittal shape of the intervertebral discs (IVD), and foot arch height. Early studies viewed IVD as the "primary shock absorbers of the lumbar spine" (Voloshin & Wosk, 1982), and lumbar lordosis as an adaptive function for dissipating shock into bending and rotational loads rather than axial compression (Adams, 2003). However, Castillo and Lieberman (2018) reported that IVD thickness and its influence on lumbar curvature may be more important for SA than its viscoelastic composition, as height accounted for nearly 20% of variation in lumbar SA across their sample. In conjunction, less lordotic spines may be more stable and less prone to injury but have less SA capacity compared to curved spines (Castillo & Lieberman, 2018). Another study also studied SA in the low back, but across a range of arch heights. The authors found that higher arches were associated with improved SA due to medial instability and internal rotation of the leg, which lengthened the duration of the impact and thus reduced loading rates (Ogon et al., 1999). These results therefore point to the load rate dependency of passive structures and highlight the multifactorial nature of SA in the low back.

Finally, with regards to footwear, numerous studies have examined the effects of footwear cushioning on SA in the distal tibia (Light et al., 1979; Sinclair, 2017; TenBroek et al., 2014). The consensus is that softer and more compliant shoes may improve SA during impact via increasing the duration over which the impact is experienced, but effects further up the kinetic chain are not well understood. Of the few studies that varied shoe condition and examined SA with respect to the back, Wosk and Voloshin (1985) noted that viscoelastic insoles were effective in reducing the transient amplitude generated during walking by over 40%, and can therefore be considered a conservative treatment for LBP symptoms, and Ogon et al. (2001) found that wearing soft shoes increased the latency of the spinal muscle response compared to jogging barefoot, leading to reduced SA. Additional details regarding the specific effects of footwear properties will be discussed in Section 2.4. However, the question remains regarding how these mechanisms interact, particularly in conjunction with trunk activation and during running.

2.4 Footwear Characteristics

Research has demonstrated the efficacy of various interventions on improving LBP among runners such as lumbar extensor strengthening programs and lower limb training (Cai et al., 2017; Gordon & Bloxham, 2016), suggesting the relationship between those impairments and LBP. Specifically, extensive research has been conducted on the interdependence of lower limb biomechanics and LBP, particularly among runners. However, these techniques demand time and program commitment, and may not be accessible for recreational runners. A simpler and more inexpensive approach would be the modification of footwear or use of insoles, which has been shown to assist with impact shock attenuation in those with LBP (Light et al., 1979; Wosk & Voloshin, 1985). Hence, over the last few decades, the characteristics of running shoes and their effects on loading during running have been areas of focus for researchers and show manufacturers.

Of particular concern is the midsole layer, which lies between the outsole that interacts with the ground and the interior lining of the shoe. Variations in midsole cushioning geometry, material, and construction have been investigated with the intention to decrease injury risk by attenuating the impact experienced during running. Two elements may be particularly related to SA and injury prevention and will be reviewed: midsole cushioning thickness and stiffness.

However, it is important to note the difficulty of evaluating each characteristic in isolation. For example, adjusting the volume of cushioning can change its midsole thickness, midsole stiffness, and shoe mass, which can have differing effects on proprioception, leg stiffness, impact loading, and muscle activation, as well as simultaneous benefits or consequences to performance. Therefore, there is a need to appropriately quantify cushioning, and be deliberate about the language used to describe its properties.

2.4.1 Midsole Thickness

The thickness of the midsole is one of the first attributes to be identified when evaluating footwear cushioning. Midsole thickness is determined by measuring the height between the bottom of the sock liner (i.e., insole), to the top of the outsole at the centre of the forefoot along the midline of the shoe using digital callipers (Law et al., 2019; Ramsey et al., 2019). Following the trends in the running community, thin and flexible midsoles with little or no heel counter have been characterized as minimal footwear or minimalist, designed to resemble barefoot running (Esculier et al., 2015; Hamill et al., 2011; Ramsey et al., 2019), and oversized midsoles with ample cushioning (>20 mm), wider toe-boxes, and minimal stability features are defined as maximalist shoes (Agresta et al., 2018).

However, midsole thickness can vary within a given shoe, owing to the offset or heel-to-toe drop (HTD) height, which is defined as the difference in height between the rear and the forefoot (Mo et al., 2021). Average HTD heights range from 0 to 16 mm, with the rearfoot height prioritized to cushion the heel (Malisoux, Chambon, Urhausen, et al., 2016). Despite both terms being used interchangeably when categorizing footwear, each characteristic can produce different effects and interactions if not accounted for (Chambon et al., 2015).

Anecdotally, a thicker midsole allows for greater deformation, or a greater distance during which the foot decelerates, and thus increased ground contact time during a step (Chambon et al., 2014; Nigg, 1986; Whittle, 1999). Increased ground contact time for the same GRF would produce a lower acceleration magnitude, and thus lower loading rate, which has been previously linked to injury risk (Davis et al., 2016; Johnson et al., 2020). Thicker midsoles also imply increased cushioning, which should better attenuate the impact and reduce injury risk (Malisoux, Delattre, Urhausen, et al., 2020; Shorten & Mientjes, 2011).

On the other hand, minimal running shoes were popularized due to their increased flexibility and lightweight benefits (Esculier et al., 2015). By mimicking barefoot running, minimal shoes encourage the adaptation of FFS patterns (Horvais & Samozino, 2013; J. H. Zhang et al., 2017), where FFS has been associated with reduced vGRF and impact transient magnitudes (Gruber et al., 2014; Lieberman et al., 2010).

However, studies comparing midsole thicknesses or shoes categorized as minimal versus maximal have produced conflicting results. Law et al. (2019) compared 6 identical shoes with midsoles ranging from 1 mm to 29 mm and found that thinner midsoles were associated with increased vGRF loading rates and shortened ground contact time, but only when the midsole was 5 mm or thinner. Protective effects against loading rates capped at a midsole height of 25 mm, with no significant differences in force metrics between the 25 and 29 mm midsoles (Law et al., 2019). Several authors also found similar results, reporting that thinner midsoles produced greater tibial accelerations (TenBroek et al., 2014) and may increase lower extremity risk of injury compared to maximal shoes (Hannigan & Pollard, 2020), particularly during initial exposure (Agresta et al., 2018), or with respect to Achilles tendon forces (Sinclair et al., 2015). At the same time, no differences in foot strike angle, cadence, or stride length were observed between shod conditions (Law et al., 2019; Mo et al., 2021; TenBroek et al., 2014), only between running barefoot versus shod (Hamill et al., 2011), while Chambon et al. (2014) showed no difference in tibial accelerations across all conditions. Lastly, Kulmala et al. (2018) found that maximalist shoes amplified rather than attenuated impact loading compared to thinner shoes, possibly due to changes in the runners' leg stiffnesses associated with the hardness of the cushioning. Notably, the shoes models used in this study were not identical, including differences in HTD heights, which calls into question the independent effect of midsole thickness on injury risk (Kulmala et al., 2018), but points to the inconsistencies across the current literature.

On the other hand, greater HTD heights provide the attenuation benefits of a thicker heel pad while simultaneously reducing the cushioning under the forefoot where ground reaction forces occur across the entire stance phase, and thus are less damaging as contact time is extended (Chambon et al., 2015). But low drop shoes, like thinner midsoles, may alter running kinematics to closer reflect that of MFS or FFS patterns, as well as reduce leg stiffness (Horvais & Samozino, 2013). In a study examining the effect of HTD height on loading rates, Chambon et al. (2015) found that greater HTD heights reduced loading rate when running on treadmill surfaces, which is consistent with studies using thicker midsoles. However, opposite effects were observed on overground running, which is noteworthy as manufacturer testing typically differs from distance running outdoors (Chambon et al., 2015). Low drop shoes were found to be associated with reduced lower limb injury risk in occasional runners, but higher risk in regular runners, suggesting that familiarization should be considered before modifications to this element (Malisoux, Chambon, Urhausen, et al., 2016). Likewise, despite changing to low drop shoes lowering the shock magnitude by 30% in a long-term intervention study, Giandolini, Horvais, and colleagues (2013) suggested that at least a month was required for runners to adjust to shoes that altered their running pattern, or risk additional pain or injury. In either case, the effects are noted only in the lower extremity, while little research has yet to be recorded at the level of the low back.

Low back pain sufferers have been historically prescribed more cushioned shoes or insoles to increase the cushioning volume, which have been shown to improve pain symptoms (Wosk & Voloshin, 1985). Ogon et al. (2001) examined the temporal response of lumbar muscles when running shod compared to barefoot and found that the presence of running shoes or insoles improved the latency between the acceleration peak and muscle response peak in the lower back, meaning the lower back experienced impact forces later. However, the authors did not examine whether such timing differences could be also attributed to changes in posture when running shod versus barefoot (Ogon et al., 2001). In a more recent study, Lee and colleagues (2018) investigated changes in lumbar kinematics and paraspinal muscle activation among habitually shod recreational runners before and after a 4-week training program where minimal footwear was progressively introduced into the runners' weekly mileage of 10 to 50 km. They found that runners adopted a more extended lumbar posture, accompanied by reduced muscle activation, when using minimal shoes compared to their regular running shoes (S. P. Lee et al., 2018). The authors suggested that these changes could be due to the reduced need to stabilize the lumbar spine when in a more upright posture, which could be an argument for inducing movement pattern changes in runners with LBP via minimal footwear (S. P. Lee et al., 2018). Unfortunately, no other study has examined the higher range of midsole thicknesses on low back loading, and it remains unknown whether a relationship exists between midsole thickness and spinal mechanics.

2.4.2 Midsole Stiffness & Hardness

A footwear characteristic that is of greater concern to researchers is the midsole cushioning stiffness. Shoe cushioning may address the hardness of the material, but impact attenuation often refers to its stiffness or compliance in response to an applied load, as deformation of the cushioning system affects the degree to which the foot and subsequent joints are loaded. In most studies of shoe cushioning, midsole hardness and stiffness are systematically varied by customizing the midsole material while all other characteristics are identical as provided by the manufacturer. In other cases, external footpads are attached to the bottom of minimal footwear to create a simulated midsole.

Hardness is often evaluated with durometer gauges, which measure indentation hardness, or the yield strength of the material's surface (Shorten & Mientjes, 2011). Shore or Asker® scales are used to quantify the resistance to indentation with higher values indicating harder materials and depending on the shape of the indenter used. Studies varying midsole hardness have used shoes in the Shore 24 to Shore 70 (Hardin et al., 2004; Kersting & Brüggemann, 2006; Nigg et al., 1987; Wakeling et al., 2002) or Asker C-40 to C-65 ranges (Baltich et al., 2015; de Wit et al., 1995; Nigg et al., 2012; Sterzing et al., 2013; Theisen et al., 2014) by adjusting the chemistry and density of the foam material. Malisoux's group (2016) used hardness measurements from the sagittal plane by cutting a subset of shoes at the metatarsal head and applying the durometer at the forefoot, but other methods may be required for experiments without a subset of shoes, or if the foam of interest exhibits dual-density characteristics at the rearfoot or outer edge (Ramsey et al., 2019). Regardless, the durometer has been considered an objective and accurate measurement of midsole hardness (Cornwall & McPoil, 2017).

However, Shorten and Mientjes (2011) have argued that indentation methods are not reflective of the impact attenuation capacity of midsole cushioning since the point of application is concentrated under the durometer rather than distributed across the footbed. Stiffness provides a more valid characterization, as under constant conditions and according to Hooke's law, a more compliant spring or softer cushioning system should increase time to force peak and decrease peak force, therefore attenuating impact.

Thus, studies have proposed using the standard impact test (ASTM F1976-06) that is employed in manufacturer testing and designed to measure the impact attenuation characteristics of the shoe's cushioning system (Ramsey et al., 2019). The test uses an 8.5 kg mass dropped from a height of 30-70 mm onto the heel or fully intact forefoot (ASTM F1976-06, 2013), where the mass mimics the effective mass of the runner's leg while the drop height produces an impact velocity of 1 m/s, resembling touchdown velocity of the foot during running (Hennig, 2011). The test specifies the total energy to be applied according to impact magnitude classification; running shoes are classified as subjected to moderate impacts since the peak GRF experienced is greater than 1.5 BW but less than 3 BW, and peak axial deceleration of the lower leg falls within 4–8 *g*, resulting in a total energy input of 5 Joules (Shorten & Mientjes, 2011).

Shorten and Mienjtes (2011) performed the standard impact test on 224 commercial running shoe heels and reported the relationships between peak impact force and time to peak or peak displacement. Despite shoe design and construction varying largely across their sample, a clear trend was observed where more compliant shoes demonstrated lower peak force, longer time to peak, and greater displacement (Shorten & Mientjes, 2011). Other studies adopting this method have recorded global heel stiffness values ranging from 51.1 ± 4.0 N/mm to 94.9 ± 5.9 N/mm, demonstrating this method's sensitivity to values within the range of stiffnesses currently utilized in commercially available shoes (~53-97 N/mm) (Malisoux, Delattre, Urhausen, et al., 2020; Malisoux et al., 2017; Theisen et al., 2014). Furthermore, the produced measurements also aligned with the authors' hardness measurements for the same shoes (i.e., harder midsoles demonstrated greater stiffness), confirming the validity of this method for quantifying midsole cushioning.

However, one argument against energy-constrained methods is that the load applied may not be representative of human loading (Mai et al., 2021; Schwanitz et al., 2010). Rather, studies examining the specific benefits provided by running shoes use forces and loading rates that are representative of the vGRF loading cycle experienced while running at given speeds. Following a similar approach as the standard impact test, force-deformation curves are then obtained to calculate the stiffness (linear slope), energy absorbed (area under loading curve), energy returned (area under unloading curve), and hysteresis or energy loss (area between loading and unloading curves) (Hoogkamer et al., 2018; Worobets et al., 2014). Worobets et al. (2014) recorded stiffness and hysteresis values of 131.2-167.0 N/mm and 20.9-31.3% respectively after 20 consecutive loading-unloading cycles of approximately 1700 N at 4250 N/s, where a minimum of 20 consecutive cycles are typically performed to ensure the shoes have reached steady state hysteresis (Schwanitz et al., 2010; Worobets et al., 2014).

Due to the debate regarding the different mechanical testing methods, Schwanitz et al. (2010) compared results from two ASTM procedures and a force-constrained test and noted that the peak forces in ASTM-based tests were well below the typical peak vGRF of 2-3 BW experienced during initial contact. Although hysteresis values in the different tests were well correlated (r=0.95-0.99), there were significant differences between the relative ranking of their shoe sample's attenuation capacities across the three methods (Schwanitz et al., 2010). Hence, while the energy-constrained method may be used commercially and provides acceleration values that pertain to the shock experienced and to be attenuated by the musculoskeletal system, the force-constrained method has the added benefit of producing responses representative of human loading, which may better inform how footwear works in conjunction with the shock attenuation mechanisms of the body (Hoogkamer et al., 2018; Worobets et al., 2014).

In either case, materials and mechanical testing have been consistent with the impact attenuation theory that more compliant midsoles better attenuate impact (Shorten & Mientjes, 2011). *In vivo* studies using force plates are less in agreement. Early studies either found higher impact peak values and loading rates with hard compared to soft shoes (de Wit et al., 1995; Nigg & Bahlsen, 2016), only changes in time to peak force (Clarke et al., 1983), or no changes at all to force variables when altering midsole hardness (Kersting & Brüggemann, 2006; Nigg et al., 1987, 1988).

Contrarily, more recent studies noted that vertical impact peaks increased and loading rates decreased as shoe midsole hardness decreased (Baltich et al., 2015; Malisoux, Delattre, Meyer, et al., 2020; Sterzing et al., 2013), although the authors speculated that such results may be influenced by landing hardness or other variables. Another study comparing midsole densities in basketball shoes also reported limited evidence for soft shoes providing better impact attenuation during landing activities via the measurement of peak GRF (S. Zhang et al., 2005). These latter observations are consistent with the "impact peak anomaly" termed by Shorten and Mientjes (2011), a consequence and limitation of using force plates. Due to the summation of non-impact, low frequency force components with the attenuated high frequency force components from impact, softer shoes appear to produce a greater impact peak in vivo compared to harder shoes (Shorten & Mientjes, 2011).

This conclusion has led to the use of impact force variables as a metric for shoe cushioning to be questioned and is further supported by a weak correlation between GRF and internal structure loading (Malisoux, Delattre, Meyer, et al., 2020; Matijevich et al., 2019). A more valid indicator of overall injury risk may be to examine participant reports of training activity and adverse events, or to compare specific limb motion while using shoes of different midsole hardness and stiffness. For example, de Wit et al. (1995) showed that running in softer midsoles produced larger eversion and pronation movements, which have been associated with the development of overuse injuries (Cavanagh & Lafortune, 1980). However, in a 5-month prospective study, Theisen et al. (2014) found that midsole hardness did not influence running-related injury (RRI) risk, suggesting that individuals adapt their running styles in response to midsole hardness. Malisoux, Delattre, Urhausen, and colleagues (2020) showed that injury risk was higher in runners wearing harder shoes with less shock absorption properties, but the overall protective effect of cushioning was only observed in runners lighter than the median weight in their sample.

With respect to shock attenuation, only a few studies have examined the effects of midsole stiffness on tibial acceleration but are limited to the context of basketball shoes and landing activities. Zhang et al. (2005) compared acceleration of the forehead and tibia in basketball shoes of different midsole densities while landing from different heights, where harder midsoles resulted in greater acceleration magnitudes at both locations. This mostly aligned with the findings by Bruce et al. (2019), who observed greater peak tibial acceleration in stiffer shoes during countermovement jump landings on various basketball court constructions, but no differences at the head.

These results point to the notion that the relationship between midsole hardness and stiffness with injury and shock attenuation must be mediated by other factors, such as leg stiffness. Baltich et al. (2015) noted that softer midsoles were associated with increased ankle and knee joint stiffness, with the ankle more sensitive than proximal joints. According to Hardin et al. (2004), these limb posture compensations for midsole hardness occur to minimize metabolic cost at the expense of increased exposure to impact shock. As joint stiffness has also been associated with various overuse injuries and degenerative diseases (Radin et al., 1973; Voloshin & Wosk, 1983), it appears that midsole hardness and stiffness may be linked to injury via leg stiffness adaptation mechanisms.

Ultimately, midsole cushioning is important in attenuating the forces and acceleration experienced during impact landing, but the relationship between these metrics and injury is still being questioned, as is the isolated effect of midsole cushioning on running-related injury risk, let alone LBP. In any case, a major gap in literature surrounds whether softer and compliant cushioning could influence the low back, as most studies do not examine mechanisms proximal to the knee.

2.5 Key Messages & Gaps in the Literature

Research on gait and running have been conducted for decades, but the current literature on running injuries and midsole cushioning pertains primarily to the function and response of the lower extremity. Despite the low back and pelvis injuries making up nearly a quarter of running-related injuries (Woolf et al., 2002), few studies have examined the role of predictive and mechanical factors related to LBP in running, as well as how interventions such as footwear cushioning influences low back muscular, postural, and loading responses.

Potential differences may be observed across runners of different sex, body mass, skill level, and injury history, as well as throughout the gait cycle and as a function of different running patterns. Thus, these factors should be controlled to accurately assess the role of footwear on potential injury and ensure the generalizability of such findings to the appropriate recreational running population. Caution should also be exercised when collecting unilateral data and assuming symmetry.

During initial contact in running, the body is subjected to ground reaction forces and accelerations that are transmitted from the foot to the head. Ground reaction forces have been particularly of interest due to the proposed relation between impact loading metrics and injury (Cavanagh & Lafortune, 1980), which are influenced by the properties of the foot-ground interface (Clarke et al., 1983). However, the use of force metrics to inform internal tissue loading have been questioned following no or inconsistent differences between peak forces and injury (Malisoux, Delattre, Meyer, et al., 2020; Matijevich et al., 2019; Shorten & Mientjes, 2011), and due to the forces measured being constrained to the lower limb. Rather, segment acceleration can be measured using accelerometers to observe how the impact is experienced throughout the body, where reduced capacities to attenuate the transient shock wave, primarily between the level of the low back and knees (Derrick, 2004), have been observed in those with LBP (Voloshin & Wosk, 1982). Both active and passive mechanisms may be involved; for instance, muscle activation assists with shock attenuation via eccentric contractions, pre-activation, or controlling leg stiffness, where differences in these variables have all been studied among LBP patients and healthy controls (Ghamkhar & Kahlaee, 2015; Hamill et al., 2009; van Dieën et al., 2003; Wakeling et al., 2003). However, no study has yet to examine the combined role of these mechanisms during running with respect to shock attenuation specifically at the low back.

Furthermore, footwear has been proposed as an intervention for addressing impact-related injuries and low back pain in running (Nigg et al., 1988; Wosk & Voloshin, 1985). Cushioning elements in the shoes may affect both ground reaction force and shock attenuation (Baltich et al., 2015; Clarke et al., 1983; Kulmala et al., 2018; Ogon et al., 2001). Current running shoes range in midsole thickness, hardness, and stiffness, which have each been related to altering lower limb injury risk (Malisoux, Delattre, Urhausen, et al., 2020; Theisen et al., 2014), but again, not yet studied with respect to LBP.

Therefore, a current gap in the literature exists at the intersection of these topics. Specifically, the role of midsole cushioning on shock attenuation in the low back may directly inform potential injury and low back pain as a function of impact loading during running.

3 Methods

3.1 Overview of Study Design

A mixed measures study design was performed. Participants (male and female) were each asked to perform running trials in the three shoe conditions on a treadmill while accelerometry, motion capture, and muscle activity were collected. Main outcome measures included sagittal joint angles, mean muscle activation levels, peak acceleration magnitudes, and shock attenuation in the low back during stance phase. Shock attenuation was calculated from accelerometers attached to the participants' lower limbs trunk, and head. Footwear cushioning properties were quantified via mechanical testing. Statistical analyses involved two-way mixed measures analysis of variance to determine whether differences existed between sex and shoe midsole stiffness conditions. Prior to participation, all participants provided informed consent. The study was approved by the Office of Research Ethics (#43973) at the University of Waterloo.

3.2 Participants

Ten male and ten female runners between the ages of 18-35 and who ran a minimum of 16 km per week were recruited from the University of Waterloo population and surrounding community (Table 3.1). Only young adults were recruited to minimize potential age-related degenerative changes in spine posture (Castillo & Lieberman, 2018) and to limit cofounding predictors of injury (Nielsen et al., 2013). Shoe size was restricted to men's US 9-10 and women's US 7-8, corresponding to three sizes per sex including half sizes. The minimum running threshold was defined based on previous studies involving recreational runners (Baltich et al., 2015; Cai & Kong, 2015; Nigg et al., 2012).

Table 3.1: Mean (SD) descriptive characteristics of study participants by sex.

Characteristic	Males	Females
Age (years)	22.90 (2.33)	22.14 (2.12)
Height (m)	1.79 (0.04)	1.70 (0.09)
Weight (kg)	69.18 (6.21)	63.73 (8.91)
BMI (kg/m^2)	21.65 (1.66)	21.92 (2.09)
Leg Length (m)	0.87 (0.04)	0.86 (0.07)

Additional exclusion criteria included 1) previous history of low back pain that required seeing a clinician or time off from work; 2) sustaining an injury within the last three months; 3) having undergone surgical interventions related to the spine or lower limbs; 4) having a BMI greater than 30 kg/m^2; 5) not adopting a RFS running pattern; and 6) not having right leg preference. Foot strike patterns were screened via self-reports from the participants during recruitment, then confirmed via experimenter observation and motion capture. Leg preference was determined by a series of tasks adapted from van Melick et al. (2017) and Chapman et al. (1987): 1) a soccer ball was placed in the centre of both feet and the participant was asked to kick the ball forward; 2) the participant was asked to write their name "in sand" on the floor; and 3) the participant was asked to "erase" their name. The leg used for two of the three tasks was considered their preferred limb. The first task has been shown to have 100% agreement between self-reported and observed preferred leg in both men and women (van Melick et al., 2017). Lastly, all participants reported that they did not regularly use orthotic insoles in their own shoes.

Demographics (age, sex) and anthropometric and measurements (height, weight, leg length, shoe size) were collected and are presented in Table 3.1.

3.3 Footwear

Three shoe conditions were purchased new through research funds and not provided by any manufacturer, and included in this study (Figure 3.1): Nike Pegasus 38 (PGS), Nike React Infinity 2 (RCT), and Nike ZoomX Invincible (ZMX; Nike Inc., Beaverton, OR, USA). The shoes were selected to represent the range of stiffnesses expected to be seen in a recreational training shoe but were similar in their composition, according to manufacturer descriptions (Table 3.2). Men's size 9, 9.5, and 10 and women's size 7, 7.5, and 8 were used.

