GAIT AND BODY COMPOSITION MECHANISMS OF SHOCK ATTENUATION AMONG RUNNERS

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This dissertation aimed to examine gait and body composition mechanisms that attenuate vibrations during running, and the influence of age and running experience on shock attenuation. Study 1 examined the association of bone mineral density, bone content, fat mass, and lean mass within the lower extremity and pelvis to shock attenuation of 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4) signals. Study 2 assessed the contribution of gait mechanics and study 3 assessed the role of age and experience in FB1-FB4 attenuation.

Fifty-eight runners ran overground ($3.35\ m\cdot s^{-1} \pm 5\%$) as motion capture and force plates recorded their gait mechanics. Inertial measurement units (IMU) at the distal tibia and low-back recorded shock attenuation along the axial and resultant directions. Body composition was assessed in thirty-eight runners using dual x-ray absorptiometry. Runners were classified as middle-aged (52.9±5.8 years-old), young (22.3±4.0 years-old), novice (running <1 year, 3-13 $miles\cdot week^{-1}$), and recreational (>1 year, 9-30 $miles\cdot week^{-1}$). IMU signals were transformed to the frequency domain using power spectral density then a transfer function of the low-back relative to the tibia quantified axial and resultant shock attenuation within FB1-FB4. Associations of gait mechanics, body composition, age, and running experience to axial and resultant shock attenuation in FB1-FB4 were tested using linear regression ($p \leq 0.05$).

Overall, gait characteristics representative of a more compliant gait (e.g., sagittal ankle and knee flexion, range-of-motion, reduced joint stiffness) were associated with greater axial attenuation within FB1, FB2, and FB4 ($p \leq 0.024$), and resultant attenuation within FB1-FB4 ($p \leq 0.045$). An increase in lower extremity fat mass decreased axial attenuation within FB2-FB4 ($p \leq 0.033$). Being middle-aged or a novice runner was associated with less resultant and axial shock attenuation within FB1-FB4 ($p \leq 0.017$). These

results indicate that middle-aged and novice runners may attenuate less shock due to their typically greater fat mass and less compliant gait than younger and more experienced runners.

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1 CHAPTER 1: Introduction

1.1 Background

According to a 2020 report by the Sports and Fitness Industry Association, 50 million Americans participate in running or jogging each year. Running provides various health benefits, including cardiovascular, musculoskeletal, and mental health improvements (Lavie et al., 2015; Lee et al., 2014; Oswald et al., 2020b). However, running-related injuries (RRI) are prevalent among the general running population, as noted previously (Van Gent et al., 2007; Videbæk et al., 2015). Notably, older runners and novice runners experience significantly more RRIs than younger and recreational runners, respectively (Linton and Valentin, 2018; McKean et al., 2006; Videbæk et al., 2015). RRIs not only hinder individuals' ability to obtain the benefits of running but also reduce their levels of physical activity (Davis and Gruber, 2019). It is unclear whether runners return to their previous physical activity levels after recovering from an RRI. This is a significant concern as physical inactivity is a leading cause of morbidity, according to WHO (Organization, 2009). Inevitably, aging leads to a reduction in musculoskeletal function among older adults (Kirkendall and Garrett, 1998). The development of an RRI among older runners may exacerbate this natural process, which is otherwise positively influenced by physical activity. Similarly, novice runners who develop an RRI may discontinue running altogether, as previously reported (Fokkema et al., 2019). To mitigate the adverse effects of RRI development among older and novice runners, it is necessary to explore factors that may increase RRI risk among these sub-populations.

Impact shock has been associated with both prospective and retrospective RRI in experienced, young recreational runners, with both injured groups displaying larger magnitudes of peak impact shock compared to uninjured runners (Davis et al., 2004; Milner et al., 2006). To examine if impact shock may be an RRI risk factor across all sub-populations of runners and not just young, recreational runners, it is crucial to determine whether older and novice runners display larger magnitudes of peak impact shock compared to younger and experienced runners. This information may address older and novice runners' heightened susceptibility to injury risk, and may guide gait retraining, footwear, or physical activity interventions that aim to reduce RRI risk in these sub-populations.

Peak impact shock does not, however, describe the body's response to impact loading. Therefore, shock attenuation should be examined instead to understand older and novice runners' response to and thereby injury risk arising from impact shock. The motion of running generates shock vibrations after foot-ground contact that have different frequencies ranging from 3-50 Hz (Gruber et al., 2014; Shorten and Winslow, 1992). As the shock vibrations are propagated from the tibia towards the head, frequencies less than 8 Hz may be actively attenuated (such as through eccentric muscle contractions and joint stiffness modulation), while those greater than 8 Hz may be passively attenuated (such as through bone and soft tissue deformation) (Shorten and Winslow, 1992). The integrity of both active and passive shock attenuation mechanisms is crucial for adequately responding to increased attenuation demands. Older runners may have poorer integrity of these mechanisms due to age, while novice runners' gait and tissues may not be well adapted due to less running exposure. Thus, their diminished ability to either actively or passively attenuate shock may put them at risk for RRI development.

Due to their gait mechanics, older and less experienced runners may rely more on passive shock attenuation mechanisms than their younger and more experienced counterparts. For example, older runners tend to exhibit smaller peak knee flexion magnitudes and knee flexion-extension ranges of motion during stance, which can reduce their ability to perform negative mechanical work and actively absorb shock (Bus, 2003b; Devita et al., 2016; Powell and Williams, 2018). Similarly, novice runners tend to display smaller magnitudes of knee flexion (Clermont et al., 2017; Clermont et al., 2019; Moore, 2016) and ankle eversion (Harrison et al., 2021) throughout stance, which can also hinder their ability to actively attenuate shock. As a result, it is likely that older and novice runners rely more on passive attenuation mechanisms.

It is currently unknown how joint kinematics and kinetics contribute to shock attenuation across the frequency spectrum during running. For instance, while there is speculation that a less stiff knee joint which permits greater negative joint work may aid active attenuation (Hamill et al., 2014), the association of joint stiffness to the attenuation of frequencies between 3-50 Hz is unknown. The attenuation of frequencies greater than 5 Hz can be increased by increasing knee flexion angles at and during ground contact, as noted during in-line skating and in a pendulum model (Edwards et al., 2012; Lafortune et al., 1996). However, it is yet to be examined if this holds true among runners. To fully understand the differences in shock attenuation capabilities between older/novice and younger/experienced runners, it is necessary to examine the contribution of joint kinematics and kinetics to shock attenuation within 3-8 Hz and beyond 8 Hz.

If older and novice runners rely more on passive attenuation mechanisms, it may lead to increased shock attenuation by musculoskeletal tissues such as bone and lean mass. Bone mineral density, which depends on mineral accrual, plays a role in bone's stiffness and thereby its ability to absorb shock by deforming (Oftadeh et al., 2015). Higher BMD, due to greater mineral accrual, is protective against bone stress injuries (Kelsey et al., 2007; Myburgh et al., 1990). Similarly, muscle and adipose tissue also contribute to shock absorption due to their viscoelastic properties (Nigg et al., 1995). However, tissue properties decrease with age (Goodpaster et al., 2006; Warming et al., 2002) and increase with long-term exposure to weight-bearing exercise (Luden et al., 2012; Ravnholt et al., 2018; Yu et al., 2022). Therefore, tissue properties of older and novice runners may explain their increased injury risk compared with younger and more experienced runners. To determine if diminished tissue properties underlie greater injury susceptibility, it must be examined if older and novice runners attenuate shock more actively or passively. The contribution of body composition parameters, such as bone mineral density and content, fat mass, and lean mass, to shock attenuation across frequencies has been assessed in animal models (Paul et al., 1978) but must also be assessed in running humans to understand the loads experienced by these structures during human running. Injury risk arising from passive shock attenuation demands can then be assessed accordingly.

Impact shock may be an important RRI risk factor because the degree of shock and how it is attenuated in the body indicates which musculoskeletal structures are most relied upon for attenuation, and therefore loaded. Novice and older runners may be particularly susceptible to impact shock as an RRI mechanism; Their gait patterns may cause impact shock to be greater compared to more experienced and younger runners, and concurrently their tissues may not be capable of optimally attenuating impact-related shock (Beck et al., 2016; Boyer et al., 2017a; Devita et al., 2016; Warming et al., 2002). Therefore, assessing if the magnitude of impact shock and shock attenuation are age and running experienced dependent, and the relationship of joint kinematics, joint kinetics, and biological tissues to the degree of shock attenuated is necessary to identify if older and novice runners' tissues are loaded differently compared with younger and more experienced runners. This information will provide insight to whether older and novice runners incur greater injury risk compared to younger and more experienced runners.

1.2 Aims and Hypotheses

1.2.1 Aim 1

Examine the associations of lower extremity and gynoid fat mass and lean mass, and lower extremity and pelvis bone mineral density and content to the degree of shock attenuation within frequency bands of 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4) among runners.

The deformation of passive musculoskeletal structures (Paul et al., 1978) is more effective at attenuating high rather than low frequencies of shock (Paul et al., 1978). Thus, the following was hypothesized:

1.2.1.1 Hypothesis

There will be an increase in the degree of shock attenuated between the tibia and lower back within FB2, FB3, and FB4 with an increase in each body composition measure, but the degree of shock attenuated within FB1 will not be associated with any of the body composition measures.

1.2.2 Aim 2

Examine the associations of sagittal knee and ankle angles at foot-ground contact, knee and ankle joint stiffness, joint range of motion at the ankle (frontal and sagittal planes) and knee (sagittal plane), peak ankle eversion, peak knee flexion, negative joint work and power at the ankle and knee during the deceleration phase of running to the degree of shock attenuation within frequency bands of 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4) among runners.

Previous evidence suggests the knee is a low-pass filter of impact shock (Edwards et al., 2012). Active sagittal knee and ankle motions and stiffnesses may dictate the amount of negative work performed and consequently energy absorbed by eccentrically contracting musculature. Ankle eversion motion may also influence shock attenuation by dictating the time occurrence of the vertical ground reaction force impact peak (Nigg and Morlock, 1987) and altering the stiffness of the foot-shoe complex at ground contact. The following was hypothesized:

1.2.2.1 Hypothesis 1

Knee and ankle joint stiffness during deceleration will be negatively associated with the degree of shock attenuation within FB1 but not within FB2, FB3, or FB4.

1.2.2.2 Hypothesis 2

Frontal and sagittal plane ankle joint ROM and sagittal plane knee ROM during deceleration will be positively associated with the degree of shock attenuation within FB1, but not within FB2, FB3, or FB4.

1.2.2.3 Hypothesis 3

Sagittal plane knee and ankle angle at contact, and peak ankle eversion and knee flexion magnitude during deceleration will be positively associated with the degree of shock attenuation within FB1, but not within FB2, FB3, or FB4.

1.2.2.4 Hypothesis 4

Sagittal plane negative joint work and power absorption at the ankle and knee during deceleration will be positively associated with the degree of shock attenuation within FB1, but not within FB2, FB3, or FB4.

1.2.3 Aim 3

Estimate the independent effects of age and running experience on impact shock and degree of shock attenuation within frequency bands 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4) among runners.

Older and middle-aged runners typically display gait patterns that may result in greater peak impact shock and reduced active shock attenuation magnitudes compared with younger runners; Older compared with younger runners display reduced peak knee flexion magnitude during stance, reduced knee flexion-extension and ankle dorsiflexion-plantarflexion ROM, and reduced energy absorption at the ankle (Bus, 2003b; Devita et al., 2016; Powell and Williams, 2018). In addition, more experienced runners display greater knee flexion during late swing and throughout stance (Clermont et al., 2017; Clermont et al., 2019; Moore, 2016) compared with less experienced runners. These findings indicate both impact shock and active shock attenuation may increase with running experience. The following was hypothesized:

1.2.3.1 Hypothesis 1

Age will have a significant positive association with time-domain peak impact shock at the tibia, signal power magnitude within FB2-FB4 at the tibia, time-domain shock attenuation, and the degree of shock attenuated within FB2-FB4, but a significant negative association with time-domain peak impact shock at the lower back and the degree of shock attenuated in FB1.

1.2.3.2 Hypothesis 2

Running experience will have a significant negative association with time-domain peak impact shock at the tibia, signal power magnitude within FB2-FB4 at the tibia, time-domain shock attenuation, and the degree of shock attenuated within FB2-FB4, but a significant positive association with time-domain peak impact shock at the lower back and the degree of shock attenuated in FB1.

1.3 Significance

Not only do injured runners engage in less physical activity post-injury (Davis and Gruber, 2019), but they are also at a greater risk for developing RRI in the future (Hulme et al., 2017). Novice and middle-aged runners may experience these worrisome consequences of RRI to a more significant extent compared with recreational and younger runners, respectively, given their higher risk for injury development (McKean et al., 2006; Nielsen et al., 2013; Videbæk et al., 2015). Per 1000 hours of running, novice runners develop 17.8 RRI compared with 7.7 RRI among recreational runners (Videbæk et al., 2015). Similarly, middle-aged runners between 45-65 years are at heightened injury risk compared with younger runners (McKean et al., 2006; Nielsen et al., 2013). Exploring how known RRI risk factors like impact shock and shock attenuation magnitude differ between groups with varied injury risk may provide insight into population-specific risk reduction strategies and interventions. To gain a better understanding of the stress experienced by various musculoskeletal structures and potential shock attenuation mechanisms used by middle-aged and novice runners that could be increasing their risk for

RRI, it is necessary to also examine the relationship between different shock frequencies and the previously proposed active and passive shock attenuation mechanisms.

1.4 Scope of the study (delimitations)

1. All subjects were free of any chronic illnesses that may disallow engagement in at least 15 minutes of low-intensity physical activity.
2. All subjects had no history of lower back or lower limb surgery in the past 10 years and had returned to normal running training volume for a minimum of 8 weeks prior to enrollment if an injury was developed in the past 6 months.
3. Young runners were between 18-30 years, and middle-aged runners were between 45-65 years.
4. Runners were regarded as novice (i.e., less experienced) if they had been running for <1 year and run 9-13 miles per week. Runners were considered recreational (i.e., more experienced) if they had been running for >1 year and running 13-30 miles per week (Honert et al., 2020).
5. Participants performed overground running trials in a laboratory setting along an 18m runway at self-selected and criterion running speed set at 3.35 ms^{-1}.
6. 3D motion capture (using Qualisys Tracking Manager) was used to capture joint and segment motions during running.
7. Inertial measurement units containing triaxial accelerometers measured body segment accelerations during running.
8. Kinematic and kinetic data were calculated for running at a single criterion running speed for all subjects.
9. Fast Fourier Transform (FFT) and Power Spectral Density (PSD) were used to perform a spectral analysis of the tibia, lower back, and head acceleration signals.
10. All participants wore lab-provided neutral footwear during the gait data collection.

11. Dual x-ray absorptiometry (iDXA) was used to measure body site-specific areal bone mineral density and body composition at the different anatomical locations.

1.5 Limitations

1. All gait data were collected in laboratory-based settings. Findings may therefore not be generalizable to gait mechanics during outdoor or free-living running.
2. Impact shock and shock attenuation were measured using surface-mounted accelerometry and may have been affected by soft tissue movement during running.
3. The criterion running speed may have differed from subjects' preferred running speed hence the mechanics observed may not be a perfect representation of those used during running training.
4. Sex-based differences in kinematics and kinetics may have existd, potentially influencing impact shock and shock attenuation magnitudes.

1.6 Assumptions

1. Subjects experienced minimum recall bias when filling out questionnaires regarding their running training and injury history.
2. Selection bias existed. For example, subjects who were presently injury-free but had a more frequent injury history and were therefore motivated to learn about their gait biomechanics and body composition may have chosen to enroll while individuals who remained relatively injury-free or were not keen on having their gait and body composition assessed chose not to reach out.
3. External placement of retro-reflective markers and IMUs did not interfere with subjects' natural running gait.
4. Subjects did not alter their gait when running in lab-provided shoes, which may have differed from their usual running shoes.

5. In-vitro accelerometry, i.e., using surface mounted accelerometry, sufficiently represented loading experienced by musculoskeletal structures in-vivo.
6. Kinematics were calculated using inverse dynamics which assumes rigidity of body segments, invariant segmental masses, and a fixed location of each segment's center of mass.
7. Subjects were well hydrated, did not participate in vigorous physical activity, or eat food two hours before their iDXA scan.

1.7 Operational definitions of terms

1. Active shock attenuation: The absorption of lower-frequency shock (3-8 Hz) between the tibia, lower back, and lower back and head.
2. Bone mineral content: Amount of bone mineral (grams) contained in a certain volume of bone.
3. Bone mineral density: Amount of bone mineral (grams) contained per square centimeter (cm^2) of bone.
4. Deceleration: Period of the stance phase that begins at initial foot-ground contact and ends when the joint of interest reaches its peak joint angular position in the sagittal plane.
5. Fat mass: Total amount of fat in the body (including subcutaneous and visceral) independent of lean muscle mass.
6. High-frequency shock: Frequency components of the tibial, lower back, and head acceleration signals within the range of 9-50 Hz.
7. Impact shock: The shock generated at foot-ground collision.
8. Joint angle at initial contact: The magnitude of a joint's angular position at the instant of foot-ground contact.
9. Joint range of motion: The difference between the maximum and minimum magnitude of a joint angle during deceleration.

10. Joint stiffness: The relationship between joint moment and joint angular displacement during deceleration.

11. Lean mass: Total amount of adipose-free skeletal mass in the body (or a specific anatomical region) excluding bone (including connective tissue, water, and organs).

12. Lower-frequency shock: Frequency components of the tibial, lower back, and head acceleration signals within the range of 4-8 Hz.

13. Negative joint work: The mechanical work performed by muscles about a joint during deceleration.

14. Passive shock attenuation: The absorption of higher-frequency shock (9-50 Hz) between the tibia, lower back, and lower back and head.

15. Peak impact shock: The maximum magnitude of resultant acceleration at each anatomical site.

16. Shock attenuation: The magnitude of the shock wave absorbed between the tibia and the lower back and the lower back.

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2 CHAPTER 2: Literature Review

2.1 General Introduction

The overall aim of this dissertation is twofold: (1) to determine the relationship of age and running experience to the frequency content of impact shock and shock attenuation during overground running, and (2) to examine which biomechanical and body composition factors contribute to active and passive shock attenuation. This chapter reviews literature that is relevant to (a) impact force and Impact shock, (b) signal properties of impact shock, (c) determinants of impact shock, (d) shock attenuation of the energy contained in 3-8 Hz, 9-20 Hz, 21-35 Hz, and 36-50 Hz frequency bands and their determinants, (e) an overview of the running gait biomechanics of different age and running experience-subgroups that are relevant to the frequency content of impact shock and attenuation, (f) body composition and musculoskeletal tissue properties relevant to the frequency content of impact shock and attenuation, and (g) current evidence for impact shock and shock attenuation as running related injury risk factors among different runner sub-populations.

2.2 Impact Force & Impact Shock

An *impact* is defined as the collision of two objects that generates a large force over a relatively short period of time. During running, this impact occurs between the foot and the ground with every step. The ground reaction force generated in response to this impact occurs within the first 50 milliseconds after ground contact and is termed *impact force* (Nigg et al., 1995). The foot-ground collision and this resulting impact force causes a vibration of the body segments, which is referred to as the *impact shock* (Nigg et al., 1995). Thus, factors that dictate the magnitude of impact force may also modulate characteristics of impact shock.

Impact forces are measured during the stance phase of running using force platforms embedded in the ground for overground running or through force-sensing treadmill belts during treadmill running. Impact forces can be measured in all three directions – vertical, anterior-posterior, and mediolateral. Upon foot-ground impact, the vertical ground reaction force-time curve displays an initial peak during the impact phase, i.e., the time period between foot-strike to when the stance-leg center of mass stops decelerating (Derrick et al., 1998). This peak is called the vertical ground reaction force impact peak (VGRF-IP). VGRF-IP is the result of the leg and foot segmental motions upon ground contact rather than that of the whole body (Bobbert et al., 1991). The magnitude of the VGRF-IP is significantly smaller or may be completely absent during forefoot running compared with rearfoot running. In the horizontal direction, the anterior-posterior ground reaction force is posteriorly directed at foot-ground contact (displaying a negative peak) as the foot applies an anteriorly directed force, helping the leg to decelerate. The mediolateral ground reaction force is either medially (Freychat et al., 1996; McClay and Manal, 1999; Nachbauer and Nigg, 1992)or laterally (Nilsson and Thorstensson, 1989) directed upon foot-ground contact depending on factors like foot-strike pattern and running speed and is highly variable (Gruber et al., 2017; Hamill et al., 1983; McClay and Manal, 1999; Nilsson and Thorstensson, 1989). Vertical, anterior-posterior, and mediolateral ground reaction force together compose the *resultant ground reaction force* (i.e., the vector sum of all three forces) experienced by the leg when the foot contacts the ground.

VGRF-IP may be linked to the development of running related overuse injuries (Davis et al., 2016; Hreljac et al., 2000) while the anterior-posterior and medio-lateral GRF variables may not be (Johnson et al., 2020). Therefore, studies examining gait have mostly focused on the vertical component of the ground reaction force. However, evidence increasingly suggests that the vertical loading rate of the VGRF rather than the VGRF-IP magnitude may be related to RRI risk (Van der Worp et al., 2016; Zadpoor and Nikooyan, 2011). Vertical loading rate is the rate of impact force development between 20% - 80% of the period between foot-strike and peak impact force during rearfoot running. During forefoot running, the VGRF-IP may be missing in which case the vertical loading rate is the rate of force development between 20% - 80% within the first 50 milliseconds of stance. Vertical loading rate describes how quickly the vertical ground reaction force is applied to the body. There are two ways of quantifying vertical load rates: Average vertical loading rate (AVLR) which is an average rate of change between the initial (20%) and final (80%) force magnitudes or Instantaneous vertical loading rate (VILR) which is an instantaneous rate of change (i.e., changes in force per unit of time) between initial and final force magnitudes. Both vertical loading rates and VGRF-IP are not, however, the same as impact shock. They are, though, related to peak impact shock, i.e., the maximal amplitude of impact shock occurring within the first 50 milliseconds after foot-strike. While VGRF-IP is moderately correlated with peak impact shock, vertical loading rate displays a strong correlation with peak impact shock (Hennig and Lafortune, 1991; Hennig et al., 1993). Therefore, alterations to vertical loading rates and VGRF-IP due to factors like running speed or joint and segment orientations at ground contact may result in the peak impact shock magnitude changing.

Impact shock is typically measured using accelerometers externally mounted to the antero-medial aspect of the distal tibia. Historically, tibial shock has been measured in the vertical or axial direction only. However, the advent of triaxial accelerometers, i.e., accelerometers measuring acceleration along the vertical, anterior-posterior, and mediolateral axes, has allowed researchers to assess tibial shock in all three directions. The mediolateral and anterior-posterior components of tibial shock have not yet been linked to running related injuries, and little is known about their relationship to shock attenuation. Therefore, this literature review will focus on the vertical component of impact shock and its attenuation. However, the resultant component of impact shock will also be explored in this dissertation because its anterior posterior and mediolateral components may contain signal characteristics that demand non-trivial shock attenuation responses (Gruber et al., 2017).

2.3 Signal properties of Impact Shock

Impact shock is a time-varying, biological signal with properties in both the time and frequency domain. In the time domain, the amplitude of impact shock can range between 5-15 g's (Shorten and Winslow, 1992). Peak amplitude of impact shock occurs within the first 50 milliseconds after foot-strike but precedes the VGRF-IP by ~5 milliseconds (Hennig and Lafortune, 1991). In the frequency domain, research has consistently shown that the impact shock is composed of frequencies ranging between 3-50

Hz (Gruber et al., 2014; Gruber et al., 2017; Hamill et al., 1995; Shorten and Winslow, 1992). The amplitude of impact shock at each frequency describes the power or energy contained in that frequency. As an example, Figure 1 displays data from Gruber et al., 2014 and Hamill et al., 1995 wherein the impact shock signal at the tibia consists of two primary power or energy spikes at 4-8 Hz and 9-20 Hz corresponding to lower and higher freqeuncy shock, respectively. Lower frequency vibrations are related to the active motion of the body during ground contact whereas higher-frequency vibrations are related to the foot-ground impact.

2.4 Determinants of Impact Shock

2.4.1 Vertical touchdown velocity

Vertical impact force (VGRF-IP) and impact shock characteristics are both influenced by the lower leg and foot vertical touchdown velocity at ground contact (Bobbert et al., 1991; Bobbert et al., 1992; Gerritsen et al., 1995; Nigg et al., 1995; Whittle, 1999). Increases in foot and tibia velocity at touchdown is associated with an increase VGRF-IP magnitude (Gerritsen et al., 1995). Greater touchdown velocities indicate a larger momentum of the foot and lower leg during prior to contact. Consequently, a larger external impulse (i.e., area under the force-time curve) is required for deceleration. Due to a larger force applied over time (i.e., during the deceleration period), the generated VGRF-IP and potentially impact shock are augmented (Bobbert et al., 1991). An increase in touchdown velocity influences both time and frequency domain characteristics of impact shock, such that a 0.3 $m.s^{-1}$ increase in touchdown velocity increases peak impact shock by 4.3g in the time domain and increases the power contained across frequencies in the frequency domain (Lafortune et al., 1996). Thus, the impact velocity is crucial to both the time and frequency domain features of impact shock.

2.4.2 Knee flexion angle at ground-contact

Impact force and impact shock characteristics are dictated by knee joint flexion angle at ground contact (i.e., knee contact angle). At larger knee contact angles (i.e., more flexed), the higher-frequencies that compose the VGRF-IP are primarily a result of larger leg and foot segmental accelerations rather than those originating from the rest of the body (Bobbert et al., 1991). Further, a knee contact angle (i.e., flexion) of 40 degrees versus 0 degrees results in an increase in impact shock magnitude by 57% and increases the signal power contained in frequencies above 30 Hz (Lafortune et al., 1996). Evidently, knee contact angle alters time-domain variables and the frequency content of both impact force and impact shock. During running, larger knee contact angles may increase the signal power contained in frequencies greater than 8 Hz. Thus, runners with greater knee flexion contact angles, such as older runners (Bus, 2003b; Devita et al., 2016), may experience more signal energy of the higher frequencies associated with impact than younger runners. Greater power contained in higher frequencies may increase the reliance on passive shock attenuation mechanisms in older adults. In essence, knee flexion contact angles influence the frequency content of impact shock.

2.4.3 Sagittal plane foot & ankle angles at ground-contact

In addition to knee contact angles, sagittal plane foot and ankle joint angles at contact may influence the magnitude and frequency content of peak impact forces and impact shock. In the time domain, musculoskeletal modeling revealed that for every degree increase in foot plantarflexion contact angle (i.e., foot orientation relative to the ground), VGRF-IP increases by 85 newtons (Gerritsen et al., 1995). The effects of foot and ankle contact angle on impact force and impact shock magnitudes have also often been assessed in the context of foot-strike patterns during running. Per a meta-analysis, the primary kinematic variables differentiating forefoot from rearfoot strike patterns are sagittal plane foot and knee angles at foot strike. Forefoot strikers run with greater ankle plantarflexion and knee flexion contact angles than rearfoot strikers who display greater ankle dorsiflexion and knee extension at contact (Almeida et al., 2015). These kinematic differences are concurrent with kinetic differences between foot strike patterns: Forefoot runners (i.e., those who have greater ankle plantarflexion at foot strike) display smaller VILR and AVLR than rearfoot runners and a decreased, if not visually absent VGRF-IP. Impact shock is also influenced by foot strike pattern. Clansey et al. (2014) found that rearfoot runners who were provided real-time feedback training to reduce impact shock did so by transitioning to a more plantarflexed ankle position at foot strike. Due to this transition, impact shock reduced by ~31% post gait retraining.

Sagittal plane foot and ankle angles at contact influence the power contained in different frequencies of impact shock. Despite the visual absence of a prominent VGRF impact force peak in the time-domain, forefoot runners (running with greater ankle plantarflexion at foot strike) display impact-related higher-frequency content during overground running. Gruber et al. (2017). In an earlier investigation, Gruber et al. (2014) also found the frequency content of impact shock to differ between foot-strike patterns during treadmill running. Within the impact-related higher-frequency range (9-20 Hz), peak energy or power occurred at the lower end of the range for forefoot runners but at the higher end of the range for rearfoot runners. This indicates rearfoot runners likely experienced a greater degree of higher-frequency vibrations related to impact than forefoot runners. Within the lower frequency range (3-8 Hz), peak power was concentrated in similar frequencies for both foot-strike patterns. In summary, alterations to sagittal plane foot and ankle contact angles modify the frequency content of both impact force and impact shock.

2.4.4 Frontal plane foot & ankle angles at ground-contact

Frontal plane ankle and rearfoot orientation at ground contact also influences the magnitude and timing of peak impact force and impact shock (Nigg and Morlock, 1987; Perry and Lafortune, 1995). For instance, a restriction in rearfoot eversion or ankle joint pronation is associated with an increase in the VGRF-IP (Lafortune et al., 1994; Nigg and Morlock, 1987; Perry and Lafortune, 1995), peak impact shock, and AVLR (Perry and Lafortune, 1995). These findings suggest implications for the frequency content of impact shock. Increases in peak impact shock magnitude with restricted eversion could increase the signal power of the frequencies associated with impact. Alternatively, Derrick et al. (2002) suggests peak impact shock measured at the tibia may increase by ~20% when the rearfoot becomes more inverted at contact, as evidenced following a fatiguing run (Derrick et al., 2002). Collectively, the presented findings suggest frontal plane ankle and foot orientation at ground contact are important mechanical variables to account for when assessing impact shock magnitude and frequency content of shock. However, the relationship between frontal plane ankle and foot motion and the frequency content of impact shock is yet to be established.