***Figure 3.1**: Women's models of the three shoes used in this study. Right: PGS; Middle: RCT; Left: ZMX*

***Table 3.2**: Men's 9.5 and Women's 7.5 shoe geometry characteristics.*

Characteristic	PGS		RCT		ZMX	
	Female	Male	Female	Male	Female	Male
Heel height (mm)	28	33	32	33	35	37
Forefoot height (mm)	18	23	23	24	27	28
Heel-to-toe Drop height (mm)	10	10	9	9	8	9
Heel Width (mm)	72	81	85	92	91	102
Toe Width (mm)	96	110	106	114	106	118
Mass (g)	252	283	244	302	253	274

Prior to the experimental protocol, stiffness quantification was performed using a servohydraulic materials testing system (Model 8872, Instron Canada, Toronto, ON, Canada) according to the force-controlled protocol described by Hoogkamer et al. (2018) and Worobets et al. (2014). Each shoe condition in men's size 9.5 and women's size 7.5 was mounted to the materials testing machine so that the sole was flat and parallel to the ground (Figure 3.2). The midsole was vertically compressed using a peak magnitude of 1700 N at a loading rate of 4250 N/s and then unloaded to 50 N. Twenty consecutive loading and unloading cycles were performed, with the final five cycles used to determine force-deformation curves. Data was recorded at a frequency of 1000 Hz. Linear equivalent stiffness (N/mm) was calculated from the slope of the line of best fit for the loading curve using a 1st degree polynomial fit. Energy absorbed by the midsole was calculated from the area under the loading curve, energy returned was calculated as the area under the unloading curve, and hysteresis was calculated as the area between both curves, or the ratio of energy lost as heat.

Figure 3.2: *Exemplar Men's ZMX shoe undergoing stiffness quantification via the materials testing system.*

Studies have shown that increased mileage is related to reduced shock attenuation capacity of the running shoe (Cook et al., 1985; Hamill & Bates, 1988; Verdejo & Mills, 2004; L. Wang et al., 2010), however more than 85% of the initial shock absorption capacity is still retained after 50 miles of running (Cook et al., 1985). Furthermore, machine-simulated running has been demonstrated to result in 25% greater reduction in SA capacity compared to actual running (Cook et al., 1985; L. Wang et al., 2010). In a study examining wear effects after 500 km, only a 5% change in the resulting GRF was observed (L. Wang et al., 2010). Given that the shoes used in this study were subjected to less than 10 km each (i.e., max 10 runners per shoe × approximately 990 m per trial at 3.3 m/s for 5 min = 9900 m or 9.9 km), no wear or deterioration effects were expected following mechanical or human testing.

3.4 Instrumentation

3.4.1 Surface Electromyography

Six channels of surface electromyography (EMG) were collected: right and left lumbar erector spinae, rectus abdominus, and external oblique (Table 3.3). Prior to electrode placement, the skin overlying the muscles of interest was shaved using disposable razors and cleaned with a light abrasive cloth (Kimwipes, Kimberley-Clark Inc., Irving, TX, USA) and alcohol solution.

A bipolar electrode configuration was adopted using two disposable silver/silver chloride electrodes (Dual Electrodes, Noraxon USA, Inc., Scottsdale, AZ, USA) placed over the middle of each muscle belly. The electrodes were oriented parallel to the muscle fibre direction with a 2 cm interelectrode distance, according to SENIAM guidelines (Hermens et al., 2000). A reference ground electrode was placed on the left rib. Specific electrode placements for each muscle are described in Table 3.3 and shown in Figure 3.3. All anatomical placements were manually palpated by the same researcher.

EMG signals were differentially amplified with a common-mode rejection ratio (CMRR) of 115 dB at 60 Hz and bandpass filtered (bandwidth 10-1000 Hz; input impedance 10 GΩ; AMT-16, Bortec, Calgary, Canada), then gained by a factor of 500-5000 depending on the individual's muscle activity during maximal voluntary isometric contractions (MVCs). The gained signal was sampled at 2500 Hz using a 16-bit A/D conversion card with a range of ±10 volts.

***Table 3.3**: Muscles recorded by surface electromyography and corresponding electrode placement.*

Muscle	**Electrode Placement** (Callaghan et al., 1999)
Lumbar Erector Spinae (LES)	Oriented vertically 3 cm bilateral to L3 spinous process
Rectus Abdominus (RA)	Oriented vertically 3 cm bilateral to the umbilicus
External Oblique (EO)	Oriented infero-medially 15 cm bilateral to the umbilicus

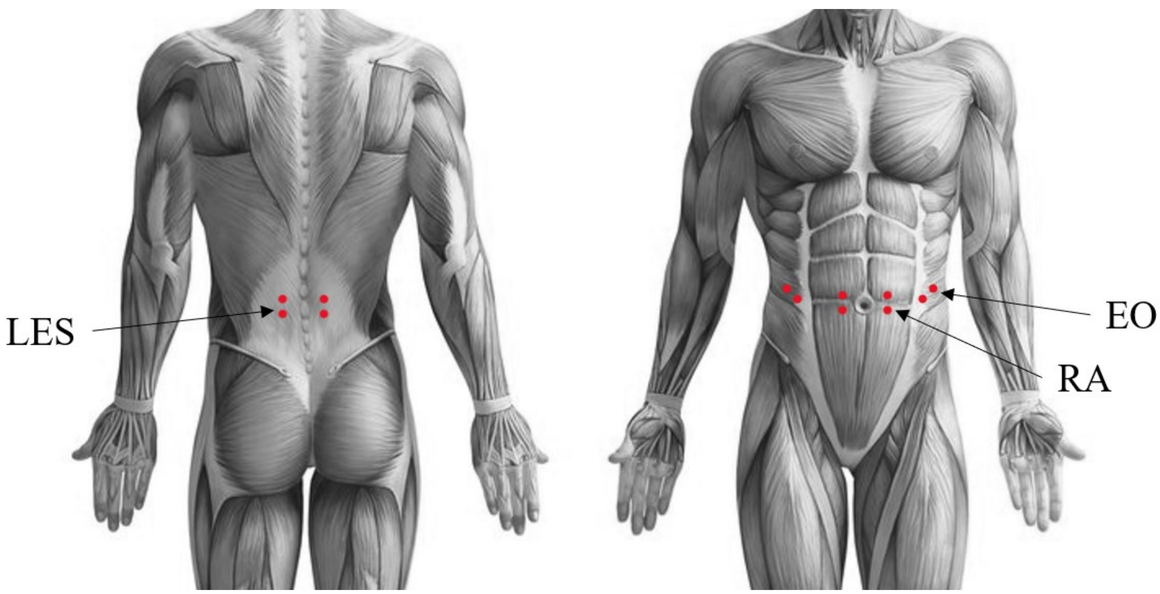

***Figure 3.3**: Electrode placements for surface electromyography. See Table 3.3 for abbreviations.*

Each signal was normalized to its corresponding MVC, where the participant was asked to gradually reach and maintain maximal effort during an isometric contraction of five seconds. A minimum interval of 2 minutes separated consecutive MVCs to prevent muscular fatigue.

The following tasks were used to elicit MVCs in the corresponding muscles and have been previously used and reported by Dankaerts et al. (2004):

- **Lumbar Erector Spinae**: Modified Biering-Sorensen Test – The participant laid prone on a table with their trunk suspended over the edge so that the table edge was aligned with the participant's anterior superior iliac spines (ASIS). Their legs were secured to the table and resisted by an experimenter. The participant was asked to cross their arms and flex their trunk, then extend upwards until parallel, during which downwards resistance was applied by a second experimenter on the subject's back. The participant was encouraged to maintain maximal effort during trunk extension.
- **Rectus Abdominus**: Resisted Curl-up – The participant sat supine on a table with their trunk flexed 45 degrees, knees flexed to 90 degrees, and hands across their chest. They were asked to flex their trunk at maximal effort while an experimenter applied manual resistance to their shoulders.
- **External Oblique**: Crossed Resisted Curl-up – The same position used for the rectus abdominus MVC was adopted. The participant was asked to rotate their trunk, with their right shoulder moving towards the left, at maximal effort while an experimenter applied resistance in the opposing direction. The same procedure was performed on the left side.

Finally, five-second rest trials were collected with the participant lying still while prone and supine to allow for the potential removal of any resting bias in the EMG signal.

3.4.2 Accelerometry

Four low-mass tri-axial piezoelectric accelerometers (ADL377, Analog Devices, Norwood, MA, USA) with a range of ±200 *g* were firmly affixed to the skin over the right anteromedial distal tibia, T12-L1 and L5-S1 vertebral levels, and to the head via an elastic headband (Table 3.4). The tibia location was selected to minimize the influence that ankle rotation may have on acceleration magnitude (Lafortune & Hennig, 1991), while the spinal locations were selected to isolate the shock attenuation within the lumbar spine (Castillo & Lieberman, 2018). The head acceleration was included as a measure of total body shock attenuation. Anatomical locations were determined by manually palpating and counting the spinous processes. The skin was first pre-loaded using medical tape (Hypafix, BSN Medical, Hamburg, Germany) to minimize the soft tissue oscillation between the bone and the sensor (Saha & Lakes, 1977). The vertical (y) and transverse (x) axes of each accelerometer were aligned with the craniocaudal (+y pointing inferiorly) and mediolateral axes of the spine (+x to the subject's left), respectively. The device was calibrated at 1 g = 9.81 m/s^2 to align with earth gravitation (Kavanagh & Menz, 2008). Analog data was A/D converted using a 16-bit card with a sampling frequency of 1250 Hz.

Table 3.4: *Accelerometer location and corresponding placement*

Accelerometer Location	Accelerometer Placement
Head	Posterior central aspect of the head via a headband
T12-L1	Between the spinous processes of T12 and L1
L5-S1	Between the spinous processes of L5 and S1
Distal Tibia	Anteromedial aspect of the right distal tibia, superior to the medial malleolus

3.4.3 Motion Capture

Three-dimensional kinematics were collected using an active optoelectronic motion capture system (Optotrak Certus, Northern Digital Inc., Waterloo, ON, Canada). Four position sensors with banks of three infrared detecting cameras each were used to capture the volume around the treadmill. The capture volume was calibrated prior to collection using a rigid cube with 16 infrared-emitting diodes (IREDS) or markers via a dynamic and static calibration trial. The dynamic trial defined the operational volume of interest and each camera's relative position to each other so that the marker position root mean square (RMS) error was less than 0.5 mm (Callaghan et al., 1999), and the static trial ensured that the position sensors were aligned to a global coordinate system

(GCS). The GCS was established in accordance with ISB recommendations: +X directed anteriorly, +Y superiorly, +Z laterally to the right (G. Wu & Cavanagh, 1995). Sampling frequency was set to 50 Hz.

Rigid marker clusters consisting of four to five markers each were used to track the participant's segments. The clusters were firmly affixed to the subject's lumbar spine, pelvis, and right thigh, shank (Table 3.5), and shoe using double-sided tape (Indoor Carpet Tape, Scotch, St. Paul, MN, USA), medical tape (Hypafix, BSN Medical, Hamburg, Germany), and Velcro straps to limit motion of the clusters relative to the skin (Table 3.5). Medical tape was also placed between the clusters and the skin to minimize effects of sweat. Participants were also asked to wear a grounding bracelet (Anti Static ESD Wrist Strap, FEITA Electronics Co., Ltd., Dongguan City, China) to limit electrostatic interference from the treadmill. Anatomical landmarks were then digitized using a four-marker probe to define segment end points and imaginary markers with respect to the segment clusters (Table 3.5). All landmarks were palpated by the same researcher to limit inter-subject error.

Table 3.5*: Location of marker clusters and digitized bony landmarks to define segment end points.*

Segment	Cluster Location	Digitized Bony Landmarks
Lumbar Spine (LUM)	L3 Spinous Process	• Bilateral 12th ribs • Bilateral iliac crests
Pelvis (PLV)	S1 Spinous Process	• Bilateral iliac crests • Bilateral anterior superior iliac spines • Bilateral posterior superior iliac spines • Bilateral greater trochanters
Right Thigh (RTH)	Right Lateral Femur	• Right greater trochanter • Right medial & lateral femoral condyles
Right Shank (RSK)	Right Lateral Fibula	• Right medial & lateral femoral condyles • Right medial & lateral malleoli

Right Foot/Shoe (RFT)	Right Anterolateral Talus	• Right medial & lateral malleoli • Right 1st and 5th metatarsal heads • Right distal phalanx of 2nd toe • Right calcaneal tuberosity

A quiet standing calibration trial was collected to define neutral joint angles, followed by dynamic functional joint trials for the right hip and knee. These dynamic trials involved movement of the leg through moderate range flexion/extension, abduction/adduction, and circumduction for the hip and flexion/extension at the knee and are used to improve prediction of the hip and knee joint centres respectively. Two more quiet standing calibrations and their associated new landmarks were collected following each footwear change.

3.5 Protocol

Participants were asked to run in each shoe at 3.3 m/s for 5 minutes on a digitally controlled treadmill at level grade. Pace and duration were based on previous experiments involving recreational runners as described in Table 3.6, which includes commonly used testing and self-selected speeds. Prior to collection, participants were given five minutes for warm up and familiarization with the assigned speed (Fellin & Davis, 2009; Sinclair, Edmundson, et al., 2013). Shoe order was randomized for each subject. Data was continuously collected after 3 minutes had elapsed to allow for sufficient metabolic and mechanical adaptation to the shoe (Fellin & Davis, 2009; Hardin et al., 2004; Hunter et al., 2019), for a total of 2 minutes per condition. Between shoe conditions, five minutes of rest were permitted to avoid fatigue effects and during which the shoes were changed and the foot segment re-digitized (Bishop et al., 2006; Pappas et al., 2017). A total of 3 trials were collected. The order of operations for a data collection session is outlined in Figure 3.4.

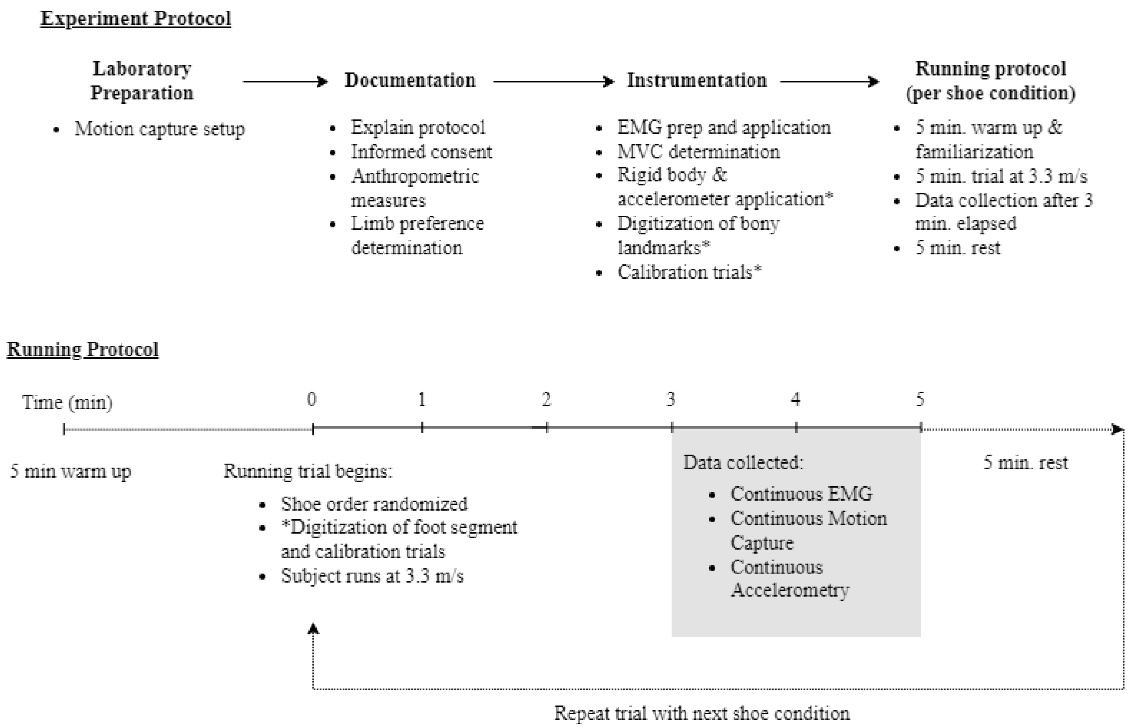

***Figure 3.4**: Outline of experimental and running protocol.* *Each shoe had its own rigid body. The foot was re-digitized and new calibration trials were collected per condition.

Table 3.6: Compilation of studies using treadmill running and their corresponding experimental protocols.

Author	Participants	Footwear Conditions (if applicable)	Warm Up Protocol	Testing pace & duration	Rest Protocol
(Addison & Lieberman, 2015)	22 healthy adults	Minimal footwear with 2 footpads differing in stiffness	Practice walking & running at prescribed speed	Walking at Froude number of 0.28 (1.48 – 1.68 m/s) Running at Froude number of 1.2 (3.06-3.48 m/s) for 30 s	*N/D*
(Bishop et al., 2006)	9 healthy adults accustomed to treadmills	2 shoes with different cushioning	*N/D*	2.23 m/s or 3.58 m/s for 5 min	5 min
(Cigoja et al., 2019)	13 male recreational runners	Control shoe and control shoe inserted with custom carbon fibre plate	Familiarization trials	3.5 m/s for 30 s (~35-45 steps)	*N/D*
(Clermont et al., 2017)	15 recreational and 20 competitive long-distance runners who all had recently participated in a 10 km/half marathon/marathon	Standard shoe	2-5 min	2.7 m/s for 60 s	N/A
(Frank et al., 2013)	10 healthy runners who ran a min. of 20 km/week	Own shoes and minimal shoe	5 min warm up at speed based on RPE	Speed equivalent to Borg RPE of 3 (3.16 ± 0.3 m/s) for 10 min	*N/D*
(Giandolini et al., 2013)	9 adult rearfoot strikers who practiced sports and running (10.3 ± 3.71 h/week)	Standard shoe	5 min at speed determined during familiarization session	Speed determined during familiarization session (3.28 ± 0.65 m/s) for 5 min	2 min
(Gruber et al., 2014)	19 RFS and 19 FFS runners who ran a min. of 16 km/week at 3.5 m/s	Neutral racing flats	Several minutes	3.5 m/s for 2 min	N/A

(Hamill et al., 2009)	3 groups of 11 subjects who ran a min. of 20 km/week over last 6 months	*N/D*	5-10 min at different speeds	3.8 m/s (based on average preferred speed per group) for 30 s	N/A
(Hardin et al., 2004)	12 rearfoot strikers who ran 20 ± 7.4 km/week	2 shoes with different midsole hardness	*N/D*	3.4 m/s for 6 min	Rested until HR < 120 bpm
(Horvais & Samozino, 2013)	12 male regular runners, rearfoot strikers, who ran min. 20 km/week	16 pairs of custom shoes with different heel height and drop height	10 min at preferred speed	3.9 m/s and 4.7 m/s for 1 min	2 min
(Law et al., 2019)	15 male rearfoot strike runners who ran a min. of 12 km/week for past 6 months	6 shoe conditions with different midsole thickness	5 min to determine test speed	Speed determined during warm-up or 3 min	Rest allowed between footwear conditions
(Malisoux, Delattre, Urhausen, et al., 2020)	802 runners, capable of performing 15 min of consecutive running	2 shoes with different global stiffness	3 min	Preferred running speed (soft group 9.8 ± 1.5 km/h, hard group 9.9 ± 1.5 km/h) for 8 min habituation followed by 2 min data collection	*N/D*
(M. A. Mercer et al., 2018)	10 adults who ran a min. of 10 miles/week	Neutral and maximally cushioned shoes	*N/D*	Self-selected speed (2.3 ± 0.4 m/s), self-selected speed + 0.447 m/s, and self-selected speed – 0.447 m/s for 5-10 min depending on time to reach steady state	*N/D*
(McLeod et al., 2020)	21 experienced male runners with a min. 10 km time of 36:00 min	6 shoe conditions with different stiffness based on number of carbon fibre layers	5 min in own shoes	2.98 m/s or 4.47 m/s for 5 min (separate sessions per speed)	2 min
(Pappas et al., 2015)	22 healthy students	N/A	2.22 m/s for 5 min	4.44 m/s for 30 s	*N/D*
(Pappas et al., 2017)	31 healthy students	N/A	2.22 m/s for 5 min, then increased to	6.67 m/s at 10 s, performed 7 times	5 min

			5.55 m/s in last 30 s		
(Sinclair, Edmundson, et al., 2013)	12 male runners who ran min. 35 km/week with previous treadmill experience	2 shoes with different shock attenuation properties	4.0 m/s for 3 min	4.0 m/s for 6 min	Rested until HR < 110 bpm
(Sinclair et al., 2016)	12 male runners who ran min. 35 km/week with previous treadmill experience	2 shoes with different energy return properties	3 min at 12 km/h (3.33 m/s)	12 km/h (3.33 m/s) for 6 min	Rested until HR < 110 bpm
(TenBroek et al., 2014)	10 recreational male runners who could run for 30 min	3 shoes with different midsole thickness	Standard treadmill warmup in own footwear	3.0 m/s for 30 min	Separate days per condition
(Worobets et al., 2014)	12 recreational athletes with running experience	2 shoes with different midsole materials	10 min warm up in own shoes	3.4 ± 0.3 m/s for 5 min (0.225 m/s below speed associated with each subject's anaerobic threshold)	5 min

HR = heart rate; N/D = no data

3.6 Data Analysis

All data processing were performed with custom scripts using MATLAB (version 2020b, The Mathworks Inc., Natick, MA, USA) unless otherwise stated. All signals were time-normalized to 100% of stance using cubic spline interpolation. Average values from fifteen consecutive strides per condition were used for statistical analysis.

3.6.1 Kinematics

Visual3D motion analysis software (Version 6.01.36, C-Motion Inc., Kingston, ON, Canada) was used to process all kinematic data. Raw marker data was processed with a maximum gap size of 10 ms (cubic spline interpolated if necessary; Howarth & Callaghan, 2010) and filtered using a 4th order zero-lag Butterworth filter with a cut-off frequency of 12 Hz (Baltich et al., 2015; Hannigan & Pollard, 2020; Kulmala et al., 2018).

A three-dimensional rigid link model consisting of the lumbar spine, pelvis, thigh, shank, and foot segments was constructed using the digitized bony landmarks as segment end points (Table 3.5). The quiet standing calibration trial was used as a reference posture, and local coordinate systems (LCS) were defined for each segment based on ISB conventions (G. Wu et al., 2002). Functional joint trials were used to improve the accuracy of hip and knee joint centre estimations based on the built-in recursive sphere-fitting algorithm.

Time-varying Euler angles were then derived using a Z-X-Y Cardan rotation sequence with flexion/extension about the local z-axis, abduction/adduction about the x-axis, and internal/external rotation about the y-axis. Ankle, knee, hip, and lumbar joint angles were calculated with respect to the proximal segment (e.g., ankle defined as angle of the foot with respect to the shank, lumbar angle defined as the angle of the lumbar spine with respect to the pelvis), with positive angles representing joint flexion (negatives were calculated for knee and lumbar flexion). Foot strike angle was identified by the sagittal angle of the foot segment with respect to the global horizontal axis at initial contact. Angles greater than 0 degrees indicated RFS (Lieberman et al., 2010), and only participants who ran with a RFS for at least 75% of their strides were included.

Fifteen consecutive strides were randomly selected from the two minutes of data capture (Delgado et al., 2013; Riley et al., 2008) and then checked to ensure that all motion capture data were visible. If data was missing, the starting stride number was re-selected. Fifteen strides were used following the protocols of Delgado et al. (2013) and Riley and colleagues (2008), where 12 to 19 strides have been previously reported as the mean stride count required for stable sagittal plane kinematics in masters-level runners (Riazati et al., 2019), who have shown to have even greater stride to stride variability than young adults (Boyer et al., 2017).

Stance phase was classified following the algorithm by Smith et al. (2015): initial contact (IC) was defined as the first maximum in vertical displacement between the ipsilateral heel and posterior superior iliac spine (PSIS) markers, and toe-off (TO) occurred at the peak vertical displacement between the right second metatarsal and PSIS markers closest to the second peak in knee extension. This algorithm was adopted based on its reliance on kinematic methods, applicability to treadmill running, and temporal accuracy compared to the gold standard of using vertical ground reaction force (see Appendix A). Using these definitions, the sagittal joint angle at initial contact, as well as the mean, minimum, maximum, and total range during stance phase were used for statistical analyses.

3.6.2 Muscle Activation

The digital time-varying EMG signals of the lumbar erector spinae, rectus abdominus, and external obliques were processed as follows:

1. Systematic bias was removed (i.e., detrended).
2. A 4^{th} order, 60 Hz band-stop Butterworth filter was applied to remove potential contamination from electromagnetic hum (Mello et al., 2007).
3. The uncontaminated data was full-wave rectified and digitally low-pass filtered using a 2^{nd} order, dual-pass Butterworth filter with a cut-off frequency of 2.5 Hz (Brereton & McGill, 1998).
4. The maximal amplitude measured during the MVC trials (processed with the same steps) for each corresponding muscle was used as the reference amplitude for normalization of each muscle's signal to allow muscle activity levels to be expressed as a percentage of maximum.

Co-activation coefficients (CCI) and the timing between all muscle pairs were then computed. This was obtained first by calculating by the CCI on normalized, linear enveloped EMG data via the equation according to Lewek et al. (2004) and Nelson-Wong et al. (2010):

$$CCI = \sum_{i=1}^{N} \left(\frac{EMG_{low_i}}{EMG_{high_i}} \right) \times \left(EMG_{low_i} + EMG_{high_i} \right) \qquad \textit{(Equation 1)}$$

where EMG_{high} and EMG_{low} refer to the muscle with the higher and lower activation levels respectively during the *i*th frame, and *N* represents the total period of data frames, set to 100 during stance phase. The CCI across all 15 strides was then averaged for each pair of muscles and expressed as %MVC.