2.4.5 Running velocity, stride length & stride frequency

Running velocity influences impact shock because longer stride lengths are concurrent with faster running velocities (Mercer et al., 2002). Likewise, running velocity is positively related to not just peak impact shock magnitude but also the power concentrated in both lower and higher shock frequencies (Mercer et al., 2002; Shorten and Winslow, 1992). Stride length increases with faster running speeds may therefore underlie the positive correlation of running speed to impact shock (Mercer et al., 2002). In their study, Mercer et al. (2005) found that the VGRF-IP and peak leg impact shock increased with running speed when stride length was allowed to vary. However, when stride length was constrained, impact shock was not significantly altered as running speed increased. The authors explain this result by suggesting lower extremity posture at ground contact may have been altered when stride length was allowed to vary but remained unaltered when stride length was constrained as speed increased. Thus, the changes in joint and segment angles at contact that coincide with changes in stride length may be the underlying factors influencing the magnitude of impact-related kinetic variables.

These results are supported by another study isolating the effect of stride length on impact shock that found increasing stride length at a given running velocity will increase the vertical velocity of the foot at touchdown (Derrick et al., 1998). Given that the vertical velocity of the foot is zero after touchdown, a larger tibial deceleration is required thereby increasing tibial shock. Stride frequency, which is coupled with stride length at given running velocity (i.e., running velocity = stride length * stride frequency), also influences the magnitude of impact shock. A recent systematic review found consistent evidence across ten studies that increased stride frequency (thereby decreased stride length) at a constant running speed was concurrent with reduced impact shock (Schubert et al., 2014). Increases in stride frequency resulted in increases in ankle plantarflexion at contact while peak knee flexion during stance decreased, potentially explaining the observed reductions in impact shock magnitude across the reviewed studies. Similarly, a 10% acute increase in stride frequency resulted in a 21.2% decrease in AVLR and a 16% decrease in VILR (Adams et al., 2018), both correlates of impact shock (Hennig et al., 1993; Van den Berghe et al., 2019c). Despite the aforementioned studies finding an inverse relationship between peak impact shock and stride frequency, impact shock may increase with stride frequency because knee flexion contact angle is positively related to stride frequency and negatively to stride length (Schubert et al., 2014). Both stride length and frequency influence the magnitude of impact shock via their influence on impact-relevant joint kinetic and kinematic parameters. The effects of stride length and stride frequency must be considered when assessing factors that influence the frequency content of impact shock.

In summary, the gait determinants of impact shock include joint and segment angles at ground contact, touchdown velocity of the lower leg and foot, and stride length and stride frequency which influence touchdown mechanics. Appendix 1 summarizes the findings of previous studies that explored the effects of these determinants on impact shock magnitude and its correlates, namely peak impact force and vertical loading rates of the vertical ground reaction force. Thus, the magnitude of impact shock experienced by runners may differ based on the nature of these determinants. However, whether and how each determinant dictates the frequency content of impact shock is unclear due to relatively less research performed in the frequency than the time-domain of impact shock.

2.5 Shock Attenuation During Running

2.5.1 General introduction

The impact shock generated at foot-ground collision contains frequency vibrations ranging between 3-50 Hz that are propagated proximally towards the head in the form of a shock or energy wave. Along the way, soft and rigid tissue like muscle, articular cartilage, and bone passively absorb the power or energy contained in higher-frequency vibrations (typically greater than 8 Hz) given their viscoelastic properties. These soft and rigid tissues are, therefore, passive shock absorption mechanisms. Shock absorption of lower frequencies (typically less than 9 Hz) is enabled via active shock absorption mechanisms like muscle actions that perform negative mechanical work and alter joint and segment positions, ranges of motion, and joint stiffness (Nigg et al., 1995). This active and passive shock absorption, or *shock attenuation*, permits only 8%-15% of the initial impact shock measured at the distal tibia to reach the head (Ratcliffe and Holt, 1997). As a result, shock attenuation protects proximal musculoskeletal structures from excessive accelerations, thereby maintaining vision and balance.

Just like impact shock, shock attenuation can also be examined in both the time and frequency domains. In the time domain, the amplitude of peak impact shock is the largest at the tibia and smallest at the head. Less than 4% of shock attenuation occurs between the shoulders and head (Ratcliffe and Holt, 1997), suggesting most of the shock is attenuated within the trunk and lower extremities. During running, the vertebral column attenuates shock, as demonstrated by a 10% increase in shock attenuation within the lumbar spine for every 1% increase in lumbar lordosis (lumbar spine curvature) (Castillo and Lieberman, 2018). However, the degree of shock attenuation that occurs within the lumbar spine is less than that occurring between the tibia and the lower back. This difference in shock attenuation between body segments may be because most active and passive shock absorption mechanisms are concentrated within the lower extremity, resulting in significant shock attenuation by the time the shock wave reaches the lower back (Castillo and Lieberman, 2018). In the frequency domain, the power (i.e., energy) contained in both lower and higher frequencies is reduced as the shock travels from the tibia to the head. The body is especially good at attenuating the power contained in higher frequencies. For instance, Lafortune et al. (1994) observed the presence of frequencies between 62.5 Hz – 100 Hz in the resultant ground reaction force signal but the absence of frequencies >30 Hz in the acceleration signal measured from the proximal tibia during running. Given that acceleration was measured at the proximal tibia, this finding indicates that the foot and lower leg may be responsible for the attenuation of power contained in frequencies exceeding 30 Hz. Such attenuation may occur due to the soft tissue in the heel fat-pad, as is the case during barefoot walking (Wearing et al., 2014), or soft tissue and bone in the tibia.

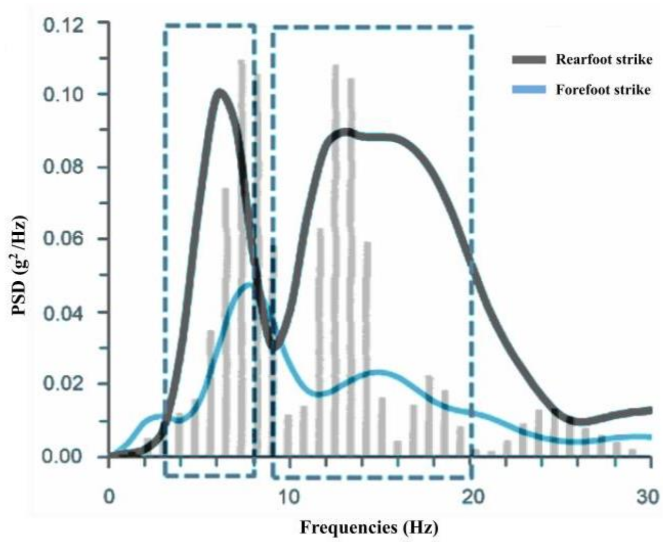

Figure 1. Tibial Power Spectral Density (PSD) adapted from Gruber et al., 2014 (rearfoot and forefoot strike) and Hamill et al., 1995 (shaded bars) demonstrating power concentrated in lower (4-8 Hz) and higher (9-20 Hz) frequencies.

2.5.2 Active shock attenuation and its mechanisms

Shock frequencies between 3-8 Hz is believed to be actively attenuated. Several mechanisms have been proposed as active attenuators of impact shock (Appendix 2). These mechanisms include negative mechanical work performed by joints (Hamill et al., 2014), knee flexion motion during ground contact (Edwards et al., 2012), joint stiffness (Hamill et al., 2014), and ankle motion in the frontal (Maharaj et al., 2016) and sagittal planes during stance (Gruber et al., 2014). However, the individual role of each proposed active mechanism in attenuating the energy contained in specific frequencies must be studied to identify the degree to which they contribute to active shock attenuation.

2.5.2.1 Negative mechanical joint work

Negative mechanical work is performed by lower extremity joints during early stance which contributes to energy absorption (Winter, 1983). Negative joint work is the mechanical work performed by the torque generated exclusively through eccentric muscle contractions; It is the time integral of mechanical power during the energy absorption phase of stance (i.e., the period between foot-strike and approximately 50% of stance). The energy absorbed in this manner is either dissipated as heat or used to perform positive work (i.e., energy generation) during concentric contractions (Garrett et al., 1987). Given that negative joint work is exclusively a function of eccentric muscle contractions, and active shock attenuation is proposed to occur via eccentric muscle contractions, negative mechanical work may aid active shock attenuation. For example, rearfoot runners experience lengthening of their ankle flexors (e.g., tibialis anterior) and forefoot runners lengthen their ankle extensors (e.g., gastrocnemius) after impact as the ankle either plantarflexes or dorsiflexes (based on foot strike pattern) to place the foot flat on the ground. At the same time, the knee extensor muscles (e.g., quadriceps) lengthen as the knee flexes to decelerate the body. Following this, the knee extensors continue to lengthen accompanied by lengthening of the ankle extensors as the knee continues to flex and the ankle dorsiflexes, respectively. These muscle actions enable us to decelerate our bodies after impact and prevent collapse into the ground. While lengthening or eccentrically contracting, these muscles perform negative mechanical work. In the process, mechanical energy is absorbed, and the power contained in lower frequencies of impact shock is potentially attenuated.

As stride length during running increases, there is a concomitant increase in the negative mechanical work done by muscles at the ankle, knee, and hip joints and the attenuation of energy contained in impact shock frequencies greater than 5 Hz (Derrick et al., 1998). Thus, shock attenuation may be achieved, at least in part, due to negative mechanical joint work (i.e., energy absorbed by muscles crossing the ankle, knee, and hip). However, the relationship of negative mechanical joint work to the attenuation of specific frequency bands (i.e., lower versus higher frequency) remains to be explored. Given that negative mechanical work is performed exclusively via active eccentric muscle contractions, we would expect negative mechanical work to aid the active attenuation of lower (3-8 Hz) frequencies to a greater degree compared with higher frequencies (>8 Hz). This information may have implications for the active attenuation demands experienced by runners of varying ages, considering lower extremity negative work declines with age (Devita et al., 2016). If negative work dictates a runner's capacity to attenuate shock actively, older runners may be less capable of active shock attenuation thereby increasing their reliance on passive attenuation mechanisms.

2.5.2.2 Knee flexion

Knee flexion is another important contributor to shock attenuation in both the time and frequency domains. An increase in knee flexion contact angle from 0 to 40 degrees is associated with a 57% increase in peak tibial shock but 45% decrease in peak head shock in the time domain, while also significantly increasing shock attenuation within frequencies above 5 Hz Lafortune et al. (1996). Similarly, Edwards et al. (2012) suggested that the knee serves as a low-pass filter of shock, allowing lower-frequency vibrations to be transmitted to the head while attenuating higher-frequency vibrations. Increasing knee flexion may shift the shock-attenuation responsibility away from passive and less deformable structures like bone to active, more compliant structures like muscles and tendons. Given that running is dynamic whereas the positions assumed by participants in Edwards et al. (2012) and Lafortune et al. (1996) were static, it remains to be seen whether their findings hold during running.

2.5.2.3 Sagittal plane ankle motion

Sagittal plane ankle motion contributes to energy absorption during early stance (Winter, 1983), presumably through eccentric muscle actions. Such energy absorption may reflect the attenuation of shock. For instance, rearfoot runners who land with a dorsiflexed ankle attenuate more energy across shock frequencies than forefoot runners who land in a plantarflexed ankle position (Gruber et al., 2014). While rearfoot runners may necessarily need to attenuate more shock due to more signal power at the tibia than forefoot runners, the mechanism for such attenuation is unclear but may involve greater negative mechanical work performed at the knee joint; ankle angle at ground contact influences sagittal plane knee motion as observed among rearfoot runners who exhibit less knee joint stiffness than forefoot runners (Hamill et al., 2014) + (Laughton et al., 2003). Consequently, rearfoot runners perform more negative mechanical work at the knee joint and thereby may attenuate more shock actively. Thus, sagittal plane ankle motion may independently and through its influence on knee motion contribute to the degree of shock attenuated actively.

2.5.2.4 Lower extremity & joint stiffness

Lower extremity stiffness and joint stiffness may also influence the degree of shock attenuated actively. Leg stiffness represents the magnitude of vertical ground reaction force generated as leg length changes; it is derived as the ratio of maximum vertical ground reaction force to change in vertical leg length (McMahon and Cheng, 1990). Since it depends on the vertical ground reaction force that represents whole-body center of mass motion, the independent contributions of each joint to lower extremity stiffness is difficult to ascertain. Joint stiffness is therefore used as an alternative and is modeled as a torsion spring, describing the joint moment (i.e., torque) produced per unit change in joint angle magnitude; it is calculated as the slope of the joint angle-moment plot using a linear regression equation (Hamill et al., 2014). Joint stiffness is related to the performance of negative mechanical work during the deceleration phase of running (i.e., approximately the first half of stance phase). Specifically, joint stiffness during the deceleration phase can be described as the negative work done by all torques acting at the joint per unit change in joint angle magnitude. Joint stiffness decreases during the deceleration phase of stance, increasing the negative mechanical work performed exclusively through eccentric contractions of the muscles spanning the joint. Decreased joint stiffness can result in greater mechanical energy absorbed by the eccentrically contracting muscles. Therefore, shock attenuation may also increase with a less stiff joint during deceleration. This phenomenon is observed among rearfoot runners who display greater ankle stiffness but lower knee stiffness than forefoot runners, and thereby also perform less negative mechanical work at the ankle but more at the knee (Hamill et al., 2014). These results suggest that joint stiffness may be negatively related to energy absorption and thereby shock attenuation.

2.5.2.5 Frontal plane ankle and rearfoot motion

Frontal plane ankle and rearfoot motion may dictate the degree of shock attenuated actively by the foot and ankle. For instance, active shock attenuation at the foot and ankle may be enabled in part via eccentric contractions of the tibialis posterior muscle-tendon unit during ankle pronation in early stance. Pronation is composed of subtalar eversion, ankle dorsiflexion, and forefoot abduction. Maharaj et al. (2016) assessed the electrical activity of and length changes to the tibialis posterior muscle-tendon unit during subtalar joint pronation when walking. During subtalar joint pronation in early stance, negative mechanical work was performed to absorb energy at the ankle joint. Concurrently, electrical activity in the tibialis posterior muscle increased and the muscle-tendon unit lengthened, presumably to resist pronation. Thus, the tibialis posterior muscle-tendon unit aided energy absorption by eccentrically contracting and performing negative work as the subtalar joint pronated. Through such a mechanism, the degree of frontal plane ankle motion may dictate the degree of shock attenuated actively. To examine the contributions of frontal plane motion to shock attenuation, the alterations to power contained in different frequencies must be examined as a function of ankle eversion range of motion or peak eversion magnitude during the deceleration phase of stance.

2.5.2.6 Active muscle actions

The role of active muscle action in shock transmission may be elucidated when examining the relationship between muscle fatigue and shock attenuation. Shock attenuation during running between the tibia and head is 12% lower after compared to before a fatiguing graded exercise test (Mercer et al., 2003a). These results suggest that fatigued muscles may not only increase impact shock magnitude but may also decrease the degree of shock attenuated via muscle actions. Conversely, however, Derrick et al. (2002) found that shock attenuation between the tibia and head increased from 74.5% to 77.5% as impact shock increased during an exhaustive run. A rise in shock attenuation with fatigue may be explained by the observed increase in peak knee flexion during stance. As discussed earlier, the quadricep muscles lengthen to enable greater knee flexion, and lengthening or eccentric muscle contractions increase energy absorption. Regardless of whether shock attenuation increases or decreases with fatigue, muscle actions play a role in shock attenuation given that their fatigue alters attenuation magnitudes. Since active muscle actions dictate joint and segment behaviors, examining the relationship of joint and segment behaviors to the attenuation of energy across shock frequencies will be valuable to better understand the musculoskeletal system's response to shock vibrations.

2.5.2.7 Stride length and stride frequency

Stride length and stride frequency determine running speed, which is in turn positively correlated with the degree of both lower (5-8 Hz) and higher (10-20 Hz) frequency shock attenuated between the leg and head (Mercer et al., 2002; Shorten and Winslow, 1992). At faster running speeds typically achieved via longer stride lengths, impact shock at the tibia is augmented, but shock magnitude at the head is maintained. Hence, shock attenuation increases in response to a larger tibial impact shock as stride length and velocity of impact increase (Derrick et al., 1998; Mercer et al., 2002). Unlike stride length, stride frequency is weakly associated with the degree of shock attenuated (Mercer et al., 2003b; Mercer et al., 2002) However, a recent review by Schubert et al. (2014) suggests that at higher stride frequencies, shock attenuation and energy absorption at the ankle, knee, and hip joints decrease. Concurrently, ankle plantarflexion at contact increases while peak knee flexion during stance decreases. Less knee flexion during stance may reduce negative work performed at the knee joint. A decrement in energy absorption may explain the observed reductions in shock attenuation across the reviewed studies. Evidently, both stride length and frequency influence the degree of shock attenuated. However, it is unknown how each of these spatiotemporal variables influences a runner's ability to attenuate shock across the frequency spectrum.

2.5.3 Passive shock attenuation and its mechanisms

2.5.3.1 General introduction

During running, the event of foot-ground contact generates an impact force. As a result, an impact shock is produced which travels up the body and is composed of energy concentrated in frequencies between 3-50 Hz (Gruber et al., 2014; Gruber et al., 2017). At the head, energy is predominantly contained in frequencies between 3-8 Hz (Derrick et al., 1998), suggesting frequencies greater than 8 Hz may be better attenuated by the body than lower frequencies. The higher frequencies are associated with the ground reaction force impact peak which occurs within the first 25 milliseconds of stance during heel-toe running (Bobbert et al., 1991). Due to such a quick occurrence of peak impact force and shock, the associated higher frequencies are likely not absorbed via active mechanisms such as muscle actions. They must therefore be attenuated via passive mechanisms like biological tissue due to their material and structural properties.

2.5.3.2 Bone, fat mass and lean mass

To our knowledge, only one study has investigated the relationship between body composition and impact shock attenuation (Schinkel-Ivy et al., 2012). Schinkel-Ivy et al. (2012) assessed the effects of leg tissues including fat mass, lean muscle mass, wobbling mass (fat mass plus lean mass), and bone mineral content on proximal tibial impact shock magnitudes, the slope of, and time to peak impact shock independent of muscle activation. Impact shock was quantified at the proximal end of the tibia. Therefore, any alterations to peak impact shock magnitudes as a function of body composition reflected the magnitude of shock attenuated. Peak impact shock and slope (rate of shock development to peak) were negatively correlated with leg lean mass, wobbling mass, and bone mineral content whereas wobbling mass was positively correlated with time to peak impact shock. Thus, as each of these tissue masses increases, so does shock attenuation within the tibia as indicated by lower peak impact shock and slope, and a longer time to peak. Attenuation of shock independent of muscle activation indicates that energy was dissipated passively due to tissue properties rather than active muscle actions. This study highlights the importance of assessing the relationship between tissue composition and degree of shock attenuation. However, the contribution of each tissue type to the attenuation of energy contained in different shock frequencies was not explored. Since the body effectively attenuates higher-frequency shock, identifying which of these tissue groups are related to higher-frequency attenuation will provide insight into our reliance on them for shock attenuation.

Bone is a known shock absorber of impulsive mechanical loading. The stress and strain-dependent behavior of bone during stress-relaxation experiments highlight viscoelasticity and thereby energy dissipation within the bone (Morgan et al., 2018). Stress refers to the intrinsic force applied by the bone to resist deformation, while strain describes the magnitude of the deformation itself. The structure and composition of bone including porosity, amount of bone mineral, orientation, and concentration of proteins such as Type I collagen lend to its mechanical property of viscoelasticity (Augat and Schorlemmer, 2006; Ojanen et al., 2015). In cortical bone, bone minerals, specifically hydroxyapatite, make up 70% of bone whereas collagen fibrils constitute 22% of bone. Bone mineral enables the effective resistance of compressive loads while collagen fibrils enable the resistance of high tensile strain (Augat and Schorlemmer, 2006), allowing bone to deform significantly and dissipate energy before reaching material failure. In trabecular bone, bone mineral is found to significantly contribute to the viscoelastic properties of bone that enables energy dissipation (Ojanen et al., 2015). The amount of bone mineral (grams) contained per square centimeter (cm^2) of bone, also known as bone mineral density, positively correlates with the elastic modulus (i.e., stiffness) of bone (Nobakhti and Shefelbine, 2018). Bone stiffness dictates the material's resistance to deformation. Therefore, greater stiffness due to a higher bone mineral density may also dictate the magnitude of energy dissipated within bone.

Animal models and cadaver studies have demonstrated the shock absorption capabilities of bone. In cadavers, 59% of the impact shock is attenuated between the tibia and femur. The degree of attenuation decreases to 54% upon the removal of cartilage and menisci, with a further decrease observed upon the removal of subchondral bone and intra-capsular soft tissue (Chu et al., 1986). Thus, shock attenuation decreases by a larger magnitude when subchondral bone versus cartilage and menisci are removed indicating that bone may be more effective at absorbing impact-related energy. Similar findings are observed in animal models. The magnitude of an externally applied peak force at the distal end of the interphalangeal joint in cows is smaller than that observed at the proximal end (i.e., point of application). This degree of attenuation is the greatest when subchondral bone is left to be the primary absorber of shock rather than cartilage and periarticular tissues (Radin et al., 1970). In the frequency domain, shock attenuation successively increases at higher frequencies in both alive and deceased rabbit tibia (Paul et al., 1978); minimal attenuation is observed between 3-18 Hz, 18% at 60 Hz, and almost 98% at 360 Hz. These results indicate bone is not an effective attenuator or lower frequency shock, but rather enables passive attenuation of high-frequency shock both in the presence and absence of muscle tissue. Extrapolating these results to human running, most of the impact-related energy is contained in higher frequencies between 9-20 Hz. Thus, bone may be relied on for higher-frequency shock attenuation up to 60 Hz, beyond which there is almost no energy relevant to running motion (Shorten and Winslow, 1992).

In addition to bone, soft tissue including both fat and fat free mass contribute to the attenuation of mechanical energy during non-running dynamic tasks. For instance, in addition to palm fat pad deformation, 70% of the total energy dissipated from the forearm during impact is accounted for by soft tissue deformation during a forearm karate strike (Pain and Challis, 2002). Similarly, 34.4% of impact-related energy is dissipated during simulated pendulum impacts to the heel fat pad and tibia when using a rigid body model, while 89.9% of the energy is dissipated when using a wobbling mass (fat mass plus lean mass) model (Pain and Challis, 2001). This finding highlights the importance of soft tissue mass to shock attenuation. Similarly, Boyer and Nigg (2006) found that the energy associated with shock vibrations immediately post impact was maximum between 10-25 Hz for the rectus femoris muscle at its proximal end but between 25-50 Hz at its distal end indicating shock attenuation. There was also significant attenuation of frequencies above 25 Hz between the distal and proximal ends of the rectus femoris. Thus, even within one soft tissue compartment, shock attenuation of higher frequency vibrations is apparent. The authors suggest attenuation to be a function of soft tissue deformation owing to viscoelastic properties of tissues as well as muscle tuning, i.e., pre-activation of muscles to reduce soft tissue vibrations. Support for the role of soft tissue deformation in shock absorption is provided by studies that investigated the magnitude of negative mechanical work (indicating energy dissipation) performed by these tissues in response to impact shock (Gittoes et al., 2006; Riddick and Kuo, 2016; van der Zee and Kuo, 2021; Zelik and Kuo, 2010). It should be noted that the composition of soft tissue was not segregated into fat mass versus lean mass in the above experiments. Fat mass (adipose tissue) and lean mass (muscle mass and connective tissue) differ in their material properties – adipose tissue is more compliant than muscle tissue (Guimarães et al., 2020). Consequently, their contribution to energy dissipation may be dissimilar.

2.5.3.3 Negative mechanical work via soft tissues

Negative mechanical work performed by soft tissue like the heel fat pad, viscera, and muscles also contributes to passive energy dissipation (van der Zee and Kuo, 2021; Zelik and Kuo, 2010). For instance, soft tissues perform aggregate negative work during running, the magnitude of which increases with increasing running speed (Riddick and Kuo, 2016). Soft tissue work during stance is approximately 19 joules, accounting for 25% of total body negative work performed during stance. Thus, both fat mass and fat-free mass, including connective tissue, likely play a role in shock attenuation during locomotion. However, precisely which frequencies these different types of tissues can attenuate is yet unknown. The reliance on each tissue type for shock attenuation coupled with each tissue's ability to absorb shock due to its material properties may dictate material failure risk

2.5.3.4 Natural frequency of biological tissues

The natural frequency of various musculoskeletal structures may also dictate their ability to dampen shock related vibrations. A structure's natural frequency is the frequency at which it will tend to vibrate when struck once due to its structural properties, specifically stiffness and mass. Stiffer materials or those with a smaller mass will have a higher natural frequency compared with more compliant or heavier materials. For instance, Bediz et al. (2010) found that participants with osteopenia and osteoporosis who had lower bone mineral density (indicative of lower bone mass) also had a lower natural frequency of tibial bone compared with healthy participants. A lower natural frequency of any material for a given input frequency ($i.e., relative\ frequency = \frac{input\ frequency}{natural\ frequency}$) implies that vibrations are less transmissible through that material or have a smaller amplitude of vibration, as seen in Figure 2B (adapted from Boyer and Nigg (2007)). Less transmissibility in turn indicates more damping or absorption of vibrations. In brittle bone like osteoporotic bone, increases in energy absorption due to a lower natural frequency and lower bone mineral density compared to healthy bone may push osteoporotic bone towards material failure. Alternatively, decreases in bone mineral density result in less bone stiffness. A less stiff (i.e., more compliant) bone will deform more easily compared to a stiff bone, thereby absorbing energy. For instance, in rat bone, mechanically stimulating training results in a 23% increase in bone mineral density accompanied by a 40% decrease in the damping factor of bone (Dimarogonas et al., 1993). This finding may indicate an inverse relationship between bone mineral density and the ability to damp vibrations, possibly due to increased stiffness. Bhattacharya et al. (2010) also found that individuals with osteoporosis had a higher natural frequency (indicating higher transmissibility) and lower viscous damping coefficient compared with healthy individuals. This may be due to a loss of Type-I collagen in the bone matrix which would reduce viscous damping due to a stiffer bone. Thus, both mass and stiffness dictate a material's damping response to vibration, either by altering its natural frequency, its ability to deform, or both. Figure 2A (adapted from Guimarães et al. (2020)) demonstrates that fat or adipose tissue is the most compliant, followed by muscle, cartilage, ligament, tendon and finally bone, in that order. Thus, for an equal amount of mass, fat tissue will have a smaller natural frequency compared with bone. These differences will have implications for the transmissibility of shock vibrations through each tissue type. During running, therefore, the ability of each tissue type to contribute to shock attenuation in the lower extremities will depend on both their mass concentration and their material stiffness.

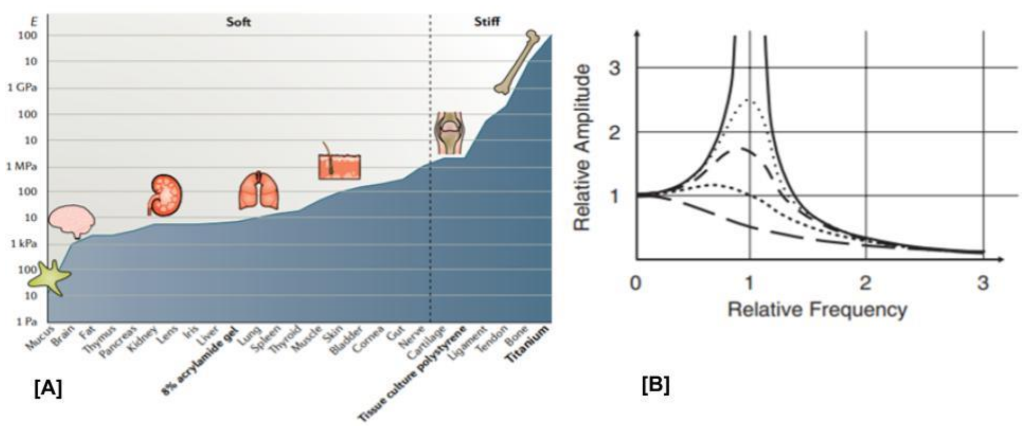

Figure 2A: Stiffness of various biological materials, sourced from Guimarães et al. (2020); **Figure 2B**: Resonance curves for input signals of a vibrating mass with different damping features, sourced from Boyer and Nigg (2007). For all degrees of damping (i.e. curves), the input frequency = natural frequency when relative frequency is "1". The relative amplitude indicates transmissibility of each relative frequency. When the natural frequency of a material decreases, relative frequency is >1, and transmissibility declines. Compliant or softer materials with smaller natural frequencies are therefore less transmissible, providing greater damping.

2.5.3.5 Tissue deformation

In addition to specific structural properties, energy dissipation is also enabled when biological tissues deform. When loaded or stressed, both soft and rigid biological tissue store energy via deformation (strain). This stored energy is known as strain energy. Strain energy is either released over time or dissipated in the form of heat when the applied stress is removed. The role of biological tissue as a passive shock attenuator, therefore, will depend on how much energy can be stored without material failure, and the proportion of that energy that can be absorbed or dissipated upon stress removal (Smeathers, 1989). Dissipation of energy within a biological material is represented using stress-strain or load-deformation curves which represent the viscoelastic nature of these tissues. Viscoelastic biological materials display both viscous and elastic behavior when loaded. Elastic elements will resist deformation (strain) based on the magnitude of applied loads or forces (stress), whereas viscous elements will resist deformation based on the rate of stress application as well (Whittle, 1999). The deformation of viscoelastic materials, and thereby energy dissipation will depend on the resistance afforded by elastic and viscous elements when external loads are applied. As an impact-related shock wave travels ground-up towards the head during locomotion, viscoelastic shock absorbers including bone, muscles, tendons, ligaments, articular cartilage, and other soft tissue structures within joint capsules either passively stretch or compress to dissipate energy. Such energy dissipation significantly reduces the amplitude of the shock wave by the time it reaches the head. Voloshin et al. (1981) demonstrated this phenomenon by studying the acceleration amplitude at the proximal tibia relative to the forehead in individuals with and without joint and/or bone degenerative disease as they performed multiple barefoot walking trials at 4.2 km/hr. Results revealed a significant attenuation of the shock wave as it traveled from the proximal tibia to the forehead in both groups, but the group with joint and/or bone degeneration attenuated 23% less shock compared with the healthy group. These results allude to the importance of bone and joint (soft tissues within joints) integrity to shock attenuation capabilities during locomotion. In terms of the frequency domain, Rubin et al. (2003) assessed the transmissibility of whole-body vibrations between 15-35 Hz within the human body. The findings revealed that the transmissibility of whole-body vibrations exceeding 25 Hz at the hip and spine relative to the feet reduced to 80% during erect stance, 60% during relaxed stance, and to 30% with 20-degree knee flexion. Thus, high-frequency vibrations were being attenuated effectively by tissues in the lower extremities and spine due to both their tissue properties during erect stance and their deformation with postural change during knee flexion. Hence, tissues within the body contribute to attenuation of various shock frequency vibrations, though the relationship of their material properties to their attenuation capabilities during running needs to be examined.