Cross-correlation was also conducted on the time-varying normalized EMG signals during the stride period according to the cross-correlation method developed by Nelson-Wong et al. (2009): One signal, $x(t)$, was held stationary while the second, $y(t)$, was incrementally shifted in time along the entire length of the signal to produce a special correlation value (R_{xy}) at each time shift (τ), with the phase lag represented by τ. The increment size for τ was defined by the relationship $\tau_{min} = 1/f_s$, where f_s is the sampling frequency, in this case 2048 Hz. This was digitally incremented via the function:

$$R_{xy}(\tau) = \frac{\frac{1}{N}\sum_{i=1}^{N}(x_i - \bar{x})\left(y_{i+\tau\cdot f_s} - \bar{y}\right)}{\frac{1}{N}\sqrt{\sum_{i=1}^{N}(x_i - \bar{x})^2 \sum_{i=1}^{N}(x_i - \bar{y})^2}} \qquad (Equation\ 2)$$

where the numerator is the sum of the product of the two time-varying signals with their respective means subtracted, the denominator is the square root of the product of squared deviations of the two signals used to normalize the coefficient value, *N* is the number of data points in the signal, τ is the discrete temporal phase shift, and f_s is the sampling frequency in Hertz. The maximum τ value was set to within ±500 ms based on the observed activation timing between paraspinal muscles during gait studies (Nelson-Wong et al., 2009). For comparisons involving the LES, the LES was selected as the reference muscle since it provided the most contribution to sagittal motion. Between the RA and EO, the RA was selected as the reference muscle. This determined that a positive phase lag would indicate that the reference muscle was activated first, while negative τ values meant the other muscle was activated first. Alternatively, a positive R_{xy} meant that the two signals were increasing and decreasing together, while a negative value indicated an inverse relationship (one as increasing in activation while the other muscle was reduced).

Mean bilateral activation per muscle during stance phase as well as the co-activation coefficient, cross-correlation, and phase lag of each muscle pair were used for statistical analyses.

3.6.3 Acceleration and Shock Attenuation

Accelerometry signal data was first calibrated at 1 g = 9.81 m/s^2, then filtered using a second order, low pass Butterworth filter with a cut-off frequency of 60 Hz and de-trended as necessary. Resultant acceleration was calculated as $\sqrt{x^2 + y^2 + z^2}$. Peak impact resultant acceleration magnitudes were calculated for four locations: distal tibia, L5-S1, T12-L1, and the head. Peak acceleration was defined as the highest peak measured during stance phase.

Shock attenuation (SA) was then calculated in the frequency domain as the ratio between the signal power of the inferior to superior sensors as a measure of incoming and outgoing shock during stance phase (Kavanagh & Menz, 2008). SA between each subsequent accelerometer was measured as well as the total SA: between the distal tibia and L5-S1; between L5-S1 and T12-L1; between T12-L1 and the head; and between the distal tibia and the head.

Fast Fourier transformations (FFT) were first used to calculate the power spectral density (PSD) of each signal, where PSDs from 0 to 30 Hz were computed based on the frequency range thought to be associated with running impact (Giandolini et al., 2019; Shorten & Winslow, 1992; TenBroek et al., 2014) and normalized to 1 Hz bins. The sum of the powers from 0 to 30 Hz were used to normalize signals to mean squared amplitudes.

SA was consequently measured in decibels (dB) via the transfer function:

$$H = 10\, log_{10}\left(\frac{PSD_{sup}}{PSD_{inf}}\right) \qquad (Equation\ 3)$$

where PSD_{sup} and PSD_{inf} represent the PSD of the signals from the superior and inferior accelerometers respectively, negative H values indicate signal attenuation, and positive values indicate signal gain. This approach has been widely used for analyses of SA during walking, running, and other dynamic tasks (Castillo & Lieberman, 2018; J. J. Chu & Caldwell, 2004;

Gruber et al., 2014; Hamill et al., 1995; Kavanagh & Menz, 2008; J. A. Mercer et al., 2003; Shorten & Winslow, 1992).

3.7 Statistical Analysis

All statistical analyses were conducted in R (Version 4.0.3, R Development Team, Vienna, Austria).

Descriptive statistics including means and standard deviations were calculated for each footwear condition per sex. The Shapiro-Wilk test was conducted on the residuals of all parameters to test for violations of normality assumptions. If found to deviate from normality, data was log transformed if possible. Two-way mixed measures analyses of variance (ANOVA) were then conducted to determine the effect of SEX and midsole cushioning properties (SHOE) on sagittal kinematics, muscle activation and co-activation, segment acceleration, and shock attenuation (Table 3.7). Mauchly's test was performed to test if the assumption of sphericity was violated, and Greenhouse-Geisser Epsilon corrections were used to correct the *p*-value if necessary. For data that violated the assumption of normality and could not be log transformed due to negative values, a non-parametric longitudinal analysis was performed in place of the two-way mixed measures ANOVA. The *nparLD* software package in R was applied following a F1-LD-F1 design, where F*x*-LD-F*y* denotes the number of whole-plot factors (*x*), longitudinal data (LD), and the number of sub-plot factors (*y*) (Noguchi et al., 2012).

Significant interactions were decomposed using simple effects, and post-hoc pairwise comparisons via Tukey's honestly significant difference (HSD) were used to determine the means driving significance for significant main effects. For all pairwise comparisons, Bonferroni corrections were applied and the adjust *p*-value was used to determine significance.

If no significant main or interaction effect of SEX was observed, data from both groups were collapsed and one-way repeated measures ANOVAs were conducted. The same post-hoc analysis was applied for any significant results.

For nonparametric analyses, decompositions were conducted using Friedman one-way ANOVAs and Wilcoxon signed rank tests with Bonferroni corrections.

Partial eta-squared (η_p^2) was used to evaluate effect size for all applicable tests, and significance was assigned to *p*-values < 0.05. Non-significant results but with *p*-values < 0.10 were also reported.

A summary of all dependent and independent variables is outlined in Table 3.7, with priority outcome measures denoted by superscripts.

Table 3.7: Summary of Dependent and Independent Variables

Dependent Variable		Independent Variable	Statistical Analysis
Acceleration	• Peak tibial acceleration[1] • Peak L5-S1 acceleration[1] • Peak T12-L1 acceleration • Peak head acceleration	• Sex (b/n) [4] • Shoe (w/n)	• Mixed Measures ANOVA
Shock Attenuation	• L5-S1/Tibia[1] • T12-L1/L5-S1[1] • Head/T12-L1 • Head/Tibia	• Sex (b/n) [4] • Shoe (w/n)	• Mixed Measures ANOVA

EMG	• %MVC:[2]
	  o LES
	  o RA
	  o EO
	• Co-activation coefficient (CCI), cross correlation (R_{xy}) & Phase-lag (τ):
	  o LES & RA
	  o LES & EO
	  o RA & EO
		• Sex (b/n)
		• Shoe (w/n)
			• Mixed Measures ANOVA

Kinematics	• Sagittal ankle joint angles • Sagittal knee joint angles[3] • Sagittal hip joint angles • Sagittal lumbar flexion angles[3]	• Sex (b/n) • Shoe (w/n)	• Mixed Measures ANOVA

b/n=between-subjects factor; w/n=within-subjects factor
[1] addresses hypothesis 1; [2] addresses hypothesis 2; [3] addresses hypothesis 3; [4] addresses hypothesis 4

4 Results

All participants completed the protocol as planned. However, data from one female had to be excluded due to being classified as a midfoot striker, which was revealed after data collection was completed. In addition, all accelerometry data from one male participant and foot kinematic data from one shoe condition for one female participant were removed due to technical issues that resulted in incomplete data. This resulted in 19 participants for further analysis, with 17 (9M, 8F) containing full datasets. Full statistical results are provided in Appendix B.

4.1 Footwear Characteristics

One-way between-groups ANOVAs were conducted on stiffness, energy absorbed, energy returned, and hysteresis to determine if the shoes were significantly different, and post-hoc pairwise comparisons with Bonferroni corrections were run if necessary to determine which means were driving significance (Appendix B). Tukey's HSD revealed that the ZMX was significantly more compliant than the PGS and RCT ($p<0.001$), but no difference was observed between the PGS and RCT ($p=0.909$). The energy absorbed by the PGS was significantly less than that of the other two shoes (PGS vs. RCT $p<0.01$; PGS vs. ZMX $p<0.001$). All shoes differed significantly in energy released ($p<0.001$ for PGS vs. RCT and PGS vs, ZMX; $p<0.05$ RCT vs. ZMX) and hysteresis ($p<0.001$). Specifications of each shoe are shown in Table 4.1.

***Table 4.1**: Men's 9.5, Women's 7.5, and combined (mean) right midsole characteristics computed across 5 loading and unloading mechanical testing cycles.*

Characteristic	PGS			RCT			ZMX		
	Female	**Male**	***Mean***	**Female**	**Male**	***Mean***	**Female**	**Male**	***Mean***
Peak Deformation (mm)	13.69	14.73	*14.21*	13.92	15.38	*14.65*	15.94	17.52	*16.73*
Stiffness (N/mm)	103.87	98.21	*101.04*	106.42	97.14	*101.78*	89.67	82.62	*86.16*
Energy Absorbed (J)	8.04	9.10	*8.57*	9.16	10.23	*9.71*	9.14	10.61	*9.87*
Energy Returned (J)	6.45	7.34	*6.89*	7.60	8.58	*8.09*	8.09	9.44	*8.76*
Hysteresis (%)	19.79	19.31	*19.55*	17.05	16.38	*16.72*	11.49	10.98	*11.23*

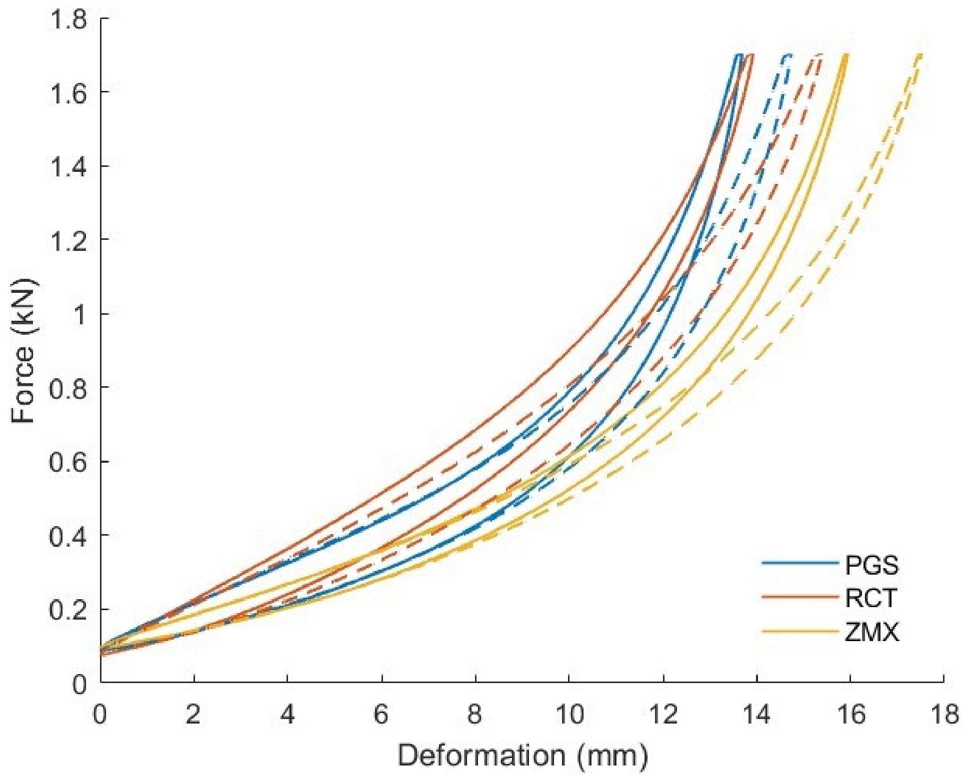

Figure 4.1: *Mean force-deformation curves obtained from the last 5 cycles of mechanical testing on Women's 7.5 (solid line) and Men's 9.5 (dashed line) shoes.*

4.2 Sagittal Angles

Nineteen participants (10M, 9F) were included in the analysis for the knee, hip, and lumbar joints, while eighteen (10M, 8F) were used for all analyses involving the ankle. Results from the Shapiro-Wilk test revealed that residuals for all levels and joints were normally distributed except for one level at initial contact in the knee (females knee angles in ZMX; $p < 0.001$) and one level for minimum hip angles (female hip angles in PGS, $p < 0.01$), thus non-parametric analyses were conducted for those variables. All kinematic variables are shown in Table 4.2, where positive values represent joint flexion for all joints.

***Table 4.2**: Mean (SD) sagittal kinematics for male and female runners in three different shoe conditions at initial contact and during stance phase. Positive values represent joint flexion. All angles are expressed relative to standing.*

	PGS		**RCT**		**ZMX**	
	Female	**Male**	**Female**	**Male**	**Female**	**Male**
At Initial Contact						
Ankle Dorsiflexion (°)	3.1 (6.0)	2.1 (4.9)	-0.4 (6.8)	2.3 (4.8)	4.3 (6.5)	2.4 (3.5)
Knee Flexion (°)	12.6 (2.6)	9.2 (5.8)	8.7 (5.2)	10.1 (5.8)	12.0 (2.8)	11.1 (8.5)
Hip Flexion (°)	24.5 (7.5)	25.0 (10.4)	24.5 (7.0)	23.7 (10.7)	24.8 (7.0)	22.5 (15.2)
Lumbar Flexion (°)	5.5 (6.6)	4.5 (7.8)	1.7 (8.2)	4.5 (7.2)	5.2 (6.8)	5.5 (8.7)
During Stance Phase						
Mean Ankle Dorsiflexion (°)	3.8 (4.4)	3.8 (3.0)	-0.4 (5.2)	2.4 (3.7)	3.6 (4.2)	2.5 (2.6)
Mean Knee Flexion (°)	26.8 (3.5)	25.8 (3.8)	25.1 (3.0)	27.0 (3.7)	27.1 (3.4)	27.1 (4.5)
Mean Hip Flexion (°)	10.0 (6.5)	15.0 (8.7)	13.5 (5.1)	16.0 (7.0)	12.8 (6.4)	14.4 (9.7)
Mean Lumbar Flexion (°)	6.1 (6.6)	5.4 (8.7)	1.4 (7.8)	4.2 (8.5)	5.0 (7.4)	6.3 (8.6)
Peak Ankle Plantarflexion (°)	-21.9 (4.4)	-20.2 (4.7)	-26.1 (7.0)	-21.6 (5.0)	-21.7 (5.0)	-22.1 (4.2)
Peak Knee Extension (°)	11.8 (2.9)	8.7 (5.5)	7.9 (4.8)	9.2 (5.9)	11.2 (2.9)	9.0 (8.8)
Peak Hip Extension (°)	-17.4 (8.2)	-9.6 (8.3)	-13.2 (6.2)	-9.4 (4.6)	-13.8 (7.6)	-12.0 (12.3)
Peak Lumbar Extension (°)	1.3 (6.3)	0.7 (8.8)	-3.2 (7.1)	0.5 (8.6)	0.1 (6.8)	1.3 (7.9)
Peak Ankle Dorsiflexion (°)	14.8 (4.1)	15.3 (3.4)	9.8 (5.3)	12.4 (3.7)	13.3 (4.2)	12.8 (3.0)
Peak Knee Flexion (°)	39.9 (5.1)	38.1 (5.5)	38.4 (4.4)	39.5 (5.4)	39.8 (4.3)	40.9 (5.9)
Peak Hip Flexion (°)	25.7 (7.3)	29.5 (10.2)	27.3 (5.3)	29.3 (8.0)	27.0 (5.8)	30.6 (9.9)
Peak Lumbar Flexion (°)	10.6 (7.1)	10.0 (9.8)	4.7 (8.3)	7.6 (8.6)	8.9 (7.9)	10.4 (9.3)
Ankle ROM (°)	37.1 (5.0)	35.4 (4.7)	36.0 (5.3)	34.1 (4.0)	34.8 (4.4)	35.0 (4.4)
Knee ROM (°)	28.9 (5.6)	29.4 (7.2)	30.3 (3.2)	30.6 (8.0)	28.7 (4.4)	32.3 (9.4)
Hip ROM (°)	42.7 (3.6)	38.3 (6.8)	40.6 (4.8)	38.1 (4.4)	40.5 (4.4)	42.1 (10.9)
Lumbar ROM (°)	9.2 (4.0)	9.3 (3.6)	7.8 (2.6)	7.2 (1.8)	8.8 (3.5)	9.0 (3.8)

4.2.1 Angles at Initial Contact

The *nparLD* function with ANOVA-type statistics revealed a significant interaction effect of SEX × SHOE on knee flexion during initial contact ($F_{(1, 1.87)} = 3.19$, $p < 0.05$). Females wearing the ZMX demonstrated significantly greater knee flexion (12.0 ± 2.8°) compared to when wearing the RCT (8.7 ± 5.2°; $p < 0.05$), but not the PGS (12.6 ± 2.6°). No difference was observed between sexes ($F_{(1, 1.87)} = 0.22$, $p = 0.637$) for the knee, nor in other joint angles at initial contact as tested using a mixed-measures ANOVA with between-subject factor of SEX and within-subject factor of SHOE. Therefore, SEX was removed from the analysis.

A final repeated-measures ANOVA with within-subject factor of SHOE revealed no significant main effects in any other joint ($F_{(2, 36)} \leq 0.80$, $p \geq 0.457$). However, violations to sphericity were observed for the ankle ($W = 0.63$, $p < 0.05$). Corrected *p*-values using the Greenhouse-Geisser Adjustment ($\varepsilon = 0.73$) were not significant ($F_{(1.46, 24.77)} = 2.25$, $p = 0.138$), but greater dorsiflexion was exhibited when running in the PGS (2.5 ± 5.3°) and ZMX (3.3 ± 5.0°) compared to the RCT (1.1 ± 5.8°).

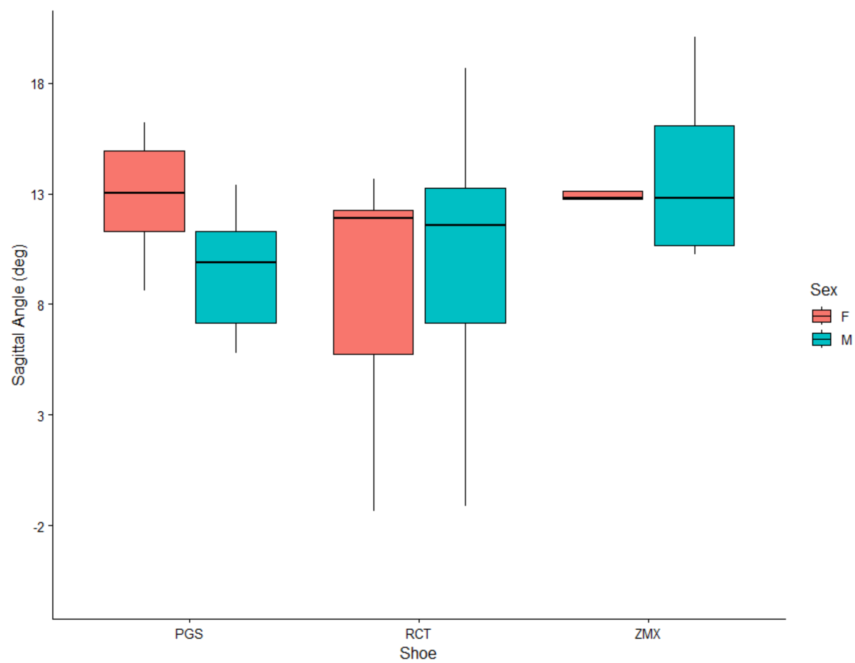

Figure 4.2*: Boxplot of sagittal knee angles (degrees) across shoe conditions and sex at initial contact. Positive values represent knee flexion. A significant difference was observed between females wearing the RCT and ZMX.*

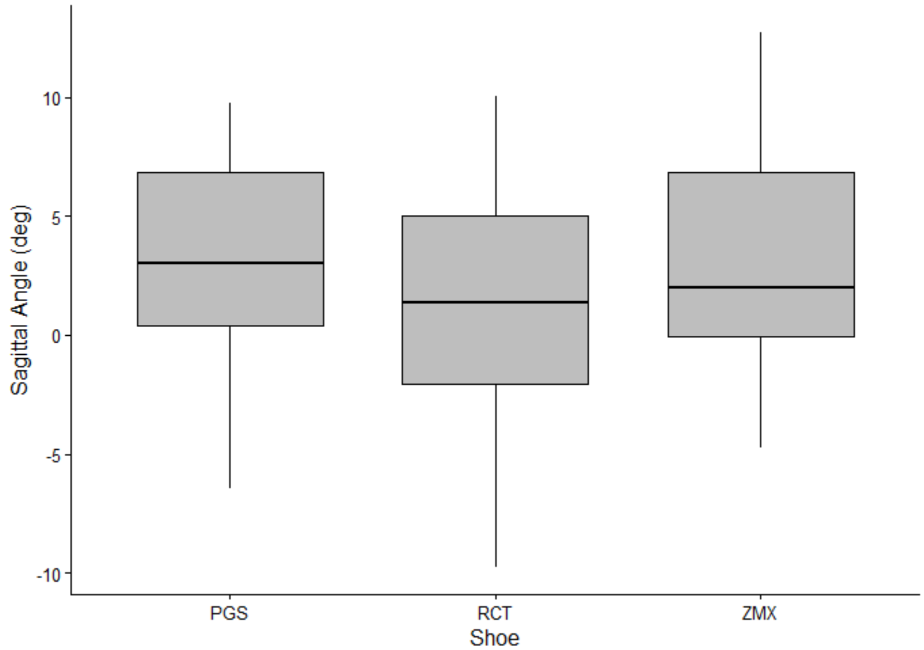

Figure 4.3: *Boxplot of sagittal ankle angles (degrees) across shoe conditions collapsed across sex at initial contact. Positive values represent ankle dorisflexion.*

4.2.2 Angle Ranges During Stance Phase

Sagittal range of motion (ROM) of each joint during stance phase was also calculated. One level of knee ROM data (males, ZMX, $p < 0.05$) violated the assumption of normality and was thus analyzed via the *nparLD* test. The analysis revealed a significant SEX × SHOE interaction in hip ROM ($F_{(1, 1.97)} = 3.57$, $p < 0.05$), however post-hoc Wilcoxon signed rank tests with Bonferroni corrections did not detect any differences. Two-way mixed measures ANOVAs on the remaining joints and levels also did not find any significant interactions or main effects in the ankle, knee, or lumbar ROM ($F \leq 2.98$, $p \geq 0.065$).

After removing SEX as a factor, final one-way repeated-measures ANOVAs did not reveal any significant differences in all joints ($F_{(2, 36)} \leq 2.92$, $p \geq 0.067$). Greater sagittal range of motion was detected when running in PGS for both ankle ($F_{(2, 34)} = 2.71$, $p = 0.081$) and lumbar ROM ($F_{(2, 36)}$

= 2.92, p = 0.067), but the difference was not statistically significant. No violations to sphericity were observed (W ≥ 0.81, p ≥ 0.159).

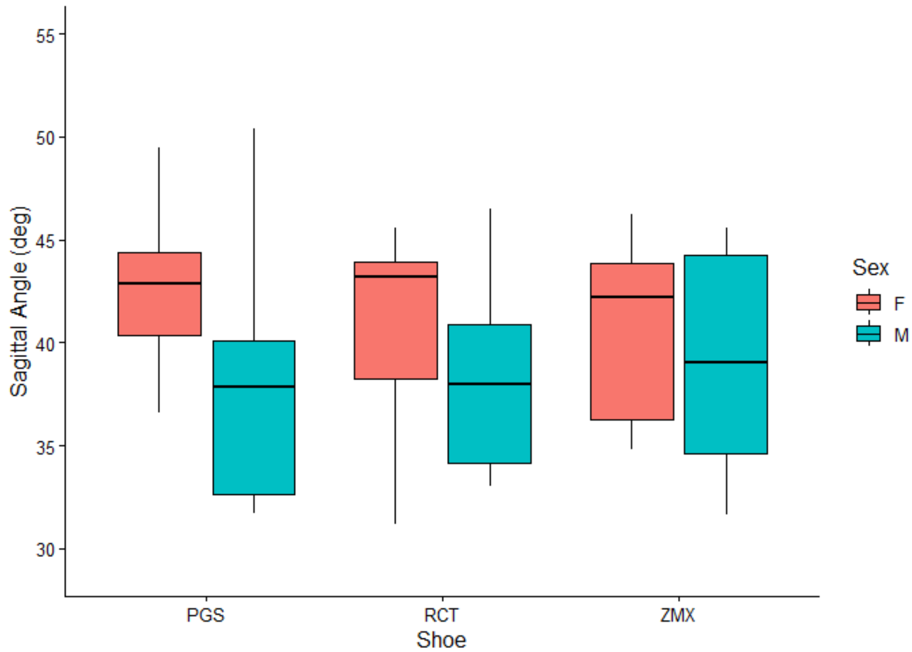

Figure 4.4*: Boxplot of sagittal hip range of motion (degrees) across shoe conditions and sex during stance phase. Positive values represent hip flexion.*

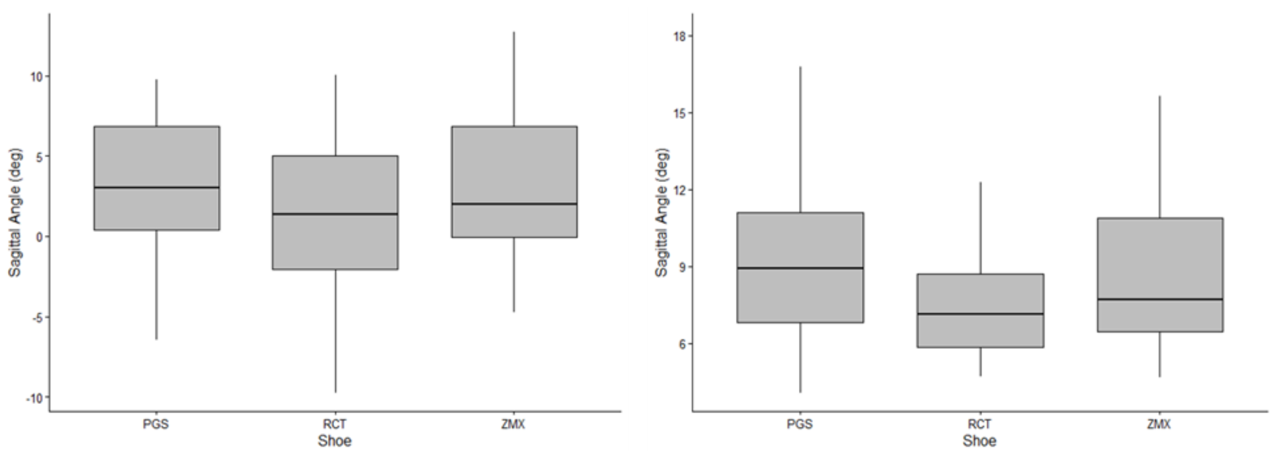

Figure 4.5*: Boxplots of sagittal ankle (left) and lumbar (right) range of motion (degrees) across shoe conditions collapsed across sex during stance phase. Positive values represent ankle dorsiflexion and lumbar flexion. Greater range of motion was demonstrated in the PGS compared to other shoes, but the difference was not statistically significant.*

4.2.3 Mean, Minimum and Maximum Angles During Stance Phase

Mean, minimum and maximum sagittal angles during stance phase were computed where minimum angles corresponded to peak ankle plantarflexion, knee extension, hip extension, and lumbar extension, and mean and maximum values represent flexion for all joints.

Mixed measures ANOVA with between-subject factor of SEX and within-subject factor of SHOE was conducted on mean ankle dorsiflexion, knee flexion, hip flexion, and lumbar flexion angles during stance phase. No significant results were found for any joint except for the ankle, where a significant main effect of SHOE was observed ($F_{(2, 32)} = 5.35$, $p < 0.05$). Following a Greenhouse-Geisser correction due to violations in sphericity ($W = 0.59$, $p < 0.05$, $\varepsilon = 0.71$), the main effect remained, and post-hoc tests showed that the average ankle posture in the PGS was more dorsiflexed (3.8 ± 3.6°) compared to the RCT (1.2 ± 4.5°), but not the ZMX (3.0 ± 3.4°). As no significant effects involving SEX were observed, it was dropped as a factor, and final one-way repeated-measures ANOVAs were conducted on the knee, hip, and lumbar angles. For all outcome measures, no significant results were found ($F < 1.25$, $p > 0.30$).

Mixed measures ANOVAs with between-subject factor of SEX and within-subject factor of SHOE revealed a significant SEX × SHOE interaction ($F_{(2, 32)} = 3.44$, $p = 0.044$) and significant main effect of SHOE ($F_{(2, 32)} = 4.68$, $p < 0.05$) on minimum ankle angles, or peak ankle plantarflexion. However, violations to sphericity were observed ($W = 0.16$, $p < 0.001$), and a Greenhouse-Geisser correction was applied ($\varepsilon = 0.54$). Corrected *p*-values only maintained the significant effect of shoe ($p < 0.05$). Post-hoc pairwise comparisons revealed that running in the RCT resulted in significantly greater plantarflexion (-23.6 ± 6.2°; $p < 0.05$) compared to the PGS (-21.0 ± 4.5°), however no difference was observed between either shoe and the ZMX ($p = 0.26$ compared to PGS, $p = 0.72$ compared to RCT; -21.9 ± 4.5°).

A significant main effect of SHOE was also observed for maximum ankle angles, corresponding to the most dorsiflexion exhibited during stance phase ($F_{(2,32)} = 11.64$, $p < 0.001$). Greenhouse-Geisser corrections were applied due to violations of sphericity ($W = 0.59$, $p < 0.05$, $\varepsilon = 0.71$), and pairwise comparisons revealed that runners demonstrated significantly greater dorsiflexion when wearing the PGS (15.1 ± 3.6°) compared to both the RCT (11.3 ± 4.5°; $p < 0.001$) and ZMX (13.0 ± 3.5°; $p < 0.01$), but no difference was observed between RCT and ZMX ($p = 0.339$).

The *nparLD* function with ANOVA-type statistics revealed a significant main effect of SEX on minimum hip angles or peak hip extension ($F_{(1, 1.84)} = 4.24$, $p < 0.05$). Pairwise comparisons via Wilcoxon signed rank tests showed that females demonstrated significantly greater hip extension (-14.8 ± 7.3°) compared to males (-10.3 ± 8.7°; $p < 0.05$).

No other significant interaction or main effects were observed across both max and min flexion angles in all joints. After removing SEX as a factor, final one-way repeated-measures ANOVAs also did not reveal any significant effects ($F_{(2,3\ 6)} < 2.17$, $p > 0.129$).

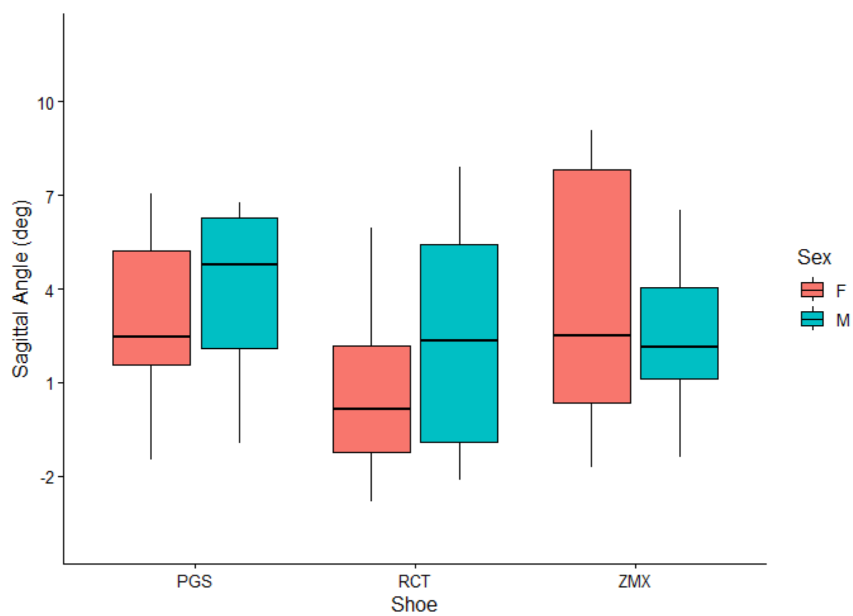

Figure 4.6: *Boxplot of mean sagittal ankle angle (degrees) across shoe conditions and sex during stance phase. Positive values represent ankle dorsiflexion. A significant difference was observed between PGS and RCT.*

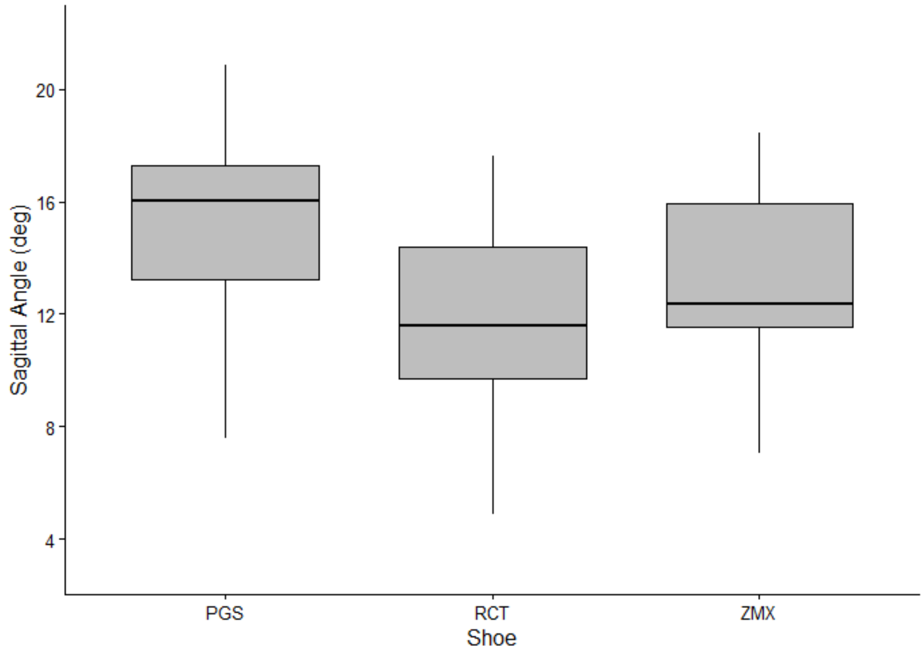

Figure 4.7: *Boxplot of peak ankle dorsiflexion angles (degrees) across shoe conditions collapsed across sex during stance phase. Positive values represent ankle dorsiflexion. A significant difference was observed between PGS and other shoes.*

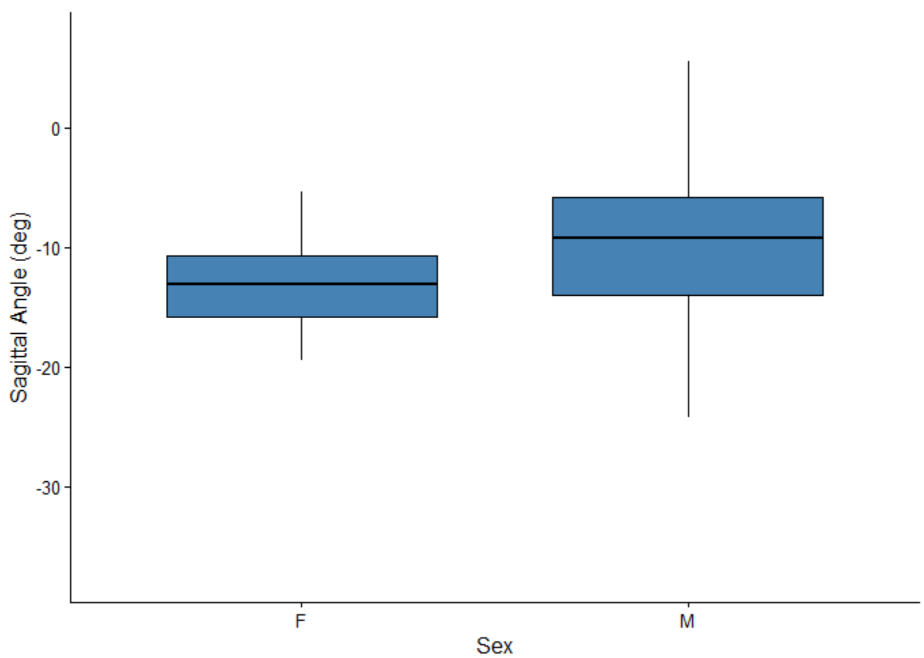

Figure 4.8: *Boxplot of peak hip extension angles (degrees) between sex collapsed across shoe conditions during stance phase. Positive angles represent hip flexion. A significant difference was observed between sex.*

4.3 Muscle Activation

***Table 4.3**: Mean (SD) muscle activation, co-activation and cross-correlation coefficients, and phase lags with peak correlation for male and female runners in three different shoe conditions.*

	PGS		RCT		ZMX	
	Female	**Male**	**Female**	**Male**	**Female**	**Male**
LES Activation (%MVC)	10.97 (4.13)	20.85 (31.63)	11.95 (4.15)	13.92 (9.77)	11.91 (5.31)	17.42 (17.01)
RA Activation (%MVC)	11.67 (3.71)	9.11 (4.59)	13.76 (4.99)	9.13 (4.11)	11.17 (3.38)	8.54 (4.39)
EO Activation (%MVC)	9.92 (6.54)	6.00 (5.43)	9.55 (7.07)	6.58 (5.54)	9.75 (8.77)	6.30 (5.74)
LES-RA						
Co-activation Coefficient (%MVC)	16.3 (5.6)	15.6 (8.1)	15.1 (5.0)	16.5 (8.7)	17.2 (7.5)	14.9 (7.7)
Correlation	0.27 (0.21)	0.34 (0.20)	0.37 (0.11)	0.34 (0.17)	0.32 (0.16)	0.34 (0.20)
Phase Lag (s)	-0.01 (0.31)	-0.04 (0.18)	-0.13 (0.20)	0.03 (0.19)	0.02 (0.27)	-0.18 (0.22)
LES-EO						
Co-activation Coefficient (%MVC)	13.1 (5.1)	10.2 (9.1)	13.1 (6.0)	11.5 (10.9)	13.0 (6.4)	11.4 (11.4)
Correlation	0.36 (0.10)	0.29 (0.11)	0.35 (0.14)	0.35 (0.12)	0.46 (0.10)	0.31 (0.09)
Phase Lag (s)	-0.11 (0.19)	-0.05 (0.26)	0.05 (0.24)	-0.05 (0.26)	-0.12 (0.18)	0.01 (0.34)
RA-EO						
Co-activation Coefficient (%MVC)	13.8 (5.5)	11.6 (10.1)	14.2 (8.7)	11.8 (9.2)	12.9 (6.4)	10.8 (8.4)
Correlation	0.37 (0.15)	0.38 (0.19)	0.35 (0.15)	0.42 (0.13)	0.39 (0.12)	0.44 (0.19)
Phase Lag (s)	0 (0.27)	-0.06 (0.23)	0.1 (0.26)	-0.04 (0.13)	0.14 (0.26)	-0.08 (0.15)

4.3.1 Mean Activation Levels

Nineteen participants (10M, 9F) were included in this analysis. Mean bilateral activation expressed as a percentage of the subject's maximal voluntary contraction was calculated in each shoe. A mixed-measures ANOVA with between-subject factor of SEX and within-subject factor of SHOE revealed a significant main effect of SHOE on RA activation ($F_{(2, 34)} = 3.90$, $p < 0.05$). However, violations to sphericity were observed ($W = 0.52$, $p < 0.01$), and a Greenhouse-Geisser correction was applied ($\varepsilon = 0.68$). Corrected *p*-values maintained the significant effect of SHOE ($F_{(1.35, 23.01)} = 3.90$, $p < 0.05$), however, no significant differences were detected during post-hoc pairwise comparisons with Bonferroni corrections.

The *nparLD* test with ANOVA-type statistics was used to assess differences in LES and EO activation due to violations of normality ($p < 0.01$). A significant main effect of SEX was observed in EO activation ($F_{(1, 1.89)} = 5.45$, $p < 0.05$). Post-hoc pairwise comparisons using Wilcoxon summed rank tests and Bonferroni corrections revealed that EO activation was significantly greater ($W = 75$, $p < 0.05$) in females (9.92 ± 6.54%) than males (6.00 ± 5.43%) when wearing the PGS. No significant main effects or interactions were observed for LES activation ($F_{(1, 1.54)} \leq 0.60$, $p \geq 0.510$), nor when SEX was removed and Friedman ANOVAs were conducted ($Q = 2$, $p = 0.368$), but male LES activation levels were consistently greater than female levels in all shoes.

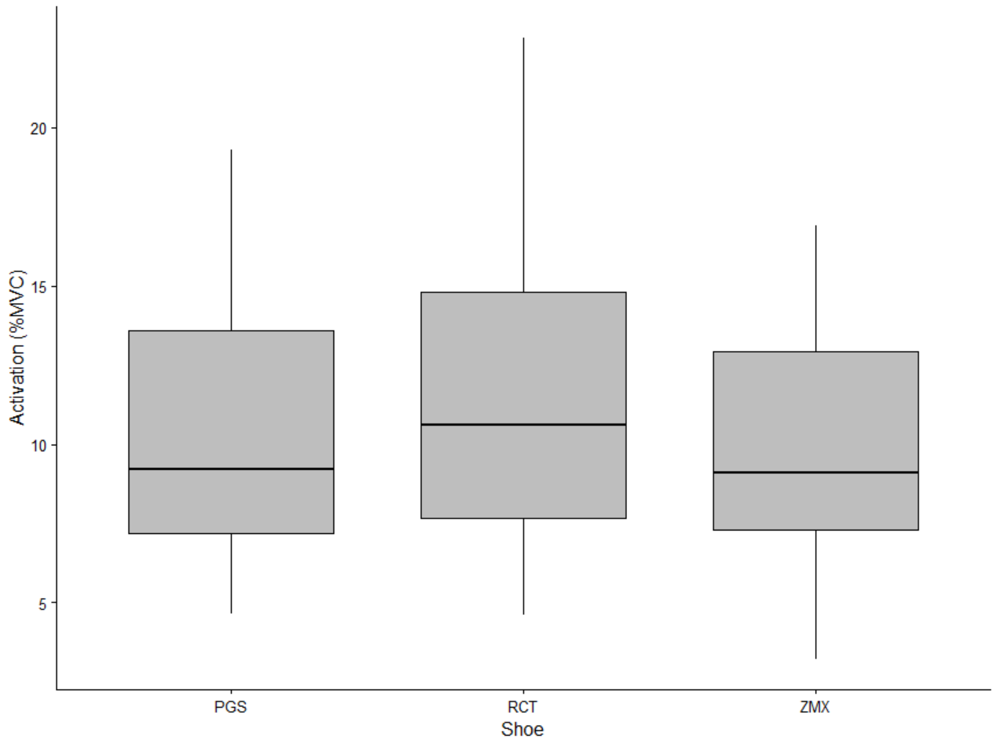

Figure 4.9: *Boxplot of mean RA activation levels (%MVC) across shoe conditions collapsed across sex during stance phase.*

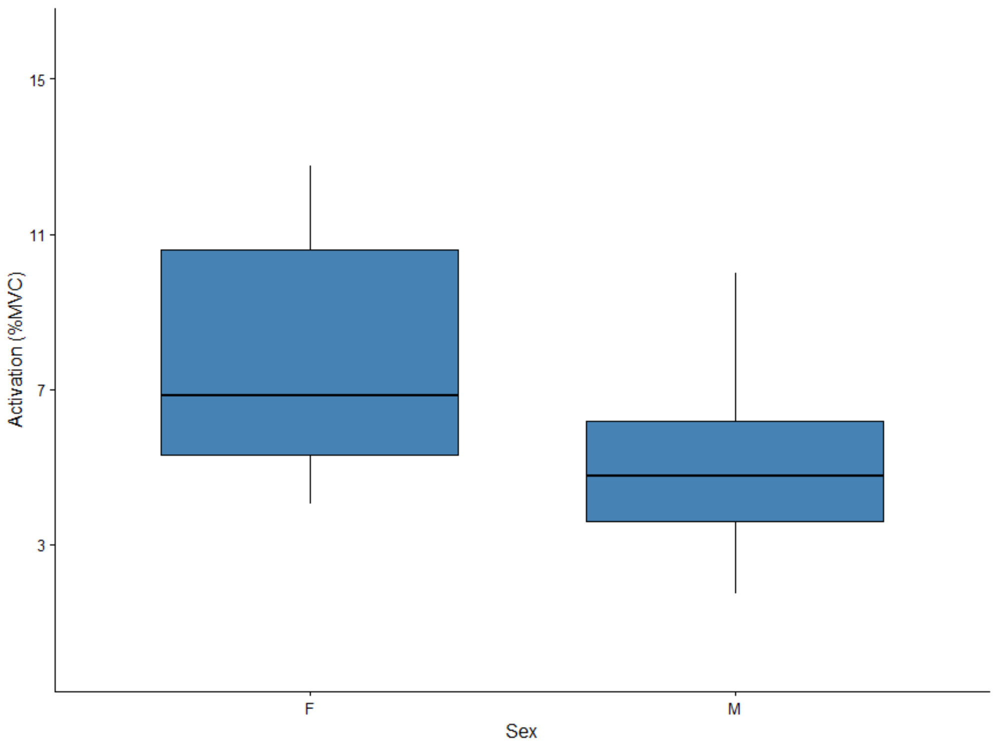

***Figure 4.10**: Boxplot of mean EO activation levels (%MVC) between sex collapsed across shoe conditions during stance phase. A significant difference was observed between sex.*

4.3.2 Co-Activation Coefficients

Co-activation coefficients (CCI) were calculated to provide a measure of the degree of co-activation between bilateral LES, RA, and EO during stance phase across each shoe condition. However, several levels were found to violate normality ($p < 0.001$) for coefficients between the LES and EO, and RA and EO. Thus, the *nparLD* test with ANOVA-type statistics were used for those two pairings. No significant interactions or main effects were observed. Females (13.00 ± 5.63%) demonstrated slightly higher LES-EO CCI than males (11.00 ± 10.20%)., but the difference was not statistically significant ($F_{(1,1.75)} = 3.40$, $p = 0.06$). After collapsing across sex, one-way Friedman ANOVAs did not reveal any significant effects either ($Q \leq 1.26$, $p \geq 0.53$). For the CCI between LES and RA, a mixed-measures ANOVA with between-subject factor of SEX and within-subject factor of SHOE did not reveal a significant interaction or any main effects

($F_{(2,34)} \leq 2.38$, $p \geq 0.108$). After removing SEX as a factor, one-way repeated-measures ANOVA also did not result in any significant differences across shoes ($F_{(2,36)} = 0.02$, $p > 0.05$).

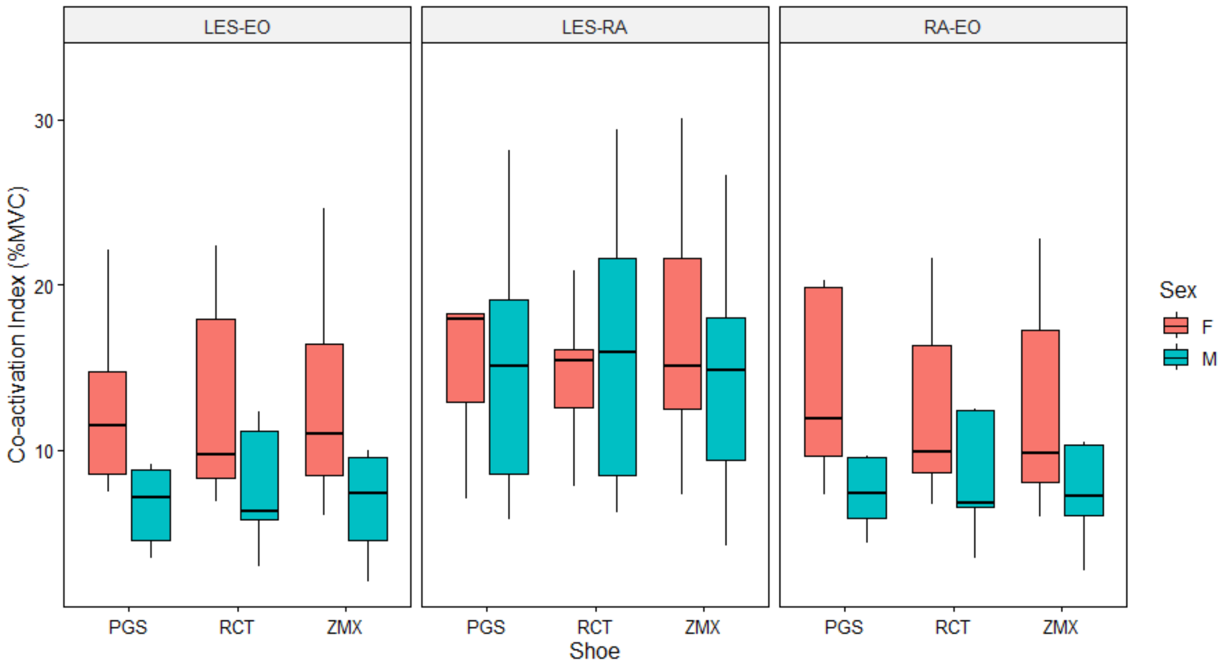

Figure 4.11: *Boxplots of muscle co-activation indices (%MVC) during stance phase across shoe conditions and sex, facetted by muscle pairing.*

4.3.3 Cross-Correlation & Phase Lag

Lastly, cross-correlations were also performed to determine the muscle sequencing relationship between bilateral LES, RA, and EO activation patterns in each shoe during stance phase (Nelson-Wong et al., 2009). Correlations (Rxy) were computed and averaged across 15 strides using the bilateral time series activation patterns per muscle. For each pairing, the phase lag that corresponded to the maximum Rxy was also calculated. All data met the assumption of normality ($p > 0.05$).