2.5.4 Summary

In summary, previous research has explored potential active and passive mechanisms of shock attenuation during various tasks. However, as seen in Appendix 2, the degree to which each mechanism contributes to the energy attenuation within different shock frequencies is unknown and must be established to understand tissue loads that may arise from a reliance on these mechanisms during running. This information will be especially useful when designing injury risk reduction interventions like gait retraining or footwear design that alter a runner's gait and may consequently alter their reliance on active and passive shock attenuation mechanisms.

2.6 Aging and Running Biomechanics

2.6.1 Gait mechanics and aging

The natural process of aging results in several age-related kinetic, kinematic, and spatiotemporal changes during running gait. Shorter stride length, higher stride frequency, and greater knee flexion contact angles are some kinematic and spatiotemporal factors that change with age (Bus, 2003b; Devita et al., 2016; Fukuchi et al., 2014; Kulmala et al., 2014) while also increasing touchdown velocity during each running step. Touchdown velocity dictates the momentum of the foot and lower leg as they contact the ground during each running step. Increases in touchdown velocity and thereby momentum will consequently heighten impact force upon ground contact and the subsequent impact shock generated. Thus, factors influencing touchdown velocity will also affect the magnitudes of impact force and impact shock. Bus (2003b) observed that when both younger (20-35 years) and middle-aged (55-65 years) runners ran at 3.3 $m.s^{-1}$, middle-aged compared with younger runners displayed significantly greater knee flexion at ground contact. Concurrently, middle-aged runners also experience a larger VGRF peak impact magnitude and maximum AVLR of the VGRF-IP. Greater knee flexion at ground contact could result in a smaller effective mass of the body, thereby increasing accelerations of segments nearest to the ground such as the tibia (Derrick, 2004). That is, tibial acceleration or shock may increase with a decrease in effective mass of the body as knee flexion increases. However, the subsequent impact shock generated by the impact force was not analyzed among either group, leaving the relationship between impact shock and age unexplored. The author also suggested that the higher impact peak and AVLR observed among the middle-aged group indicate a loss of shock attenuation capacity with age, however, shock attenuation was not quantified.

While impact shock and shock attenuation decrease with stride frequency, these variables are seen to increase with stride length (Schubert et al., 2014). However, the directionality of these relationships depends on joint posture at initial contact. For instance, if the knee flexion angle at contact increases, which it does at higher stride frequencies and shorter stride lengths (Heiderscheit et al., 2011;

Schubert et al., 2014), then impact shock may consequently increase due to a reduced effective mass. Devita et al. (2016) found that stride length was inversely correlated with age. Similarly, Kulmala et al. (2014) observed a shorter stride length and a higher stride frequency among older compared with younger athletes running at similar speeds. In the context of how stride length and frequency influence impact shock, a shorter stride length, higher stride frequency, and thereby a larger knee flexion contact angle with age may result in higher magnitudes of impact shock among older compared with younger runners. Indeed, older and middle-aged compared with younger runners have previously displayed greater knee contact angles, less peak knee flexion during stance, and less knee flexion-extension range of motion (Bus, 2003b; Devita et al., 2016) suggesting they may not only experience greater impact shock but may also attenuate a smaller degree of shock actively. Alterations to peak impact shock magnitudes with age implies that the frequency content of shock and thereby attenuation demands may change accordingly. Therefore, it is crucial to identify the associations between age, impact shock, and degree of shock attenuated actively and passively to assess if older runners are indeed at a greater risk for certain types of injuries (e.g., soft tissue versus bony) than younger runners.

Age-related alterations to knee joint kinematics potentially underlie age-related changes in mechanical energy absorption at the knee. Devita et al. (2016) findings that stride length decreases with age indicates stance-phase knee flexion may concurrently reduce, leading to less shock attenuation by active muscle actions among older compared with younger runners. Indeed, when running at preferred speeds, older runners (50-59 years old) display 36.8 degrees of peak knee flexion whereas younger runners (20-29 years old) display 40.6 degrees of peak knee flexion during running stance (Devita et al., 2016). Similarly, Bus (2003b) also observed diminished knee flexion-extension range of motion throughout stance among older compared with younger male runners. Notably, the older compared with younger cohort displayed 9.5 degrees less knee flexion range of motion during initial stance when shock absorption typically occurs. Smaller knee flexion magnitudes and less knee flexion-extension range of motion among older adults, especially during the shock absorption phase of stance, may result in reduced stretching of the knee extensor muscles, thereby performing less negative work, absorbing less energy, and attenuating less shock actively.

Not only do older compared with younger runners display reduced knee joint kinematics, but they also display diminished ankle joint function. Older runners (~60 years old) displayed 35.94 degrees of ankle dorsiflexion-plantarflexion excursion whereas younger runners (~30 years old) displayed 38.85 degrees of ankle excursion indicating the range of motion at the ankle reduces with age (Fukuchi et al., 2014). Decrements in ankle joint range of motion with age may indicate that musculature around the ankle joint like the gastrocnemius, soleus, tibialis anterior, and tibialis posterior perform less joint work, both positive and negative. Since negative joint work reflects the energy absorbed by muscles about a joint, diminishing negative work with age may suggest less active shock attenuation at the ankle joint among older adults. An age-related decline in energy absorption at the ankle joint, and in the lower extremity overall has indeed been evidenced. Total lower extremity negative power production is predicted to decrease by approximately 31% between the ages of 20-60 years, with an additional 47% reduction predicted after age 80. (Devita et al., 2016). Similarly, Kulmala et al. (2014) observed that younger athletes (~26 years old) absorbed 31% more power or energy via the ankle plantarflexors compared with older athletes (~78 years old) when running at similar speeds. Such age-related changes in ankle joint kinetics may reflect diminished muscle strength and power or a gradual loss of Achilles tendon elasticity with age among other factors. The reduced ability of muscles and tendons to perform negative mechanical work and thereby absorb energy may have implications for the aging body’s ability to attenuate impact shock. Therefore, the association between age, impact shock, and the degree of shock attenuation must be explored as a step towards understanding factors that may contribute to a runner’s ability to adapt to externally applied loads.

2.6.2 Age-related alterations to biological tissues

As discussed previously, eccentric muscle contractions are energy absorption mechanisms. Upon lengthening, or eccentrically contracting, muscles absorb energy. Subsequently, this energy is either returned when performing positive work or dissipated as heat when acting as a shock absorber (Lindstedt et al., 2001). The ability of muscles to eccentrically contract, generate force, and thereby perform negative mechanical work depends on several properties of muscle tissue. Muscle mass, i.e., the number, length, and size of muscle fibers dictates a muscle's ability to generate force and perform mechanical work, both positive and negative. It is well known that muscle mass diminishes with age, a phenomenon termed sarcopenia. Muscle size reduces by ~40% between ages 24-80 years, with especially large declines recorded post the 50th decade of life. Muscle fiber count also diminishes by ~35% between the ages of 52-77 years (Deschenes, 2004). Sarcopenia has implications for a muscle's ability to perform mechanical work. Longitudinal evidence suggests that concomitant with a 23.7%-29.8% decline in isokinetic knee extensor strength, a 16.1% reduction in muscle cross sectional area, and a 20% decrease in the percent concentration of type I (slow-twitch) muscle fibers, total mechanical work performed by the knee extensors diminishes by 2.3% (Frontera et al., 2000). Similarly, previous evidence suggests eccentric muscle strength, specifically knee extensor or quadriceps strength, declines at the rate of ~9 newtons per decade of life (Hortobágyi et al., 1995). Despite a lower rate of decline compared with concentric or isometric quadricep strength (~30 newtons decline per decade), a loss of eccentric muscle strength with age is reflected in the diminished capacity to perform negative mechanical joint work and thereby absorb energy actively. Consequently, older compared with younger adults may rely less on active shock attenuation mechanisms, attenuating less lower-frequency shock. However, this hypothesis needs to be confirmed.

Due to inverse relationships of age to negative joint work, joint ranges of motion, and eccentric muscle function, reliance on passive mechanisms to attenuate shock may increase with age. Components of lean body mass like bones, muscles, tendons, and ligaments undergo morphological and mechanical changes with age. For example, knee extensor muscles are more compliant among older compared with younger adults (Kubo et al., 2003b). Similarly, older adults, both runners, and non-runners, display a less stiff quadriceps femoris tendon and aponeurosis compared with younger adults (Karamanidis and

Arampatzis, 2006). Greater compliance increases the energy storage capacity of tissues and may therefore aid shock absorption. Older adults also tend to have increased concentrations of non-contractile (fat or connective tissue) versus contractile tissue. While more fat compared with muscle mass reduces older adults' force production capacity (Williams et al., 2002), it may aid shock attenuation due to the negative work performed by soft tissue at foot-ground contact (van der Zee and Kuo, 2021; Zelik and Kuo, 2010). However, older adults, both men, and women tend to have reduced muscle thickness of the vastus lateralis (Kubo et al., 2003a) and gastrocnemius muscles compared to younger adults (Karamanidis and Arampatzis, 2006). Vastus lateralis muscle cross-sectional area and the number of muscle fibers also decrease with age (Lexell et al., 1988), resulting in muscle mass decrements. Thinner muscle tissue and fewer muscle fibers, i.e., less muscle mass, may reduce its passive shock attenuation or energy dissipation capabilities. Therefore, increased reliance on connective tissue unable to meet attenuation requirements may result in tissue overload, and eventually heightened injury risk. Examining differences in the ability of older versus younger runners to attenuate energy contained in frequencies >8 Hz is therefore crucial to understanding why they may be at differential risk for RRI development.

Bone mineral density (BMD) at various anatomical regions like the femoral neck and trochanter among others is seen to decline with age (Burger et al., 1994; Warming et al., 2002); depending on anatomical location and sex, yearly declines in BMD range between 0.22%-1.07%. As discussed previously, bone tissue is a potential shock absorber due to its viscoelastic properties that derive from collagen content in bone, bone mineral density, and bone stiffness among other factors (Augat and Schorlemmer, 2006; Ojanen et al., 2015). Bone mineral density positively correlates with bone stiffness (Nobakhti and Shefelbine, 2018). Therefore, the material stiffness of bone may decline as its mineral density decreases with age. A less stiff bone may deform to a greater extent, thereby dissipating more energy compared with a stiff bone. Hence, greater passive shock attenuation of energy or power in higher frequency vibrations may be observed among older compared with younger adults. However, cortical bone specifically loses strength and becomes more brittle with age (Tommasini et al., 2005).

Consequently, a greater reliance on bone for passive attenuation may also mean heightened bony injury risk among older adults.

2.7 Running Experience and Running Biomechanics

2.7.1 Gait mechanics and running experience

So far, only one study has explored the relationship between tibial shock and running experience. Camelio et al. (2020) found that novice runners who underwent a 3-week training program (~20% increase in weekly volume) displayed an increase in peak impact shock post-training. 31.90% of this increase was explained by training volume. Although this correlation was not significant, a positive relationship between weekly running volume and peak impact shock highlights the possibility that running experience may be positively related to peak impact shock, and potentially the energy (or power) contained in different shock frequencies though this is yet to be substantiated.

Evidence regarding impact-relevant kinetic and kinematic differences between runners with varying running experience is lacking. Of the few existing studies on this topic, some studies have found no significant differences in running biomechanics between groups (Agresta et al., 2018; Schmitz et al., 2014), while others suggest group differences do exist (Boyer et al., 2014; Clermont et al., 2017; Clermont et al., 2019; Hafer et al., 2019b; Harrison et al., 2021; Quan et al., 2021a; Quan et al., 2021c). For instance, kinematics like impact-relevant sagittal plane knee angle and foot-strike angle, and correlates of impact shock such as VILR, AVLR, and the VGRF-IP were not previously associated with running experience (Agresta et al., 2018). Similarly, among a group of female runners, VGRF-IP and AVLR were not significantly different between more experienced recreational runners and active novice runners (Schmitz et al., 2014). These results suggest that running experience may not influence impact shock magnitude, though impact shock was not measured in either study. Similar magnitudes of impact shock do not imply that injury risk among more and less experienced runners is the same. The susceptibility to injury may depend not only on the magnitude of impact shock but also on the frequency content of shock. The power or energy contained in different shock frequencies may change with running experience. Attenuation demands on the body may consequently vary. Eventually, the ability of each group to attenuate the power contained in various frequencies may depend on the integrity of their shock attenuation mechanisms like eccentric muscle actions or bone mass. The integrity of these attenuation mechanisms may in turn be dictated by exposure to mechanical loading, i.e., exposure to running. Recently, Quan et al. (2021a) found that although the VGRF-IP was similar between novice and experienced runners, novice runners displayed a smaller AVLR. A smaller rate of force application to the body among novice compared with recreational runners may lead to differences in the power contained in various shock frequencies between groups. Therefore, it is necessary to identify if power contained in various impact shock frequencies and their attenuation differ between runner sub-populations as a function of running experience.

Studies using machine learning techniques have been successful at classifying runners into high and low running experience groups based on certain running kinematic parameters which were also found to significantly differ between groups. For example, upon successfully classifying runners into competitive and recreational, between-group differences revealed that competitive runners, who had spent more years running, displayed greater knee flexion magnitude throughout the stance phase compared to recreational runners (Clermont et al., 2017). This finding suggests impact shock and active shock attenuation may increase with running experience. Kinetics were, however, not measured in this study. Similarly, among male runners, higher mileage compared with lower-mileage runners displayed a significantly more flexed knee during late swing and throughout stance suggesting knee flexion at contact may have also been greater among this group (Clermont et al., 2019). Considering a more flexed knee at contact and during stance may increase impact shock and active attenuation, respectively, it is possible that both impact shock and active shock attenuation increase with running experience, at least among male runners. Similarly, Suda et al. (2020) classified running experience levels based on foot-ankle kinematics during barefoot running. Although impact-related ground reaction force variables were similar between groups, group comparisons revealed less experienced runners displayed greater foot plantarflexion and toe dorsiflexion angle magnitudes throughout stance compared to more experienced runners. These movement patterns may imply that foot structures such as the toe flexors, plantar fascia, and tendons experienced more tension and thereby passive energy dissipation among less experienced runners. Hence, it can be hypothesized that less experienced compared with more experienced runners are more reliant on passive rather than active shock mechanisms to attenuate shock.

Kinematic differences between running experience groups have been found in secondary planes of motion as well. Recreational runners display significantly more frontal and transverse plane ankle range of motion among than competitive runners (Quan et al., 2021b), suggesting potentially greater energy dissipation at the ankle among less experienced runners. Clermont et al. (2019) found that lower compared with higher-mileage runners displayed a significantly more abducted foot during mid-swing. The authors suggested that foot abduction angle during mid-swing may be predictive of VGRF-IP magnitude, a correlate of impact shock, at ground contact due to foot contact posture. However, the mechanisms via which mid-swing abduction angles influence impact shock magnitudes and frequencies remains unknown. Similarly, among female runners, knee abduction angles and excursion, and knee internal rotation angles were significantly greater among novice compared with recreational runners throughout stance (Harrison et al., 2021). Previously, lower mileage runners have displayed greater knee adduction excursion during initial stance (Boyer et al., 2014) which includes the event of foot-ground impact. Not much is known about the contribution of frontal and transverse plane knee motion to shock attenuation, but knee joint excursion in any plane could influence eccentric muscle actions that may contribute to shock attenuation. Recent evidence suggests novice runners may indeed run with greater ankle and knee joint excursions in the sagittal plane (Quan et al., 2021a). Greater joint excursions could indicate an increase in negative work via eccentric muscle actions. Negative work is an indicator of energy absorption, therefore novice compared with recreational runners may actively attenuate more shock.

Experienced runners run with a more everted ankle and internally rotated tibia throughout stance compared with novice runners (Harrison et al., 2021). Greater ankle eversion at ground contact is previously found to both increase (Derrick et al., 2002) and decrease (Nigg and Morlock, 1987; Stacoff et al., 1988) peak impact force. A heightened peak impact force could increase impact shock in the time domain. However, it could also increase the power contained in higher frequency vibrations in the frequency domain. Alternatively, lower impact shock with greater eversion could mean heightened shock attenuation at the ankle. Ankle eversion and tibial internal rotation allow the knee to flex. Hence, differences in the coordinated motion of ankle eversion-tibial internal rotation between experience groups could have implications for between-group differences in knee flexion magnitude as well. Knee flexion magnitude is positively related to energy attenuation within frequencies greater than 5 Hz (Edwards et al., 2012; Lafortune et al., 1996). Previous studies have found that joint and/or segment coordination patterns and variability differ between running experience levels. Less-experienced runners display significantly lower coordination variability compared with more-experienced runners for the sagittal thigh versus shank and shank versus foot couples during early and mid-stance (Hafer et al., 2019a). Although the relationship of impact shock to coordination patterns and variability is unknown, less variable coordination between the thigh and shank, and shank and foot, may influence the ability of the knee and ankle to respond to shock. Variability is suggested to represent the body's ability to respond to perturbation. A more variable system might be more adaptive in its response to perturbation. Thus, more experienced runners with their greater coordination variability may be more effectively able to respond to shock by adapting their coordination.

In addition to kinetic and kinematic differences, spatiotemporal parameters relevant to impact shock also vary with running experience. Highly trained runners display a higher stride frequency and shorter stride length compared with novice and untrained runners (De Ruiter et al., 2014; Gómez-Molina et al., 2017; Slawinski and Billat, 2004). As Schubert et al. (2014) suggest, increased stride frequency (thereby decreased stride length) at a constant running speed could result in decreased impact shock, shock attenuation, and energy absorption at the ankle, knee, and hip joints. García-Pinillos et al. (2019), however, found the opposite results whereby elite runners ran with a lower stride frequency and longer step lengths compared with novice runners. It stands to reason that differences in impact shock and shock attenuation may exist between running experience groups owing to known impact-relevant spatiotemporal differences. Identifying what these group differences are, in both the time and frequency domains, will enable us to better understand how a runner’s body is responding to shock per their current musculoskeletal health.

2.7.2 Biological tissue adaptation with exposure to running

Shock attenuation depends not only on the active modulation of gait patterns, muscle activation, and muscle contractions, but also on the tissue properties of various musculoskeletal structures. For example, soft tissue mass in the shank and the heel fat pad significantly contribute to energy dissipation during impact (Pain and Challis, 2001). Soft tissue properties like viscoelasticity enable them to dissipate energy and thereby potentially attenuate shock vibrations. During locomotion, soft tissues may also passively perform negative work at foot-ground contact, thereby absorbing energy (van der Zee and Kuo, 2021; Zelik and Kuo, 2010). Therefore, any change to soft tissue material properties, including mass, may have implications for their energy dissipation abilities. A meta-analysis of 35 studies revealed body fat percentage decreases as healthy inactive individuals are exposed to running training (Hespanhol Junior et al., 2015). In support of these findings, a 10-week beginner running program found decreased body-mass index and body fat percentage post-training among the novice runners’ sample (Stevinson et al., 2020). Likewise, 12-weeks of both interval and continuous running training is found to reduce body fat percentages among moderately fit but untrained individuals (Thomas et al., 1984). Among recreational runners, those who completed a greater number of marathons in their lifetime had a lower body fat percentage compared to those with fewer marathon finishes (Nikolaidis et al., 2021). Consequently, the contribution of body fat to passive attenuation may also decrease with running experience. On the other hand, lean body mass may increase with running training (Ravnholt et al., 2018). Luden et al. (2012)

observed an ~8% increase in the vastus lateralis type I muscle fiber size among novice runners post a 16-week marathon training program. Increases in muscle fiber size among larger muscle groups like the quadriceps could augment muscle mass and its contribution to passive shock attenuation. Morphology and mechanical properties of other connective tissue like the Achilles tendon are also responsive to mechanical loading. Experienced endurance runners display less elongation and compliance of the vastus lateralis muscle-tendon complex compared with untrained individuals when performing a vertical jump (Kubo et al., 2000). A recent narrative review revealed runners have a stiffer Achilles tendon with a larger cross-sectional area compared with non-runners (Yu et al., 2022). Similarly, habitual runners' Achilles tendon has a 36% larger cross sectional area at its distal end than non-runners (Magnusson and Kjaer, 2003). Due to a stiffer, less compliant tendon, experienced runners compared with novice or non-runners may be transmitting more muscle force than dissipating energy. These morphological and mechanical changes in tendon properties with running experience suggest the reliance on passive shock attenuation via tendon tissue may decrease with increased running experience. Such an inverse relationship between running experience and passive shock attenuation demands could heighten injury risk among less experienced runners because their tendons have a smaller cross-sectional area (Rosager et al., 2002). A smaller tendon cross sectional area is typically accompanied by a less stiff tendon, indicating a reduced capacity to resist deformation upon force application and thereby possibly heightened injury risk.

The material properties of bone like its mineral density, stiffness, and viscoelasticity dictate its ability to deform and dissipate energy, especially that contained in higher frequency shock (Paul et al., 1978). The material properties of both soft and rigid tissues are altered through exposure to axial mechanical loading such as running. For instance, female elite athletes engaging in non-aquatic sports like football display a higher bone mineral density in the proximal femur compared with sedentary controls and female elite athletes engaging in aquatic sports like swimming (Bellver et al., 2019). Similarly, untrained individuals who underwent 7 weeks of high-intensity intermittent running training displayed increased bone mineral density (Ravnholt et al., 2018) indicating osteogenic responses of bone to high-impact mechanical loading. Among runners, volumetric bone mineral density at the proximal tibia increases with weekly running volume up until 30 km/week, after which it plateaus. Moreover, bone mineral density is positively associated with peak impact shock measured at the proximal tibia for higher weekly running volumes (>30 km/week) (Dériaz et al., 2010). In terms of shock attenuation, these findings could mean that as running exposure increases, bone mineral density increases. However, despite increased bone mineral density, passive shock attenuation within the tibia decreases with running experience. This phenomenon is plausible since bone mineral density is positively associated with the elastic modulus of bone (Nobakhti and Shefelbine, 2018). As bone mineral density increases, the bone becomes stiffer and may thereby deform less, dissipating smaller amounts of energy. Hence, passive shock attenuation may be negatively related to running experience partly due to increased bone mineral density.

2.8 Impact Shock, Shock Attenuation, and Running Related Injury

The relationship of impact shock to running-related injury development has not been as extensively studied as that of its correlates, VGRF-IP, and vertical loading rates (average vertical loading rate: AVLR, instantaneous vertical loading rate: VILR). Of the few studies exploring the role of impact shock as a potential RRI risk factor, one cross-sectional study found that runners with a history of tibial stress fracture display larger magnitudes of tibial shock compared to control runners when running overground (Milner et al., 2006). In the same study, those who ran with larger magnitudes of tibial shock were more likely to be classified as injured with 70% accuracy.

Correlates of impact shock, specifically the VGRF-IP and vertical loading rates have been extensively studied as RRI risk factors. Previously, runners with a history of one or more RRIs displayed larger magnitudes of the VGRF-IP and vertical loading rate compared to runners with no history of RRIs (Hreljac et al., 2000). Similarly, female runners with a history of plantar fasciitis are seen to display larger VILR compared to age and mileage matched control runners (Pohl et al., 2009). The VGRF-IP was not, however, significantly different between groups. Similarly, (Bennell et al., 2004) found no differences in the VGRF-IP between female runners with and without a history of tibial stress fracture. These results are corroborated by a recent systematic review and meta-analysis of thirteen studies exploring the relationship of tibial and metatarsal stress fracture history to ground reaction force variables (Zadpoor and Nikooyan, 2011). Meta-analysis results revealed no significant differences between the stress fracture and control groups for the VGRF-IP, but significantly higher average and VILR for the stress fracture group. The studies reviewed so far were either cross-sectional or retrospective in nature, making it difficult to evaluate the contributions of impact-related ground reaction force variables to RRI risk. A systematic review including only prospective studies found strong evidence suggesting VGRF-IP is not significantly related to RRI development (Ceyssens et al., 2019). Vertical loading rate variables, on the other hand, may be a potential RRI risk factor though prospective evidence is limited and conflicting (Ceyssens et al., 2019). A randomized control trial with a six-month follow-up period did not find VGRF-IP and loading rate to be significantly associated with prospective injury development (Malisoux et al., 2022). Similarly, a pooled meta-analysis of five high-quality prospective studies provided moderate evidence for no significant differences in the VGRF-IP between injured and uninjured runners (Vannatta et al., 2020). In the same study, a pooled analysis of two high-quality studies on female recreational runners (Davis et al., 2016; Napier et al., 2018) provided conflicting evidence for the role of AVLR and VILR in RRI development. However, a randomized control trial utilizing gait retraining to reduce vertical loading rates found significantly fewer prospective RRIs during a one-year follow-up in the intervention compared to the control group (Chan et al., 2018). More such gait retraining studies with an injury tracking follow-up period are needed to clarify the role of vertical loading rates and peak impact force in RRI development with current evidence unable to provide conclusive findings.

Despite being limited, existing evidence suggests vertical loading rate rather than peak impact force may be a potential RRI and is found to be higher among both retrospectively and prospectively injured compared with uninjured runners (Chan et al., 2018; Van der Worp et al., 2016). Since peak impact shock is positively correlated with vertical loading rates (Hennig et al., 1993; Tenforde et al., 2020; Van den Berghe et al., 2019c), it stands to reason that it may be higher among injured compared with uninjured runners, and thereby a potential RRI as also previously suggested (Milner et al., 2006). Older compared with younger runners previously displayed larger AVLR (Bus, 2003b). Based on the positive association between vertical loading rate and impact shock, older runners may display larger magnitudes of impact shock. Heightened demand on the musculoskeletal system arising from impact-related shock with age may be a reason for the RRI rate being higher among older compared to younger runners (McKean et al., 2006). Novice runners are also at a greater risk of injury development compared with recreational runners (Linton and Valentin, 2018; Videbæk et al., 2015). However, only one study has directly measured the relationship between running experience and impact shock, finding a positive association between the two variables (Camelio et al., 2020). As such, not much is known about how age and running experience relate to impact shock. To better understand why older and novice runners develop more RRIs, it is worth exploring how RRI risk factors like impact shock vary with age and running experience.

The frequency content of impact shock potentially dictates which shock attenuation mechanisms are most relied upon. For a comprehensive understanding of impact shock as an RRI mechanism, therefore, impact shock must be assessed in the frequency domain. High-frequency shock is supposedly attenuated by passive shock attenuation mechanisms like soft and rigid musculoskeletal tissues. Hence, if older compared to younger runners display a larger peak impact shock with more energy contained in frequencies greater than 8 Hz, the shock attenuation demands on their muscle mass, bones, ligaments, tendons, and adipose tissue may increase. Considering that the integrity of several of these structures declines with age, injury risk to these structures may subsequently augment. The frequency content of impact shock, and thereby shock attenuation demands may also influence the type of injuries developed within each runner sub-population. For example, older runners who may experience greater energy content in higher shock frequencies may thereby rely on bone tissue to attenuate it, subsequently experiencing a greater proportion of bony injuries. Alternatively, younger runners may experience a greater proportion muscular injuries if they experience larger energy content within lower-frequency shock and must thereby attenuate it actively via eccentric muscle contractions. Similarly, recreational runners with potentially larger magnitudes of impact shock and higher-frequency energy compared to novice runners may experience more bony injuries. Hence, exploring the frequency content of impact shock and the degree of higher and lower-frequency shock attenuation among each runner sub-population will provide actionable insights to injury risk reduction interventions. These interventions, be it footwear design or training regimens directed towards improving tissue health, can be customized for each sub-population based on their shock attenuation demands and capabilities.

2.9 Summary

Kinematics and kinetics relevant to impact shock magnitude and frequencies are age and running experience dependent. Particularly, joint angle orientations at ground contact, stride length, stride frequency, and VGRF load rates are altered with age and running experience. Not only might these alterations influence the magnitude and frequency content of impact shock, but they may also dictate the degree of shock attenuated actively. Mechanical and morphological properties of various musculoskeletal tissues that lend to the passive attenuation of shock also change with age and exposure to mechanical loading. Yet, little is known about the independent and combined effects of age and running experience on the frequency content of impact shock and the degree of frequency domain shock attenuation. Additionally, the relationship of select active and passive shock attenuation mechanisms to the degree of shock attenuated within various frequency ranges during running is unclear. Overall, understanding which mechanisms enable the attenuation of lower and higher frequency energy, and how the energy contained and absorbed within different shock frequencies varies with age and running experience will allow for effective injury risk reduction interventions among runner sub-populations.

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3 CHAPTER 3: Methods

3.1 General Introduction

This chapter details the characteristics of each independent and dependent variable for each aim and states the methods used to measure and quantify them. As stated previously, the specific aims are:

Aim 1: Examine the associations of lower extremity and gynoid fat mass and lean mass, and lower extremity and pelvis bone mineral density and content to the degree of shock attenuation within frequency bands of 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4) among runners.

Aim 2: Examine the associations of sagittal knee and ankle angles at foot-ground contact, knee and ankle joint stiffness, joint range of motion at the ankle (frontal and sagittal planes) and knee (sagittal plane), peak ankle eversion, peak knee flexion, negative joint work and power at the ankle and knee during the deceleration phase of running to the degree of shock attenuation within frequency bands of 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4) among runners.

Aim 3: Estimate the independent effects of age and running experience on impact shock and degree of shock attenuation within frequency bands 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4) among runners.

3.2 Sample size estimates

All a-priori power calculations were performed in G*Power (Release 3.1.9.7, 2020) for linear regression.

3.2.1 Aim 1

A-priori power calculations using data on lower leg wobbling mass and bone mineral content from Schinkel-Ivy et al. (2012) revealed a sample of N=22 was required to achieve 80% power with an error probability of 0.05 (calculated effect size $f^2 = 0.94$). Therefore, N=38 runners were recruited to improve the generalizability of results across the runner population.

3.2.2 Aim 2

A-priori power calculations using shock attenuation data from Edwards et al. (2012) revealed a sample of N=29 is required to achieve 80% power with an error probability of 0.05 (calculated effect size f^2 = 1.00). Therefore N=58 runners were recruited to make the results generalizable across the runner population.

3.2.3 Aim 3

For the independent effects of age, a-priori power calculations revealed a total of N=40 subjects were required to obtain 80% power with an error probability of 0.05 using the vertical ground reaction force impact peak effect size from Bus (2003a) (calculated Cohen's d = 1.060). Therefore, N=44 runners were recruited. For the independent effects of running experience, a-priori power calculations revealed a total of N=19 subjects were required to obtain 80% power with an error probability of 0.05 using the peak tibial shock data from Camelio et al. (2020) (calculated f^2 effect size = 0.46). Therefore, N=38 runners were recruited to include a wide range of running experiences.

3.3 Inclusion and Exclusion Criteria

3.3.1 General recruitment criteria across all three aims.

A total of 58 runners were recruited for this dissertation. Inclusion criteria required subjects to 1) be between 18-30 years old (young group) or 45-65 years old (middle-aged group), 2) be either novice or recreational runners (defined below), 3) have no history of lower back or lower limb orthopedic surgery in the past 10 years, 4) if an injury was developed in the past 6 months, they must have returned to normal running training volume for ≥8 weeks prior to enrollment, and 5) be free of chronic illnesses that may disallow engagement in ≥15 minutes of low to moderate-intensity physical activity. Tobacco users and individuals with a history of cardiovascular disease, pulmonary disease, or neurological dysfunction were excluded from the study. Women pregnant at the time of the study were also excluded. Inclusion and exclusion were assessed via questionnaire responses obtained during telephone or email screenings. Additionally, subjects' responses to the American College of Sports Medicine's modified Physical

Activity Readiness Questionnaire (PAR-Q) (Appendix 3) also dictated eligibility. All testing procedures were approved by the Indiana University Institutional Review Board (IRB protocol 10585, Appendix 4). All subjects gave written informed consent before participation (Appendix 4).

3.3.1.1 Definition of running experience groups

The literature lacks a consensus on definitions for different levels of running experience. Establishing such definitions is difficult as experience can be viewed from various perspectives. For instance, a skill-based perspective would be based on how exposure to running affects movement coordination patterns (Boyer et al., 2014; Hafer et al., 2019b), while a tissue-adaptation perspective would focus on how running influences mechanical loads on body tissues and their adaptation to those loads (Luden et al., 2012; Ravnholt et al., 2018; Yu et al., 2022). Both perspectives are likely to influence shock attenuation and magnitude. Although novice and recreational runners are commonly classified, their definitions vary across studies. To address this, Honert et al. (2020) formulated a consensus definition based on input from 142 experts, including running coaches, physicians, scientists, and running shoe bloggers. According to this definition, novice runners are those who have been running for less than a year, with a weekly mileage of 3-13 miles and running three or fewer times per week. Recreational runners, on the other hand, are those who have been running for more than a year, with a weekly mileage of 9-30 miles and running one to five times per week. These consensus definitions were adapted for recruitment purposes since they are the only established consensus definitions to date.

3.3.1.2 Definition of age groups

Previous studies examining the relationship between age and gait mechanics have used varying definitions for group classification. The age-range definitions for young individuals have varied within 18-40 years, while that for middle-aged individuals has varied within 45-70 years (Boyer et al., 2012; Boyer et al., 2017; Krupenevich et al., 2021; Melaro et al., 2020; Phinyomark et al., 2014). Additionally, a MEDLINE/PubMed search using Medical Subject Headings indicated that “middle aged” is defined as individuals between 45-64 years old (PubMed). Therefore, runners between 18-30 years old were recruited as younger runners, while runners between 45-65 years old were recruited as middle-aged runners to enable comparisons to previous biomechanics studies.

3.3.2 Aim 1

38 runners from the larger pool of 58 runners completed the body composition scan and were therefore included in this analysis. Dual x-ray absorptiometry (iDXA) was used to assess body composition, which required a separate IRB protocol (IRB Protocol 10953, Appendix 7) from the gait analysis procedures of aims 2 and 3 (IRB Protocol 10585, Appendix 4). This was done to allow subjects the option of refusing the radiation exposure of the iDXA scan. However, to be included in the procedures for aim 1, subjects had to be included in the gait analysis protocol (IRB Protocol 10585, Appendix 4). If subjects chose to complete the iDXA scan on a separate day from the gait analysis, they were excluded if they experienced pain or injury since their visit for the gait analysis, as this may have resulted in significant changes to or stoppage of running training. The Indiana University Institutional Review Board (IRB Protocol 10953, Appendix 5) approved all testing procedures for this additional iDXA protocol, and all subjects provided written informed consent before participation.

3.3.3 Aim 2

For this aim, 57 runners were recruited without classifying them based on running experience or age group. Instead, the inclusion criteria required them to be between 18-65 years old and have been running between 3-30 miles per week, while all other inclusion and exclusion criteria remained the same as in aim 1. The testing procedures were approved by the Indiana University Institutional Review Board (IRB protocol 10585, Appendix 4), and all subjects provided written informed consent before participating.

3.3.4 Aim 3

Based on the general recruitment criteria, 58 runners were recruited and classified into three groups: middle-aged (n=20), young recreational (n=24), and young novice (n=14), using the definitions of age and running experience.

Four subjects had exactly one year of running experience. Since the consensus definition by Honert et al. (2020) did not indicate which group these runners should be classified into, they were classified as either novice (n=3) or recreational (n=1) based on if their weekly mileage values were within ±1 standard deviation of the group mean calculated upon excluding them. Additionally, eight runners qualified as recreational per the “years” criteria but not “weekly mileage” criteria while the opposite was true for one runner. These runners were classified as recreational if they had been running for >1 year (n=8) and novice if they ran for <1 year (n=1) because they did engage in regular running and previous studies have classified novice runners as those with no regular running experience (Harrison et al., 2021; Quan et al., 2021a).

3.4 Study Design

For all aims, a cross-sectional study design was employed to analyze the associations between exposures and outcomes at the same time point. This study design provides valuable information that can be used to design a longitudinal study in the future. However, conducting a longitudinal study at present was not feasible due to the need to follow subjects for a significant period of time to observe changes in exposures such as age, running experience, and body composition variables.

3.5 Study Protocol

3.5.1 Informed Consent and Questionnaires

First, all subjects provided written informed consent before enrolling in the study. Then, they completed the modified Physical Activity Readiness Questionnaire (PAR-Q) from the American College of Sports Medicine (Appendix 3) to determine their readiness for physical activity. Additionally, questionnaires were administered to collect information about the subject's running training history (Appendix 6) and injury history (Appendix 6). The running training history questionnaire aimed to gather information about the subject's running experience by inquiring about the number of years they had been running and their current weekly running mileage.

3.5.2 Bioelectrical Impedance Analysis (BIA) (Aim 1)

After completing the informed consent procedures and questionnaires, subjects who agreed to complete the body composition analysis (Aim 1) and underwent the iDXA scan on a separate day from the gait analysis also underwent a bioelectrical impedance analysis (BIA) to assess if there were changes in body composition between visits. The subject's height and weight were measured using a stadiometer and weighing scale, respectively, before BIA analysis. Percent fat mass and lean mass (lbs) were measured bilaterally for the legs using the hand-to-leg bioelectric impedance analysis (BIA) machine (Tanita, MC-780U plus P) which sends a small undetectable electrical signal through the body for ~5-10 seconds. Subjects stood on the BIA machine without socks and shoes, and their body mass was measured. Then, the subject's height, age, sex, and body type were entered into the BIA machine. The machine defines an "athletic" body type as a person involved in intense physical activity for ≥12 hours per week. Since the inclusion criteria for this study included only novice and recreational runners and excluded competitive runners and required subjects to run for a maximum of 30 miles per week, it was expected that all subjects had a "standard" body type. Subjects then grasped the metal handles, and the readings of leg percent fat mass and lean mass were recorded bilaterally after the completion of the electrical signal.

3.5.3 Gait Analysis (all aims)

After completing the informed consent procedures, questionnaires, and BIA procedures, subjects changed into form-fitting clothes and laboratory standard running shoes (Reebok; One X CrossFit Cushion 3.0) to standardize the effects of shoes on shock attenuation. Subjects then warmed up on a treadmill for approximately 5-8 minutes at their self-selected comfortable running speed while ensuring light exertion, which was defined as a rating of perceived exertion of less than 11 on Borg's RPE scale (Borg, 1998).

Subjects were affixed with external retroreflective markers bilaterally per published standards (Sutherland, 2002) to quantify joint and segment kinematics and kinetics during running. Calibration markers were placed on the following palpable bony anatomical landmarks: iliac crest, posterior superior iliac spine, anterior superior iliac spine, greater trochanter, medial and lateral femoral epicondyles (knees), medial and lateral malleoli (ankle), first and fifth metatarsal heads, top of the foot, and toe. The calibration markers were placed at the endpoint of body segments and were used to build the anthropometric model of a subject. This model defines a subject's body segment lengths, segmental masses, position of each segment's center of mass, body segment inertial parameters or properties, and the estimation of joint center locations. These parameters are necessary to compute because the relationship between kinematics (angular and linear acceleration and velocity) and kinetics (torques and forces) depends specifically on segmental masses and the moment of inertia about their center of mass. The estimation of joint center locations is crucial because they serve as the origins of joint coordinate systems (i.e., reference systems within which joint motion can be described). Additionally, clusters of four markers mounted on rigid plastic plates were secured to the thigh and tibia to track each segment's position and orientation as it moved in space during running movement. A cluster of three markers were placed on the shoe over the posterior aspect of the heel to track the position and orientation of the calcaneus as it moves in space. These marker clusters allow us to establish segmental coordinate systems (i.e., local reference frames) used to describe segment position and orientation in the laboratory space. The calibration markers and the heel cluster were secured using duct tape whereas the marker clusters on the thigh and tibia were secured using elastic bandages wrapped around each segment.

To measure impact shock and shock attenuation during running, inertial measurement units (IMUs) were used. These IMUs contained triaxial accelerometers and had a frequency of 1600 Hz with a capture range of ±20g. The IMUs were lightweight, weighing 12 grams, and were attached bilaterally on the antero-medial aspect of the distal tibia and the lumbosacral joint (i.e., lower back at the L5/S1 location). These anatomical sites were chosen because they have minimal soft tissue vibration (Wosk and Voloshin, 1981). The IMUs were secured using adjustable elastic straps and self-adhesive tape, which were tightened firmly, but comfortably for the subject. A firm mounting of the IMU was necessary because compliant attachment can amplify the magnitude of the acceleration signal, thereby contaminating the signal. The axes of the IMU were consistent for all anatomical locations, with the X-axis as axial (upward = positive), Y-axis as mediolateral (right = positive), and Z-axis as anterior-posterior (anterior = positive). The acceleration data collected from the accelerometers was stored on-board the sensors themselves using IMeasureU's Capture.U mobile application.

Prior to performing overground running trials, a standing calibration trial was recorded first. During this trial, subjects stood stationary with their feet parallel to each other and shoulder-width apart while their arms were crossed across their chest. The positions of calibration markers were recorded and averaged over 5 seconds. These markers allowed for the modeling of body segments as a system of linked, rigid segments, each with its own coordinate system. The calibration markers at the medial and lateral femoral epicondyles, medial and lateral malleoli, and the first and fifth metatarsal heads were removed prior to running trials as they were not required for tracking running motion. During the running trials, subjects ran on an 18-meter runway embedded with three force plates (OR-6-2000; AMTI Inc, Watertown, MA, USA) sampling at 1200 Hz. Their marker positions were captured by 13 motion capture cameras (Oqus 400; 500 Qualisys AB, Göthenburg, Sweden) at 240 Hz.

Subjects were instructed to perform a jump before starting each trial with the command "3, 2, 1 Jump". This jump enabled temporal alignment of the acceleration signal from the sensors with the motion capture data during data processing. To minimize the effects of speed on running mechanics, Subjects completed a minimum of 8 overground running trials at a criterion speed of 3.35 m/s ±5%. Running speed was monitored using two optoelectrical timing gates (TC-Gait, Brower Timing System, Draper, UT, USA) placed 6 meters apart. Trials that fell outside the allowable timing range of 3.35 m/s ±5% were discarded, and subjects were instructed to either speed up or slow down to ensure adherence with the running speed requirements. A trial was considered successful if subjects ran within ±5% of the given speed condition, did not noticeably modify their gait to target the force platforms (visual assessment), contacted the force platforms with their entire foot, and did not increase or decrease their running speed between timing gates as assessed using the optoelectrical timing gates. Subjects indicated their perceived exertion via the Borg's Ratings of Perceived Exertion (RPE) scale (Borg, 1998) every 10 trials to ensure they were not fatiguing (i.e., RPE <13). If subjects reported an RPE of 13 or greater, they were allowed to rest until they reported an RPE of <13. Upon completion of a minimum of 8 successful trials bilaterally, the retroreflective markers and sensors were removed, and any borrowed laboratory clothing and footwear was returned by the subject.

3.5.4 Dual X-ray Absorptiometry (Aim 1)

If subjects decided to undergo the dual x-ray absorptiometry (iDXA) scan on the same day as their gait analysis, they were taken to a room in the biomechanics laboratory for their scan. Otherwise, they were required to visit the laboratory within 30 days (about 4 and a half weeks) after their gait analysis. Prior to the scan, subject information such as age, height, weight, and sex were entered into the iDXA software (enCore, version 17). Subjects were required to remove any clothing and jewelry containing metal, as metal can impact scan results. If necessary, subjects were provided with scrubs. They lay supine on the iDXA platform in a demarcated rectangular area with their arms by their side. The researcher positioned the subject in the center of the rectangular region, ensuring that the mid-line of their body was aligned with that of the rectangular region. Velcro straps were then placed around the subject's knees and ankles to prevent any movement during the scan, which lasted approximately seven minutes.

3.6 Independent and Dependent Variables

3.6.1 Aim 1

3.6.1.1 Independent variables

To identify the degree to which body composition variables contribute to shock attenuation, the following independent variables were chosen: lower extremity bone mineral density and content, pelvic bone mineral density and content, lower extremity fat mass and lean mass, and gynoid fat mass and lean mass.

3.6.1.2 Dependent variables

Degree of axial and resultant shock attenuation between the tibia and lower back within FB1, FB2, FB3, and FB4.

3.6.2 Aim 2

3.6.2.1 Independent variables

To examine the mechanisms that contribute to active shock attenuation, the following biomechanical variables were chosen: sagittal plane ankle and knee angles at foot-ground contact, sagittal plane knee (flexion-extension) excursion during deceleration, sagittal plane ankle (dorsiflexion-plantarflexion) excursions during deceleration, frontal plane ankle (inversion-eversion) excursion during deceleration, peak knee flexion during stance, peak ankle eversion during stance, knee joint stiffness, ankle joint stiffness, sagittal plane knee joint negative mechanical work during deceleration, sagittal plane ankle joint negative work during deceleration, sagittal plane knee joint power absorption during deceleration, and sagittal plane ankle joint power absorption during deceleration.

3.6.2.2 Dependent variables

Degree of axial and resultant shock attenuation between the tibia and lower back within frequency bands of FB1, FB2, FB3, and FB4.

3.6.3 Aim 3

3.6.3.1 Independent variables

Age and Running experience; To assess the independent effects of age and running experience on impact shock and shock attenuation in time and frequency domains, both age and running experience were treated as categorical variables.

3.6.3.2 Dependent variables

In the time domain, the outcome variables for the axial and resultant components were magnitude of impact shock at the tibia, magnitude of impact shock at the lower back, and shock attenuation between the tibia and lower back. In the frequency domain, the outcome variables for the axial and resultant components were signal power magnitude for the tibia and lower back and degree of shock attenuation between the tibia and lower back within frequency bands of FB1, FB2, FB3, and FB4. The impact shock and shock attenuation variables were chosen as they will provide insight to the reliance on different musculoskeletal structures to attenuate shock. This information may be helpful in assessing which musculoskeletal structures may be at high injury risk among different runner sub-populations.

3.7 Data Processing

3.7.1 Aim 1

3.7.1.1 Independent variables

. To examine the degree to which signal power contained in different shock frequencies was attenuated, dual x-ray absorptiometry (iDXA) was used. iDXA is an imaging procedure that sends two low-dose or low-energy x-ray beams through the body of the subject as they lay supine. One dose is absorbed by soft tissue and the other by bone. The degree of absorption by bones and soft tissue depends on their respective densities. A detector placed underneath the subject's body then detects the amount of x-rays for both doses that come through the body to the other side. This residual amount of energy that is not absorbed by bone and soft tissue indicates each tissue's mass and density. The iDXA software subsequently yields mass and density values for various anatomical locations in the body (Figure 1). Thus, the following variables were extracted for the lower extremity and pelvis: bone mineral density (BMD; defined as the amount of bone mineral in bone tissue and represented in the units of grams/cm^2), bone mineral content (BMC; defined as the amount of mineral contained in a given volume of bone and represented in the units of grams), absolute fat mass (lbs), and lean mass (lbs). Additionally, fat mass and lean mass will also be extracted for the gynoid region.

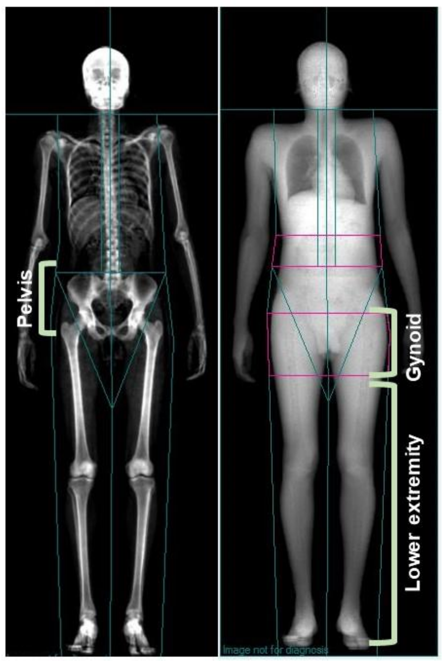

Figure 1. A sample iDXA scan result displaying the anatomical regions analyzed to extract measures of bone mineral density and content (lower extremity and pelvis), and fat mass and lean mass (lower extremity and gynoid).

3.7.1.2 Dependent variables

All dependent variables in both time and frequency domains were calculated using the axial and resultant acceleration signals following the procedures described below. These procedures are in line with previous guidelines for signal processing (Sheerin et al., 2019).

3.7.1.2.1 Axial Acceleration

Three-dimensional acceleration at the tibia and lower back was measured during running using tri-axial accelerometers embedded in IMUs. All data captured and stored on board the accelerometers was downloaded via IMeasureU's desktop application called Capture.U. The IMUs were oriented such that the X-axis (positive upward) of the accelerometer was aligned with the long axis of the tibia (i.e., axially). To calculate peak impact shock and shock attenuation from the axial acceleration signal, the X-axis which is the axial axis was extracted for processing in MATLAB (Release 2022a, MathWorks, Natick, MA, USA). As a first step, the axial acceleration signal was converted from units of meters per second squared (m/s^2) to gravity (i.e. 1 g = 9.81 m/s^2). Next, a least-squares best-fit line was subtracted from each signal to remove any existing linear trend (i.e., detrending). A linear trend in acceleration signals is not characteristic to non-fatiguing overground running at a constant speed. Therefore, any systematic trends may be due to errors like drifts in the accelerometer which must be corrected. Signal offset was removed such that the signal oscillates about zero. Given that the accelerometers collect 1g of signal at rest, the 1g offset was removed from the signal during the offset removal step. Unfiltered signals were used for frequency domain analyses.

3.7.1.2.2 Resultant Acceleration

Given that the accelerometers in the IMUs were tri-axial, they measured accelerations along three axes – X (axial with upward as positive), Y (mediolateral, positive to the right) and Z (anterior-posterior, positive in the forward direction). The signal measured along each axis at the tibia and lower back was processed using the same procedures described for the axial acceleration signal. That is, the signal along each axis was converted to units of gravity, detrended, and signal offset was removed. The detrended and offset-removed signals were used to calculate resultant acceleration. At the tibia and lower back, the resultant acceleration was calculated as follows:

$$a_{Res} = \sqrt{a_x^2 + a_y^2 + a_z^2}$$

Where a_{Res} is the resultant acceleration signal, x, y, and z describe segmental accelerations (“a”) at the tibia or lower back along the axial, mediolateral, and anterior-posterior directions, respectively. Filtering was not performed for the frequency domain analyses.

3.7.1.2.3 Stance Phase Detection

Stance phases were detected for both the tibia and lower back acceleration signals. Stance phase detection was required to decompose the acceleration signals into the frequency domain. A frequency domain decomposition allows us to examine the signal power contained in frequencies of impact shock as well as how this power is attenuated between the tibia and lower back (i.e., frequency domain outcome variables). The portion of the axial and resultant acceleration signal occurring during the stance phase was detected using a footstrike detection algorithm adapted from Aubol and Milner (2020). Step detection was performed on the axial acceleration signal at the tibia and the obtained gait events of heel-strike and toe-off were subsequently applied to the lower back and tibia for both axial and resultant acceleration components.

Prior to implementing the step detection algorithm, it was necessary to temporally align the 3D motion capture system with the IMUs and temporally align the tibial and lower back IMUs. Alignment of the motion capture system with the IMUs was necessary to (1) ensure the step used to assess running kinematics and kinetics (i.e., active shock attenuation mechanisms in aim 2) was the same step used to calculate shock attenuation, and (2) the step identified in the acceleration signals was performed at constant running speed to remove potential speed effects on impact shock and attenuation. It was also necessary to align the tibia and lower back signals to ensure lag in start times did not influence the observed shock attenuation. The following procedures and Figure 2 outline the methods used to temporally align the motion capture system and IMUs and the step detection algorithm subsequently employed:

1. Temporal alignment of the tibia and lower back IMU data

Despite simultaneous initiation of data collection across IMUs, sensor data were aligned based on Unix time stamps to account for possible lag in start times; The location of the first common Unix timestamp in both signals (i.e., lower back and tibia) was identified, and designated as the start of both signals. Then, both signals were linearly interpolated to 1600 Hz using the "resample" function in MATLAB because each sensor sampled at a slightly different sampling frequency (right leg: 1598.151 Hz, left leg: 1604.322, lower back: 1587.432 Hz). These two steps ensured signal alignment between the tibia and lower back.

2. Temporal alignment of the IMU data with motion capture data

Figure 2 schematically explains the procedures detailed here to temporally align the IMU data with the motion capture system. To identify the gait-related active shock attenuation mechanisms in Aim 2 and ensure the running steps identified in aims 1 and 3 were run at standardized speed, identified the same running step in the axial acceleration signal and motion capture data were identified. To do so, the time elapsed (QTM period) between the manually flagged jump-landing in QTM and foot-strike on a force plate (T_2) (i.e., when the force surpassed 20 newtons) was calculated. The location of the jump-landing (T_1) was also identified in the tibial acceleration signal as the location of the first peak with a minimum magnitude of 3g. The manually flagged jump landing in QTM and T_1 were expected to occur in proximity of each other therefore the QTM period was overlaid onto the tibial acceleration signal starting at T_1. The running step in the tibial acceleration signal immediately following the QTM period was the one coinciding with the force-plate contact for that limb.

A supplementary analysis (Appendix 7) was performed to identify the time lag or lead relative to T1, since the jump landing in QTM was manually flagged. The analysis revealed that the manual flag preceded T1 by ~0.16 seconds. However, this time lag of 0.16 seconds was small enough to ensure that the step in the acceleration signal preceding the one coincident with force-plate contact was excluded. This ensured that the same step was identified in both motion capture and acceleration data. A validated foot-contact detection algorithm adapted from Aubol and Milner (2020) was utilized to identify foot-strike and toe-off for the step immediately following the QTM period. Data were filtered using a 4th order Butterworth lowpass filter with a cutoff frequency of 70 Hz for the sole purpose of step detection before implementing this algorithm. Step detection was performed using only data along the axial axis from the tibial sensor, and the locations of foot-strike and toe-off were subsequently applied to all axes, including the resultant, in both the tibial and lower back sensors.

Per the foot-contact detection algorithm, peaks in the jerk (i.e., derivative of acceleration) of the axial acceleration signal that occurred within 150 milliseconds of a peak in the axial acceleration signal shortly following foot-contact were identified. The local minima in the axial acceleration signal occurring within 75 milliseconds of each jerk peak were identified, however local minima that occurred after a axial acceleration peak which was less than 20% in magnitude of the third acceleration highest peak were removed. Finally, the location of the earliest local minima occurring before the axial acceleration peak was identified and designated as foot strike. The foot-contact detection algorithm only identified the location of foot-strike but not toe-off. Thus, toe-off was identified using contact time. Contact time was calculated via force-plate data as the time between foot-strike and toe-off. The application of contact time to foot-strike location identified via step detection yielded the location for toe-off. The locations of foot-strike and toe-off thus identified were subsequently applied to all axes of the unfiltered tibial and lower back sensor data, thereby detecting individual stance phases.

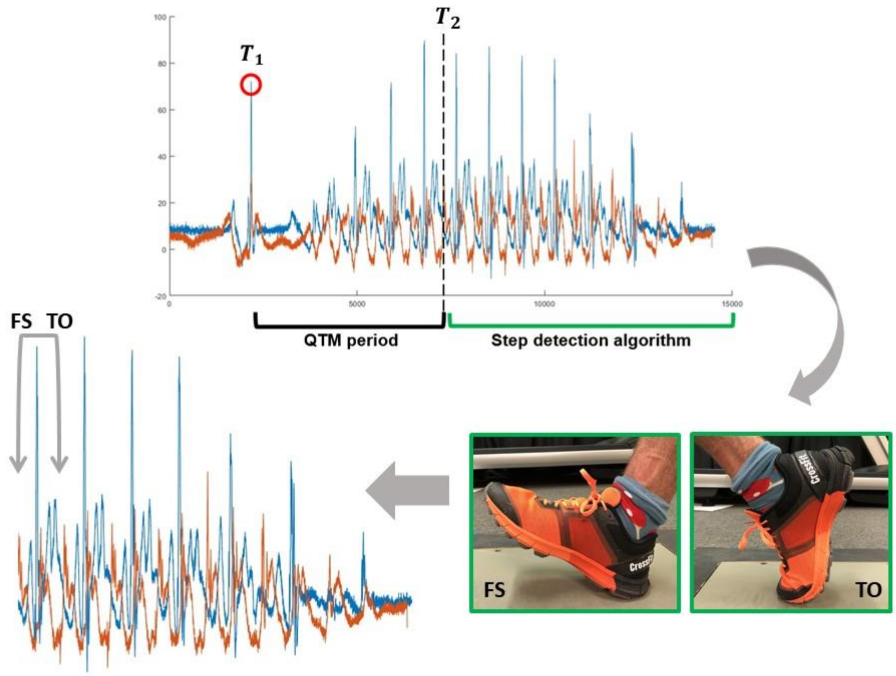

Figure 2. Method used to align the motion capture data with the IMU data so that the same step was identified in both data types. The QTM period was the time elapsed between the jump landing at T_1 and foot-strike on a force plate at T_2. The black vertical dashed line in the top panel indicates the end of QTM period at T_2. The signal before T_2 was discarded while the signal after it was entered into the step detection algorithm to detect the gait events of foot strike (FS) and toe off (TO) that were subsequently applied to all acceleration components. T_1: location of the jump landing manually flagged in QTM and observed in the acceleration signal.

3.7.1.2.4 Frequency Domain Analyses

The signal power contained in different shock frequencies was calculated by transforming the unfiltered stance-phase (detected per the procedure described above) in the time-domain to the frequency domain. This transformation was achieved by calculating its power spectral density (PSD) that utilizes the Fast Fourier Transform (FFT) algorithm (Shorten and Winslow, 1992). The PSD was calculated for each running trial recorded at both the tibia and lower back. The Fourier Transform is a mathematical tool that decomposes a periodic signal into its sinusoidal components, each characterized by a frequency, amplitude, offset, and phase angle. The frequency represents how rapidly a signal oscillates per second, and the amplitude refers to the magnitude of these oscillations while the offset is the mean value of the signal. The phase angle refers to the shift or time delay in the signal. Using this amplitude and frequency information, the power contained at each frequency was assessed as described below.

First, the time domain stance-phase signal was used to calculate the PSD for each frequency to determine the power contained at each frequency. The Nyquist sampling frequency, which is at least twice as high as the highest frequency present in the signal, is required for the FFT algorithm. In biomechanics studies, the sampling rate is typically five to ten times the highest frequency present in the signal. Shock frequencies during running have been known to range up to 50 Hz (Shorten and Winslow, 1992), so a minimum sampling rate of 500 Hz is necessary. The tibia and lower back stance phases were sampled at a rate of 1600 Hz, which meets the Nyquist criterion.

Each stance phase signal was padded with zeros at both ends to make the length of the signal 1024 data points because the FFT requires the number of samples that constitute the signal to be a power of two. The FFT also assumes that the time-domain signal used represents one period of a periodic signal and has a circular topology, meaning that its endpoints are connected, and the signal is made up of an integer number of periods. However, if these assumptions are not met, the endpoints of each signal entered into the FFT will be discontinuous, leading to artificial high-frequency components in the frequency-domain signal. These components can be much larger than the Nyquist frequency and appear in the spectrum between 0Hz to Nyquist frequency. To reduce such leakage, a Hanning windowing technique was used, which reduces the amplitude of the discontinuities at endpoints of each frequency-domain signal, making the endpoints of the signals meet and form a continuous or circular topology.