Mixed-measures ANOVAs with between-subject factor of SEX and within-subject factor of SHOE revealed no significant main effects or interactions for all muscle pairing cross-correlations and phase lags ($F_{(2, 34)} \leq 3.12$, $p \geq 0.057$). After SEX was dropped from the analysis, one-way repeated-measures ANOVAs also did not find any statistically significant differences in correlations or phase lags ($F_{(2, 36)} \leq 1.61$, $p \geq 0.214$).

Additionally, there were no strong ($r > 0.7$) correlations for any muscle pairing or shoe. Most correlations were identified as moderate ($0.3 < r < 0.7$) or low ($r < 0.3$; Ratner, 2009). Phase lags between the LES and RA, and LES and EO were negative at -0.053 ± 0.23 s and -0.040 ± 0.25 s respectively, indicating that the LES was activated after the abdominal muscles, while the 0.006 ± 0.23 s lag between RA and EO suggest that these muscles were activated nearly simultaneously.

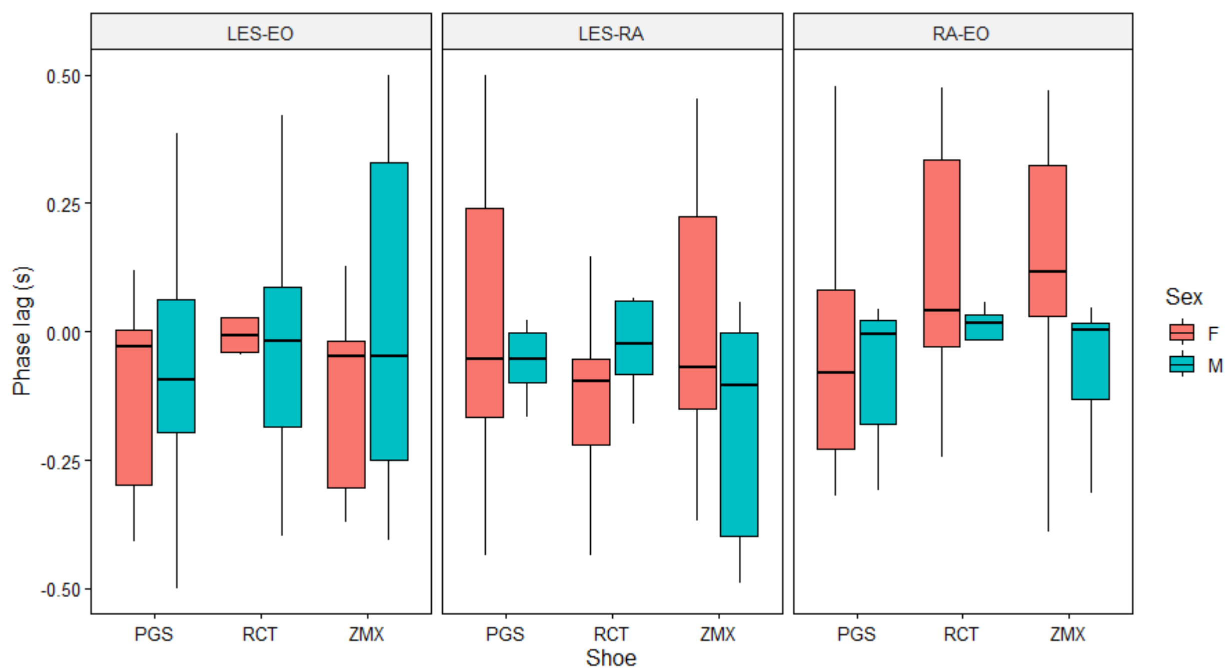

***Figure 4.12**: Boxplots of phase lags (s) between muscles during stance phase across shoe conditions and sex, facetted by muscle pairing.*

4.4 Shock Acceleration Magnitude & Attenuation

Peak resultant acceleration magnitudes and shock attenuation ratios were analyzed for eighteen participants (9M, 9F). Shapiro-Wilk tests revealed that all data met the assumption of normality except head acceleration data for the female levels ($p < 0.001$), thus a non-parametric test was applied for one variable, and two-way mixed measures ANOVAs with between-subject factors of SEX and within-subject factors of SHOE were conducted on the remaining outcome measures.

The *nparLD* function revealed a significant main effect of SHOE on peak head acceleration ($F_{(1, 1.59)} = 3.57$, $p < 0.05$). However, post-hoc pairwise comparisons using Wilcoxon's signed rank test with Bonferroni corrections did not find any significant differences between shoes, though head acceleration magnitudes measured in both the RCT (2.70 ± 1.24 *g*, p = 0.062) and ZMX (2.65 ± 1.50 *g*, p = 0.08) were greater than that in the PGS (2.46 ± 1.12 *g*) when collapsed across SEX.

Mixed-measures ANOVAs also did not find any significant main effects or interactions for peak acceleration at the tibia, L5, or L1 levels ($F \leq 2.86$, $p \geq 0.072$). After removing SEX as a factor, one-way repeated measures ANOVAs also did not reveal any significant differences ($F_{(2, 34)} \leq 1.28$, $p \geq 0.291$).

***Table 4.4**: Mean (SD) shock acceleration magnitude and shock attenuation for male and female runners in three different shoe conditions.*

	PGS		RCT		ZMX	
	Female	**Male**	**Female**	**Male**	**Female**	**Male**
Tibial Acceleration (*g*)	9.80 (2.99)	8.92 (3.59)	10.20 (2.80)	9.60 (4.35)	9.16 (2.47)	8.95 (3.79)
L5 Acceleration (*g*)	5.17 (2.13)	4.95 (2.51)	5.54 (2.28)	4.83 (2.37)	5.72 (2.56)	4.90 (2.22)
L1 Acceleration (*g*)	5.61 (1.55)	5.03 (2.33)	5.7 (2.73)	4.82 (2.34)	4.95 (1.57)	4.93 (1.88)
Head Acceleration (*g*)	2.79 (1.53)	2.13 (0.27)	3.09 (1.64)	2.31 (0.49)	3.11 (2.04)	2.19 (0.40)
Tibia-L5 Shock Attenuation (dB)	-5.80 (3.22)	-4.93 (2.61)	-5.05 (3.52)	-5.24 (2.58)	-4.31 (3.15)	-4.58 (2.54)
L5-L1 Shock Attenuation (dB)	-0.13 (3.93)	-0.32 (2.34)	-0.90 (4.02)	-0.22 (2.20)	-0.33 (4.08)	-0.18 (2.42)
L1-Head Shock Attenuation (dB)	-4.03 (5.28)	-4.32 (2.81)	-4.63 (5.07)	-3.87 (2.52)	-4.33 (5.47)	-3.95 (2.35)
Tibia-Head Shock Attenuation (dB)	-9.71 (3.37)	-8.93 (2.11)	-8.79 (3.37)	-8.09 (2.21)	-8.97 (3.76)	-8.34 (2.52)

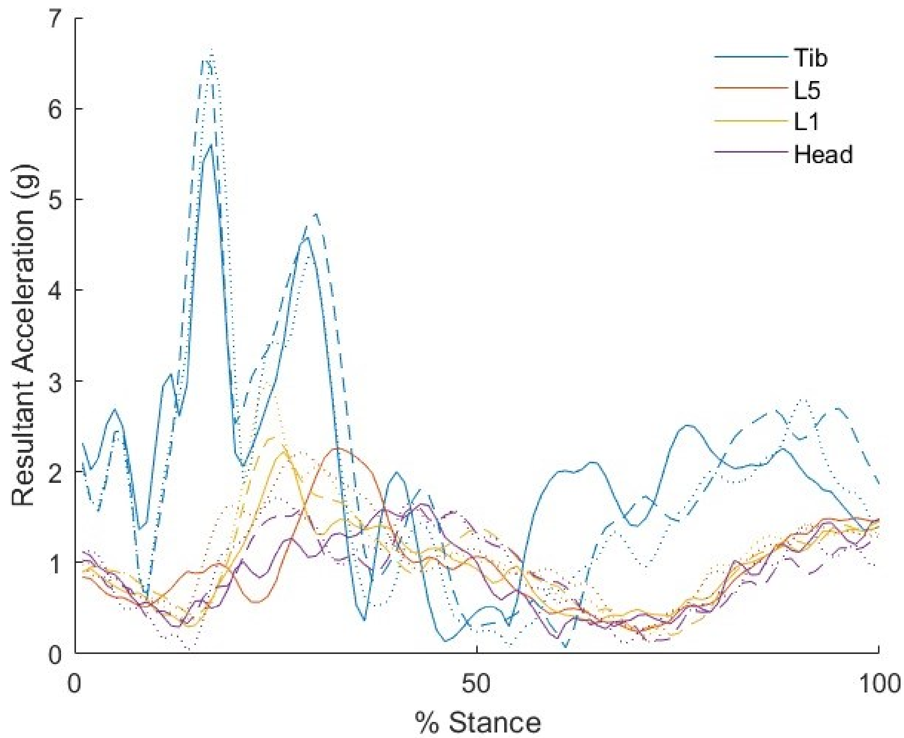

Figure 4.13: *Typical resultant acceleration (g) curves from each accelerometer during stance phase across PGS (solid lines), RCT (dashed lines), and ZMX (dotted lines).*

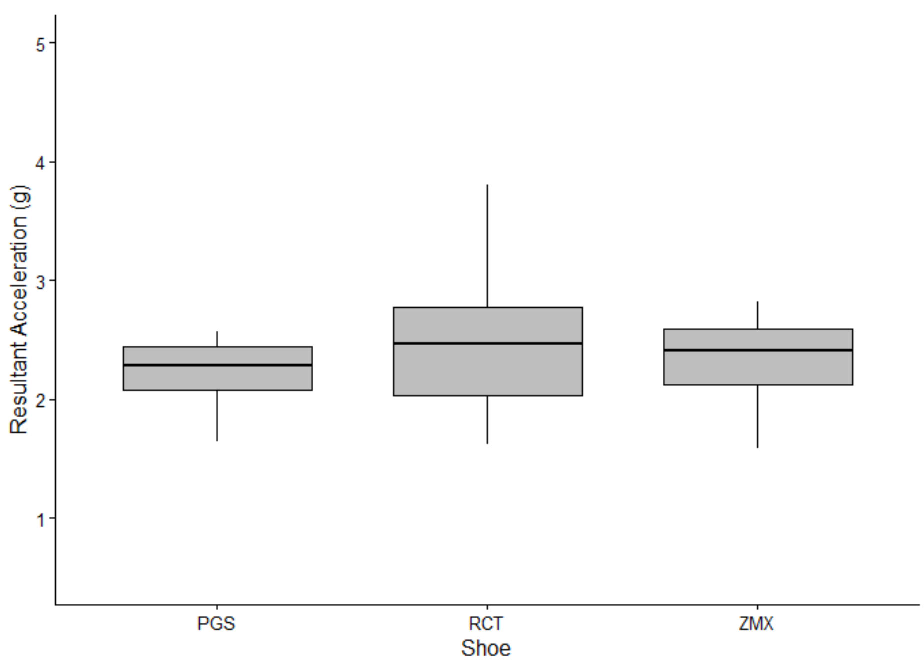

Figure 4.14: *Boxplot of peak resultant head acceleration (g) across shoe conditions collapsed across sex during stance phase.*

Shock attenuation (SA) was calculated as a ratio of the power spectral densities computed between 0 to 30 Hz for each sensor. Two-way mixed measures ANOVAs revealed no significant main or interaction effects of SEX, thus it was dropped from the analysis. Final, one-way repeated measures were conducted, and also revealed no significant differences between shoes. Tibia to L5 SA appeared to be greater in the PGS (-5.37 ± 2.88 dB) and RCT (-5.15 ± 3.00 dB) compared to ZMX (-4.44 ± 2.78 dB), but the effect was not significant ($F_{(2, 34)} = 3.06$, $p = 0.06$). No other effects were statistically significant, and no violations to sphericity were observed ($W \geq 0.71$, $p \geq 0.065$).

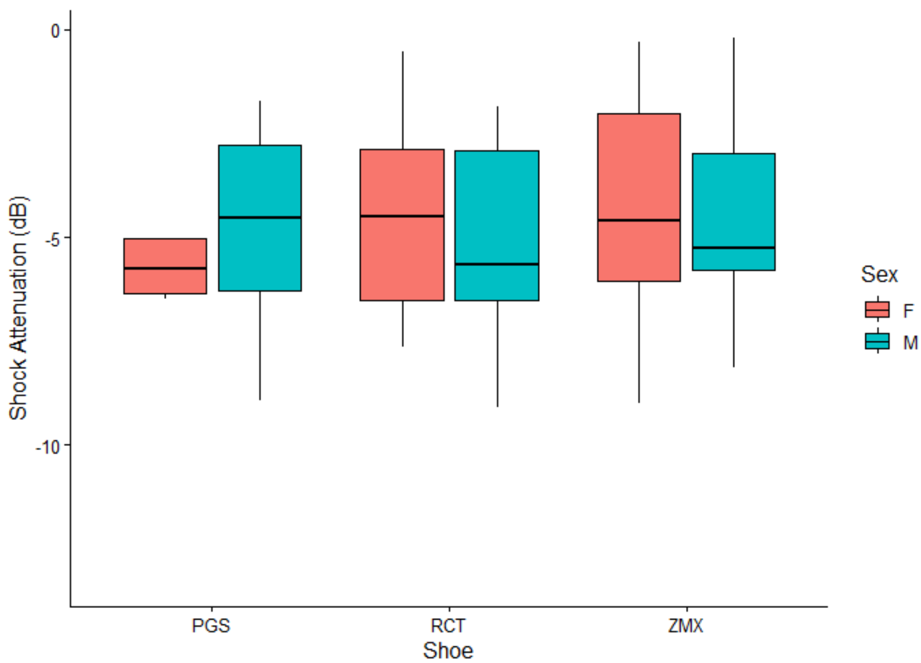

Figure 4.15: *Boxplot of mean shock attenuation (dB) between the tibia and L5 across shoe conditions and sex during stance phase. Negative values denote signal attenuation.*

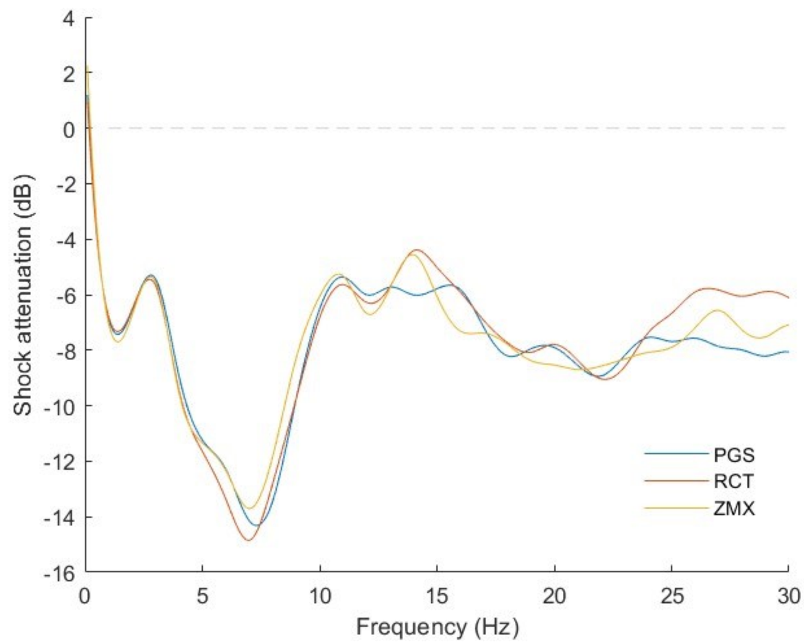

Figure 4.16*: Transfer function depicting the mean shock attenuation (dB) from 0 to 30 Hz between the tibia and L5 acceleration signals across shoe conditions. Negative values represent signal attenuation.*

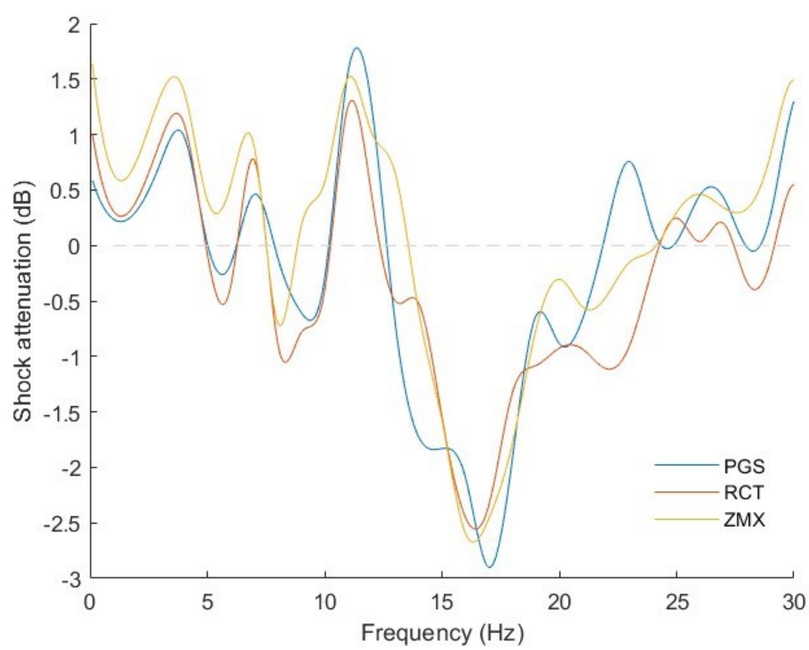

Figure 4.17*: Transfer function depicting the mean shock attenuation (dB) from 0 to 30 Hz between L5 and L1 acceleration signals across shoe conditions. Negative values represent signal attenuation.*

5 Discussion

Midsole cushioning stiffness has been suggested as one footwear component that may have the most influence on the degree of shock or "impact" that a runner experiences with each ground contact. Softer, more compliant midsoles may allow for greater energy dissipation and deformation, thus redirecting the force of impact into the shoe, rather than employing than the musculoskeletal system. However, previous studies have only focused on local effects on the ankle, and it remained unclear whether midsole cushioning affected the body further up the kinetic chain – particularly at a primary site of shock absorption and attenuation, the low back.

Therefore, this study sought to answer the overarching question of whether midsole cushioning stiffness affects shock attenuation in the low back. In general, the results from this study suggest that midsole cushioning stiffness does not substantially affect shock attenuation in the low back during running. Most variables were not sensitive to differences in footwear, particularly beyond the lower limb, of which only knee flexion during initial contact and ankle dorsiflexion across stance phase differed between shoe conditions. Neither lumbar shock magnitudes nor attenuation varied across midsole stiffness. A sex difference was observed in peak hip extension and mean rectus abdominus activation, but not in any shock attenuation metrics.

Further details and specific findings within each outcome measure are elaborated on in the following sections. Firstly, characteristics of the footwear used in this study are provided and compared to existing work on midsole stiffness. The research questions pertaining to sagittal postures (RQ4) and muscle activation patterns (RQ3) are then discussed prior to answering the overarching research questions (RQ1-2) on whether midsole stiffness affects acceleration and shock attenuation in the low back to provide sufficient context and evidence for the mechanisms of shock attenuation adopted during running. Finally, whether such mechanisms differed across sex is reviewed (RQ5-6). Additional *a posteriori* results are located in Appendix C.

A summary of the primary research questions and corresponding hypotheses and conclusions are presented in Table 5.1.

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Table 5.1: *Overview of research questions, hypotheses, conclusions, and main supporting evidence.*

Question	Hypothesis	Conclusion	Supporting Evidence
How does midsole stiffness affect shock transmission and attenuation in the low back during running?	Running in softer and more compliant midsoles will result in increased acceleration magnitudes at the low back. Running in softer and more compliant midsoles will result in decreased shock attenuation in the low back.	Rejected Rejected	No differences in magnitude were observed in the lumbar spine between midsole conditions, nor in shock attenuation in the low back across the entire frequency spectrum. However, *a posteriori* analyses demonstrated greater low frequency shock attenuation in the lower extremity in more compliant midsoles, providing support for partial acceptance.
How does trunk and low back muscle activation change with midsole stiffness during running?	Running in softer and more compliant midsoles will increase trunk and low back muscle activation.	Rejected	Although there is potential for partial acceptance due to the significant main effect observed in the rectus abdominus, no consistent differences were observed regarding peak trunk and low back muscle activation between shoes.
How do low back and lower limb kinematics change with midsole stiffness during running?	Running in softer and more compliant midsoles will result in less knee flexion and lumbar spine flexion at initial contact during running.	Partially Accepted	Greater knee extension was demonstrated in female runners when wearing shoes with softer midsoles compared to stiffer footwear, however no differences between midsoles were observed in males or at in lumbar flexion for both sexes.
Are there sex differences in shock attenuation when running in midsoles of different stiffness?	Females will experience greater segment acceleration in the lower extremity compared to males during running. Females will experience greater overall shock attenuation compared to males during running.	Rejected Rejected	No significant interactions were detected between males and females for all acceleration or shock attenuation measures.

5.1 Footwear

The shoes selected for this study were intended to represent the range of midsole stiffness typically found in recreational running shoes. However, a surprising lack of difference was observed between the PGS and RCT. After averaging across men's and women's shoes, the RCT was the stiffest shoe measured during this protocol at 101.78 N/mm, which differed only by 0.74 N/mm from the PGS at 101.04 N/mm. The ZMX was the least stiff at 86.16 N/mm. A noteworthy consideration is due here though as these measures are conditional on best estimating the line of best fit to the loading curve. Thus, peak deformation must also be examined, particularly during a force-controlled test, which points to the midsole of the PGS being the stiffest, or most resistant to deformation under an applied load.

Regardless, these stiffness values fall within the range of midsole stiffness of commercially available shoes (Shorten & Mientjes, 2011), but are lower than that of some previous work. Using the same approach, Worobets et al. (2014) recorded stiffness values of 129.7-186.1 N/mm across their conventional running shoes which had ethylene vinyl acetate (EVA) and expanded thermoplastic polyurethane (TPU) midsoles, while Hoogkamer et al. (2018) tested shoes with an embedded carbon fiber plate in a polyether block amide foam midsole, and produced a deformation of 11.9 mm under 2000 N, approximating to 168.1 N/mm. Again, these comparisons must be interpreted with caution, as it is unclear how the researchers determined their line of best fit for calculating linear equivalent stiffness. As evident in the force deformation curves of Figure 4.1 and Figure 5.1, the presence of a toe-region was more pronounced in the present study, which points to the elastic deformation of the material used in these midsoles. The line of best fit for this data was calculated for the linear region as determined by the region following the first abrupt change in slope. It is possible that both Worobets' (2014) and Hoogkamer's (2018) groups performed their linear fits to the entire loading curve, which could lead to the overestimation of linear deformation and therefore higher stiffnesses. Despite the presence of the carbon fiber plate, the shoes used by Hoogkamer et al. (2018) have been marketed to be made with softer foam that provides greater energy return, so it was unexpected to note that their shoes were both stiffer (168.1 N/mm) and returned less energy (87.0%) than the shoes used in this study (stiffness ≤ 101.78 N/mm; energy returned ≤ 88.77%, calculated as $100 - \text{hysteresis}$).

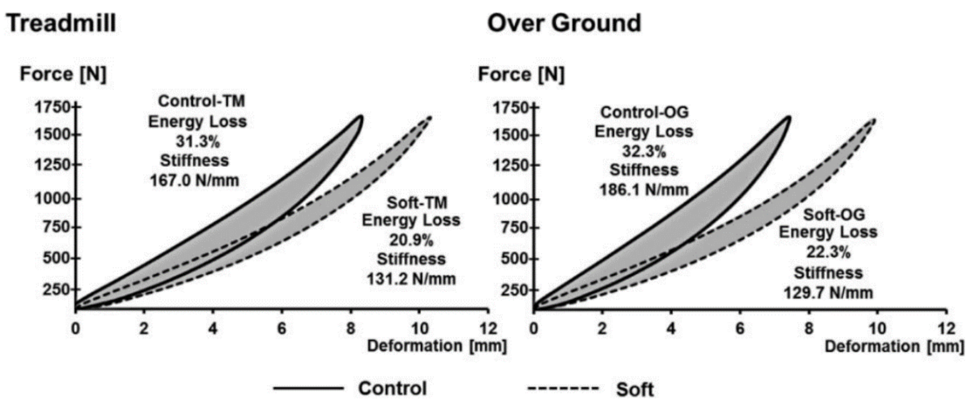

***Figure 5.1**: Mean force-deformation curves obtained from the 20th cycles of three mechanical testing sessions for soft and control shoes, adopted from* (Worobets et al., 2014).