The processed signal was used to calculate the amplitude-frequency spectrum for each stance phase via the FFT, describing the amplitude of each frequency component. Subsequently, the power contained at each frequency was calculated by squaring the amplitude at that frequency. Zero-padding the signal adds zeros to the signal length, which decreases the power contained at all frequencies. To correct for this, the power at each frequency was multiplied by $(N + L) \div N$, where N was the number of non-zero values in the signal and L was the number of padded zeros (Shorten and Winslow, 1992). The power contained in FB1, FB2, FB3, and FB4 was then assessed by calculating the power spectral density (PSD).

To compare signal power at different frequencies across signals of varying lengths, the PSD is a useful tool. The width or resolution of each frequency bin, and thus the power contained at each frequency, is dictated by the sampling time or length of the signal. However, during running, there can be inter- and intra-subject variability in contact time (i.e., time taken to complete a stance phase), which can affect the sampling time and signal power observed for each subject's stance phase signal. To account for this, the PSD was normalized to 1 Hz bins for frequencies ranging from 0Hz to Nyquist (i.e., half of the sampling rate) (Edwards et al., 2012; Gruber et al., 2014; Hamill et al., 1995).

3.7.1.2.5 Shock Attenuation in the Frequency Domain

In the frequency domain, the degree of shock attenuated between the tibia and lower back within each frequency band was quantified using a transfer function (Hamill et al., 1995). The transfer function is a ratio between the PSD values of the lower back and tibial signals calculated was follows:

$$Shock\ attenuation = 10\ log_{10}\ (PSD\ lower\ back / PSD\ tibia)$$

Where shock attenuation is the signal gain or attenuation in decibels and PSD is the power spectral density for the lower back and tibia. The transfer function was calculated for each frequency bin, with positive values indicating a gain in signal power and negative values indicating attenuation of signal power. The degree of shock attenuation was calculated by integrating the transfer function result over each frequency band, as done previously (Gruber et al., 2014). The transfer function and shock attenuation were calculated in this manner for both the axial and resultant components (Figure 3).

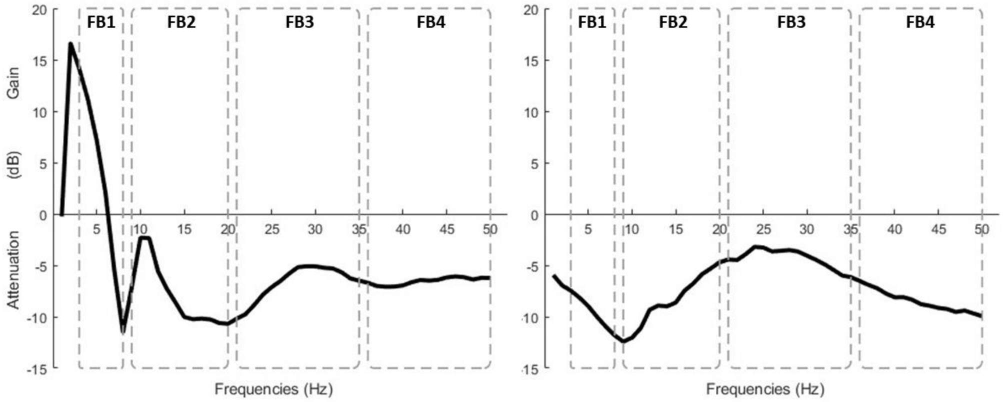

Figure 3. Transfer function for the entire sample (solid black line) for the axial (left) and resultant (right) acceleration signals. Positive values indicate a gain in signal energy between the tibia and lower back while negative values indicate attenuation. The dashed boxes demarcate the four frequency bins in order from left to right: 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4). Shock attenuation is the area under each curve within each frequency bin. dB: decibel; Hz: hertz

3.7.2 Aim 2

3.7.2.1 Dependent variables

Both the axial and resultant components were used to calculate shock attenuation within FB1, FB2, FB3, and FB4, following the same method used for Aim 1.

3.7.2.2 Independent variables

Three-dimensional marker positions were tracked using Qualisys Tracking Manager (Qualisys AB, Gothenburg, Sweden). The axes of the global coordinate system were defined as follows: x-axis was medio-lateral with the positive direction defined to the right of the origin; y-axis was anterior-posterior with the positive direction defined as anterior, and z-axis was vertical with the positive direction defined as upward. Joint coordinate systems were calculated using joint center estimation and segment coordinate system transformations (Wu et al., 2002). The axes of the joint coordinate system were oriented in the same way as those of the global coordinate system. Force platform data was used to identify the gait events of heel strike and toe-off that defined the stance phase for the data derived from the motion capture and force plates. Heel-strike was identified as the time when the ground reaction force exceeds 20 Newtons, while toe-off was identified as the time when the ground reaction force fell below 20 Newtons.

Raw marker and force plate data were filtered in Visual 3D (C-Motion Inc, Rockville, MD, USA) using a 4^{th} order, zero-lag Butterworth lowpass filter with a 12 Hz cutoff frequency for marker data. The identical cutoff frequency for both force plate and marker data ensured that impact-related artifacts were minimized when calculating joint moments (Derrick et al., 2020; Kristianslund et al., 2012).

3.7.2.2.1 Joint Center Estimation

Joint centers are used to define the endpoints of body segments to define segmental lengths. Joint centers were calculated for each joint as follows: the hip joint center was defined using regression equations from Bell et al. (1989), the knee joint center was defined mid-way between the medial and lateral femoral epicondyle markers, and the ankle joint was defined mid-way between the medial and lateral malleoli markers.

3.7.2.2.2 Body Segment Creation

Segmental masses from Dempster (1955) and segmental inertial properties from Hanavan Jr (1966) were used to model the lower extremities as a rigid linked segment system. In conjunction with these model properties, the pelvis segment was constructed using the distal and proximal pelvic radii. The proximal pelvic radius was defined using the positions of the right and left iliac crest, while the distal pelvic radius was defined using the right and left greater trochanter. The origin of the pelvis segment coordinate system was estimated using both inertial and geometric properties. Using the greater trochanter marker, the proximal radius for the thigh segment was calculated while the lateral and medial femoral epicondyle markers were used to calculate the distal radius. The inertial and geometric properties of the segment were used to define the coordinate system for the thigh. Using the markers at the lateral and medial femoral epicondyles, the proximal radius of the shank segment was calculated. The markers at the medial and lateral malleoli were used to calculate the distal radius. The coordinate system for the tibia segment was then defined based on its inertial and geometric properties. The foot segment was modeled in two different ways. The first method, which was used to calculate kinetics, involved defining the proximal radius using the lateral and medial malleoli and calculating the distal radius using the first and fifth metatarsal heads. Segment inertial properties and geometries were used to define the foot coordinate system. However, the coordinate system of the foot defined in this manner is rotated due to the orientation of the foot not being parallel to the ground, which results in joint kinematics that are not clinically meaningful. Therefore, the second method was used to calculate kinematics. In this method, the proximal radius was calculated using the medial and lateral malleoli projected to the floor, and the distal radius was calculated using the first and fifth metatarsal heads projected to the floor. This method created a "virtual foot" that had its axes oriented with those of its proximal segment, which was the tibia.

3.7.2.2.3 Joint angles and ranges of motion

Three-dimensional lower limb kinematics were calculated using filtered data and model properties. Kinematics were calculated per the right-hand rule with an x-y-z Cardan sequence. Sagittal plane knee and frontal plane ankle joint angle time-series were negated such that knee flexion, ankle dorsiflexion, and ankle eversion were all positive magnitudes. Table 1 displays the directionality of joint and segment angles:

Table 1. Directionality of joint angles of interest.

Joint, Motion Plane	Motion (Polarity)
Knee, sagittal plane	Flexion (positive), Extension (negative)
Ankle, sagittal plane	Dorsiflexion (positive), Plantarflexion (negative)
Ankle, frontal plane	Inversion (negative), Eversion (positive)

Knee and ankle joint angles were calculated with reference to their proximal segments; knee joint angles were calculated with reference to the thigh segment, and ankle joint angles were calculated with reference to the shank segment. The thigh, shank, and foot segment angle were calculated relative to the global coordinate system.

Sagittal knee and ankle angles at contact were calculated as the angle magnitude at foot-strike in each respective joint-angle time-series. Peak knee flexion and peak ankle eversion were identified as the maximum magnitudes occurring in their respective joint-angle time series (Figure 4). Deceleration (i.e. the energy absorption) period during stance was defined as starting at the beginning of stance and terminating when ankle dorsiflexion angle in the sagittal plane reached its peak magnitude. Ankle dorsiflexion was chosen because it was previously used to define the end of the energy absorption phase during stance (Hamill et al., 2014). Knee and ankle joint ranges of motion in the sagittal plane (knee and ankle) and frontal plane (ankle) were calculated as the difference between the maximum and minimum magnitudes in the joint-angle time series during deceleration (Figure 4). Joint angles at initial contact, peak joint angles during stance, and ranges of motion during deceleration were calculated for each running trial for each limb and then averaged across trials for each subject per limb.

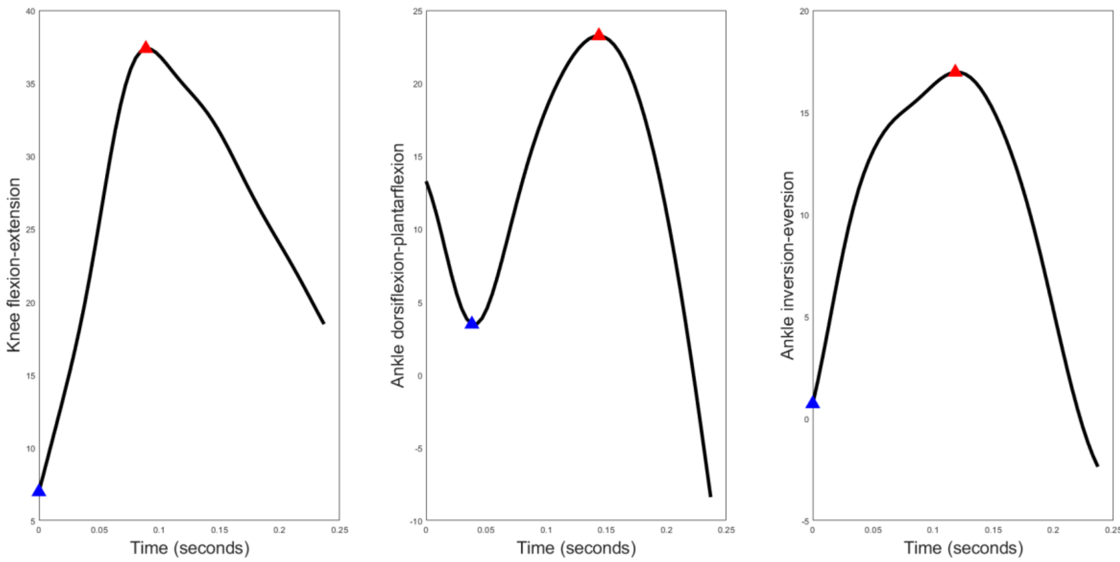

Figure 4. Joint angle time-series for the sagittal plane knee and ankle, and frontal plane ankle angles (from left to right). The red marker indicates the local maxima for each time-series during deceleration, while the blue marker indicates the local minima for each time-series during deceleration.

3.7.2.2.4 Joint Moments, Joint Powers, and Joint mechanical work

As is routinely done, joint moments (i.e., torques) were computed using Newtonian inverse dynamics and expressed in their respective joint coordinate systems (i.e., reference system) (Derrick et al., 2020) (Figure 5). They were calculated using marker data and force plate data, both of which were filtered at 12 Hz to avoid impact related artifacts in joint moment calculations (Kristianslund et al., 2012). Joint powers were calculated as the product of joint angular velocities and joint moments (Figure 5). Sagittal plane ankle and knee joint powers for each trial were exported to MATLAB (Release 2018b, MathWorks, Natick, MA, USA) to calculate maximum negative joint power and negative mechanical joint work. Negative joint power was extracted as the local minima (i.e., negative peak) along the joint power-time curve occurring during deceleration. The negative regions under the power-time curve were numerically integrated over the deceleration period to obtain a measure of negative mechanical joint work during deceleration. Negative mechanical work at the knee and ankle were for each running trial for each limb and then averaged across trials for each subject per limb.

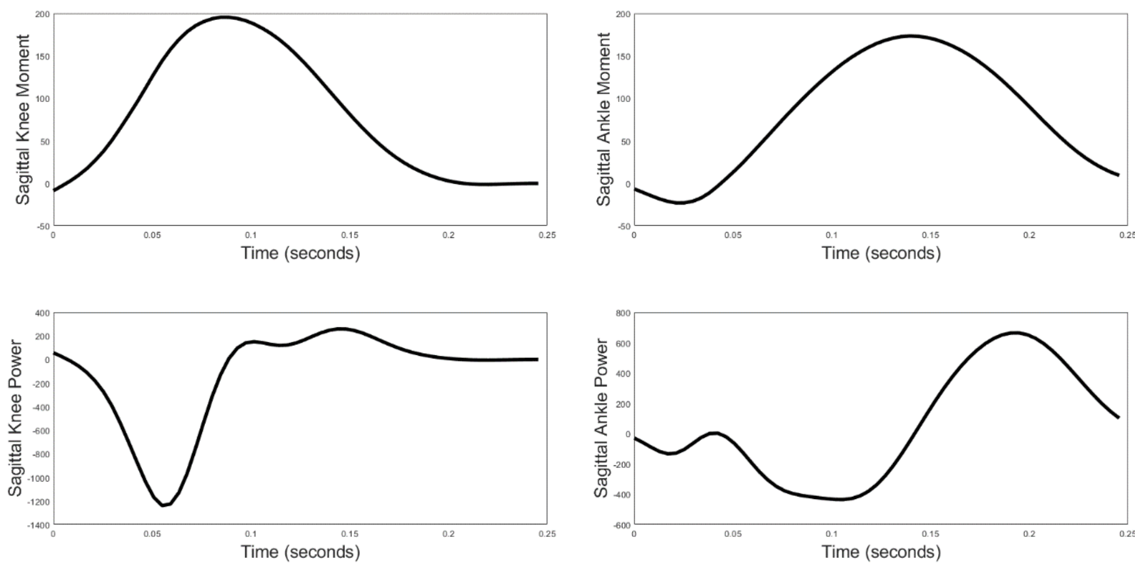

Figure 5. Sagittal knee and ankle joint moments (top row) and sagittal knee and ankle joint powers (bottom row). Positive values indicate an extensor moment and power generation, while negative values indicate a flexor moment and power absorption at each joint. The negative area under the power-time curve quantifies negative joint work for both the ankle and knee.

3.7.2.2.5 Joint Stiffness

Joint stiffness for the ankle and knee in the sagittal plane was calculated using a joint moment-angle plot. A linear regression equation was fit to the plot using data during the energy absorption phase beginning at initial contact and ending at the instant when ankle dorsiflexion reached its maximum magnitude (Hamill et al., 2014). Ankle and knee joint stiffness were calculated for each running trial for each limb and then averaged across trials for each subject per limb.

3.7.3 Aim 3

3.7.3.1 Dependent variables

3.7.3.1.1 Shock Attenuation and Signal Power Magnitude in the Frequency Domain

Shock attenuation within FB1, FB2, FB3, and FB4 were calculated in the same way as done for Aim 1 for both axial and resultant components. Signal power magnitude for each component at the tibia and lower back was calculated as the integral of the power contained within FB1, FB2, FB3, and FB4 (Figure 6).

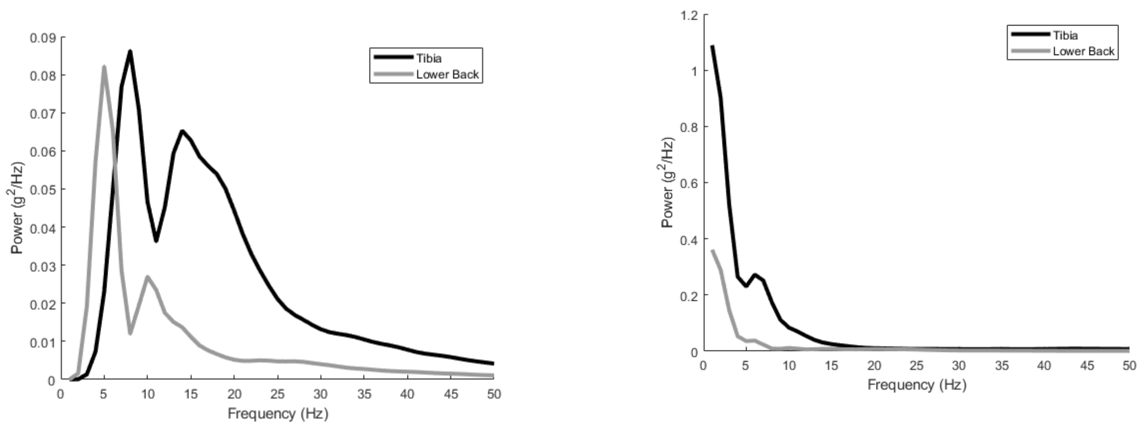

Figure 6. Axial (left) and resultant (right) Power spectral density calculated for the tibia and lower back during running stance. The PSD was integrated over 3-8 H, 9-20 Hz, 21-35 Hz, and 36-50 Hz to obtain the signal power magnitude within each of these frequency bins for both the tibia and the lower back.

1. Peak Impact Shock Detection in the Time Domain

For time domain analyses, tibial and lower back axial and resultant acceleration signals were filtered using a 4th order, zero-lag Butterworth low-pass filter with a cut-off frequency of 70 Hz. The choice of cut-off frequency was based on both prior literature and a power spectrum analysis. Previous literature has used a cut-off frequency between 40 Hz-100 Hz (Sheerin et al., 2019), however, resonance arising from accelerometer mass and mounting techniques typically occurs between 60-90 Hz and can amplify the magnitude of acceleration to a degree uncharacteristic of human running (Shorten and Winslow, 1992). Hence, a power spectrum analysis was performed using the MATLAB's *'obw'* function to quantify the frequency bandwidth within which 95% of signal power resided. This power analysis was performed on all trials across subjects bilaterally and revealed that 95% of total signal power at the tibia was contained below 65.88 Hz. Hence, a cut-off frequency of 70 Hz was deemed appropriate and applied to both the tibia and lower back signals. For each stance phase along both axial and resultant components, peak impact shock at the tibia was identified as the local maxima in the acceleration signal occurring within the first 50 milliseconds of stance with a magnitude of >3g (Figure 7); 3g is a threshold used to detect footstrike in the past suggesting peak impact shock must exceed this minimum threshold (Van den Berghe et al., 2019b). Similarly, peak impact shock at the lower back was identified as the local maxima in the acceleration signal immediately following peak tibial shock (Figure 7).

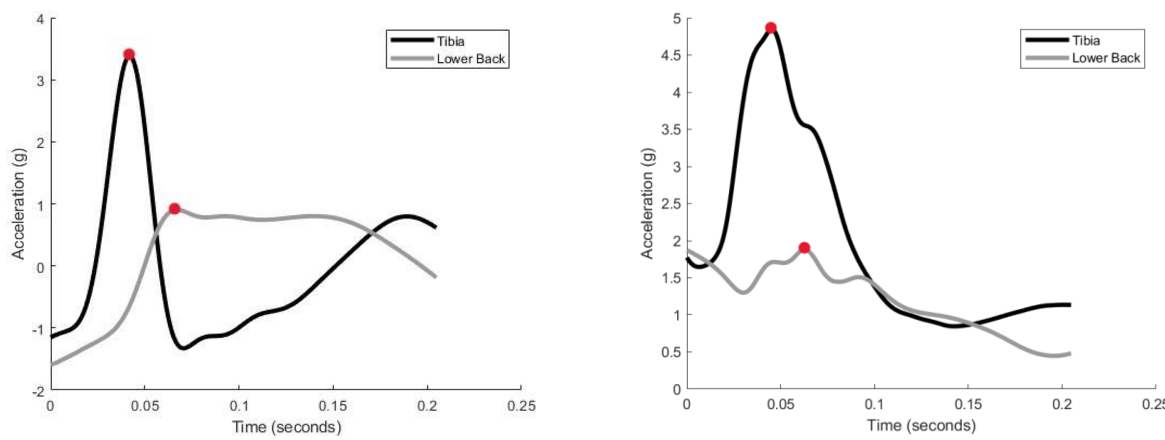

Figure 7. Detection of peak impact shock (red dot) in the time-domain along the axial (left) and resultant (right) acceleration components.

3.7.3.1.2 Shock Attenuation in the Time Domain

In the time domain, the degree of shock attenuation between the tibia and lower back was calculated using a transfer function as follows (Mercer et al., 2002; Reenalda et al., 2019):

$$Shock\ attenuation = \left(1 - \frac{Peak\ impact\ shock - lower\ back}{Peak\ impact\ shock - tibia} * 100\right)$$

Axial and resultant shock attenuation in the time domain was calculated for each subject as an average across trials for each limb.

3.7.3.2 Independent variables

Both age and running experience were treated as categorical variables. For age, the young group was treated as the reference group and coded "0" while the middle-aged group was coded "1". For running experience, the recreational group was treated as the reference group and coded "0" while the novice group was coded "1". Subjects were classified into the middle-aged group if they were between 45-65 years old and into the young group if they were between 18-30 years old. Subjects were classified as recreational if they had been running for more than one year and ran between 9-30 miles per week, and novice if they had been running for less than a year and ran between 3-13 miles per week (Honert et al., 2020).

The running experience definitions used by Honert et al. (2020) did not provide guidelines to classify runners who ran precisely for one year or those who met either but not both classification criteria of years and weekly mileage. To ensure 80% power, individuals who ran for exactly one year were included as either novice (n=3) or recreational (n=1) based on if their weekly mileage values were within ±1 standard deviation of the group mean calculated upon excluding them. Additionally, eight runners qualified as recreational per the "years" criteria but not "weekly mileage" criteria, while the opposite was true for one runner. These runners were classified as recreational if they had been running for >1 year (n=8) and novice if they ran for <1 year (n=1) because they did engage in regular running, and previous studies have classified novice runners as those with no regular running experience (Harrison et al., 2021; Quan et al., 2021a)

3.8 Statistical Design

3.8.1 Bilateral Differences

As a reminder, Table 2 lists the independent and dependent variables for each aim. Independent and dependent variable data for Aims 1 and 2, and dependent variable data for Aim 3 was collected bilaterally. Therefore, bilateral differences in these measures were identified for the entire sample using repeated measures t-tests for parametric data and Wilcoxon signed-rank tests for non-parametric data, with normality assessed via Shapiro-wilk tests ($\alpha \leq 0.05$). Data was not collapsed across limbs for variables that were significantly different between limbs, while all other data were collapsed across limbs for the regression analysis in each aim.

3.8.2 Aim 1

Unadjusted linear regression ($\alpha \leq 0.05$) was used to assess the relationship of each dependent variable, i.e., degree of shock attenuation within FB1, FB2, FB3, and FB4 to lower extremity and pelvis BMD and BMC, and lower extremity and gynoid lean mass and fat mass.

3.8.3 Aim 2

To examine the associations between shock attenuation and active attenuation mechanisms, one unadjusted regression model was implemented per independent-dependent variable pairing. Given that shock attenuation includes negative values, positive beta coefficients indicated a reduction in shock attenuation while negative beta coefficients indicated an increase in shock attenuation with every unit increase in the independent variable. Each regression model was checked for violations of linearity, independence, homoscedasticity, and normality of residuals.

3.8.4 Aim 3

Unadjusted linear regression was used to assess the independent effects of age and running experience on each dependent variable in both time and frequency domains along the axial and resultant components. For the independent effects of age and running experience, the young and the recreational groups were treated as the reference groups, respectively. One regression model for each independent-dependent variable pairing was implemented. If dependent variables were significantly different between limbs, then one model was implemented for each limb separately. Linear regression assumption violations were assessed. The dependent variable was log-transformed for models that violated the assumed normality of residuals as assessed using Q-Q plots. Given that shock attenuation includes negative values, positive beta coefficients indicated a reduction in shock attenuation while negative beta coefficients indicated an increase in shock attenuation relative to the reference group.

Table 2. List of independent and dependent variable for each aim

Aim	Dependent variable	Independent variable
Aim 1	Time domain: Peak impact shock, degree of shock attenuation between the tibia and lower back for the axial and resultant acceleration components.	Age, Running Experience
	Frequency domain: Signal power magnitude at the tibia and lower back and the degree of shock attenuated within 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4).	Age, Running Experience
Aim 2	Degree of shock attenuated within 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4).	Sagittal plane knee angle at initial contact, Sagittal plane ankle angle at initial contact, Sagittal plane knee (flexion-extension) excursion during deceleration, Sagittal plane ankle (dorsiflexion-plantarflexion) excursions during deceleration, Frontal plane ankle (inversion-eversion) excursion during deceleration, Peak knee flexion during stance, Peak ankle eversion during stance, Knee joint stiffness, Ankle joint stiffness, Sagittal plane knee joint negative mechanical work during deceleration, Sagittal plane ankle joint negative work during deceleration, Sagittal plane knee joint power absorption during deceleration,

		Sagittal plane ankle joint power absorption during deceleration.
Aim 3	Degree of shock attenuated within 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4).	Leg bone mineral density, leg lean mass, leg fat mass, and leg fat percentage.

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4 CHAPTER 4: Study 1

Manuscript Title: Passive Mechanisms of Shock Attenuation During Human Running

Intended Journal: Journal of Experimental Biology

4.1 Abstract

Background. Impact shock frequencies >8 Hz may be passively attenuated via biological tissue deformation during running. However, the degree of shock attenuated by bone, fat tissue, and lean tissue across frequencies relevant to running (i.e., 3-50 Hz) must be ascertained as it may reveal the demands placed on each tissue type with implications for interventions designed to reduce tissue loads.

Purpose. To examine the associations between lower extremity body composition measures to the degree of shock attenuated within 3-8 Hz, 9-20 Hz, 21-35 Hz, and 36-50 Hz during human running.

Methods. Thirty-eight healthy runners (age 31.55±14.94 years; 17 female, 22 male) running 3-30 miles·week^{-1} were enrolled. Impact shock was measured using triaxial accelerometers at the distal tibia and lower back as they ran down an 18-meter level platform at 3.35±5% m·s-^{1}. Power spectral density (PSD) transformed the time domain acceleration signals to the frequency domain. Shock attenuation (dependent variable) was calculated along the axial and resultant acceleration components using a transfer function of the lower back PSD relative to the tibia PSD within 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4). Bone mineral density (BMD) and content (BMC) in the lower extremity and pelvis, and fat and lean mass in the lower extremity and gynoid (independent variables) were measured via dual x-ray absorptiometry. Linear regression ($\alpha<0.050$) was used to test for unadjusted associations between each independent-dependent variable pair.

Results. An increase in lower extremity fat mass decreased axial shock attenuation within FB2 (p-value=0.014, beta coefficient (β)=0.015, 95% confidence interval (CI)=0.003-0.028) and FB4 (p=0.033, β=0.032, CI=0.002-0.061). An increase in gynoid fat mass decreased axial shock attenuation within FB2

(p=0.008, β=0.014, CI=0.003-0.024), FB3 (p=0.034, β=0.021, CI=0.001-0.041), and FB4 (p=0.012, β=0.031, CI=0.007-0.055).

Conclusions. BMD, BMC, and lean mass may not contribute to shock attenuation across the frequency spectrum during running, highlighting the need to explore structural tissue properties in addition to mass. Fat mass may decrease axial shock attenuation in the lower extremities and gynoid, potentially by decreasing lean muscle mass which should be examined.

4.2 Introduction

The mechanical energy associated with ground reaction forces and subsequent tibial accelerations during running stance is predominantly contained in frequencies between 3-50 Hz (Gruber et al., 2014; Gruber et al., 2017; Nigg, 1997; Shorten and Winslow, 1992). As these frequencies are propagated from the tibia to the head, they are absorbed either by active mechanisms like eccentric muscle actions or passive mechanisms like the deformation of bone or soft tissues along the way (Nigg et al., 1995). Such absorption is referred to as shock attenuation.

The predominant frequency content observed at the head is between 3-8 Hz (Derrick et al., 1998), shock frequencies greater than 8 Hz may be better attenuated by active and massive mechanisms than lower frequencies. Frequencies greater than 8 Hz are linked to the impact force found in the vertical ground reaction force, with the peak impact force typically occurring within the initial 25 milliseconds of stance (Bobbert et al., 1992). Due to such a quick occurrence of peak impact force and subsequently peak tibial acceleration (i.e., peak impact shock), it is unlikely that the associated higher frequencies are absorbed via active mechanisms (Bobbert et al., 1992). These higher frequencies must therefore be attenuated passively via biological tissues when they deform to dissipate energy. However, current evidence regarding the mechanisms of passive shock attenuation is based on animal models (Paul et al., 1978; Radin et al., 1970), cadaver data (Chu et al., 1986), simulation studies (Pain and Challis, 2001), and static positions among living individuals (Schinkel-Ivy et al., 2012), with very limited evidence of these mechanisms among dynamic activities (Boyer and Nigg, 2006; Pain and Challis, 2002) like running.

Animal models and cadaver data suggest that bone deformation is an effective shock attenuation mechanism (Chu et al., 1986; Paul et al., 1978) of high frequencies (Paul et al., 1978) as it is able to dissipate energy due to its viscoelastic nature (Morgan et al., 2018). Therefore, during running, characteristics that influence bone viscoelasticity including, but not limited to, bone mass or mineral density (Augat and Schorlemmer, 2006; Ojanen et al., 2015) may contribute to one's ability to passively attenuate high-frequency shock. Bone mineral content may also aid shock attenuation independent of muscle activation during running, as evidenced by a pendulum experiment (Schinkel-Ivy et al., 2012).

In addition to bone, attenuation via other soft tissue deformation increases with an increase in impact-related frequencies (Paul et al., 1978) suggesting both lean mass and fat mass may contribute to passive energy dissipation during running. In other dynamic tasks like a forearm karate strike, 70% of the energy dissipated during impact is accounted for by fat and lean mass (Pain and Challis, 2002). Similarly, when shank is modeled as a wobbling mass to account for fat and lean mass rather than a rigid body, the energy dissipation at the shank during impact increases from 34.4% to 89.9% indicating a significant contribution of soft tissues to shock attenuation (Pain and Challis, 2001).

Overall, findings to date suggest both soft and rigid tissues contribute to high-frequency shock attenuation and may therefore aid the attenuation of frequencies greater than 8 Hz during running. However, the relative degree to which each of these tissue types contributes to passive attenuation during running may vary owing to differences in their structural, material, and mechanical properties. For instance, adipose tissue is more compliant than muscle tissue (Guimarães et al., 2020) which may result in its greater deformation and thereby greater passive shock attenuation than muscle deformation.

Identifying how each tissue type contributes to shock attenuation may provide insight to the demands placed on each tissue type and the tissues' injury risk. For example, if the amount of bone mass is positively related to higher-frequency energy absorption, gait retraining protocols can aim to alter gait patterns among individuals with low bone mass such that the energy contained in high-frequency vibrations is reduced. Subsequently, bony injury risk due to the inability of low bone mass to withstand high-frequency shock may be reduced. This information may be valuable for interventions like gait retraining protocols that target a change in the input signal (i.e. peak impact shock) (Clansey et al., 2014; Crowell and Davis, 2011; Crowell et al., 2010) because that change may subsequently alter the requirements placed on different soft and rigid tissues to attenuate shock.

Uncovering the degree to which bone, fat mass, and lean mass contribute to the attenuation of different frequencies will enable intervention designs targeted at reducing tissue loads during running. The purpose of this study was to examine the relationship of body composition (lower extremity and pelvis bone mineral density [BMD] and content [BMC], lower extremity and gynoid fat mass, and lower extremity and gynoid lean mass) to the degree of shock attenuation between the tibia and lower back within 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4) among runners. It was hypothesized that the degree of shock attenuated between the tibia and lower back within FB2, FB3, and FB4 would increase with an increase in each body composition measure, but the degree of shock attenuated within FB1 would not be associated with any of the body composition measures because biological tissue may be more effective at attenuating high rather than low frequencies of shock (Paul et al., 1978) .

4.3 Methods

4.3.1 Sample Characteristics

A-priori power calculations (G*Power Version 3.1.9.4) using data on lower leg wobbling mass and bone mineral content from Schinkel-Ivy et al. (2012) revealed N=22 subjects were required to detect a relationship between X and Y of size Z with 80% statistical power with an error probability of 0.05.

Therefore, a convenience sample of N=38 runners were recruited to improve generalizability of results across ages and running exposure levels.

Inclusion criteria required runners to 1) be between 18-65 years old, 2) have been running between 3-30 miles per week, 3) have no history of lower back or lower limb orthopedic surgery in the past 10 years, 4) if an injury was developed in the past 6 months, they must have returned to normal running training volume for $\geq$8 weeks before enrollment, and 5) be free of chronic illnesses that may disallow engagement in $\geq$15 minutes of low to moderate-intensity physical activity. Tobacco users, individuals with a history of cardiovascular disease, pulmonary disease, or neurological dysfunction, and females pregnant at the time of the study were excluded from the study. All subjects gave their written informed consent prior to participation and all testing procedures were approved by the Indiana University Institutional Review Board (IRB protocol #10953).

4.3.2 Experimental Equipment

4.3.2.1 Body composition measurements

Upon enrollment, subjects completed their gait analysis procedures and body composition measurements either on the same day or different days. A bioelectrical impedance analysis (BIA) was performed on each day to ensure body composition was consistent between days for subjects who chose to complete procedures on different days. A hand-to-leg BIA machine (Tanita, MC-780U plus P) was used to measure lower extremity total fat percentage and lean mass.

Lower extremity and pelvis BMD and BMC, lower extremity and gynoid fat mass, and lower extremity and gynoid lean mass were the independent variables obtained via a dual x-ray absorptiometry scan (iDXA; Lunar Prodigy, enCORE version 17, GE Medical Systems, Madison, WI, USA).

4.3.2.2 Gait analysis measurement

Shock attenuation during running was the dependent variable measured using inertial measurement units (IMUs) containing triaxial accelerometers (Blue Trident IMU, IMeasureU, Auckland, New Zealand; sampling frequency: 1600 Hz, capture range: ±20g, weight: 12 grams). Running trials were performed along an 18-m runway embedded with three 1200 Hz force plates (OR-6- 2000; AMTI Inc, Watertown, MA, USA). The force plates were located between two optoelectrical timing gates (TC-Gait, Brower Timing System, Draper, UT, USA) placed 6 meters apart at the center of the runway to monitor running speed. The force plates were synced with Qualisys Track Manager software (QTM; Qualisys AB, Göthenburg, Sweden) and were used to record ground reaction forces during each running trial.

4.3.3 Experimental Protocol

Subjects first completed a physical activity readiness questionnaire to ensure no contraindications for light to moderate physical activity. Subjects reported details about their current weekly running mileage, total years spent running, and running injury history via questionnaires. Subjects' height and weight were measured using a stadiometer and weighing scale, respectively. Then, if body composition measures were completed on a separate day as the shock attenuation analysis, BIA was completed as the subject stood on the scale barefoot and in form fitting clothing.

Subjects wore laboratory standard running shoes (Reebok; One X CrossFit Cushion 3.0) for the running gait analysis to standardize the effects of footwear on shock attenuation. Subjects warmed up for ~5-8 minutes on a treadmill at their preferred pace that would elicit ≤11 rate of perceived exertion (RPE) on the 6-20 scale (Borg, 1998). Then, IMUs were affixed bilaterally to the antero-medial distal tibia and the lower back (lumbosacral joint) and were secured with double sided tape and elastic bandages that were tightened firmly yet to the subject's comfort. Each IMU was oriented such that the X-axis was axial (upward = positive), Y-axis was mediolateral (right = positive), and Z-axis was anterior-posterior (anterior = positive).

Subjects performed 3-5 practice runs across the 18 m level-ground platform at the target running speed of 3.35 m/s ±5% to acclimate to the speed and running in the lab. The data collection began following the practice trials. At the beginning of each data collection trial, subjects performed a jump then immediately commenced the running trial at the target speed as the IMUs captured acceleration data. All subjects ran at the target running speed to standardize the effects of running speed on shock attenuation as it is affected by running speed (Mercer et al., 2002). Eight successful trials for each limb were retained for analysis. A successful trial was one where subjects ran within ±5% of the target running speed, contacted at least one force plate with their entire foot, running speed was maintained between timing gates, and subjects did not visibly modify their gait to target force plate contact. The jump performed at the beginning of each trial served to temporally align the IMUs during subsequent data processing. Additionally, the jump landing was manually flagged in QTM and subsequently used during data processing to isolate individual running stance phases in the acceleration data.

For the iDXA scan, subjects were required to remove their socks, shoes, and any clothing and jewelry containing metal per testing requirements. Subjects were provided with scrubs when necessary. They were instructed to lay supine on the iDXA platform within a demarcated rectangular area with their arms by their side. The researcher positioned the subject in the center of the rectangular region ensuring the mid-line of their body was congruent with that of the rectangular region. Once positioned, velcro straps were placed around the subject's knees and ankles to ensure no movement occurred during the scan. The scan lasted approximately 7 minutes.

4.3.4 Data processing

Acceleration data along the X, Y, and Z axes for both the lower back and tibia were exported from each IMU to MATLAB (Release 2022a, MathWorks, Natick, MA, USA). Each signal was converted from m/s^2 to the units of gravity (i.e. $1g = 9.81\ m/s^2$). Each signal was detrended by subtracting a least-squares best-fit line from the signal. The bias (or offset) in each signal, including the 1g collected by the accelerometers at rest, was removed such that the signal oscillated about zero.

4.3.5 Step Detection

Each IMU sensor started collecting data at slightly different times and was sampled at different frequencies (right leg: 1598.151 Hz, left leg: 1604.322, lower back: 1587.432 Hz) resulting in their misalignment. Therefore, it was necessary to align the lower back and tibial acceleration signals so that the frequency analysis could be performed on individual stance phases. To ensure that the start time for all sensors was identical, the location of the first Unix timestamp common to both the tibia and lower back signals was identified. This common timestamp was designated as the start of each signal. Then, each signal was resampled to 1600 Hz via linear interpolation utilizing the "*resample*" function in MATLAB.

To analyze the frequencies in the acceleration signals, individual stance phases performed at a constant running speed were isolated following the alignment of the tibial and low-back acceleration signals. This involved two steps: identifying the step run at constant speed and detecting the gait events of foot-contact and toe-off to isolate the stance phase (see Figure 1 for a schematic representation). The step that coincided with force plate contact was performed at a constant speed because the force plates were positioned between the timing gates where running speed was monitored. To identify this specific step in the acceleration signal, the landing of the jump performed at the beginning of each running trial was identified in the tibial acceleration signal (T_1) as the location of the first peak acceleration exceeding 3g. This peak was temporally aligned with the manually flagged jump landing in QTM. The instant of foot-contact on the force plate was identified within the QTM data as the instant when the vertical ground reaction force surpassed 20 Newtons. The QTM period was defined as the time elapsed between the manually flagged jump landing and the instant of foot-contact on the force plate. The manual flag and T_1 were expected to occur at the same time, therefore the QTM period was applied to the tibial acceleration signal starting at T_1. The step in the tibial acceleration signal that immediately followed the QTM period coincided with the force-plate contact for that limb. Thus, this step was identified as the step run at constant speed and was subsequently used to detect the gait events of foot-contact and toe-off to isolate the stance phase.

The manual flag in QTM may be prone to human error that may have resulted in mis-identification of the step coincident with force plate contact. Therefore, a supplementary analysis (supplemental material 1) was performed to identify the lag or lead time of the manual flag relative to $T_{1.}$ This analysis revealed that the manual flag preceded T_1 by ~0.16 seconds which was small enough because it ensured that the step in the acceleration signal preceding the one coincident with force-plate contact was excluded while the one coincident with force-plate contact was chosen.

Upon identifying the step at constant speed that needed to be isolated, the gait events of foot-contact and toe-off were determined in the acceleration signals using a validated foot-contact detection algorithm adapted from Aubol and Milner (2020) applied to the axial component of the tibial acceleration signal. Tibial acceleration data were first filtered using a zero-lag lowpass 4th order Butterworth filter with a cutoff frequency of 70 Hz. Then the derivative of acceleration determined the signal jerk. For only the tibial acceleration signal following force platform contact, peaks of jerk that occurred within 150 milliseconds of a peak in the tibial axial acceleration signal were identified. The local minima in the axial acceleration signal occurring within 75 milliseconds of each jerk peak were determined, while the local minima that occurred after a tibial acceleration peak which was <20% in magnitude of the third highest axial acceleration peak were removed. Finally, the location of the earliest local minima in the jerk signal occurring prior to the tibial acceleration peak was identified and marked as foot-contact. The adapted algorithm only identified the location of foot-contact but not toe-off. Therefore, the location of toe-off was identified using contact time which was calculated via force-plate data as the time between foot-contact and toe-off. The locations of foot-contact and toe-off thus identified were subsequently applied to all axes of the unfiltered tibial and lower back acceleration signals yielding individual stance phases for the frequency analysis.

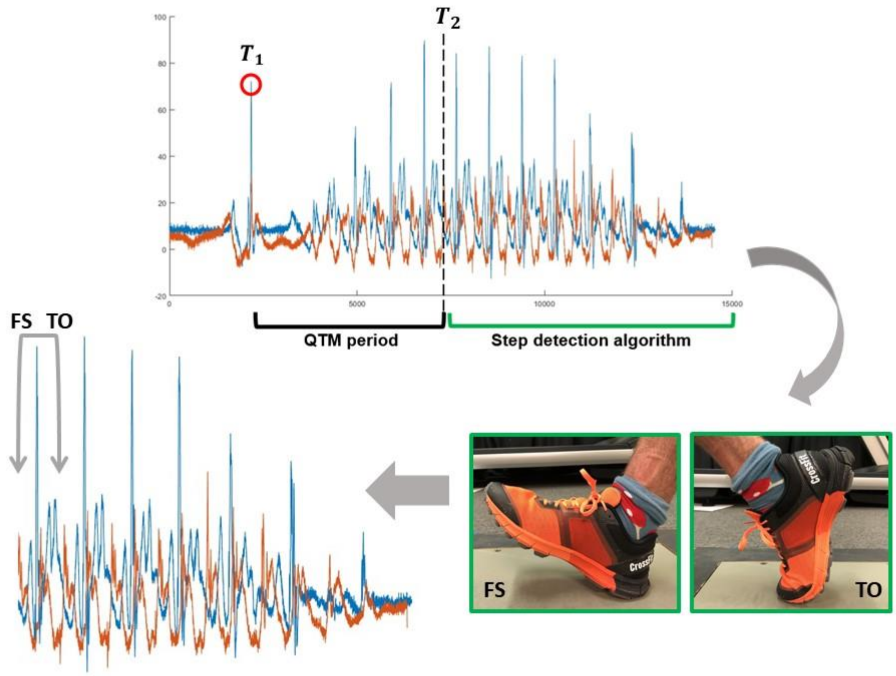

Figure 1. Method used to identify the same step in the force-plate and IMU data. The QTM period was the time elapsed between the jump landing at T_1 and foot-contact on a force plate at T_2. The black vertical dashed line in the top panel indicates the end of QTM period at T_2. The signal after T_2 was entered into the step detection algorithm to detect the gait events of foot strike (FS) and toe off (TO) subsequently applied to all acceleration components. T_1: location of the jump landing manually flagged in QTM and observed in the acceleration signal.

4.3.6 Frequency Domain Calculations

Frequency domain calculations were performed on individual unfiltered stance phases of the tibial and lower back axial and resultant acceleration signals. The resultant was calculated as $\sqrt{x^2 + y^2 + z^2}$. Each signal was zero-padded at both ends (Shorten and Winslow, 1992). Power spectral density (PSD) was computed using a Hanning window to yield measures of signal power. The PSD was normalized to 1Hz bins for frequencies between 0Hz and 0.5*sampling frequency (Edwards et al., 2012; Gruber et al., 2014; Hamill et al., 1995). The degree of shock attenuation was calculated using a transfer function as follows:

$$Shock\ attenuation = 10\ \log_{10}\left(\frac{PSD_{lower\ back}}{PSD_{tibia}}\right)$$

where *Shock attenuation* is the shock attenuated (negative values) or gained (positive values) between the tibia and lower back, and $PSD_{lower\ back}$ and PSD_{tibia} are the power spectral densities for the lower back and tibia, respectively.

4.3.7 iDXA data

Body composition measures for both lower extremities (BMC, BMD, lean mass, fat mass), pelvis (BMD), and gynoid (lean mass and fat mass) were exported from the iDXA software to R Studio (version 2022.07.1) for statistical analysis. The measures of bone mass yielded by the iDXA were areal rather than volumetric measurements.

4.4 Statistical Analysis

4.4.1 Bilateral differences

Data for the degree of shock attenuation within FB1, FB2, FB3, FB4 and lower extremity BMD, BMC, fat mass, and lean mass were collected and computed bilaterally to account for potential asymmetries in shock attenuation and body composition between limbs. To test for bilateral differences, bilateral data for lower extremity BMD, BMC, fat mass, lean mass, and each shock attenuation variable were first assessed for normality using the Shapiro-Wilk test ($\alpha \leq 0.05$). Repeated measures t-tests were used to assess differences between subjects' right and left limbs for normally distributed variables, while Wilcoxon Signed-rank tests ($\alpha \leq 0.05$) were used for non-normally distributed variables. This between-limb comparison revealed no significant differences between limbs ($p>0.05$) and therefore lower extremity, BMC, fat mass, lean mass, and each shock attenuation variable was averaged across limbs for each subject for regression analyses.

4.4.2 Regression Analysis

Unadjusted linear regression was used to assess the relationship of shock attenuation within FB1-FB4 to lower extremity and pelvis BMD and BMC, lower extremity and gynoid lean mass and fat mass. Each regression model was assessed for assumption violations. All statistical analyses were performed in

Rstudio (version 2022.07.1). Unadjusted models were used because a sensitivity analysis adjusting each model for age and sex, variables that influence body composition and running mechanics (Besson et al., 2022; Devita et al., 2016; Ponti et al., 2020), revealed no alterations to the associations between body composition and shock attenuation upon adjustment (supplemental content 1). Given that shock attenuation contains negative magnitudes, positive beta coefficients indicate a decrease in shock attenuation with an increase in the body composition variable, and negative coefficients indicate an increase in shock attenuation with an increase in the body composition variable.

4.5 Results

Anthropometrics and demographics (mean±1 standard deviation) for the 38 subjects are reported in Table 1. Summary statistics for each body composition variable and the frequency characteristics of impact shock and shock attenuation are presented in Table 2.

Table 1. Mean ± 1 SD anthropometric and demographic data (N=38).

	Group Mean ± 1 SD
Age (years)	31.55±14.94
Sex distribution	17 female, 21 male
Height (m)	1.71±0.08
Body mass (kg)	68.36±10.35
Years spent running	10.75±12.38
Current weekly mileage (miles/week)	12.56±5.95

SD: standard deviation

4.5.1 Axial Shock Attenuation

An increase in lower extremity fat mass significantly decreased shock attenuation in FB2 (p-value=0.014, beta coefficient (β)=0.015, 95% confidence interval (CI)=0.003-0.028) and FB4 (p=0.033, β=0.032, CI=0.002-0.061). Similarly, an increase in gynoid fat mass significantly decreased shock attenuation within FB2 (p=0.008, β=0.014, CI=0.003-0.024), FB3 (p=0.034, β=0.021, CI=0.001-0.041), and FB4 (p=0.012, β=0.031, CI=0.007-0.055). There were no other significant associations between body composition and axial shock attenuation (p>0.050) (Table 3).

4.5.2 Resultant Shock Attenuation

Resultant shock attenuation across frequency bands was not significantly associated with body composition variables ($p>0.050$) (Table 3).

4.5.3 Sensitivity analysis

Thirteen subjects chose to complete the iDXA scan on a separate day than their gait analysis visit (number of days between visits: 10.84±9.13). On average, there was a net increase in BIA derived %fat mass in the lower extremity (right leg: +5%±11%, left leg: +4%±10%) and a corresponding net decrease in BIA derived lean mass in the lower extremity (right leg: -2%±2%, left leg: -1%±2%) for these thirteen subjects. Wilcoxon Signed-rank tests ($\alpha \leq 0.05$) revealed that the BIA derived lean mass was significantly different between visits for the right limb ($p = 0.025$, CI = 0.00002-0.25), but not for the left limb. However, the absolute mean difference in lean mass between visits for the right (0.16 kg) and for the left (0.13 kg) was similar to that found in a previous systematic review (0.15 kg) (Talma et al., 2013), suggesting these differences may be due to measurement error rather than a meaningful change in body composition between visits. Lower extremity fat percentage was not significantly different between visits for right limb ($p=0.076$, CI = -2.13 – 0.12) nor the left limb ($p=0.094$, CI = -2.15 – 0.30). Thus, a sensitivity analysis was performed to identify if the significant change in lean mass influenced the associations between leg lean mass and shock attenuation. For the sensitivity analysis, the thirteen subjects were excluded and unadjusted linear regression models were implemented as described above using the remaining 25 subjects. One regression model was run for each lean mass and dependent variable paring, then the results of these models with 25 subjects were compared to the model that included the entire sample (N=38). There was no difference in model results regardless of whether the entire sample or the subsample excluding those performing the gait and iDXA procedures on separate days. Specifically, lean mass remained unassociated with axial and resultant shock attenuation within frequency bands FB1 (axial: $p=0.427$, $\beta=-0.001$, CI=-0.006-0.002; resultant: $p=0.169$, $\beta=0.003$, CI=-0.001-0.007), FB2 (axial: $p=0.427$, $\beta=-0.004$, CI=-0.014-0.005; resultant: $p=0.440$, $\beta=0.004$, CI=-0.006-0.015), FB3 (axial:

p=0.406, β=0.006, CI=-0.010-0.024; resultant: p=0.632, β=0.003, CI=-0.010-0.016), and FB4 (axial: p=0.400, β=0.009, CI=-0.013-0.031; resultant: p=0.073, β=0.017, CI=-0.001-0.011) with similar beta coefficients compared to the primary analysis.

Table 2. Group means ± 1 standard deviation for each body composition measure and shock attenuation for the tibia and lower back for both vertical and resultant acceleration components.

Body composition	**Group mean ± 1 SD**
Lower extremity BMD (gm/cm^2)	1.30±0.14
Lower extremity BMC (gm)	511.86±95.15
Lower extremity fat mass (gm	3028.41±902.85
Lower extremity lean mass (gm	8356.55±1683.52
Pelvis BMD (gm/cm^2)	1.01±0.14
Pelvis BMC (gm)	354.82±74.77
Gynoid fat mass (gm)	3004.98±1068.64
Gynoid lean mass (gm)	7631.09±1534.59
Axial shock Attenuation	**Group mean ± 1 SD**
3-8 Hz (dB)	-5.02±18.37
9-20 Hz (dB)	-94.94±36.24
21-35 Hz (dB)	-98.53±66.99
36-50 Hz (dB)	-103.30±84.01
Resultant shock Attenuation	**Group mean ± 1 SD**
3-8 Hz (dB)	-52.97±15.52
9-20 Hz (dB)	-54.33±44.07
21-35 Hz (dB)	-45.72±49.85
36-50 Hz (dB)	-104.14±70.22

SD: standard deviation; gm: grams; g: gravity; dB: decibel

Table 3. Regression model results for each independent-dependent variable pairing.

	β	**p-value**	**95% CI**
Axial Acceleration			
Lower extremity BMD (grams/cm^2)			
SA 3-8 Hz	-7.50	0.730	-51.32 – 36.30
SA 9-20 Hz	22.40	0.602	-108.64 – 63.84
SA 21-35 Hz	-2.62	0.974	-162.68 – 157.42
SA 36-50 Hz	101.80	0.306	-95.98 – 299.50
Lower extremity BMC (grams)			
SA 3-8 Hz	0.005	0.875	-0.06 – 0.07

SA 9-20 Hz	-0.020	0.749	-0.14 – 0.10
SA 21-35 Hz	0.08	0.463	-0.14 – 0.32
SA 36-50 Hz	0.18	0.206	-0.10 – 0.47
Lower extremity lean mass (grams)			
SA 3-8 Hz	-0.001	0.928	-0.003 – 0.003
SA 9-20 Hz	-0.001	0.650	-0.008 – 0.005
SA 21-35 Hz	0.006	0.297	-0.006 – 0.02
SA 36-50 Hz	0.01	0.135	-0.004 – 0.02
Lower extremity fat mass (grams)			
SA 3-8 Hz	0.0009	0.777	-0.005 – -.007
SA 9-20 Hz	0.015	0.014*	0.003 – 0.028
SA 21-35 Hz	0.021	0.077	-0.002 – 0.045
SA 36-50 Hz	0.032	0.033*	0.002 – 0.06
Pelvis BMD (grams/cm^2)			
SA 3-8 Hz	-21.28	0.306	-62.88 – 20.31
SA 9-20 Hz	-17.48	0.672	-100.56 – 65.59
SA 21-35 Hz	-65.05	0.392	-217.45 – 87.35
SA 36-50 Hz	105.98	0.264	-83.75 – 295.71
Pelvis BMC (grams)			
SA 3-8 Hz	-0.04	0.304	-0.12 – 0.03
SA 9-20 Hz	-0.04	0.574	-0.20 – 0.11
SA 21-35 Hz	-0.02	0.844	-0.33 – 0.27
SA 36-50 Hz	0.23	0.207	-0.13 – 0.60
Gynoid fat mass (grams)			
SA 3-8 Hz	0.0003	0.898	-0.005 – 0.006
SA 9-20 Hz	0.01	0.008*	0.003 – 0.02
SA 21-35 Hz	0.02	0.034*	0.001 – 0.04
SA 36-50 Hz	0.03	0.012*	0.007 – 0.05
Gynoid lean mass (grams)			
SA 3-8 Hz	-0.001	0.404	-0.005 – 0.002
SA 9-20 Hz	-0.004	0.213	-0.01 – 0.002
SA 21-35 Hz	0.001	0.830	-0.01 – 0.016
SA 36-50 Hz	0.009	0.307	-0.008 – 0.02
Resultant Acceleration			
Lower extremity BMD (grams/cm^2)			
SA 3-8 Hz	1.42	0.938	-35.67 – 38.51
SA 9-20 Hz	-9.66	0.853	-114.92 –96.586
SA 21-35 Hz	0.19	0.997	-118.90 – 119.29
SA 36-50 Hz	7.79	0.925	-159.94 – 175.53
Lower extremity BMC (grams)			
SA 3-8 Hz	0.008	0.752	-0.04 – 0.06

SA 9-20 Hz	-0.012	0.876	-0.16 – 0.14
SA 21-35 Hz	-0.005	0.946	-0.18 – 0.17
SA 36-50 Hz	0.09	0.423	-0.14 – 0.34
Lower extremity lean mass (grams)			
SA 3-8 Hz	0.001	0.266	-0.001–0.004
SA 9-20 Hz	0.001	0.679	-0.007 – 0.01
SA 21-35 Hz	0.002	0.653	-0.007 – 0.01
SA 36-50 Hz	0.01	0.114	-0.002 – 0.02
Lower extremity fat mass (grams)			
SA 3-8 Hz	0.004	0.109	-0.001 – 0.01
SA 9-20 Hz	0.006	0.405	-0.009 – 0.02
SA 21-35 Hz	0.01	0.231	-0.007 – 0.02
SA 36-50 Hz	0.005	0.673	-0.020 – 0.03
Pelvis BMD (grams/cm^2)			
SA 3-8 Hz	-5.43	0.758	-41.07 – 30.20
SA 9-20 Hz	2.37	0.962	-98.92 – 103.67
SA 21-35 Hz	28.34	0.618	-85.82 – 142.51
SA 36-50 Hz	-6.78	0.939	-168.16 – 154.58
Pelvis BMC (grams)			
SA 3-8 Hz	0.01	0.693	-0.05 – 0.08
SA 9-20 Hz	0.07	0.472	-0.12 – 0.26
SA 21-35 Hz	0.05	0.643	-0.17 – 0.27
SA 36-50 Hz	0.08	0.582	-0.22 - 0.40
Gynoid fat mass (grams)			
SA 3-8 Hz	0.004	0.056	-0.001-0.009
SA 9-20 Hz	0.008	0.222	-0.005 – 0.02
SA 21-35 Hz	0.01	0.080	-0.001 – 0.02
SA 36-50 Hz	0.007	0.468	-0.01 – 0.03
Gynoid lean mass (grams)			
SA 3-8 Hz	0.001	0.513	-0.002 – 0.004
SA 9-20 Hz	0.0003	0.949	-0.009 – 0.01
SA 21-35 Hz	-0.001	0.763	-0.01 – 0.009
SA 36-50 Hz	0.005	0.462	-0.009 – 0.02

β: beta coefficient; CI: confidence interval; BMD: bone mineral density; BMD: bone mineral content; Hz: hertz; SA: shock attenuation within the given frequency band; *statistically significant at the p<0.050 level.

4.6 Discussion

This study aimed to examine the associations of lower extremity and pelvis BMD, lower extremity and gynoid lean mass, and fat mass to the degree of axial and resultant shock attenuated within 3-8 Hz, 9-20 Hz, 21-35 Hz, and 36-50 Hz during running. In line with our hypothesis, the degree of shock attenuated within 3-8 Hz was not significantly associated with any body composition measure for either the shock attenuated along the axial or resultant components. These null associations are consistent with previous evidence that frequencies less than 5 Hz propagate towards the head while those between 5-8 Hz and above may be attenuated actively through, for example, increasing peak knee flexion at and during ground contact (Edwards et al., 2012; Lafortune et al., 1996). Such active attenuation and proximal propagation may eliminate the need to rely on passive tissues for the attenuation of 3-8 Hz.

Given that the attenuation by bone and soft tissue increased with an increase in frequencies from 3 Hz-500 Hz in an animal model (Paul et al., 1978), it was hypothesized that increasing lower extremity and pelvis BMD and BMC, and leg and gynoid lean mass and fat mass would enable more shock attenuation of frequencies greater than 8 Hz during running in humans. However, BMD, BMC, and lean mass variables were not associated with shock attenuation. Interestingly, an increase in lower extremity fat mass was associated with reduced axial shock attenuation across 9-50 Hz. While the attenuation of frequencies >8 Hz is proposed to be a function of passive mechanisms, active mechanisms like knee flexion at and during ground contact may contribute to attenuating frequencies >5 Hz (Edwards et al., 2012). Thus, the null associations of BMD, BMC, and lean mass to both axial and resultant shock attenuation may be explained by the active attenuation of impact-related high frequencies occurring through muscle activation prior to foot-contact or activation immediately following impact (Connick and Li, 2014; Novacheck, 1998) which may have masked the contributions of passive tissues. The quadriceps, for example, are activated approximately 78 milliseconds before initial contact and eccentrically contract to prevent excessive knee flexion at contact; eccentric muscle actions aid shock absorption post-impact due to the energy absorbed via the elongation of elastic elements in the muscle-tendon complex (Garrett et al., 1987). Muscle activity may mask the passive contributions of lean mass to shock attenuation because lean mass includes tendons, cartilage, connective tissues, and muscle mass. To isolate the effects of lean mass from muscle activation effects, it may be necessary to account for muscle activity in future work.