On the other hand, using the ASTM standard impact test, Shorten and Mientjes (2011) calculated linear equivalent stiffnesses of 55-152 N/mm for three conventional running shoes with EVA midsoles. Similarly, Theisen and colleagues (2014) saw stiffness values of 51.1 ± 4.0 and 58.7 ± 2.8 N/mm for their hard and soft shoes respectively, and Malisoux et al. (2020) compared shoes with heel stiffnesses of 61.3 ± 2.7 and 94.9 ± 5.9 N/mm. Thus, the spread of the midsole stiffnesses included in this study fall within the range of stiffnesses produced across all quantification methods. These stiffness differences have also been previously associated with changes in force metrics (Malisoux, Delattre, Meyer, et al., 2020; Shorten & Mientjes, 2011) and running economy (Hoogkamer et al., 2018; Worobets et al., 2014). For instance, a midsole stiffness difference of 33.6 N/mm resulted in a significant difference in impact peak force of 0.09 times body weight

(Malisoux et al., 2020a). Similarly, the work by Worobets et al. (2014) showed that a shoe with a midsole stiffness of 35.8 N/mm less than its comparison significantly reduced oxygen consumption during running by 0.4 mL/kg/min. The stiffness range included in this study is 15.6 N/mm, which is narrower than the range in the aforementioned studies and thus may not be sufficiently different to elicit changes in muscle activity, joint angles, and shock attenuation. However, this also reflects the range in midsole stiffness that are the most common across shoes used by recreational runners.

Specifically, the footwear used in this study can be classified as "conventional" shoes – shoes that do not meet the criteria for a "super shoe" and used specifically for racing. Such "super" racing shoes are characterized by the presence of a full-length carbon fiber plate embedded in the midsole, have a high stack height, and often contain a lightweight and highly compliant and resilient midsole foam (Hébert-Losier & Pamment, 2023; Joubert & Jones, 2022) as exemplified in the shoe tested by Hoogkamer et al. (2018). As these shoes are placed at high price points and less durable than shoes with firmer midsoles, they are typically reserved for elite athletes and unlikely to be worn during typical training runs.

The differences in geometry and other design features between shoe conditions should also be addressed. Namely, the shoes used in this study varied slightly in heel, forefoot, and heel-to-toe drop heights (Table 3.2), which could contribute towards differences in outcome measures. However, no previous difference has been observed in kinematics and foot strike patterns across shoes varying in midsole thickness from 0 to 16 mm (Chambon et al., 2014), nor in spatiotemporal parameters between shoes with thicknesses of 5 to 29 mm (Law et al., 2019). Widths of the heel and toe boxes could prompt different running patterns, however effects are typically demonstrated in the frontal and transverse planes such as increased pronation or ankle eversion (Hannigan &

Pollard, 2020), and therefore should have limited influence on the dependent variables evaluated in the present study.

Overall, the midsoles and conventional shoes used in this study were intentionally and effectively selected from commercially available models to allow the results to be generalized to the recreational runner.

5.2 Sagittal Posture

One of the primary strategies of shock attenuation is to adjust joint angles in response to changes in environmental conditions such as surface irregularities and uncertainties in order to maintain total body stiffness and reduce potential injuries from high impact forces. Stiffer surfaces and harder midsoles have been shown to warrant increased knee flexion angles (Derrick, 2004; Hardin et al., 2004) and exaggerated lumbar lordosis (Castillo & Lieberman, 2018; Delgado et al., 2013). Therefore, one initial research question that this study sought to answer was whether midsole cushioning would change low back and lower limb running kinematics. It was hypothesized that running in softer and more compliant midsoles would result in less knee flexion and less lumbar spine flexion, particularly during initial contact in running.

Several measures were collected to evaluate this hypothesis: sagittal joint angles at initial contact, mean and peak flexion and extension angles during stance phase, and the overall range of motion during stance phase.

Taken together, the results of this study partially supported this hypothesis. Greater knee extension was demonstrated at initial contact when wearing the ZMX, which had the lowest stiffness, compared to the RCT, albeit only in female runners. However, no differences were found across all shoes for males, nor were differences were observed in lumbar flexion angles at initial contact both between sexes and across all midsole conditions.

Furthermore, no differences were observed between shoe conditions for ankle and hip flexion during initial contact, nor in the overall range of motion for all joints. However, running in the RCT did result in greater peak ankle plantarflexion, while the greatest peak and average dorsiflexion was demonstrated in the PGS, despite no significant differences between the midsole cushioning stiffness of these two shoes. Thus, it appears that only local differences in the distal limb were detected during stance phase, while neither the hip nor lumbar spine were sensitive to changes in footwear.

There are several potential explanations for these findings, including the observed sex interaction.

First, the midsole stiffnesses of the PGS and RCT were not significantly different, so any effect of midsole cushioning on the musculoskeletal system may only be observable in comparisons involving the ZMX. While differences were found between the PGS and RCT regarding peak and mean ankle dorsiflexion, it is possible that these findings may be attributed to differences in the general construction of the shoe rather than differences in the midsole. The heel cushioning of the PGS extends further posteriorly compared to the RCT and ZMX, and while precautions were taken to ensure angle definitions were consistent across conditions, these slight variances could affect the runners' ankle postures, particularly during heel strike. However, Chambon et al. (2014) investigated kinematics and foot strike patterns across shoes of the same model and construction but midsole thickness varying from 0 to 16 mm and did not observe any significant differences across the shod conditions. This suggests that shoe thickness and these differences in construction should not drastically affect running kinematics.

Even if the midsole stiffnesses were sufficiently different, it is possible that no observable difference would be detected in all sagittal postures. When comparing sandals with materials that differed by over tenfold in compressive stiffness (calculated as a function of the area of force application, sandal thickness, and Young's modulus of the material), Holowka et al. (2022) did not observe any significant differences in longitudinal arch stiffness, translating to no significant differences in total leg stiffness. As their sandals had a midsole thickness of 1.2 cm, they concluded that shoes with even thicker midsoles, or traditional running shoes, are likely too stiff to produce significant changes in leg stiffness (Holowka et al., 2022). In addition, since the magnitude of low back flexion varies according to the degree of leg stiffness to maintain the total system stiffness (Delgado et al., 2013; Hamill et al., 2009), no changes in leg stiffness would imply that no changes would be required or observed at the low back. In other words, discernible differences in low back postures would depend first on differences occurring in leg stiffness components, but the available range of midsole stiffness materials used for standard running shoes needs to first differ by more than what is commercially available in order to produce such effects.

With respect to the differences in across sex, it has been well reported that male and females exhibit different knee kinematics, such as greater knee extensor moments in females (Sinclair & Selfe, 2015; Stearns et al., 2013). Weaker muscular strength, such as poor hamstring-to-quadricep ratios (Myer et al., 2009) and lack of neuromuscular control during dynamic activities (Mizuno et al., 2001; Stefanik et al., 2011) have been thought to contribute to greater knee extension in female runners particularly to help absorb landing impacts (Ford et al., 2011; Stearns et al., 2013). Thus, the results from this study are supported by previous literature, where heightened knee extension is needed in response to midsole compliance and exaggerated by the sex factor. Similar findings have also been recorded when comparing knee joint stiffness while running in shoes of differing midsole hardness, where only female participants appeared to be sensitive to softer midsoles (Baltich et al., 2015). Hence, the softer and more compliant midsoles used in this study may have been within the range that demanded greater knee joint stiffness, and thus greater knee extension, for females but not males. These differences might also only be noticeable at initial contact since that is the instance which requires the greatest adaptation and shock attenuation from the knee joint (Derrick, 2004; Edwards et al., 2012).

These results both align and contrast with findings from Nigg et al. (2012). When applying a principal component analysis on shoe midsole hardness, sex, and age, they concluded that midsole hardness affected individual movements similarly regardless of sex and age. Sagittal plane movements were the most responsive compared to other running movements, but sex only influenced the less dominant movements (Nigg et al., 2012). As it was not specified in the study, it is possible that knee flexion during initial contact is one of the less dominant movements that is sensitive to sex.

However, previous studies seem to agree with the shoe effects on knee kinematics observed in this study. Hardin et al. (2004) tested different treadmill surface stiffnesses and midsole hardnesses and reported significant differences at the hip and knee across surfaces, but only the ankle changed in response to midsole hardness. Chambon et al. (2014) did not see differences in hip or knee flexion angles at touchdown when comparing midsoles of varying thickness, though they used overground running for this experiment. Lastly, Bishop et al. (2006) also did not observe significance differences in knee angles at touchdown across shoes that differed in market cost and material stiffness. While these studies did not employ sex comparisons, these results appear to suggest that knee contact angles may not be sensitive to changes in footwear, so the results from this study may be driven by differences in kinematics between sexes.

With regards to the second component of this hypothesis, previous studies have shown that the lumbar spine flexes slightly during touchdown in response to the impact, then extends by midstance (MacWilliams et al., 2014; Schache et al., 2002; J. Seay et al., 2008). This flexion movement, in tandem with the amplitude of lumbar lordosis, has been proposed to help attenuate shock during running depending on the degree of attenuation required (Castillo & Lieberman, 2018). During a four-week training program in minimal footwear, Lee and colleagues (2018) showed that runners gradually adopted a more extended lumbar posture over time, likely to compensate for the stiffer interface that they ran on.

Thus, it is possible the lack of differences in this study in sagittal lumbar posture during initial contact and across stance phase could be related to the latency in the low back response. It could be that minor differences may only occur following changes in the ankle, knee, and hip joints, and given that Lee et al. (2018) only saw a change in total range of motion of approximately 1 degree after 4 weeks, the 5-minute protocol used in this study may not be sufficient to elicit longer changes in lumbar posture. However, it is important to note that many runners in Lee et al.'s (2018) study adopted forefoot strike landing patterns when training in minimal shoes, and Delgado et al. (2013) has shown that lumbar range of motion may differ with FFS compared to RFS running due to shorter stride lengths leading to pelvic positional changes.

On the other hand, the same authors found no difference in mean lumbar flexion or extension, so it is possible that no true difference exists in lumbar posture across footwear. In alignment with the findings from Delgado et al. (2013), differences in the distal joints suggest that the lower extremities may sufficiently accommodate surface interface changes to allow for similar lumbar lordosis levels across all modes of running. When comparing lumbar lordosis across participants during running and walking, Castillo and Lieberman (2018) concluded that individual differences in lumbar curvature contributed to differences in SA. Thus, the variability in natural lordosis shape across individuals in this study may already be greater than the variability in lumbar posture in response to midsole stiffness.

Interestingly, Schache et al. (2001) also observed that the lumbar spine was slightly more extended at initial contact during overground running compared to treadmill running. While the authors warned that such differences could be due to differing methods for identifying initial contact with overground (via force platforms) and treadmill running (via kinematic events), these results agree with the general consensus that treadmill running may provide a softer surface, thus warranting increased extension among all joints to maintain the total system stiffness. This might also explain how the angles in this study compare to previous work.

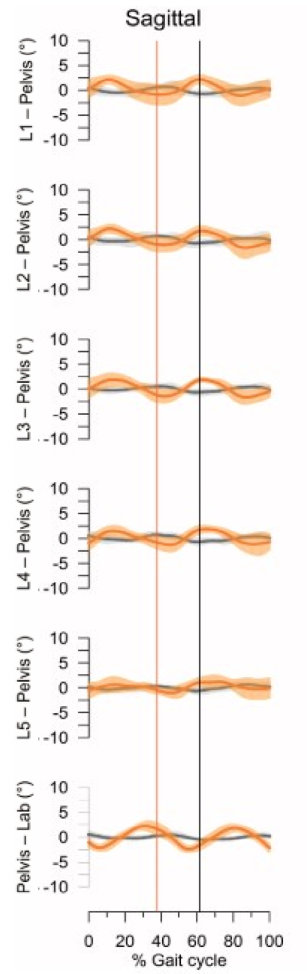

***Figure 5.2**: Running (orange) and walking (gray) angles of each lumbar vertebra relative to the pelvis, adopted from MacWilliams et al. (2014). Error bands represent ± 1 SD from mean. Positive angles denote forward flexion. Vertical lines denote foot off (38% run, 61% walk), with gait cycle defined as right foot initial contact to next right initial contact.*

A note of caution is due here since the placement of the marker cluster used to track the lumbar spine in this study differs from most experiments collecting lumbar or trunk kinematics. Due to limitations with the location of the L1 accelerometer, the lumbar cluster was placed at L3, whereas most studies will examine motion via L1 relative to the pelvis to capture an estimate of the sum of all lumbar intervertebral movements. Subsequent analyses were conducted without accelerometry to determine the expected difference between angles calculated from L1 and L3, and revealed that angles in this study may be up to 6 degrees less than that measured at L1. However, using indwelling bone pin markers to track direct bone motion without soft tissue artefact, MacWilliams et al. (2014) concluded that the sagittal motion of the superior lumbar vertebrae (L1 to L3) did not differ significantly between walking and running at 2.7 ± 0.3 m/s, nor were major differences observed graphically between angles measured at L1 or L3 relative to the pelvis (MacWilliams et al., 2014; Figure 5.2). This led the authors to conclude that the inferior vertebrae must provide most of the shock attenuation in the lumbar spine and offers justification for isolating the movement of the inferior lumbar vertebrae.

Though there is some variance in previous literature (Table 5.2) , with studies reporting lumbar (L1) ranges from 4.50 ± 7.38° (MacWilliams et al., 2014) to 6.59 ± 2.1° (J. Seay et al., 2008), means of 13.3° (Schache et al., 2002), or L3 ranges of 13.3 ± 2.4° (S. P. Lee et al., 2018), the total ROM observed in this study seems to be consistent at 8.55 ± 3.3°. Any main differences are likely due to most previous work reporting ranges over the entire gait cycle compared to only stance phase kinematics reported here, or differences in running speed and stride length (J. Seay et al., 2008).

Table 5.2*: Comparison of lumbar kinematics between the current study and previous literature. Values are collapsed across sexes and all shoe conditions for the current study and taken from control groups for other studies. Values are presented in Mean ± SD.*

	Lumbar flexion at initial contact (°)	Mean lumbar flexion during stance phase (°)	Max lumbar flexion during stance phase (°)	Max lumbar extension during stance phase (°)	Lumbar range of motion during stance phase (°)
Current study, collapsed across all factors and levels	-4.50 ± 7.38	-4.77 ± 7.82	8.72 + 8.43	0.15 ± 7.48	8.55 ± 3.26
(Schache et al., 2001)	(L) -14.9 ± 7.8 (R) -16.9 ± 8.2	-	-	-	(L) 33.9 (R) 35.9
(Schache et al., 2002)	-20.7 ± 6.6	-22.9 ± 6.2 *	-	-	13.3 ± 3.8 *

(J. F. Seay et al., 2011) †	(T12-L1) -15.8 ± 9.8 (L5-S1) 1.1 ± 4.4	-	-	-	(L5-S1) 6.59 ± 2.1
(Delgado et al., 2013) ‡	-	-	6.8 ± 6.1 *	15.2 ± 6.7 *	22.1 ± 5.1 *
(MacWilliams et al., 2014) §	-	-	2.7 ± 0.6	-2.0 ± 1.8	4.7 ± 1.9
(S. P. Lee et al., 2018) ‡	-	1.9 ± 15.3	8.6 ± 15.7	4.8 ± 14.3	13.3 ± 2.4

* across entire gait cycle; † angles expressed relative to the proximal segment (T12-L1 relative to thoracic segment; L5-S1 relative to lumbar spine); ‡ based on electrogoniometer placed at L2 (Delgado et al., 2013) or L3 (S. P. Lee et al., 2018); § based on indwelling bone pins

The differences seen across peak and average ankle plantar and dorsiflexion supports existing research. The primary finding that the shoe with the stiffest midsole elicited greater peak and average ankle dorsiflexion is not surprising given that previous work has shown that increased ankle dorsiflexion assists with shock attenuation (J. J. Chu & Caldwell, 2004), particularly with harder shoes (Baltich et al., 2015; Hardin et al., 2004). However, it was interesting that differences were not observed at initial contact, but later in the stance phase. This could be attributed to the time it takes for the ankle joint to accommodate to the shoe stiffness and for the body weight to be transferred fully onto the stance leg, as the amplitude of dorsiflexion adjusts during early stance phase depending on speed and surface conditions (Novacheck, 1997). While this contradicts some previous research that showed greater dorsiflexion in maximal shoes compared to minimal cushioning, the authors also concluded that this trend may be due to changes in foot strike pattern or the rocker construction of shoes used in that study (Hannigan & Pollard, 2020).

Overall, it appears that the results from this study are mostly in agreement with previous work. Knee and ankle kinematic differences can be observed across midsole conditions, but the hip and low back are less or not sensitive to stiffness changes, at least in the stiffness range examined. These findings may have some implications regarding the role of footwear on the lumbar spine;

there is little effect of midsole cushioning stiffness on low back posture, but distal joints appear to adjust in response to even slight differences in cushioning stiffness. Whether these effects have an influence on shock attenuation will be discussed further in Section 5.4.

5.3 Muscle Activation

A second research question that was investigated was whether trunk and low back muscle activation changed with midsole stiffness during running. Several main outcome measures were evaluated: mean muscle activation, normalized to the participant's maximal voluntary isometric contraction; the co-activation coefficient (CCI) over stance phase, which was interpolated to 100 time points; and the correlation and timing between muscle group activation patterns to identify timing relationships.

Contrary to initial expectations, this study did not show any differences in mean activation levels across shoe conditions when examined via pairwise comparisons. However, given the lack of significant differences in lumbar kinematics, these results were not surprising. Previous work has shown that lumbar extensor activation peaks during forward trunk flexion and decreases with extension (Kienbacher et al., 2015; Thorstensson et al., 1982), and as no changes were observed in lumbar posture across the differing shoe conditions, differences in mean activation levels were not likely to be exhibited.

This contradicts previous work on muscle tuning, where there is evidence that muscle activation levels adjust to compensate for shoe hardness to minimize soft tissue vibrations during running (Boyer & Nigg, 2007; Nigg & Liu, 1999; Nigg & Wakeling, 2001). However, it is possible that these differences may only be detectible either during the initial exposure (i.e. within the first minute) to the new shoe condition, or after prolonged exposure where the body has been provided ample time for adaptation. Muscle tuning refers to the fine adjustment of activation levels in response to the amplitude and frequency of the input signal and is specific to the input conditions detected by the central nervous system during the previous initial ground contact (Nigg & Wakeling, 2001). Since the conditions of each step is unlikely to vary much following three minutes in a five-minute treadmill training session, differences in trunk muscle activation may not truly exist in non-fatigued running.

In studies that did show changes in muscle activity across footwear, the effects were only observed after four weeks of training (S. P. Lee et al., 2018; Wakeling et al., 2002), and even then the reduction in lower paraspinal muscle activity was not statistically significant (S. P. Lee et al., 2018). Alternatively, Ogon and colleagues (2001) showed differences in lumbar erector spinae activation latency, but only across shod and barefoot conditions, which differ drastically in multiple aspects. Wang et al. (2017) compared lower extremity muscle activation in basketball shoes of differing midsole cushioning during the landing phase of box jumps, but also did not find differences. Lastly, two studies showed differences in trunk activation levels during a single standing session, but their interventions involved unstable shoes which were characterized by a rounded sole to promote instability in the anterior-posterior direction, and thus challenge the trunk musculature to activate even while standing (Buchecker et al., 2013; Lisón et al., 2016).

Regardless, the activation levels recorded in this study seemed to align with previous work, though caution should be exercised in comparing the values averaged over stance phase in this study with full gait cycles in other work. Nonetheless, it can only be presumed that the LES, EO, and RA are simply not sensitive to differences in shoe cushioning, and muscle tuning in the trunk does not differ when the same foot strike pattern is adopted and all running is performed under shod conditions.

Table 5.3: *Comparison of trunk muscle activity between the current study and previous literature. Values are collapsed across sexes and all shoe conditions for the current study and taken from control groups for other studies. Values are presented in Mean ± SD.*

	LES Activation (%MVC)	RA Activation (%MVC)	EO Activation (%MVC)
Current study, collapsed across all factors and levels	14.7 ± 15.6	10.5 ± 4.4	7.9 ± 6.5
(Behm et al., 2009) *†	34.1 ± 36.2	-	21.4 ± 16.4
(Lisón et al., 2016) ‡	-	2.7 ± 3.2	5.1 ± 6.9
(S. P. Lee et al., 2018)	47.0 ± 34.0	-	-
(Callaghan & McGill, 2001) *‡	8 ± 3	7 ± 3	6 ± 4

* across entire gait cycle; † during 60% heart rate reserve run (corresponding to 2.31-3.61 m/s); ‡ during walking (fast cadence for Callaghan & McGill, 2001)

Another observation was that there was a sex effect for external oblique activation and a trend approaching significance for both lumbar erector spinae and rectus abdominus activation. This finding is likely explained by sex differences in kinematics, where females exhibit greater frontal and transverse plane motion than males in running due to anatomical differences such as wider pelvises and larger Q angles (Ferber et al., 2003; Nigg et al., 2012; Phinyomark et al., 2014), and EO and RA both assist with the axial rotation of the trunk. The greater LES activation among males also aligns with this rationale since males demonstrate greater sagittal plane motion and lumbar extensors help to oppose sagittal trunk flexion. Previous work has also reported that males likely adopt different abdominal activation strategies during landing, where transverse abdominus and internal oblique activation – which were not included in this study due to lesser roles in shock attenuation – were significantly greater among males during box jump landings (Kulas et al., 2006). Thus, the trends in muscle activation between sex seen in this study support the existing literature on how males and females differ in running.

5.3.1 Muscle Co-Activation

Co-activation allows for increased joint stiffness during the landing and support phase (Bobbert et al., 1992) and helps improve joint stability particularly when the input conditions are compromised (Hubley-Kozey et al., 2008), such as in the case of highly compliant or soft midsoles. However, co-activation in the trunk has also been identified as a predisposing factor for low back pain during standing (Ghamkhar & Kahlaee, 2015; Nelson-Wong & Callaghan, 2010). Thus, a secondary investigation in this study sought to determine whether running in shoes with different midsole cushioning warranted differing levels of co-activation across the trunk and spinal musculature to determine whether midsole stiffness might have any implication on low back pain and loading.

Contrary to expectations, no differences were detected across the different shoes regarding the co-activation coefficient, nor the magnitude of correlation and sequential timing between activation patterns. Furthermore, only moderate correlations were observed, with the highest correlation being between the RA and EO (r = 0.391 ± 0.15, across all conditions). This was surprising as both these muscles perform similar functions and peak in activity around foot strike, but agrees with previous work that did not find any discernible patterns in both RA and EO activity during gait (Callaghan et al., 1999; White & McNair, 2002).

These overall results could be explained simply by the likely fact that no true difference exists. Since no changes were detected in mean activation levels across shoe conditions, it is unlikely that the timing and onset of activation would change either. Put simply, it is possible that trunk stiffness may not be a priority, especially if stability is not compromised. Changes in trunk co-activation during running have only been recorded due to changes in age and impairment, where the cost of locomotion and lack of stability are heightened (H. J. Lee et al., 2017) When comparing co-activation of the erector spinae and rectus abdominus while walking at 0.4 to 1.5 m/s, no differences were found (van der Hulst, Vollenbroek-Hutten, Rietman, & Hermens, 2010). Studies have shown that the abdominal muscles activate just prior to landing during typical running (Cappozzo, 1983; Cromwell et al., 1989), but when running – which was considered an instability-inducing exercise – was compared to prone callisthenic exercises such as curl-ups, no difference was detected in EO activation, suggesting that this muscle group was not actually sensitive or responsive to increases in instability (Behm et al., 2009). In the lower limb, Apps et al. (2016) reported greater co-activation in the gastrocnemius medialis and tibialis anterior (TA) when walking in unstable or randomly perturbing midsoled shoes. However, another study examining the gastrocnemius lateralis and TA when running in minimally and traditionally cushioned reported no differences in pre- and co-activation (Udin et al., 2023). Thus, while co-activation is a mechanism that can be utilized by the musculoskeletal system for increasing stiffness, the low back and abdominal musculature are either not compromised or sensitive to differences in footwear cushioning, or generally reduced in favour of elastic energy absorption during landing and propulsion at the end of stance phase during higher velocity locomotion (Moore et al., 2014; Tam et al., 2017).

An important warning is due here for making comparisons, as there are numerous methods of reporting co-activation in the literature, and each equation may not provide identical information. Lee and colleagues (2018), Tam's group (2017), and Apps et al. (2016) all used a unitless co-activation index that reported the magnitude of simultaneous activation between antagonistic muscles, where certain values denoted equal activation, greater relative agonist activity, or greater activity of the antagonist muscle. Udin et al.'s (2023) coactivation was defined as the time when both muscles activity levels were over 5 standard deviations from the resting EMG baseline, and expressed as a percentage of stride duration. However, all these methods specify that a muscle must be either an agonist or antagonist, which limits the interpretations permitted for synergistic activation, and are typically dependent on the length of the summation window, where more co-activity whether due to greater activation levels or longer intervals can result in the same increased co-activation index (Viggiani et al., 2018). The CCI used here is also dependent on the summation window but allows for standardized comparison as it was time-normalized to 100 data points during stance phase. Expressing the CCI relative to the subjects' maximum voluntary contraction also allows for normalized comparisons by providing a magnitude which can be directly compared to the activation level of a single muscle but restricts comparisons against CCIs expressed as a ratio.