Another reason for the null association of BMD and BMC to axial and resultant shock attenuation may be that the iDXA derived properties of bone did not include architecture or structure, which also contribute to bone mechanical properties (Nigg and Herzog, 2007). Bone mass, as measured by the iDXA, is a clinical correlate of bone strength and stiffness which contribute to bone's ability to deform and dissipate energy (Carbonare and Giannini, 2004). Thus, it was hypothesized that bone mineral density and content, which constitute bone mass, would be associated with shock attenuation. However, bone cross sectional area, width, thickness, porosity, volume, shape, trabeculae number, and trabeculae connectivity dictate the mechanical properties of bone in addition to bone mass (Carbonare and Giannini, 2004; van der Linden and Weinans, 2007). Thus, future studies can use techniques like computed tomography, magnetic resonance imaging, and finite element modeling to assess the material properties of biological tissue which may complement iDXA measurements to provide insights about a runner's ability to attenuate shock.

The heel fat pad and wobbling mass in the tibia, comprising both fat and lean mass, are known to dissipate significant impact-related energy, as demonstrated by animal models and cadaver studies (Pain and Challis, 2001; Paul et al., 1978). Therefore, it was expected that shock attenuation would increase with higher fat mass in addition to lean mass. However, the current data revealed that an increase in lower extremity fat mass led to reduced axial shock attenuation within the 9-20 Hz and 36-50 Hz frequency ranges. Similarly, an increase in gynoid fat mass resulted in reduced axial shock attenuation within the 9-20 Hz, 21-35 Hz, and 36-50 Hz ranges. When normalized to height, fat mass can predict the relationship between skeletal muscle and adipose tissue, indicating that an increase in fat mass is associated with a decline in skeletal muscle relative to adipose tissue (Schautz et al., 2012). This decline in skeletal muscle due to increased fat mass may diminish the active contributions of eccentric muscle actions that may facilitate attenuation of frequencies greater than 5 Hz (Edwards et al., 2012). Although the current sample did not show strong correlations between lower extremity fat mass and lean mass ($p=0.096$, Pearson's $r = -0.273$), gynoid fat mass was negatively correlated with gynoid lean mass ($p=0.038$, Pearson's $r = -0.337$). Consequently, the concentration of fat mass in the gynoid region may decrease the axial shock attenuation capabilities for frequencies greater than 8 Hz, potentially due to its negative association with skeletal muscle.

It was expected that fat mass would contribute to shock attenuation due to its material compliance and previously observed role in energy dissipation (Guimarães et al., 2020; Pain and Challis, 2001; Paul et al., 1978). However, it is possible that the negative correlation between gynoid fat mass and lean mass contributed to the unexpected decrease in axial shock attenuation of 9-50 Hz with an increase in gynoid fat mass. Previously, an increase in intermuscular fat tissue has been linked to limitations in muscle power production potentially due to muscle activation deficits (Reid et al., 2014). Therefore, a decline in lean and thereby muscle mass due to an increase in fat mass may reduce the degree of muscle pre-activation which may otherwise contribute to shock absorption post-impact.

4.7 Conclusion

Bone mineral density and content in the lower extremity and pelvis, and fat mass and lean mass in the lower extremity and gynoid regions were not associated with the attenuation of axial and resultant shock between 3-8 Hz. Therefore, it is concluded that the lower-frequency energy may not be passively attenuated. However, other structural tissue properties in addition to tissue mass may contribute to passive attenuation and should be taken into consideration. The effects of fat mass on the contribution of active muscle to high-frequency attenuation must be explored as it may underlie the observed inverse associations between fat mass and axial attenuation of 9-50 Hz in this study,

4.8 Supplemental Content

Supplemental Content 1. Regression model results for each independent-dependent variable pairing while adjusting for age and sex. The bolded text indicates the results differed from the unadjusted linear regression models.

	β	p-value	95% CI
Axial Acceleration			
Legs BMD (grams/cm^2)			
SA 3-8 Hz	11.21	0.711	-49.891 – 72.325
SA 9-20 Hz	17.03	0.771	-100.879 – 134.959
SA 21-35 Hz	-20.60	0.821	-204.502 – 163.301
SA 36-50 Hz	78.59	0.554	-188.952 – 346.147
Legs BMC (grams)			
SA 3-8 Hz	0.07	0.185	-0.038 – 0.191
SA 9-20 Hz	0.05	0.596	-0.166 – 0.286
SA 21-35 Hz	-0.01	0.942	-0.367 – 0.341
SA 36-50 Hz	0.09	0.701	-0.418 – 0.615
Legs lean mass (grams)			
SA 3-8 Hz	0.004	0.271	-0.003 – 0.012
SA 9-20 Hz	0.002	0.712	-0.012 – 0.018
SA 21-35 Hz	0.004	0.695	-0.019 – 0.028
SA 36-50 Hz	0.01	0.467	-0.022 – 0.047
Legs fat mass (grams)			
SA 3-8 Hz	-0.0007	0.840	-0.008 – 0.007
SA 9-20 Hz	0.01	0.037*	9.125 – 0.029
SA 21-35 Hz	0.02	0.037*	0.001 – 0.045
SA 36-50 Hz	0.045	0.004*	0.014 – 0.075
Pelvis BMD (grams/cm^2)			
SA 3-8 Hz	-13.76	0.574	-63.092 – 35.557
SA 9-20 Hz	14.02	0.767	-81.407 – 109.456
SA 21-35 Hz	-54.98	0.454	-202.699 – 92.726
SA 36-50 Hz	112.50	0.293	-101.597 – 326.616
Pelvis BMC (grams)			
SA 3-8 Hz	-0.03	0.564	-0.145 – 0.080
SA 9-20 Hz	0.02	0.853	-0.198 – 0.238
SA 21-35 Hz	-0.09	0.577	-0.433 – 0.245
SA 36-50 Hz	0.21	0.376	-0.275 – 0.709
Gynoid fat mass (grams)			
SA 3-8 Hz	-0.001	0.617	-0.008 – 0.005
SA 9-20 Hz	0.01	0.033*	0.001 – 0.025
SA 21-35 Hz	0.01	0.043*	0.0006 – 0.039

SA 36-50 Hz	0.04	0.002*	0.016 − 0.068
Gynoid lean mass (grams)			
SA 3-8 Hz	-0.0009	0.803	-0.009 − 0.007
SA 9-20 Hz	-0.006	0.387	-0.021 − 0.008
SA 21-35 Hz	-0.005	0.643	-0.029 − 0.018
SA 36-50 Hz	0.004	0.819	-0.031 − 0.039
Resultant Acceleration			
Legs BMD (grams/cm^2)			
SA 3-8 Hz	1.20	0.958	-45.981 − 48.392
SA 9-20 Hz	30.66	0.659	-109.537 − 170.858
SA 21-35 Hz	35.15	0.661	-126.431 − 196.734
SA 36-50 Hz	-92.64	0.370	-299.996 − 114.714
Legs BMC (grams)			
SA 3-8 Hz	-0.01	0.662	-0.110 − 0.071
SA 9-20 Hz	0.01	0.900	-0.254 − 0.287
SA 21-35 Hz	-0.004	0.978	-0.316 − 0.308
SA 36-50 Hz	-0.19	0.332	-0.591 − 0.205
Legs lean mass (grams)			
SA 3-8 Hz	0.003	0.260	-0.002 − 0.009
SA 9-20 Hz	0.01	0.166	-0.005 − 0.030
SA 21-35 Hz	0.01	0.298	-0.010 − 0.031
SA 36-50 Hz	0.004	0.748	-0.023 − 0.031
Legs fat mass (grams)			
SA 3-8 Hz	0.004	0.087	-0.007 − 0.010
SA 9-20 Hz	0.003	0.690	-0.014 − 0.021
SA 21-35 Hz	0.009	0.334	-0.010 − 0.030
SA 36-50 Hz	0.01	0.432	-0.016 − 0.036
Pelvis BMD (grams/cm^2)			
SA 3-8 Hz	-0.93	0.960	-39.126 − 37.253
SA 9-20 Hz	41.91	0.455	-70.941 − 154.766
SA 21-35 Hz	69.75	0.279	-59.119 − 198.632
SA 36-50 Hz	-29.64	0.724	-199.174 − 139.879
Pelvis BMC (grams)			
SA 3-8 Hz	0.02	0.639	-0.066 − 0.107
SA 9-20 Hz	0.21	0.093	-0.037 − 0.461
SA 21-35 Hz	0.15	0.304	-0.143 − 0.447
SA 36-50 Hz	-0.04	0.826	-0.430 − 0.346
Gynoid fat mass (grams)			
SA 3-8 Hz	0.004	0.083	-0.0006 − 0.009
SA 9-20 Hz	0.004	0.557	-0.010 − 0.020
SA 21-35 Hz	0.01	0.164	-0.005 − 0.029

SA 36-50 Hz	0.009	0.416	-0.013 – 0.032
Gynoid lean mass (grams)			
SA 3-8 Hz	0.002	0.407	-0.003 – 0.008
SA 9-20 Hz	0.009	0.282	-0.008 – 0.027
SA 21-35 Hz	-0.001	0.916	-0.022 – 0.020
SA 36-50 Hz	0.002	0.407	-0.003 – 0.008

β: beta coefficient; CI: confidence interval; BMD: bone mineral density; BMD: bone mineral content; Hz: hertz; SA: shock attenuation within the given frequency band; *statistically significant at the p<0.050 level.

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5 CHAPTER 5: Study 2

Manuscript Title: Active Mechanisms of Shock Attenuation During Human Running

Intended Journal: Journal of Experimental Biology

5.1 Abstract

BACKGROUND. A frequency-domain analysis of impact shock attenuation between the tibia and lower back may indicate the mechanisms used to absorb shock and thereby running-related injury risk. Identifying the degree to which previously proposed gait-related mechanisms contribute to shock attenuation across the frequency spectrum may reveal the reliance on and thereby injury risk to underlying musculoskeletal structures. **PURPOSE**. Identify the associations between previously proposed gait-related shock attenuation mechanisms and the degree of shock attenuation within 3-50 Hz. **METHODS**. 57 runners aged 32.66±15.32 years ran overground at 3.35 $m \cdot s^{-1}$ as three-dimensional motion capture and triaxial accelerometers recorded their gait mechanics and tibial and lower back accelerations, respectively. Peak sagittal knee and frontal ankle angles during stance, and ankle and knee joint angles, stiffnesses, negative work, and power absorption during deceleration (independent variables) were calculated during deceleration. The power spectral density of each acceleration signal along the axial and resultant components transformed it into the frequency domain and shock attenuation (dependent variables) within each frequency bin (FB1=3-8, FB2=9-20, FB3=21-35, FB4=36-50 Hz) was calculated using a transfer function of the lower back relative to the tibia. Unadjusted linear regression ($\alpha > 0.050$) was used to examine the associations between each independent and dependent variable. **RESULTS**. In FB1, axial shock attenuation increased with knee joint compliance (increased knee range of motion (ROM) and decreased knee stiffness), ankle dorsiflexion at contact, and peak ankle eversion during stance. Resultant attenuation in FB1 increased with peak knee flexion and sagittal ankle range of motion. Axial attenuation in FB1 decreased with sagittal ankle range of motion, while resultant

attenuation decreased with ankle dorsiflexion at contact and sagittal knee range of motion. In FB2, axial attenuation increased with knee joint compliance and ankle dorsiflexion at contact. Resultant attenuation in FB2 increased with peak knee flexion, knee stiffness, and sagittal ankle range of motion. Axial attenuation in FB2 decreased with sagittal ankle range of motion, while resultant attenuation decreased with sagittal knee range of motion and ankle dorsiflexion at contact. In FB3, resultant attenuation increased with peak knee flexion during stance. In FB4, axial attenuation increased with peak knee flexion during stance, while resultant attenuation increased with sagittal ankle range of motion. Resultant attenuation in FB4 decreased with ankle dorsiflexion at contact. **CONCLUSIONS**. Gait-related mechanisms of shock attenuation are not limited to attenuating frequencies <8 Hz, as previously thought. Individuals who run with a more flexed knee and dorsiflexed ankle at contact, and low knee and ankle stiffness may demand less attenuation contribution from passive tissues which may influence injury type, though the associations between gait-related mechanisms and the degree of shock attenuated depends on the acceleration component considered.

5.2 Introduction

Peak impact shock is a running-related injury (RRI) risk factor commonly analyzed in the time domain (Davis et al., 2004; Milner et al., 2006). While such an analysis is informative of the magnitude of impact shock experienced at the lower leg, it does not explain the body's response to the shock. Rather, frequency domain analysis of shock attenuation (i.e., absorption) enables a quantification of the body's response to shock by examining the mechanisms used to attenuate shock as it is transmitted from the lower leg proximally.

The events of foot-ground collision generate shock vibrations between 3-50 Hz (Gruber et al., 2014; Hamill et al., 1995; Shorten and Winslow, 1992) that are propagated proximally towards the head. Along the way, shock is absorbed both actively and passively such that only 8%-15% of the initial impact shock measured at the distal tibia is permitted to reach the head (Ratcliffe and Holt, 1997). As a result, shock attenuation protects proximal musculoskeletal structures between the tibia and the head from excessive accelerations, thereby maintaining vision and balance. Shock attenuation within the 3-8 Hz and 9-20 Hz frequency bands is analyzed most often because the vibration-related energy spikes within these frequency ranges. However, the presence of frequencies up to 50 Hz during running warrants an analysis of frequencies between 21-50 Hz as well because the body necessarily absorbs them. Frequencies in the range of 3-8 Hz are proposed to be attenuated by active mechanisms involving eccentric muscle contractions. On the other hand, frequencies above 8 Hz are thought to be attenuated by passive mechanisms involving soft and rigid musculoskeletal tissue deformation (Nigg et al., 1995; Shorten and Winslow, 1992). Consequently, the degree of shock attenuated within 3-8 Hz and >8 Hz may indicate the reliance on and thereby the stress experienced by active and passive shock attenuation mechanisms, giving us insight into RRI risk.

The proposed gait-related mechanisms for active impact shock attenuation (Figure 1) occurring during stance include negative mechanical work performed by joints (Derrick et al., 1998; Hamill et al., 2014), knee flexion motion during ground contact (Edwards et al., 2012; Lafortune et al., 1996), knee and ankle joint stiffness (Hamill et al., 2014), and ankle motion in the frontal (Maharaj et al., 2016) and sagittal planes (Gruber et al., 2014; Gruber et al., 2017; Nigg and Morlock, 1987; Perry and Lafortune, 1995). Although active mechanisms are suggested to attenuate the energy within 3-8 Hz, the role of each proposed active mechanism in attenuating specific frequencies during running within or beyond this range is unknown. Identifying the degree to which proposed active mechanisms contribute to shock attenuation may reveal the kinematic and kinetic strategies used to respond to impact shock.

Previous human pendulum experiments revealed larger knee flexion angles at and during foot-ground contact increase the attenuation of energy above 5 Hz (Lafortune et al., 1996), and between 15-35 Hz during in-line skating (Edwards et al., 2012). Thus, the knee may act a low-pass filter, permitting the proximal transmission of lower frequencies while attenuating frequencies >5 Hz. More knee flexion than extension at contact orients the line of action of the impact force vector away from the knee joint center, resulting in augmented shock attenuation achieved by the elongation of elastic components in muscle-tendon units as leg segments rotate (Derrick, 2004). During running, therefore, knee flexion angle at initial contact and its peak during mid-stance may be shock attenuation mechanisms. Knee flexion during stance is enabled via eccentric knee extensor muscle actions that perform negative mechanical work during deceleration, therefore negative mechanical work at the knee may hypothetically contribute to attenuation too, as observed with longer stride/step lengths (Baggaley et al., 2020; Derrick et al., 1998).

Negative mechanical work and knee flexion during stance are inversely associated with knee joint stiffness indicating that low knee stiffness may also be related to shock attenuation. Groucho running, a style characterized by exaggerated knee flexion and low knee stiffness, results in greater shock attenuation than normal running (McMahon et al., 1987). Since a compliant knee is also likely to have a larger range of motion than stiff knee, knee flexion-extension range of motion could additionally dictate the degree of shock actively attenuated which may shift attenuation responsibilities away from passive musculoskeletal structures.

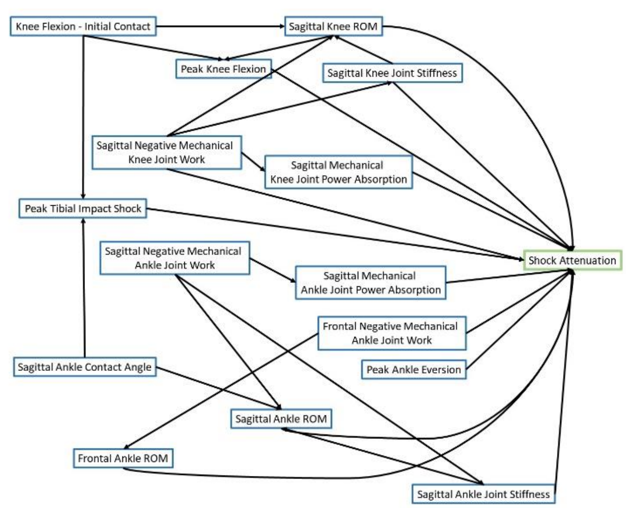

Figure 1. Web diagram for the knee and ankle joint kinematic and kinetic variables proposed to be mechanisms of active shock attenuation.

The ankle also may also play a significant role in shock attenuation. Sagittal plane ankle motion contributes to mechanical energy absorption during early stance (Winter, 1983). Energy absorption may be promoted by a less stiff ankle joint (Hamill et al., 2014) that is characterized by a larger range of motion and greater negative mechanical work compared to a stiffer joint. However, the degree to which each of these ankle joint kinematic and kinetic variables contribute to the attenuation of signal energy associated with shock vibrations across the frequency spectrum is unknown. Further, rearfoot runners attenuate more shock than forefoot runners (Gruber et al., 2014) suggesting ankle angle at initial contact may dictate the degree of shock attenuated possibly due to its influence on ankle and knee joint kinematics and kinetics during deceleration (Almeida et al., 2015; Hamill et al., 2014). In the frontal plane, subtalar joint pronation (i.e., eversion, dorsiflexion, and abduction) during early stance is accompanied by negative mechanical work performed at the ankle and lengthening of the tibialis-posterior muscle-tendon unit (Maharaj et al., 2016), all of which may contribute to shock attenuation. Identifying the degree to which each aforementioned variable contributes to shock attenuation of different frequencies will be an important first step towards understanding the demands placed on the underlying musculoskeletal structures.

The associations of previously postulated joint kinetic and kinematic shock attenuation mechanisms with shock attenuation across frequencies must be established to understand the body's response to impact shock. This understanding may be relevant to examining how a runner's gait may underlie their reliance on active versus passive shock attenuation mechanisms and thereby injury risk to the underlying musculoskeletal structures. Therefore, this study aimed to examine the relationship of sagittal ankle and knee contact angles, ankle and knee joint stiffness, joint ranges of motion at the ankle (frontal and sagittal planes) and knee (sagittal plane), negative mechanical joint work at the ankle and knee during deceleration, and peak ankle eversion and knee flexion during running stance to the degree of shock attenuated within frequency bands of 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4). Considering that impact shock is three-dimensional (LaFortune, 1991;

Sheerin et al., 2019; Van den Berghe et al., 2019a), this study aimed to assess the aforementioned associations along both its vertical and resultant acceleration components. The following hypotheses were formulated: (1) Knee and ankle joint stiffness during deceleration will be negatively associated with the degree of shock attenuation within FB1 but not within FB2, FB3, or FB4; (2) Frontal and sagittal plane ankle joint ROM and sagittal plane knee ROM during deceleration will be positively associated with the degree of shock attenuation within FB1, but not within FB2, FB3, or FB4; (3) Sagittal plane knee and ankle angle at contact, and peak ankle eversion and knee flexion magnitude during deceleration will be positively associated with the degree of shock attenuation within FB1, but not within FB2, FB3, or FB4; and (4) sagittal plane negative joint work and power absorption at the ankle and knee during deceleration will be positively associated with the degree of shock attenuation within FB1, but not within FB2, FB3, or FB4.

5.3 Methods

5.3.1 Sample Characteristics

A-priori power calculations were performed using data from Edwards et al. (2012) revealed that a sample size of N=29 was required to detect an association between gait biomechanics and shock attenuation of size $f^2 = 1.00$ with 80% statistical power (alpha=0.05). Therefore, N=57 runners were voluntarily enrolled in the study to include a wide range of ages and running experience levels, making results generalizable across runners of different ages and exposures to running. Inclusion criteria required runners to 1) be between 18-65 years old, 2) have been running between 3-30 miles per week for either less than or greater than 1 year, 3) have no history of lower back or lower limb orthopedic surgery in the past 10 years, 4) if an injury was developed in the past 6 months, they must have returned to normal running training volume for ≥8 weeks before enrollment, and 5) be free of chronic illnesses that disallowed engagement in ≥15 minutes of low to moderate-intensity physical activity. Tobacco users and females pregnant at the time of the study were excluded. All subjects gave their written informed consent before participation and all testing procedures were approved by the Indiana University Institutional Review Board (IRB Protocol #10585).

5.3.2 Experimental Protocol

Before their three-dimensional gait analysis, subjects completed the physical activity readiness questionnaire (PARQ) and self-reported their weekly running mileage, total years spent running, and running injury history via questionnaires. Subjects' height and weight were measured using a stadiometer and weighing scale. Subjects were required to wear laboratory-standard running shoes (Reebok; One X CrossFit Cushion 3.0) to standardize the effects of footwear on impact shock and attenuation. Next, subjects warmed up for ~5-8 minutes on a treadmill at their self-selected comfortable pace, ensuring light exertion, i.e., ratings of perceived exertion <11 on a 6-20 scale (Borg, 1998). To quantify running kinematics, subjects were affixed with retroreflective markers at various anatomical landmarks bilaterally, per published standards (Sutherland, 2002). Markers were attached at the following anatomical locations, with those on the foot affixed to the shoe: first and fifth metatarsal heads, top of the foot, medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanter, iliac crest, anterior-superior iliac spine, and posterior-superior iliac spine. In addition, marker clusters mounted on rigid plastic shells were affixed to the rearfoot, thigh, and shank segments to track segmental motion during running. Inertial measurement units (IMUs) (Blue Trident IMU, IMeasureU, Auckland, New Zealand; 1600 Hz sampling frequency, ±20g capture range, 12 grams) containing triaxial accelerometers were affixed to the anteromedial distal tibia bilaterally, and the lower back (L5/S1) to measure shock and shock attenuation during running. The IMUs were secured with double-sided tape and elastic bandages tightened firmly. The orientation of IMU axes at all anatomical locations was as follows: X-axis as vertical (upward = positive), Y-axis as mediolateral (right = positive), and Z-axis as anterior-posterior (anterior = positive).

Prior to the overground running trials, subjects performed 3-5 practice trials followed by a minimum of eight successful overground running trials bilaterally at 3.35 m/s ±5% along an 18-m runway embedded with three 1200 Hz force plates (OR-6- 2000; AMTI Inc, Watertown, MA, USA). Thirteen motion capture cameras recorded marker positions at 240 Hz (Oqus 400; 500) using Qualisys Track Manager (QTM) software (Qualisys AB, Göthenburg, Sweden). Running speed was monitored via two optoelectrical timing gates (TC-Gait, Brower Timing System, Draper, UT, USA) placed six meters apart at the runway's center. A trial was deemed successful if subjects ran within ±5% of the target running speed, did not seem to alter their speed between timing gates, contacted at least one force plate with their entire foot, and did not visibly modify their gait to target force plate contact. Subjects performed a small vertical jump at the beginning of each running trial which was manually flagged as an event in the motion capture software. The event flag and acceleration peak generated from the jump were used to temporally algin the IMU and motion capture data during data processing.

5.3.3 Data processing and analysis

Three-dimensional marker positions were tracked with motion capture software then three-dimensional kinetic and kinematic variables were subsequently computed in Visual3D (C-Motion Inc, Rockville, MD, USA). Both marker trajectories and force plate data used to calculate joint moments were filtered at 12 Hz using a 4^{th}-order, zero-lag Butterworth lowpass filter (Kristianslund et al., 2012). Segment inertial properties from Hanavan Jr (1966) and segmental masses from Dempster (1955) were used to model the lower extremity as a rigid-linked segment system. The global coordinate system was defined as X-axis (mediolateral) - positive to the right of the origin, Y-axis (anterior-posterior) - positive anteriorly, and Z-axis (vertical) - positive upward. Joint center estimation was performed for each joint as follows: the knee and ankle joint centers were estimated as the mid-point between the medial and lateral femoral epicondyles and the medial and lateral malleoli, respectively, while the hip joint center was estimated using regression equations from (Bell et al., 1989).

Joint angles were computed per the right-hand rule, with reference to the proximal segment. An x-y-z cardan rotation sequence was used. Sagittal plane internal joint moments were computed using Newtonian inverse dynamics and expressed in their respective joint coordinate systems. Joint power was calculated as the product of joint angular velocity and joint moment. A stance phase was identified as beginning at foot strike (i.e., when the vertical ground reaction force exceeded 20 newtons) and ending at toe-off (i.e., when the vertical ground reaction force fell below 20 newtons). The timing of these gait events of heel-strike and toe-off was also exported to MATLAB to aid subsequent step detection in the IMU data.

Joint power, joint moment, and joint angle-time series for eight stance phases bilaterally were exported to MATLAB (Release 2022a, MathWorks, Natick, MA, USA) to extract peak joint angles and compute joint ranges of motion, negative joint work, joint power absorption, and joint stiffness. Sagittal plane knee and frontal plane ankle angle time-series were inverted such that knee flexion and ankle eversion motions had positive magnitudes. Sagittal plane knee and ankle angles at initial contact were calculated as the magnitude of the first data point in the knee flexion-extension and ankle dorsiflexion-plantarflexion time-series, respectively, during stance. Peak knee flexion and ankle eversion were computed as the maximum value occurring during stance. Sagittal plane knee and ankle, and frontal plane ankle joint ranges of motion were calculated as the difference between the maximum and minimum values in the joint-angle time series during the deceleration phase of running stance. The deceleration phase was defined as the period of the stance phase that began at initial foot-ground contact and ended when the ankle reaches peak dorsiflexion during stance (i.e., end of energy absorption or the deceleration phase) (Davis IV and Gruber, 2021; Hamill et al., 2014). Peak power absorption at the ankle and knee in the sagittal plane was computed as the local minima of the power-time series occurring during the deceleration phase. The negative regions under the power-time curve were numerically integrated over the deceleration period to obtain negative mechanical joint work at the ankle and knee in the sagittal plane. Mechanical work and power are scalar values, with negative values indicating energy absorption or loss. Therefore, negative mechanical work and power absorption were represented as absolute values. Joint stiffness was computed as the slope of a least-squares line fit to joint angle-moment data from foot-strike to the end of the deceleration phase (Davis IV and Gruber, 2021; Hamill et al., 2014). All gait variables were calculated for each running trial bilaterally and averaged across 8 trials for each limb per subject.

Acceleration data from the IMUs along the vertical (x-axis), anterior-posterior (z-axis), and mediolateral (y-axis) axes were extracted and exported to MATLAB for processing. Each signal was converted from m/s^2 to units of gravity (g). To remove any existing linear trend in the signal, detrending was performed by subtracting a least-squares best-fit line from each signal. Since the accelerometers collected 1g at rest, the signal offset was removed such that each signal oscillated about zero. No filtering was performed. The resultant acceleration signal was calculated as $\sqrt{x^2 + y^2 + z^2}$.

Given the aim to associate shock attenuation between the IMU-based accelerations at tibia and lower back to gait-related active shock attenuation mechanisms obtained from motion-capture, it was necessary to ensure temporal alignment of the IMU and motion-capture data. The following steps were followed to ensure temporal alignment between IMU sensors and the motion capture data:

a. Temporal alignment of the tibia and lower back IMU data

IMU data were aligned based on Unix time stamps to account for possible lag in start times. The location of the first common Unix timestamp in both signals (i.e., lower back and tibia) was identified, and designated as the start of both signals. Then, both signals were linearly interpolated to 1600 Hz using the "resample" function in MATLAB because each sensor sampled at slightly different sampling frequencies despite being set to 1600 Hz in the software (right leg: 1598.151 Hz, left leg: 1604.322, lower back: 1587.432 Hz). These two steps ensured signal alignment between the tibia and lower back.

b. Temporal alignment of the IMU data with motion capture data

Figure 2 schematically explains the procedures to temporally align the IMU's acceleration data with the corresponding step measured by motion capture. To do so, we first identified the instant of the jump-landing (T_1) in the tibial acceleration signal as the location of the first peak with a minimum magnitude of 3g. Then, the time elapsed (QTM period) between the manually flagged jump-landing in QTM and foot-strike on a force plate (T_2) (i.e., when the force surpassed 20 newtons) was calculated. The manually flagged jump landing in QTM and T_1 were expected to occur at the same time therefore the QTM period was overlaid onto the tibial acceleration signal starting at T_1. The running step in the tibial acceleration signal immediately following the QTM period was the step coinciding with the limb contacting the force-plate. Since the jump landing in QTM was manually flagged, however, a supplementary analysis (Appendix 7) was performed to identify the time lag or lead relative to $T_{1.}$ The analysis revealed that the manual flag preceded T_1 by ~0.16 seconds. However, 0.16 seconds was small enough because it ensured that the step in the acceleration signal preceding the one coincident with force-plate contact was excluded ensuring the same step was identified in both motion capture and acceleration data.

A validated foot-contact detection algorithm adapted from Aubol and Milner (2020) was utilized to identify foot-strike and toe-off for the step immediately following the QTM period. Before implementing this algorithm, data were filtered using a zero-lag lowpass 4th order Butterworth filter with a cutoff frequency of 70 Hz for the sole purpose of step detection. Only data along the axial axis from the tibial sensor was used for step detection and the locations of foot-strike and toe-off were subsequently applied to all axes, and thereby the resultant, in both the tibial and lower back sensors. Per the foot-contact detection algorithm, peaks in the jerk (i.e., derivative of acceleration) of the axial acceleration signal that occurred within 150 milliseconds of a peak in the axial acceleration signal shortly following foot-contact were identified. The local minima in the axial acceleration signal occurring within 75 milliseconds of each jerk peak were identified, however local minima that occurred after an axial acceleration peak which was less than 20% in magnitude of the third acceleration highest peak were removed. Finally, the location of the earliest local minima occurring before the axial acceleration peak was identified and designated as foot strike. The foot-contact detection algorithm identified only the instant of foot-strike, and so toe-off was identified using contact time. Contact time was calculated via force-plate data as the time between foot-strike and toe-off. The application of contact time to the instant of foot-strike identified via step detection yielded the instant toe-off because toe-off marked the end of contact time. The instances of foot-strike and toe-off thus identified were subsequently applied to all axes of the unfiltered tibial and lower back sensor data, thereby isolating individual stance phases.