The cross-correlation method presented by Nelson-Wong et al. (2009) used in this study is more unique as it provides timing information. This method not only provides the phase lags between activity of the relevant muscles but is also not dependent on the summation window nor activation level. This allows for both outcome variables to be directly compared to previous work regarding trunk activation sequences in patients with LBP during standing, or healthy adults in walking, though it is dependent on the specific muscle groups collected. Nelson-Wong et al. (2010) observed a bottom-up activation strategy where the trunk muscles were activated following glute medius activation that may be an identifying factor for LBP development in standing, and Prince and colleagues (1994) showed phase lags around -5 ms in L2 to L4 paraspinal muscles that suggest a top-down strategy for upper body balance. While no difference was observed between sex or shoe conditions in this study, the primarily negative phase lags for LES-RA and LES-EO suggest that the lumbar erector spinae may be driving any co-activation patterns observed in the trunk, which might have important implications on how the low back assists in shock attenuation (elaborated on in Section 5.4).

5.4 Shock Attenuation

The primary research questions this study sought to answer was whether shock acceleration and shock attenuation during running changed with midsole stiffness. Peak resultant acceleration magnitudes and the ratio between the inferior to superior sensors were compared across sex and shoe conditions.

The results from this study were surprising: for peak acceleration, only differences in peak head magnitudes were observed, but post-hoc analyses did not reveal which means were driving the significance. It appeared that running in the RCT and ZMX resulted in higher peak head acceleration magnitudes. The head acceleration was included as a measure of total outgoing shock, or the remainder of the impact that is not attenuated by any mechanisms. No differences were observed between the input acceleration magnitudes at the tibia; therefore this points to some differences occurring somewhere between the input and outputs, which may speak to the shock attenuation mechanisms at play (discussed later). However, while the acceleration values observed here align with previous recordings of head shock under simulated impact (Lafortune et al., 1996) or typical treadmill running (S. P. Lee et al., 2018; TenBroek et al., 2014), this overall finding that no differences existed elsewhere contradicts with existing work on midsole thickness and softness.

There has mostly been agreement on the fact that softer, more compliant, or more cushioned midsoles result in lower distal tibial acceleration magnitudes (O'Leary et al., 2008; TenBroek et al., 2014; Xiang et al., 2022; S. Zhang et al., 2005). Studies that did not find a difference or reported higher tibial acceleration peaks with cushioned shoes mostly attributed their results to changes in foot strike pattern which superseded any effects of footwear (Chambon et al., 2014; Mo et al., 2021; Sinclair, 2017). However, all shoes resulted in a rearfoot strike pattern in this study, so it is uncertain why no differences were observed at the distal tibia or in the lumbar spine.

In the same way, no differences in shock attenuation were observed between all sensor pairs across shoes. While greater SA was observed in the PGS and RCT compared to the ZMX between the tibia and the sensor at L5, the lack of statistical significance results in rejecting the hypothesis that softer shoes would result in decreased SA but provides some evidence in support of it. These results were consistent with one previous study: TenBroek et al. (2014) showed that changing midsole thickness did not result in changes to full body (tibia to head) SA. However, Zhang et al. (2005) found less SA between the tibia and the head when wearing basketball shoes with softer midsoles, and Sinclair (2017) reported greater SA in maximalist shoes, although their comparisons involved changes in foot strike pattern between different shoe conditions. Interestingly, Xiang's group (2022) examined SA between the distal and proximal tibia while running in minimalist compared to maximalist shoes and observed that there were no differences in low frequency (3-8 Hz) shock attenuation, but higher frequency (9-20 Hz) SA was greater in maximalist shoes. Thus, *a posteriori* analyses were conducted to examine if differences existed at these smaller frequency ranges (Appendix C).

Results revealed significant differences between shoes in SA between the tibia and L5, and between the tibia and the head in the low frequency (3-8 Hz) but not high frequency (9-20 Hz) range. For both pairings, the difference was driven by greater SA in the RCT compared to the ZMX. As low frequencies represent voluntary motion of the lower extremity, therefore hinting to the active mechanisms of SA (Gruber et al., 2014; Shorten & Winslow, 1992), it could be suggested that the role played by active mechanisms between the proximal tibia and the L5 differ across footwear. This is consistent with previous work and the results from this study on knee joint posture, where greater knee extension – corresponding to greater leg stiffness and therefore associated with lower SA (Bishop et al., 2006; Kulmala et al., 2018) – was demonstrated in the most compliant shoe (ZMX). Though not investigated in this study, eccentric contraction around the knee may also differ with more compliant shoes (S. Zhang et al., 2008). These results conflict with that of Xiang et al. (2022), who saw no differences in low frequency SA across shoe conditions. However, they corresponded high frequency SA to impact experienced at the ankle, low frequency SA to changes at the proximal tibia, and only examined acceleration up to that point, which restricts their conclusions to SA mechanisms below the knee (2022). No significant differences in the 9-12 Hz range also points to the lack of difference in the role played by the passive mechanisms across the footwear conditions. It could be speculated that structures such as the menisci or soft tissue likely perform the same function regardless of stiffness. Ultimately, these results add to the evidence that softer and more compliant midsoles decrease the need for active SA in the lumbar spine as it is performed and compensated by active attenuation in the lower limb. While passive SA may occur, it is unlikely to vary across shoes of differing midsole stiffness. This provides further support of the initial hypothesis, but caution must be exercised to not use such evidence to reject the null hypothesis as these analyses were conducted *a posteriori*.

Intriguingly, a signal gain was observed at low frequencies (3-8 Hz) between the L5 and the L1, but attenuation occurred between 9 and 20 Hz. This could be assumed to be from the voluntary motion of the lumbar spine, as voluntary motion is typically less than 10 Hz, and provides further evidence of lumbar flexion in response to initial contact. Attenuation in the higher frequencies may indicate that passive structures of the lumbar spine are contributing most to the general SA, since passive mechanisms have been shown to respond better to high frequency shock (Gruber et al., 2014). This agrees with the conclusions of Castillo and Lieberman (2018), who conducted a partial regression to determine which factors out of intervertebral disc height, standing lordosis, and running lordosis were the strongest predictors of SA, and observed that the sagittal shape of the spine was a strong predictor of SA. Taken together, this may have future implications on the focus of lumbar posture for mitigating impact.

A potential explanation that may address all accelerometry results involves the various differences in sensor placement and methods between this study and others. The distal tibia has been widely used a surrogate measure of impact shock that is not influenced by ankle motion (Hennig & Lafortune, 1991), but the selection in head accelerometer placement has varied from traditional bite bars (Lafortune et al., 1995; Light et al., 1979) to different locations on the head. The decision to use the posterior aspect of the head instead of the central anterior aspect of the forehead may have contributed to differences between the findings from this study and those of Zhang et al. (2005) or TenBroek et al. (2014). However, studies have also reported that most accelerations from ground contact – whether during running or drop landings – are attenuated before they reach the torso (McErlain-Naylor et al., 2021; Shorten & Winslow, 1992), 2021; Shorten & Winslow, 1992), and therefore this difference should not have drastic implications.

The use of resultant versus uniaxial measures may also contribute to inconsistencies in what is reported in the literature versus the magnitudes reported in this study. Resultant accelerations have predominantly been used, but some studies which have reported differences across footwear employed uniaxial sensors at some or all locations (Bruce et al., 2019; Dufek et al., 2009) or did not specify their methods (Sinclair, 2017). Uniaxial measurements have the limitation that sensor alignment is critical and any misplacement may drastically affect the resulting amplitudes, while resultant measures provide greater redundancy to counter this effect.

Another plausible rationale may be that the shoes are simply not different enough to elicit differences in acceleration magnitude nor shock attenuation. One study comparing acceleration across barefoot and shod conditions of differing midsole thickness showed no difference between all shod conditions (Ogon et al., 2001). Sinclair (2017) saw differences in shock attenuation between minimalist and maximalist shoes, but their minimalist shoes resembled barefoot running with only a single rubber layer under the sole which led to major changes across foot strike and stride characteristics. The shoes used in this study could all be characterized as conventional footwear with a midsole cushion, and while this was intended to allow any findings to be generalized to the population of typical recreational runners, the shoes and cushioning stiffness may have been too similar to elicit differences in shock attenuation mechanisms.

Table 5.4*: Comparison of acceleration magnitudes and shock attenuation between the current study and previous literature across footwear conditions. Values are presented in Mean ± SD.*

	Distal Tibia Acceleration (*g*)	Proximal Sensor Acceleration (*g*)		Shock Attenuation (dB)	
		L5	***Head***	***Tibia-L5***	***Tibia-Head***
Current study, collapsed across sex	(PGS) 9.36 ± 3.24 (RCT) 9.90 ± 3.56 (ZMX) 9.06 ± 3.10	(PGS) 5.06 ± 2.26 (RCT) 5.19 ± 2.28 (ZMX) 5.31 ± 2.36	(PGS) 2.46 ± 1.12 (RCT) 2.70 ± 1.24 (ZMX) 2.65 ± 1.50	(PGS) -5.37 ± 2.88 (RCT) -5.15 ± 3.00 (ZMX) -4.44 ± 2.78	(PGS) -9.32 ± 2.78 (RCT) -8.84 ± 2.77 (ZMX) -8.99 ± 3.12
		Head			
(TenBroek et al., 2014)	(THIN) 6.04 ± 1.10 (MED) 5.895 ± 1.10 (THICK) 5.72 ± 1.10	(THIN) 1.36 ± 0.31 (MED) 1.29 ± 0.31 (THICK) 1.25 ± 0.31		(THIN) -9.19 ± 2.67 (MED) -9.44 ± 2.67 (THICK) -9.49 ± 2.67	
		Sacral			
(Sinclair, 2017)*	(MIN) 4.55 ± 0.84 (CON) 5.29 ± 1.02 (MAX) 6.09 ± 1.23	(MIN) 3.15 ± 0.82 (CON) 2.98 ± 0.71 (MAX) 3.29 ± 0.87		(MIN) 31.04 ± 13.44 (CON) 44.40 ± 16.21 (MAX) 45.07 ± 19.22	
		Proximal Tibia			
(Xiang et al., 2022)	(MIN) 8.52 ± 1.75 (CON) 7.13 ± 1.37 (MAX) 6.58 ± 0.91	(MIN) 5.7 ± 1.35 (CON) 5.32 ± 1.10 (MAX) 5.02 ± 0.90		(MIN) -32.36 ± 21.28 (CON) -28.12 ± 23.12 (MAX) -24.61 ± 23.76	

* shock attenuation calculated as a ratio of inferior to superior acceleration magnitude (%) where 100% represents complete attenuation; THIN/MEDIUM/THICK = midsole thickness conditions used by TenBroek et al., 2014; MIN = minimalist, CON = conventional shoes, MAX = maximalist shoes, denoting footwear conditions used by Sinclair, 2017 and Xiang et al., 2022

5.4.1 Synthesis of Results

There are other explanations that may better address these results when interpreted along with the findings observed in sagittal posture and muscle activation. Firstly, the difference across footwear in knee and ankle kinematics along with the observed trend in tibial-to-L5 SA suggest that the distal joints are responsible for the majority of SA during landing. Greater tibial to L5 SA was observed in the stiffer shoes, along with greater ankle dorsiflexion and knee flexion, thus supporting the idea that increased lower extremity joint flexion is needed to assist with increased shock dissipation when the landing surface is stiffer. This work largely corroborates with other studies investigating Groucho running (McMahon et al., 1987) or other lower limb kinematic strategies for shock attenuation (J. J. Chu & Caldwell, 2004), where the ankle and knee have been deemed "the locus for shock attenuation" (Derrick et al., 1998; Hamill et al., 2009) due to their roles in adjusting leg stiffness under different surface compliance conditions.

However, this effect is not maintained further up the kinetic chain. Xiang et al. (2022) examined acceleration and SA at the distal and proximal tibia and noted that any cushioning function from the shoes is more pronounced at the distal tibia but was already dissipated by the time it reaches the proximal tibia and knee. Likewise, McErlain-Naylor et al. (2021) measured accelerations from the distal tibia to the medial femoral condyle, L5, and C6, and showed no difference from L5 upwards, where resultant peak *g*'s were reduced by an average of 93 ± 4% at L5 and 93 ± 3% at C6. This is consistent with the fact that no differences in peak acceleration magnitudes, SA, posture, or muscle activation were observed in the lumbar spine in this study. It can therefore be speculated that the lumbar spine is not differentially affected by changes in midsole cushioning, so long as the lower limb is effective at attenuating the shock prior to it reaching the torso. These findings contradict that of Ogon et al. (2001), who reported differences in trunk musculature and SA with shod and barefoot conditions, but could again be explained by the presence of a barefoot comparison rather than evaluating across conventional footwear. Regardless, the values from this study align well with the work by Castillo and Lieberman (2018), who showed that lumbar lordosis does assist with SA during running. Using accelerometers affixed to T12/L1 and L5/S1, they observed mean resultant SA values of -0.77 ± 3.07 dB while running at 3.00 ± 0.18 m/s, while mean L5-L1 SA of -0.24 ± 3.14 dB was observed in the current study.

When examined in unison, these findings may have some implications on the relationship between low back loading or pain development and running. Studies have shown that footwear affects the low back (Ogon et al., 2001), and the use of insoles may help mitigate the impact to be attenuated at the level of the lumbar spine (Wosk & Voloshin, 1985). However, the present study provides evidence that different levels of midsole cushioning stiffness, particularly in conventional running shoes, may not drastically affect how the lumbar spine responds. Thus, it can be speculated that pain-free runners should be able to select any pair of traditional running shoe without concern for changes to their back pain status, provided that their foot strike pattern remains consistent. Differences may be observed regarding whole-body shock transmission, where stiffer shoes may result in changes to lower limb sagittal posture and total head acceleration, but the role of the lumbar spine is unlikely to vary. Furthermore, it is not expected that spinal and abdominal muscle activity patterns would differ between cushioning stiffness. This is important for minimizing energy expenditure and increasing performance, but also for not destabilizing the trunk and demanding higher co-activation levels which would increase the loading on the low back (Ghamkhar & Kahlaee, 2015; van der Hulst, Vollenbroek-Hutten, Rietman, & Hermens, 2010). Ultimately, though the trunk does assist with SA via flexion and muscle activation during initial contact, it appears that these muscles do not respond differently to softer, more compliant shoes, so shoe cushioning should have limited interaction with LBP during running.

5.5 Sex Comparisons

The final research questions involved whether differences existed in acceleration magnitudes and shock attenuation between sexes when running in midsole cushioning of different stiffness. Several measures showed differences between sex. Namely, females demonstrated greater knee extension when running in softer shoes, but no differences were observed across the men's knee angles at initial contact. Greater hip range of motion was also observed in females compared to males, driven by greater peak hip extension. Activation of the external obliques were greater in females than males when wearing the stiffest shoe, though trends were also evident in the other shoe conditions, and lastly, mean LES levels were greater among males. However, despite these contrasts, no significant difference was observed in acceleration, nor SA between any sensor pairs between sexes. Hence, the hypotheses that females would exhibit greater lower extremity acceleration and greater overall SA was rejected. This contradicts the conclusions from both Sinclair (2016) and Dufek et al. (2009), who each reported differences in SA across genders. Interestingly though, their findings also disagree, where Sinclair (2016) demonstrated that females had lower SA from the tibia to the sacrum, while Dufek's group (2009) saw higher total body SA values in females, along with a surface effect when comparing soft, medium, and hard treadmill beds.

Taken together, there are several plausible explanations for these findings. The first involves the differences in methodology between this study and previous work. For instance, Sinclair's (2016) participants ran at 4.00 m/s ± 5% along a runway towards a force platform, while Dufek et al.

(2009) had their subjects run at an average pace of 2.66 ± 0.36 m/s. Studies have shown that some differences exist between overground and treadmill running (García-Pérez et al., 2014; Riley et al., 2008), where higher SA might be observed during overground running due to the stiffer surfaces (Hines & Mercer, 2004). Likewise, research comparing spatiotemporal parameters between sexes have shown that females demonstrate different contact and flight times, stride lengths, step frequencies, and step angles, particularly at velocities ≥ 14 km/h (3.89 m/s; García-Pinillos et al., 2020), and it is well known that increased stride frequency and running velocity is associated with higher impacts (Hamill et al., 1995). Thus, females in these studies may be running at a faster pace than preferred, and therefore need to attenuate greater levels of shock compared to males. However, the present study used a 3.3 m/s pace and found no differences between sex, which could potentially suggest the presence of a sex-speed interaction where at some optimal speed, both sexes demonstrate similar SA capabilities, but at higher velocities, males employ greater SA, and vice versa.

Table 5.5*: Comparison of anthropometrics between sex between the current study and previous literature. Values are presented in Mean ± SD (where applicable).*

	Height (m)			**Mass (kg)**		
	Male	**Female**	**Difference**	**Male**	**Female**	**Difference**
Current study	1.79 (0.04)	1.70 (0.09)	0.09	69.18 (6.21)	63.73 (8.91)	5.45
(Sinclair, 2016)	1.77 (0.1)	1.66 (0.1)	0.11	73.2 (6.5)	64.3 (6.4)	8.9
(Dufek et al., 2009)	1.75 (0.72)	1.70 (0.52)	0.05	80.6 (8.0)	67.0 (3.6)	13.6

Sex groups in these previous studies were also less similar in anthropometrics compared to the present study, especially with regards to mass (Table 5.5). This larger mass gap could contribute to greater differences in joint kinetics and shock magnitudes to be attenuated (Nin et al., 2016), leading to more prominent sex differences in SA. The closely matched groups in this study may provide a truer comparison of SA with all other factors being equal.

Another possibility is that different sexes use different SA mechanisms throughout the body, but attenuation values are dependent on the location of the accelerometers and only provide a global indication of the degree of attenuation. The disagreement between the results of Sinclair (2016) and Dufek et al. (2009) could point to females exhibiting reduced SA capacity between the tibia to the sacrum, but overall greater SA when measured at the head. However, no differences between sexes were observed in any of the sensor pairings in this study, implying that any dissimilarity in SA mechanisms are not sufficient to elicit differences in the signals. Regardless, these conclusions still align with the kinematic and muscle activation results: greater knee and hip extension were observed among females, owing to potential poorer SA in the lower limb, but greater abdominal activation, along with potentially higher soft tissue dissipation in the upper body (Challis & Pain, 2008), could provide females with better overall SA.

5.6 Limitations

There are several limitations within this study regarding the methodologies employed.

First, this study used treadmill running rather than overground running, which have been purported to produce different kinematics (Nigg et al., 1995; Sinclair, Richards, et al., 2013) or even provide increased energy return (Winter, 1978) compared to overground methods, which may limit the extrapolation of these findings. However, recent work has shown that most spatiotemporal, kinematic, kinetic, and muscle activity measures were comparable if surface stiffnesses and treadmill or overground running experience were similar, sufficient treadmill motor power was available, and belt speed was consistent (Riley et al., 2008; Schache et al., 2001; Van Hooren et al., 2020). The selection to use a treadmill also allowed for more and consecutive strides to be analyzed due to the limited collection volume required, which is an advantage over the greater variability presented when participants must target a force platform during overground runway protocols (Challis, 2001).

Another limitation is the collection of unilateral data and assumption of symmetry, as well as omitting frontal and transverse plane kinematics. There is a possibility that differences in SA mechanisms are missed due to disregarding these kinematic parameters. However, Nigg et al. (2012) showed that midsole hardness primarily influenced movements in the sagittal plane, and Malisoux's group (2022) reported that no difference was observed in frontal plane ankle kinematics across shoes of differing stiffness As running – specifically, lumbar motion during running – primarily occurs in the sagittal plane, the focus of this study was on shock attenuation through changes in sagittal joint angles. Likewise, studies examining asymmetry indexes in running have not found significant differences in injury rate or kinematic variables such as leg stiffness, concluding that it was acceptable to collect unilateral data, particularly for the purposes of limiting encumbrance to the participant, if the variables with the lowest asymmetry index were the desired outcome measures (Carpes et al., 2010; Pappas et al., 2015). As a further prevention measure, all participants were restricted to having a right-side limb preference.

Regarding the exclusion criteria, this study only recruited runners who adopted rearfoot strike patterns. This decision was intentional to limit differences in SA being attributed to changes in foot strike pattern, which have been shown to result in different movement patterns and muscle activation (Almeida et al., 2015b; Delgado et al., 2013; Gruber et al., 2014), and may therefore interfere with any effects of midsole cushioning. However, majority of the recreational running population are rearfoot strikers (Larson et al., 2011), thus this sample allows for the most generalizability.

The sample was also restricted to runners who could fit into a US men's shoe size of 9 to 10 or women's 7 to 8. This exclusion allowed for better control over the differences in footwear characteristics (e.g. mass, length, stiffness), but since foot length is scaled to height and stature, the full range of the population may not have been captured with this sample. Hence, there is a possibility that differences in body anthropometrics, combined with a pre-selected speed, may elicit differences in shock attenuation or between sexes not captured by this study.

Similarly, the shoes selected for this study were not identical in geometry. Differences in design features such as shoe weight, heel and forefoot heights, toe box widths, outsole patterns and rubbers, and materials or construction of the upper may contribute towards differences in running mechanics and thereby confound results. However, Hoogkamer et al. (2016) reported that shoe mass differences of greater than 100 g is needed to significantly influence metabolic cost and running economy by approximately 1%. Horvais and Samozino (2013) showed that midsole heel height and drop height were positively correlated with foot angle at ground contact, but all within the classification of rearfoot striking, and only when observed across heel height and drop height ranges of 0 to 30 mm and 0 to 15 mm respectively. Furthermore, when comparing six different shoes with midsole thicknesses of 0 to 25 mm, Law et al. (2019) only observed significant differences in foot strike angle and spatiotemporal parameters between the 0 and 25 mm conditions, while Chambon et al. (2014) saw no differences in ground reaction force or tibial acceleration variables across five shoes with thicknesses ranging from 0 to 16 mm. In contrast, the heel heights used in this study ranged from 28 to 37 mm (difference of 8 mm) across both male and female shoe models, while the drop heights ranged from 8 to 10 mm (difference of 2 mm;

Table 3.2). Thus, while non-identical in construction, the present footwear conditions did not vary substantially across the ranges available. Regardless, such a limitation should be addressed and may influence the generalization of these results to other shoe models.

Lastly, the current results should not be extrapolated to a fatigued state. Various studies have shown that fatigue can influence running kinematics (Derrick et al., 2002), muscle activation and co-activation (Kellis et al., 2011), and impact acceleration and attenuation during running (Coventry et al., 2006; García-Pérez et al., 2014; Mizrahi et al., 2000). Fatigue may also affect runners differently based on sex (Bazuelo-Ruiz et al., 2018). Participants in this study ran for a total time of 20 minutes including warm-up, with a minimum of 5 minutes rest between 5 minute runs in each shoe condition. It was not expected that the present protocol would induce effects of fatigue, therefore this study does not inform how fatigue may alter any effects of shoe cushioning on running biomechanics and low back impact shock attenuation.

6 Conclusion

The aim of the present research was to examine the effects of midsole cushioning stiffness on shock attenuation in the low back of healthy recreational runners. General results showed that contrary to expectations, midsole cushioning stiffness did not have a strong influence on lumbar spine posture, muscle activation, or shock attenuation, but did contribute to differences in lower limb posture and magnitudes of shock attenuation. Specifically, softer, more compliant midsoles resulted in greater ankle plantarflexion and lower knee flexion angles at initial contact and throughout stance phase, which may have contributed to the lower shock attenuation experienced throughout the lower limb. This supports the idea that leg stiffness allows for sufficient accommodation to variation in impact shock experienced at the distal tibia, so that no differences are evident or necessary by the time the shock wave reaches the low back. These findings may have implications on the role of manipulating midsole cushioning to address lower limb loads and injuries during running, and provides evidence against using footwear as a solution or intervention for low back pain, but also encourages the notion that low back pain should not be exacerbated by footwear.

This study was also valuable by investigating both male and female recreational runners. While previous work has primarily focused on male runners, the inclusion of sex comparisons provide insight into the different mechanisms that each sex may employ. Greater knee and hip extension in females, along with higher abdominal activation, may contribute to challenges in how the lower limb and low back attenuate impact shock. However, there is still much to be examined, and further research is necessary to better understand how these attenuation mechanisms may relate to potential overloading of the musculoskeletal structures and differences in injury rates across sex.

Collapsing measures across sex should not be done where possible, and a larger sample size would provide better insight on this area.

Future directions should also explore a greater range in midsole stiffness to allow such results to be generalized across everyone from elite sprinters to barefoot runners, who each have drastically different running characteristics and specialized footwear. Another avenue is to investigate whether impairment of these mechanisms could result in injury in these populations, such as by comparing across groups with different lower limb ailments or low back injuries. Finally, these results could be better supplemented by the addition of kinetic and three-dimensional kinematic analyses to provide a holistic understanding on how midsole cushioning interacts with posture, muscle activation, and shock attenuation.

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Appendices

A. Initial Contact Definitions

Table 7.1: Comparison of kinematic methods for identifying initial contact and toe-off during treadmill running.

Author	Definition	Mean ± SD Error with respect to Gold Standard (ms)
	vGRF > 20 N	***Gold standard***
(Alton et al., 1998) *	IC: peak downward position of the lateral malleolus marker on the heel TO: instance of increase in vertical position of the 2nd metatarsal head	IC: 32.0 ± 14.1 TO: -3.8 ± 4.2
(Hreljac & Stergiou, 2000) *	IC: local minima of foot and shank angular acceleration in sagittal planes closest to the first peak in knee extension TO: local minima of foot and shank angular acceleration closest to second peak in knee extension	IC: -20.0 ± 8.6 TO: 34.1 ± 50.7
(Dingwell et al., 2001) *	IC: first occurrence of peak knee extension TO: second occurrence of peak knee extension	IC: -27.4 ± 7.4 TO: 13.8 ± 10.5
(O'Connor et al., 2007)	IC: minima in vertical velocity of the foot centre (midpoint of heel and toe marker locations) that is <35% of the range in heel heights encountered during the trial TO: peak in vertical velocity of the foot centre	IC: 16 ± 15 TO: 9 ± 15
(Zeni et al., 2008) *	IC: maximum anterior-posterior displacement between heel and sacrum markers TO: maximum anterior-posterior displacement between toe and sacrum markers	IC: 18.7 ± 81.4 TO: 91.4 ± 22.5
(Smith et al., 2015) *	IC: first maxima in vertical displacement between heel and PSIS markers TO: maxima in vertical displacement between 2nd metatarsal and PSIS closest to second peak in knee extension	IC: 1.2 ± 17.1 TO: 4.7 ± 5.9

* errors calculated by Smith et al., 2015; vGRF = vertical ground reaction force; IC = initial contact; TO = toe-off; PSIS = posterior superior iliac spine

B. Overall ANOVA Results

***Table 7.2**: One-way between-groups ANOVA results for shoe characteristics (conducted across five testing cycles per each male and female shoe, N = 30).*

	DF_B	DF_E	F-Value	*p*-Value	Significance	η_p^2
Stiffness (N/mm)	2	27	49.434	9.43e-10	*	0.785
Energy Absorbed (J)	2	27	12.168	0.000171	*	0.474
Energy Returned (J)	2	27	27.010	3.61e-07	*	0.667
Hysteresis (%)	2	27	1904.5	8.73e-30	*	0.993

***Table 7.3**: Mixed-measures ANOVA results for sagittal joint angles at initial contact.*

	DF_B	DF_E	F-Value	*p*-Value	Significance	η_p^2
Ankle Dorsiflexion (°)						
Sex	1	16	0.000947	0.976		0.0000592
Shoe	2	32	3.110	0.058		0.163
Sex × Shoe	2	32	2.975	0.065		0.157
Knee Flexion (°) †						
Sex	1	N/A	0.223	0.637		N/D
Shoe	1.87	N/A	3.295	0.040	*	N/D
Sex × Shoe	1.87	N/A	3.295	0.044	*	N/D
Hip Flexion (°)						
Sex	1	17	0.047	0.832		0.003
Shoe	2	34	0.122	0.886		0.007
Sex × Shoe	2	34	0.200	0.820		0.012
Lumbar Flexion (°)						
Sex	1	17	0.068	0.798		0.004
Shoe	2	34	0.839	0.441		0.047
Sex × Shoe	2	34	0.539	0.588		0.031

† non-parametric analysis; N/A = not applicable; N/D = no data due to *nparLD* function limitations

***Table 7.4**: Mixed-measures ANOVA results for sagittal joint ranges of motion during stance phase.*

	DF_B	DF_E	F-Value	*p*-Value	Significance	η_p^2
Ankle ROM (°)						
Sex	1	16	0.303	0.589		0.019
Shoe	2	32	2.986	0.065		0.157
Sex × Shoe	2	32	1.651	0.208		0.094
Knee ROM (°)						
Sex	1	17	0.252	0.622		0.015
Shoe	2	34	1.259	0.297		0.069
Sex × Shoe	2	34	1.936	0.160		0.102
Hip ROM (°) †						

Sex	1	N/A	1.417	0.234		N/D
Shoe	1.97	N/A	0.836	0.432		N/D
Sex × Shoe	1.97	N/A	3.565	0.029	*	N/D
Lumbar ROM (°)						
Sex	1	17	0.007	0.936		0.000386
Shoe	2	34	2.708	0.081		0.137
Sex × Shoe	2	34	0.144	0.866		0.008

† non-parametric analysis; N/A = not applicable; N/D = no data due to *nparLD* function limitations

***Table 7.5**: Mixed-measures ANOVA results for mean sagittal angles during stance phase.*

	DF_B	**DF_E**	**F-Value**	***p*-Value**	**Significance**	η_p^2
Mean Ankle Dorsiflexion (°)						
Sex	1	16	0.132	0.722		0.008
Shoe	2	32	5.354	0.010	*	0.251
Sex × Shoe	2	32	2.513	0.097		0.136
Mean Knee Flexion (°)						
Sex	1	17	0.036	0.851		0.002
Shoe	2	34	1.402	0.260		0.076
Sex × Shoe	2	34	2.340	0.112		0.121
Mean Hip Flexion (°)						
Sex	1	17	1.442	0.246		0.078
Shoe	2	34	0.0634	0.537		0.036
Sex × Shoe	2	34	0.375	0.690		0.022
Mean Lumbar Flexion (°)						
Sex	1	17	0.189	0.670		0.011
Shoe	2	34	1.239	0.303		0.068
Sex × Shoe	2	34	0.342	0.713		0.020

***Table 7.6**: Mixed-measures ANOVA results for peak flexion angles during stance phase.*

	DF_B	**DF_E**	**F-Value**	***p*-Value**	**Significance**	η_p^2
Peak Ankle Dorsiflexion (°)						
Sex	1	16	0.273	0.608		0.017
Shoe	2	32	11.639	0.000159	*	0.421
Sex × Shoe	2	32	1.858	0.172		0.104
Peak Knee Flexion (°)						
Sex	1	17	0.004	0.948		0.00025
Shoe	2	34	2.215	0.125		0.115
Sex × Shoe	2	34	2.484	0.098		0.128
Peak Hip Flexion (°)						
Sex	1	17	1.134	0.302		0.063
Shoe	2	34	0.194	0.824		0.011
Sex × Shoe	2	34	0.136	0.873		0.008

Peak Lumbar Flexion (°)						
Sex	1	17	0.191	0.668		0.011
Shoe	2	34	1.801	0.180		0.096
Sex × Shoe	2	34	0.276	0.760		0.016

***Table 7.7**: Mixed-measures ANOVA results for peak extension angles during stance phase.*

	DF_B	DF_E	F-Value	*p*-Value	Significance	η_p^2
Peak Ankle Plantarflexion (°)						
Sex	1	16	0.787	0.388		0.047
Shoe	2	32	4.679	0.017	*	0.226
Sex × Shoe	2	32	3.444	0.044	*	0.177
Peak Knee Extension (°)						
Sex	1	17	0.388	0.542		0.022
Shoe	2	34	1.205	0.312		0.066
Sex × Shoe	2	34	1.935	0.160		0.102
Peak Hip Extension (°) †						
Sex	1	N/A	4.245	0.039	*	N/D
Shoe	1.84	N/A	0.709	0.481		N/D
Sex × Shoe	1.84	N/A	0.973	0.372		N/D
Peak Lumbar Extension (°)						
Sex	1	17	0.284	0.601		0.016
Shoe	2	34	0.766	0.473		0.043
Sex × Shoe	2	34	0.559	0.577		0.032

† non-parametric analysis; N/A = not applicable; N/D = no data due to *nparLD* function limitations

***Table 7.8**: Mixed-measures ANOVA results for muscle activation and co-contraction variables during stance phase.*

	DF_B	DF_E	F-Value	*p*-Value	Significance	η_p^2
LES Activation †						
Sex	1	N/A	0.102	0.749		N/D
Shoe	1.53	N/A	0.927	0.343		N/D
Sex × Shoe	1.53	N/A	0.597	0.507		N/D
RA Activation						
Sex	1	17	3.205	0.091		0.159
Shoe	2	34	3.901	0.030	*	0.187
Sex × Shoe	2	34	2.078	0.141		0.109
EO Activation †						
Sex	1	N/A	5.454	0.020	*	N/D
Shoe	1.88	N/A	0.655	0.510		N/D
Sex × Shoe	1.88	N/A	1.480	0.228		N/D
LES-RA CCI						
Sex	1	17	0.027	0.871		0.002
Shoe	2	34	0.044	0.957		0.003

Sex × Shoe	2	34	2.384	0.108	0.123
LES-RA Correlation					
Sex	1	17	0.097	0.759	0.006
Shoe	2	34	1.435	0.252	0.078
Sex × Shoe	2	34	1.075	0.353	0.059
LES-RA Phase Lag					
Sex	1	17	0.193	0.666	0.011
Shoe	2	34	0.313	0.733	0.018
Sex × Shoe	2	34	3.118	0.057	0.155
LES-EO CCI †					
Sex	1	N/A	3.406	0.064	N/D
Shoe	1.75	N/A	0.013	0.980	N/D
Sex × Shoe	1.75	N/A	0.708	0.475	N/D
LES-EO Correlation					
Sex	1	17	4.413	0.051	0.206
Shoe	2	34	1.905	0.164	0.101
Sex × Shoe	2	34	2.700	0.082	0.137
LES-EO Phase Lag					
Sex	1	17	0.479	0.498	0.027
Shoe	2	34	0.598	0.555	0.034
Sex × Shoe	2	34	0.704	0.501	0.040
RA-EO CCI †					
Sex	1	N/A	2.076	0.150	N/D
Shoe	1.91	N/A	0.682	0.499	N/D
Sex × Shoe	1.91	N/A	0.702	0.489	N/D
RA-EO Correlation					
Sex	1	17	0.492	0.493	0.028
Shoe	2	34	0.521	0.599	0.030
Sex × Shoe	2	34	0.547	0.584	0.031
RA-EO Phase Lag					
Sex	1	17	3.326	0.086	0.164
Shoe	2	34	0.755	0.478	0.043
Sex × Shoe	2	34	0.759	0.476	0.043

† non-parametric analysis; CCI = co-activation coefficient; N/A = not applicable; N/D = no data due to *nparLD* function limitations

Table 7.9: *Mixed-measures ANOVA results for peak acceleration magnitudes during stance phase.*

	DF_B	DF_E	F-Value	*p*-Value	Significance	η_p^2
Tibial Acceleration (*g*)						
Sex	1	16	0.147	0.706		0.009
Shoe	2	32	1.219	0.309		0.071
Sex × Shoe	2	32	0.188	0.830		0.012
L5 Acceleration (*g*)						
Sex	1	16	0.298	0.592		0.018
Shoe	2	32	0.429	0.655		0.026
Sex × Shoe	2	32	0.713	0.498		0.043
L1 Acceleration (*g*)						
Sex	1	16	0.293	0.596		0.018
Shoe	2	32	0.702	0.503		0.042
Sex × Shoe	2	32	0.825	0.447		0.049
Head Acceleration (*g*) †						
Sex	1	N/A	3.261	0.071		N/D
Shoe	1.59	N/A	3.573	0.038	*	N/D
Sex × Shoe	1.59	N/A	0.040	0.932		N/D

† non-parametric analysis; N/A = not applicable; N/D = no data due to *nparLD* function limitations

Table 7.10: *Mixed-measures ANOVA results for mean shock attenuation during stance phase.*

	DF_B	DF_E	F-Value	*p*-Value	Significance	η_p^2
Tibia-L5 SA						
Sex	1	16	0.011	0.918		0.000677
Shoe	2	32	3.126	0.058		0.163
Sex × Shoe	2	32	1.351	0.273		0.078
L5-L1 SA						
Sex	1	16	0.00000435	0.998		0.000000272
Shoe	2	32	1.483	0.242		0.085
Sex × Shoe	2	32	1.383	0.265		0.080
L1-Head SA						
Sex	1	16	0.022	0.883		0.001
Shoe	2	32	0.085	0.919		0.005
Sex × Shoe	2	32	1.754	0.189		0.099
Tibia-Head SA						
Sex	1	16	0.102	0.754		0.006
Shoe	2	32	2.264	0.120		0.124
Sex × Shoe	2	32	1.086	0.350		0.064

C. *A Posteriori* Analyses Results

Two-way mixed measures ANOVAs were conducted on shock attenuation in the low frequency (3-8 Hz) and high frequency (9-20 Hz) ranges for all sensor pairings across shoe condition and sex. Shapiro-Wilk tests revealed that all data met the assumption of normality ($p > 0.05$). Results revealed significant main effects of SHOE in 3-8 Hz SA between the tibia and L5 ($F_{(2,32)} = 4.397$, $p < 0.05$) and tibia and the head ($F_{(2,32)} = 3.719$, $p < 0.05$). For both sensor pairings, post-hoc pairwise comparisons showed that wearing the RCT (tibia-L5 -8.83 ± 2.64 dB; tibia-head -9.68 ± 2.40 dB) resulted in significantly greater SA compared to the ZMX (tibia-L5 -7.86 ± 3.00 dB, $p < 0.05$; tibia-head -8.78 ± 2.67 dB, $p < 0.01$). No other significant differences were observed in either L5 to L1 or L1 to head SA of the low frequency range, or all variables in the high frequency range ($F \leq 1.929$, $p \geq 0.162$).

Table 7.11*: Mean (SD) high and low frequency range shock attenuation for male and female runners in three different shoe conditions.*

	PGS		RCT		ZMX	
	Female	**Male**	**Female**	**Male**	**Female**	**Male**
3-8 Hz						
Tibia-L5 Shock Attenuation (dB)	-8.09 (3.02)	-8.9 (2.23)	-8.71 (2.74)	-8.94 (2.68)	-8.06 (3.18)	-7.67 (2.98)
L5-L1 Shock Attenuation (dB)	0.41 (1.41)	0.12 (0.90)	0.31 (1.36)	0.17 (1.93)	0.58 (1.63)	0.58 (1.83)
L1-Head Shock Attenuation (dB)	-0.33 (2.85)	-0.33 (2.37)	-0.39 (2.79)	-0.83 (2.13)	-0.23 (2.71)	-0.45 (1.55)
Tibia-Head Shock Attenuation (dB)	-8.82 (1.84)	-9.34 (2.50)	-9.42 (2.14)	-9.94 (2.74)	-8.87 (2.30)	-8.70 (3.14)
9-20 Hz						
Tibia-L5 Shock Attenuation (dB)	-4.80 (3.86)	-3.15 (2.98)	-3.97 (4.06)	-3.56 (3.48)	-3.07 (4.25)	-3.00 (3.30)
L5-L1 Shock Attenuation (dB)	-1.05 (4.74)	-0.83 (2.07)	-0.99 (4.68)	-1.06 (2.00)	0.02 (4.11)	-1.01 (2.14)
L1-Head Shock Attenuation (dB)	-5.41 (5.47)	-4.91 (2.71)	-5.33 (5.41)	-4.55 (2.84)	-5.04 (5.42)	-4.64 (2.24)
Tibia-Head Shock Attenuation (dB)	-9.15 (4.16)	-7.22 (2.27)	-8.31 (4.28)	-7.06 (2.48)	-8.13 (4.71)	-6.63 (2.50)

***Table 7.12**: Mixed-measures ANOVA results for mean high and low frequency range shock attenuation during stance phase.*

	DF_B	DF_E	F-Value	*p*-Value	Significance	η_p^2
3-8 Hz						
Tibia-L5 SA						
Sex	1	16	0.029	0.868		0.002
Shoe	2	32	4.397	0.021	*	0.216
Sex × Shoe	2	32	1.664	0.205		0.094
L5-L1 SA						
Sex	1	16	0.063	0.805		0.004
Shoe	2	32	0.505	0.608		0.031
Sex × Shoe	2	32	0.077	0.926		0.005
L1-Head SA						
Sex	1	16	0.043	0.838		0.003
Shoe	2	32	0.302	0.742		0.019
Sex × Shoe	2	32	0.140	0.870		0.009
Tibia-Head SA						
Sex	1	16	0.067	0.799		0.004
Shoe	2	32	3.719	0.035	*	0.189
Sex × Shoe	2	32	0.721	0.494		0.043
9-20 Hz						
Tibia-L5 SA						
Sex	1	16	0.189	0.670		0.012
Shoe	2	32	1.929	0.162		0.108
Sex × Shoe	2	32	1.379	0.266		0.079
L5-L1 SA						
Sex	1	16	0.034	0.856		0.002
Shoe	2	32	0.907	0.414		0.054
Sex × Shoe	2	32	1.210	0.311		0.07
L1-Head SA						
Sex	1	16	0.043	0.838		0.003
Shoe	2	32	0.302	0.742		0.019
Sex × Shoe	2	32	0.140	0.870		0.009
Tibia-Head SA						
Sex	1	16	0.970	0.339		0.057
Shoe	2	32	1.569	0.224		0.089
Sex × Shoe	2	32	0.280	0.758		0.017

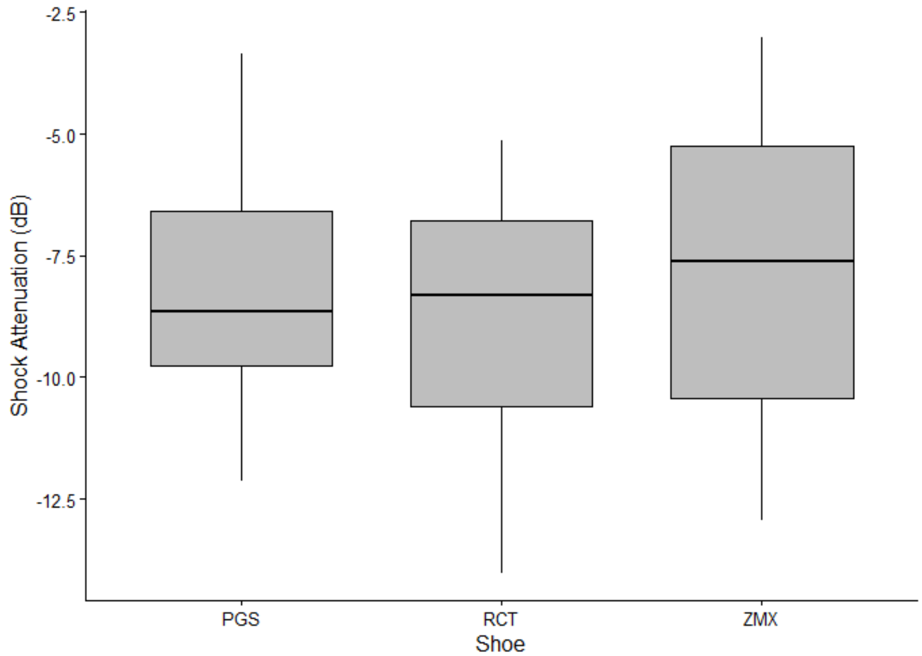

Figure 7.1: *Boxplot of mean 3-8 Hz shock attenuation (dB) between the tibia and L5 across shoe conditions collapsed across sex during stance phase. Negative values denote signal attenuation. Significant differences were observed between the RCT and ZMX.*

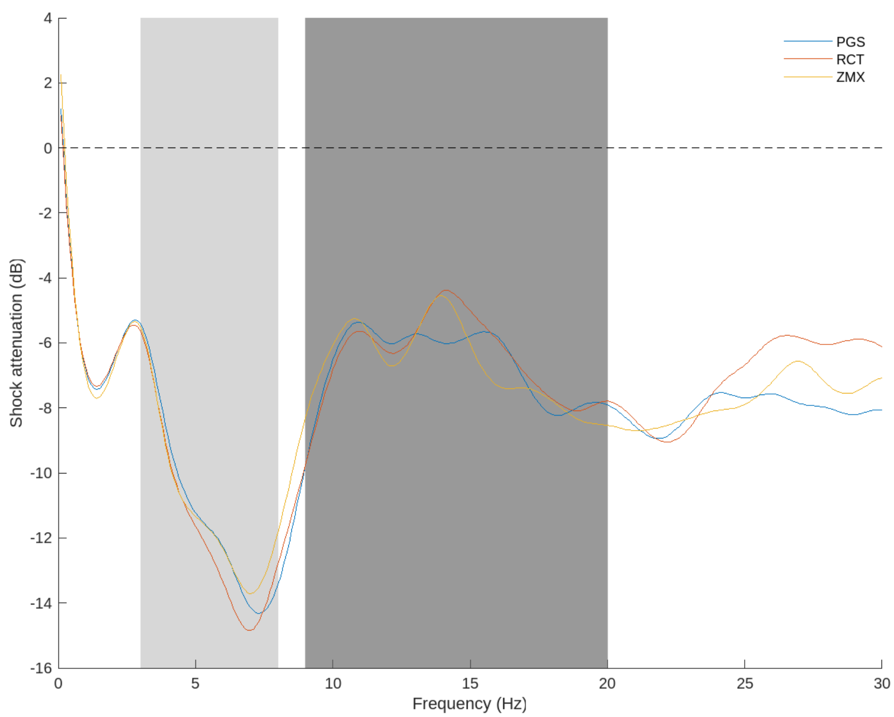

Figure 7.2*: Transfer function depicting the mean shock attenuation (dB) across low frequency (light gray) and high frequency (dark gray) ranges between the tibia and L5 acceleration signals across shoe conditions. Negative values represent signal attenuation.*

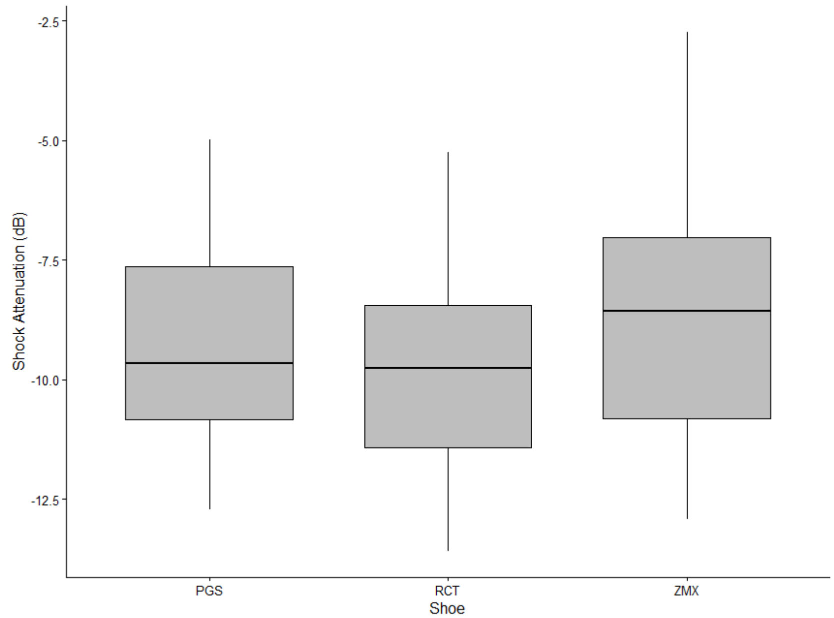

Figure 7.3: *Boxplot of mean 3-8 Hz shock attenuation (dB) between the tibia and the head across shoe conditions collapsed across sex during stance phase. Negative values denote signal attenuation. Significant differences were observed between the RCT and ZMX.*

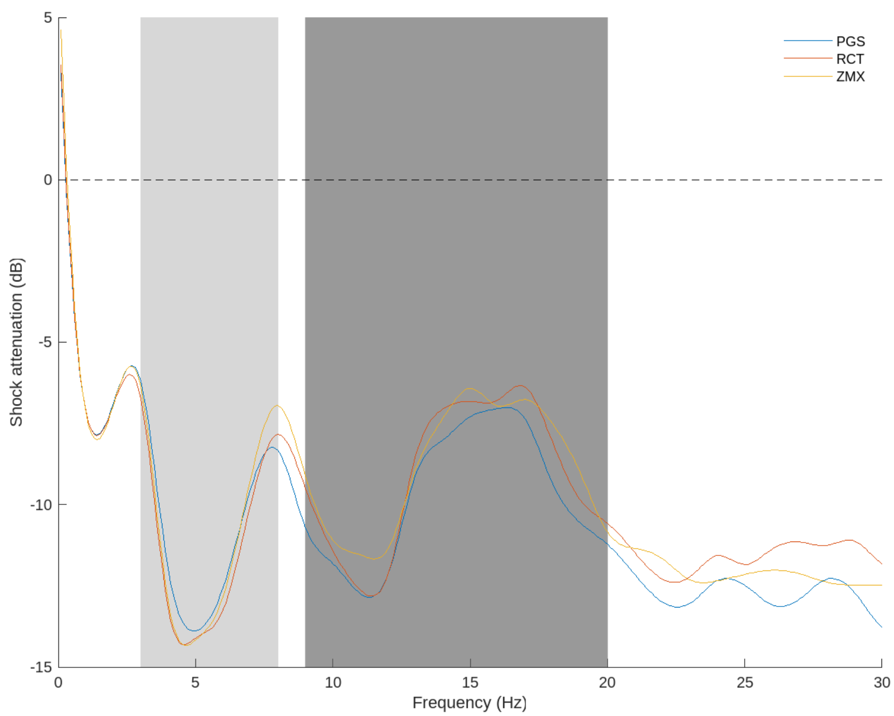

Figure 7.4: *Transfer function depicting the mean shock attenuation (dB) across low frequency (light gray) and high frequency (dark gray) ranges between the tibia and the head acceleration signals across shoe conditions. Negative values represent signal attenuation.*