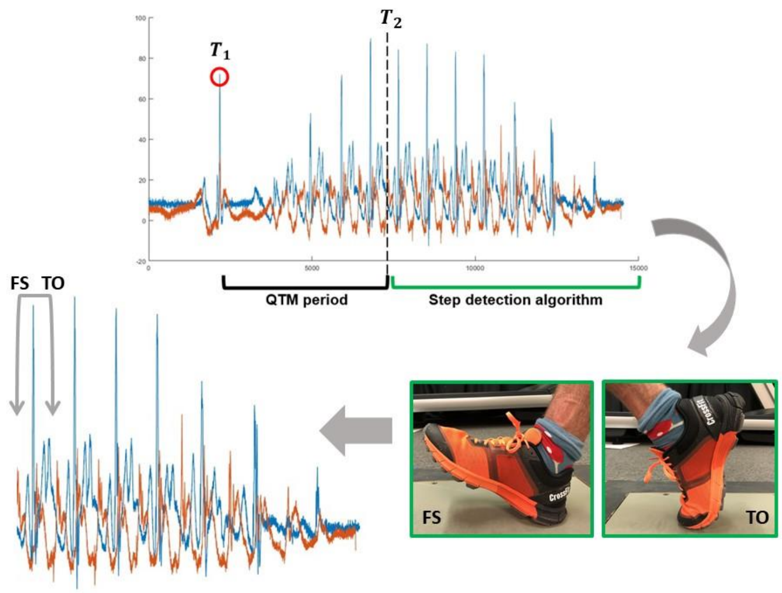

Figure 2. Method used to align the motion capture data with the IMU data so that the same step was identified in both data types. The QTM period was the time elapsed between the jump landing at T_1 and foot-strike on a force plate at T_2. The black vertical dashed line in the top panel indicates the end of QTM period at T_2. The signal before T_2 was discarded while the signal after it was entered into the step detection algorithm to detect the gait events of foot strike (FS) and toe off (TO) that were subsequently applied to all acceleration components. T_1: location of the jump landing manually flagged in QTM and observed in the acceleration signal.

For the frequency analysis, the detrended, unfiltered tibia and lower back acceleration signals from each stance phase were padded with zeros at both ends such that the signal length equaled a power of two, ensuring periodicity (Shorten and Winslow, 1992). The signal power (i.e., energy) contained in the tibial and head acceleration signals was determined by computing the power spectral density (PSD) using a Hanning window. The PSD was normalized to 1Hz bins for frequencies between 0Hz and Nyquist (Edwards et al., 2012; Gruber et al., 2014; Hamill et al., 1995).

The degree of shock attenuated between the tibia and lower back acceleration signals was quantified for each stance phase in the frequency domain along both the axial and resultant components using a transfer function (Hamill et al., 1995) calculated as follows:

$$\text{Shock attenuation} = 10\ \log_{10}\left(\frac{PSD_{lower\ back}}{PSD_{tibia}}\right)$$

The magnitude of shock attenuated or gained was calculated as the integral of the transfer function within FB1, FB2, FB3, and FB4, with negative values (in decibels) indicating attenuation and positive values indicating a gain (Figure 3) in the shock signal as it traveled from the tibia to the lower back.

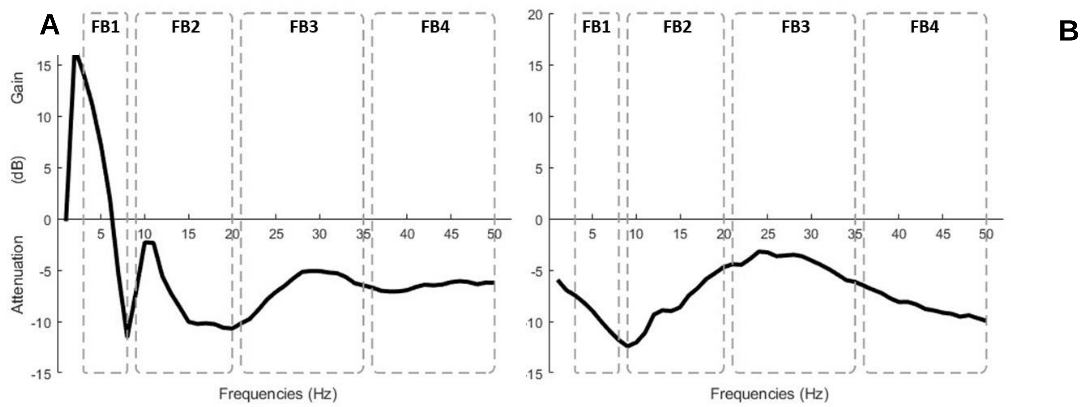

Figure 3. Transfer function for the entire sample (solid black line) for the axial (A) and resultant (B) acceleration signals. Positive values indicate a gain in signal energy between the tibia and lower back while negative values indicate attenuation. The dashed boxes demarcate the four frequency bins in order from left to right: 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4). dB: decibel; Hz: hertz

5.3.4 Statistical analysis

Knee and ankle joint kinematic and kinetic variables were the independent variables whereas shock attenuation within FB1, FB2, FB3, and FB4 along both the axial and resultant components were the dependent variables. All data were an average across 8 trials for each limb of each subject.

5.3.4.1 Bilateral differences

There is no evidence to suggest that shock attenuation across the frequency spectrum is significantly different between limbs during running therefore it was sought to collapse the data across both right and left limbs for each subject. However, first each independent and dependent variable value was compared between limbs within each subject to examine if it was appropriate to collapse data across limbs or treat each limb separately in subsequent regression analyses. FB1, sagittal knee, and ankle ROM, and negative work and power absorption at the ankle were non-normally distributed per Shapiro-Wilk tests (α>0.050) therefore Wilcoxon signed-rank tests (i.e., repeated measures) were used to identify bilateral differences in these variables while repeated measures t-tests were used for the other variables which were normally distributed. Only knee joint stiffness was significantly different between limbs (p-value=0.003, 95% confidence interval (CI)=0.212-1.031, Cohen's d effect size: 0.316). Therefore, limbs were treated as separate subjects when using knee joint stiffness as the independent variable in subsequent regression analyses whereas data was averaged across limbs for all other independent and dependent variables because there were no significant differences between limbs (p>0.050).

5.3.4.2 Regression Analysis

To examine the associations between shock attenuation and active attenuation mechanisms, one unadjusted regression model ($\alpha \geq 0.050$) was implemented for each exposure-outcome pairing. Shock attenuation is negative so, as the exposure magnitude increases, positive beta coefficients indicate a reduction in shock attenuation while negative beta coefficients indicate an increase in shock attenuation. Each regression model was checked for violations of linearity, independence, homoscedasticity, and normality of residuals. Given that the models with knee joint stiffness treated each limb as a separate subject resulting in two observations from the same subject, autocorrelation was examined using the Durbin-Watson test (α>0.05) to ensure that the residuals were independently distributed. Linear regression assumptions were met for all models.

5.4 Results

Anthropometric and demographic data for the fifty-seven included subjects are presented in Table 1.

Table 1. Anthropometric and demographic data for the entire sample.

	Group Mean ± 1 SD
Age (years)	32.66 ± 15.32
Sex distribution	24 female, 33 male
Height (meters)	1.72 ± 0.07

Body mass (kilograms)	69.52 ± 9.82
Current weekly mileage (miles/week)	12.66 ± 6.82

SD: standard deviation; Current weekly mileage was self-reported as miles per week on average in the three months preceding study enrollment.

5.4.1 Axial acceleration

Table 2 outlines overall results and Figure 4 displays scatter plots for each significant association. Within FB1, axial attenuation increased with an increase in sagittal knee ROM during deceleration (Figure. 4A), sagittal ankle angle at contact (i.e., more dorsi- than plantarflexion) (Figure. 4B), and peak ankle eversion during stance (Figure. 4C) but decreased with an increase in sagittal ankle ROM (Figure. 4D) and knee joint stiffness (Figure. 4E). In FB2, axial attenuation increased with an increase in sagittal knee ROM (Figure, 4F) and sagittal ankle angle at contact (Figure. 4G) while attenuation decreased with an increase in knee joint stiffness (Figure. 4H) and sagittal ankle ROM (Figure. 4I). There were no statistically significant associations found within FB3. In FB4, axial attenuation increased with an increase in peak knee flexion during stance (Figure. 4J) but decreased with sagittal contact angle (Figure. 4K).

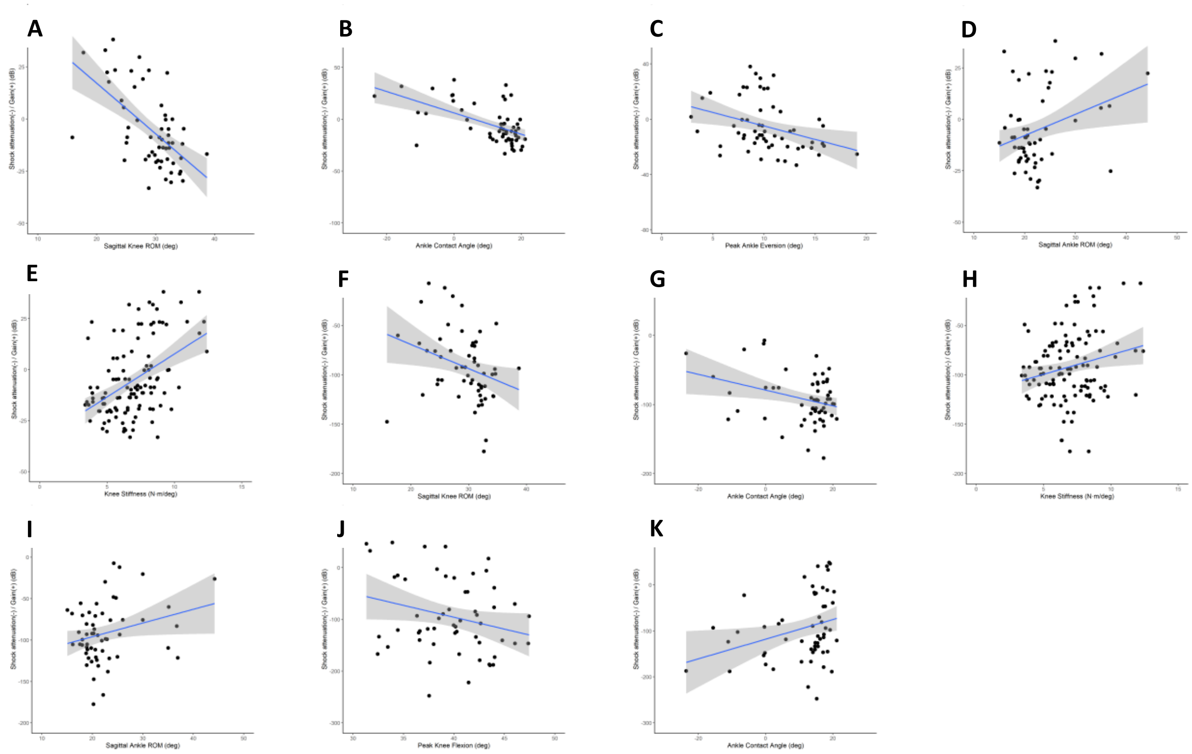

Figure 4. This figure displays the analyses performed on the vertical acceleration component. Each plot displays a significant association between the exposure (x-axis) – outcome (y-axis) pairing obtained identified via unadjusted linear regression (α>0.050). A-E: Outcome = FB1 (3-8 Hz); F-I: Outcome = FB2 (9-20 Hz); J-K: Outcome = FB4 (36-50). dB: decibel.

5.4.2 Resultant Acceleration

Results are displayed in Table 2, while Figure 5 visualizes the significant associations found. Within FB1, shock attenuation increased with an increase in peak knee flexion (Figure. 5A), sagittal ankle ROM (Figure. 5B), and frontal ankle ROM (Figure. 5C) while shock attenuation decreased with an increase in sagittal knee ROM (Figure. 5D), sagittal ankle angle at contact (i.e., more dorsi- than plantarflexion) (Figure. 5E), and ankle stiffness (Figure. 5F). In FB2, shock attenuation increased with an increase in sagittal knee contact angle (Figure. 5G), peak knee flexion (Figure. 5H), knee stiffness (Figure. 5I), and sagittal ankle ROM (Figure. 5J) while shock attenuation decreased with an increase in sagittal knee ROM (Figure. 5K), sagittal ankle angle at contact (Figure. 5L), and ankle stiffness (Figure. 5M). An increase in peak knee flexion was associated with an increase in shock attenuation within FB3 (Figure. 5N). Within FB4, shock attenuation increased with an increase in sagittal knee contact angle

(Figure. 5O), sagittal ankle ROM (Figure. 5P), and frontal ankle ROM (Figure. 5Q) while shock attenuation decreased with an increase in sagittal ankle angle at contact (Figure. 5R) and ankle stiffness (Figure. 5S).

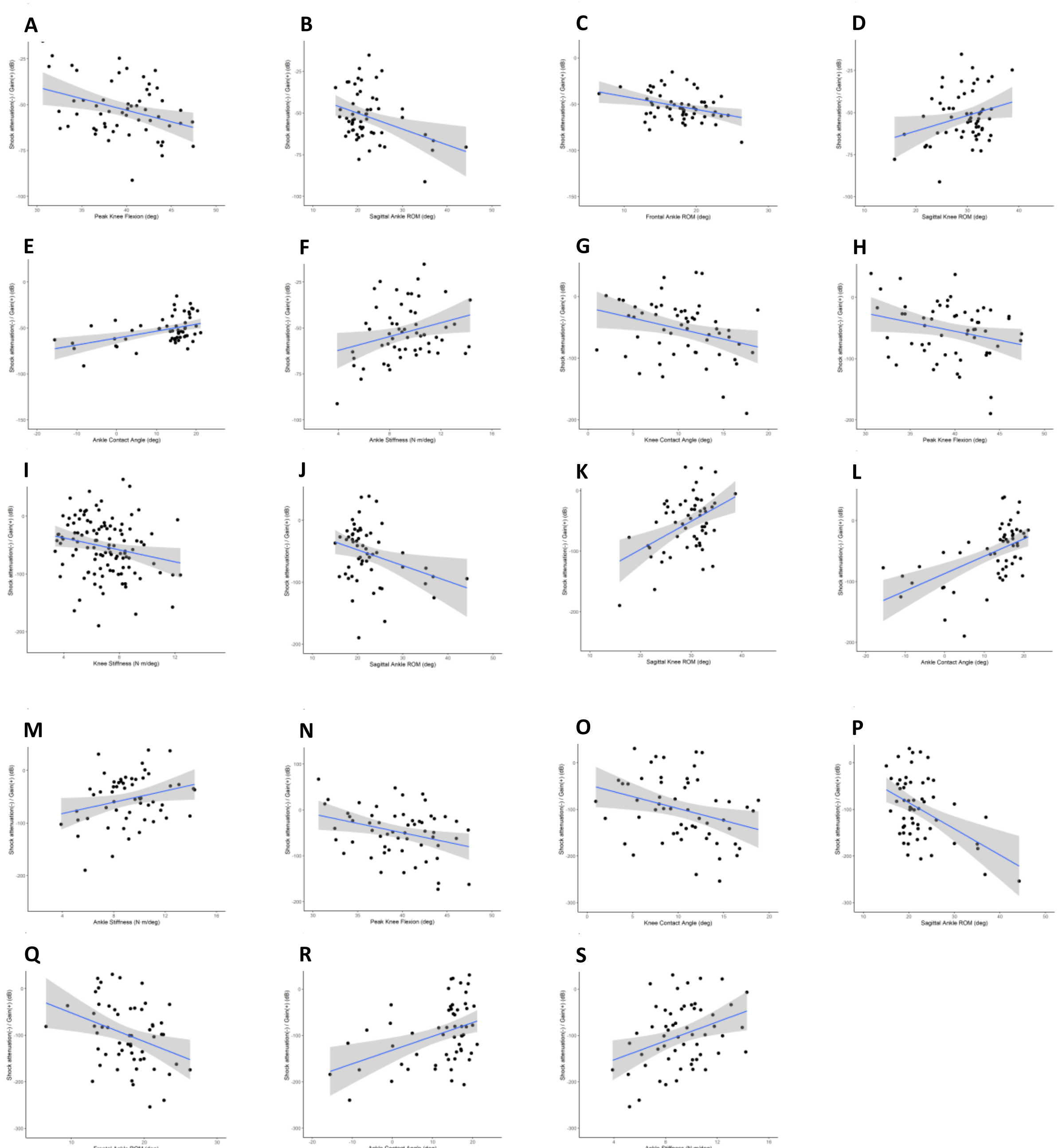

Figure 5. This figure displays the analyses performed on the resultant acceleration component. Each plot displays a significant association between the exposure (x-axis) – outcome (y-axis) pairing obtained identified via unadjusted

linear regression (α>0.050). A-F: Outcome = FB 1 (3-8 Hz); G-M: Outcome = FB2 (9-20 Hz); N: Outcome = FB3 (21-35 Hz); O-S: Outcome = FB4 (36-50). dB: decibel.

Table 2. Regression model results for each exposure-outcome pairing.

	β	**p-value**	**95% CI**
Axial Acceleration			
Sagittal knee angle IC (deg)			
SA 3-8 Hz	1.01	0.066	-0.06 – 2.09
SA 9-20 Hz	0.20	0.848	-1.93 – 2.34
SA 21-35 Hz	-2.72	0.171	-6.65 – 1.21
SA 36-50 Hz	-4.07	0.085	-8.73 – 0.58
Peak knee flexion (deg)			
SA 3-8 Hz	< -0.01	1.000	-1.18 – 1.18
SA 9-20 Hz	-2.06	0.067	-4.27 – 0.15
SA 21-35 Hz	-3.55	0.091	-7.71 – 0.59
SA 36-50 Hz	-6.42	0.009*	-11.22 – -1.62
Knee joint stiffness (N·m/deg)			
SA 3-8 Hz	4.63	<0.001*	3.01 – 6.26
SA 9-20 Hz	3.96	0.024*	0.51 – 7.40
SA 21-35 Hz	3.32	0.350	-3.68 – 10.33
SA 36-50 Hz	1.15	0.728	-7.15 –9.46
Sagittal knee ROM (deg)			
SA 3-8 Hz	-2.42	<0.001*	-3.318 – -1.521
SA 9-20 Hz	-2.44	0.019*	-4.47 – -0.415
SA 21-35 Hz	0.22	0.911	-3.77 – 4.21
SA 36-50 Hz	1.96	0.409	-2.77 – 6.71
Sagittal knee negative mechanical work (J)			
SA 3-8 Hz	-0.12	0.518	-0.49 – 0.25
SA 9-20 Hz	-0.47	0.190	-1.18 – 0.24
SA 21-35 Hz	-0.25	0.705	-1.61 – 1.09
SA 36-50 Hz	0.63	0.430	-0.97 – 2.25
Sagittal knee power absorption (W)			
SA 3-8 Hz	< -0.01	0.623	-0.01 – 0.01
SA 9-20 Hz	-0.01	0.380	-0.04 – 0.01
SA 21-35 Hz	< -0.01	0.848	-0.06 – 0.05
SA 36-50 Hz	0.03	0.350	-0.03 – 0.09
Sagittal ankle angle IC (deg)			
SA 3-8 Hz	-1.03	<0.001*	-1.44 – 0.62
SA 9-20 Hz	-1.41	0.014*	-2.04 – -0.23
SA 21-35 Hz	0.53	0.551	-1.24 – 2.30
SA 36-50 Hz	2.30	0.027*	0.26 – 4.33

Peak ankle eversion (deg)			
SA 3-8 Hz	-1.95	0.008*	-3.37 – -0.52
SA 9-20 Hz	-1.51	0.301	-4.41 – 1.38
SA 21-35 Hz	-0.74	0.785	-6.21 – 4.72
SA 36-50 Hz	1.53	0.640	-5.00 – 8.06
Ankle joint stiffness (N·m/deg)			
SA 3-8 Hz	-1.19	0.263	-3.31 – 0.92
SA 9-20 Hz	-1.42	0.468	-5.52 – 2.67
SA 21-35 Hz	1.52	0.69	-6.16 – 9.21
SA 36-50 Hz	7.19	0.114	-1.80 – 16.19
Sagittal ankle ROM (deg)			
SA 3-8 Hz	1.03	0.015*	0.20 – 1.86
SA 9-20 Hz	1.65	0.045*	0.03 – 3.27
SA 21-35 Hz	-0.01	0.996	-3.15 – 3.13
SA 36-50 Hz	-2.49	0.182	-6.19 – 1.20
Frontal ankle ROM (deg)			
SA 3-8 Hz	1.11	0.083	-0.15 – 2.38
SA 9-20 Hz	1.65	0.185	-0.81 – 4.11
SA 21-35 Hz	-1.05	0.654	-5.72 – 3.62
SA 36-50 Hz	-2.69	0.334	-8.25 – 2.85
Sagittal ankle negative mechanical work (J)			
SA 3-8 Hz	0.04	0.815	-0.36 – 0.46
SA 9-20 Hz	-0.22	0.582	-1.02 – 0.57
SA 21-35 Hz	-0.52	0.481	-2.02 – 0.96
SA 36-50 Hz	-0.47	0.596	-2.26 – 1.31
Sagittal ankle power absorption (W)			
SA 3-8 Hz	0.01	0.477	-0.02 – 0.04
SA 9-20 Hz	< -0.01	0.829	-0.07 – 0.07
SA 21-35 Hz	-0.05	0.359	-0.18 – 0.06
SA 36-50 Hz	-0.05	0.429	-0.21 – 0.09
Resultant Acceleration			
Sagittal knee angle IC (deg)			
SA 3-8 Hz	-0.71	0.115	-1.59 – 0.17
SA 9-20 Hz	-2.79	0.042*	-5.48 – -0.09
SA 21-35 Hz	-2.44	0.122	-5.57 – 0.67
SA 36-50 Hz	-4.99	0.013*	-8.90 – -1.09
Peak knee flexion (deg)			
SA 3-8 Hz	-1.25	0.007*	-2.16 – -0.35
SA 9-20 Hz	-2.94	0.045*	-5.81 – -0.06
SA 21-35 Hz	-4.07	0.014*	-7.29 – -0.85
SA 36-50 Hz	-3.93	0.070	-8.20 – 0.33

Knee joint stiffness (N·m/deg)			
SA 3-8 Hz	-0.58	0.441	-2.06 – 0.90
SA 9-20 Hz	-5.15	0.022*	-9.57 – -0.73
SA 21-35 Hz	1.11	0.688	-4.25 – 6.49
SA 36-50 Hz	2.28	0.516	-4.67 – 9.24
Knee flexion-extension ROM (deg)			
SA 3-8 Hz	0.92	0.038*	0.05 – 1.79
SA 9-20 Hz	4.65	<0.001*	2.15 – 7.14
SA 21-35 Hz	1.29	0.418	-1.87 – 4.45
SA 36-50 Hz	2.65	0.195	-1.40 –6.70
Sagittal knee negative mechanical work (J)			
SA 3-8 Hz	0.11	0.436	-0.18 – 0.42
SA 9-20 Hz	-0.43	0.364	-1.37 – 0.51
SA 21-35 Hz	-0.66	0.216	-1.73 – 0.40
SA 36-50 Hz	0.14	0.833	-1.25 – 1.54
Sagittal knee power absorption (W)			
SA 3-8 Hz	<0.01	0.495	-0.01 – 0.01
SA 9-20 Hz	-0.02	0.272	-0.06 – 0.01
SA 21-35 Hz	-0.02	0.190	-0.07 – 0.01
SA 36-50 Hz	<0.01	0.929	-0.05 – 0.06
Sagittal ankle angle IC (deg)			
SA 3-8 Hz	0.69	<0.001*	0.34 – 1.05
SA 9-20 Hz	2.48	<0.001*	1.43 – 0.52
SA 21-35 Hz	0.73	0.300	-0.67 – 2.14
SA 36-50 Hz	3.28	<0.001*	1.67 – 4.89
Peak ankle eversion (deg)			
SA 3-8 Hz	0.26	0.667	-0.97 – 1.50
SA 9-20 Hz	3.36	0.074	-0.34 – 7.08
SA 21-35 Hz	0.34	0.874	-4.02 – 4.71
SA 36-50 Hz	-0.62	0.823	-6.27 – 5.01
Ankle joint stiffness (N·m/deg)			
SA 3-8 Hz	1.87	0.028*	0.20 – 3.54
SA 9-20 Hz	5.36	0.042*	0.18 – 10.55
SA 21-35 Hz	3.71	0.225	-2.35 – 9.77
SA 36-50 Hz	10.23	0.007*	2.78 – 17.67
Sagittal ankle ROM (deg)			
SA 3-8 Hz	-0.95	0.005*	-1.61 – -0.28
SA 9-20 Hz	-2.53	0.018*	-4.62 – -0.44
SA 21-35 Hz	-0.73	0.558	-3.23 – 1.76
SA 36-50 Hz	-5.61	<0.001*	-8.41 – -2.74
Frontal ankle ROM (deg)			

SA 3-8 Hz	-1.39	0.006*	-2.39 – -0.40
SA 9-20 Hz	-2.52	0.12	-5.72 – 0.67
SA 21-35 Hz	-1.84	0.323	-5.55 – 1.85
SA 36-50 Hz	-6.14	0.008*	-10.68 – -1.60
Sagittal ankle negative mechanical work			
SA 3-8 Hz	-0.05	0.729	-0.39 – 0.28
SA 9-20 Hz	-0.49	0.345	-1.53 – 0.54
SA 21-35 Hz	0.53	0.368	-0.65 – 1.72
SA 36-50 Hz	0.06	0.936	-1.48 – 1.60
Sagittal ankle power absorption (W)			
SA 3-8 Hz	-0.01	0.543	-0.03 – 0.01
SA 9-20 Hz	-0.04	0.296	-0.13 – 0.04
SA 21-35 Hz	0.02	0.644	-0.07 – 0.12
SA 36-50 Hz	-0.01	0.790	-0.14 – 0.11

β: beta coefficient; CI: confidence interval; Hz: hertz; SA: shock attenuation within the given frequency band; J: joules; W: watts; *statistically significant at the $p<0.050$ level.

5.5 Discussion

This study aimed to identify the associations of previously proposed active shock attenuation mechanisms to the degree of axial and resultant impact shock attenuated within 3-8 Hz, 9-20 Hz, 21-35 Hz, and 36-50 Hz bins. Overall, the findings of this study confirm the current hypotheses that active joint and segment motions contribute to the attenuation of 3-8 Hz, while revealing active contributions to the attenuation of frequencies >8 Hz. Thus, during human running, active contributions to shock attenuation may extend beyond the 3-8 Hz frequency range.

The degree of shock attenuated via previously proposed active mechanisms was not limited to the 3-8 Hz band as hypothesized in the current study. For instance, in addition to attenuating shock within 3-8 Hz, a more compliant (i.e., less stiff) knee joint and greater sagittal knee ROM were both associated with greater attenuation of 9-20 Hz along the axial acceleration component. These results confirm our hypotheses that knee joint stiffness and knee ROM may be negatively and positively related to active shock attenuation, respectively, and provide novel information regarding their contribution to the attenuation of 9-20 Hz along the axial acceleration component during running. As previously evidenced during non-running tasks, sagittal plane knee motion contributes to the attenuation of frequencies >5 Hz, possibly due to the underlying eccentric muscle contractions that absorb mechanical energy (Edwards et al., 2012; Lafortune et al., 1996). Combined with this prior evidence, the present results indicate that knee compliance contributes to axial shock attenuation of 3-20 Hz. However, these conclusions are not supported by the resultant acceleration component.

In contrast to the axial component, an increase in sagittal knee ROM was associated with a decrease in shock attenuation within 3-8 Hz and 9-20 Hz along the resultant, which was unexpected but may be due to some frequencies within 3-8 Hz experiencing a gain rather than attenuation. For example, lower extremity joint excursions during deceleration include the horizontal movement of the trunk, arms, and head during deceleration that add to the energy observed at the lower back (Chu and Caldwell, 2004; Nigg et al., 1995; Shorten and Winslow, 1992). Due to such added energy at the lower back, the resulting gain rather than attenuation of low frequency signal components may have potentially decreased shock attenuation along the resultant.

The relationship between knee joint stiffness and shock attenuation differs between axial and resultant components within the 9-20 Hz frequency range. While axial attenuation decreased with increasing knee joint stiffness, resultant attenuation increased. One possible explanation for this discrepancy is the non-trivial contribution of the anterior-posterior component to the resultant acceleration signal (Giandolini et al., 2016; Van den Berghe et al., 2019a). The magnitude of this component, and its contribution to the resultant, may be influenced by knee stiffness. For instance, rearfoot runners typically run with a stiffer knee than forefoot runners (Hamill et al., 2014; Laughton et al., 2003), while also exhibiting a larger anterior-posterior peak impact shock magnitude (Van den Berghe et al., 2019a). These findings suggest knee stiffness may be positively associated with peak impact shock along the anterior-posterior component which may in-turn increase the energy concentrated in 9-20 Hz along the resultant component. Therefore, an increase in knee stiffness may necessitate a substantial resultant attenuation response due to its potential positive association with the anterior-posterior peak impact shock. Such a substantial attenuation response could increase the risk for bony or soft tissue injuries. Previous research has implicated knee joint stiffness as a potential risk factor for running-related injuries (Messier et al., 2018), although the relationship is non-linear. Individuals with unusually high knee stiffness (>9 N·m/deg) may be particularly susceptible to injury development (Davis IV and Gruber, 2021). To fully understand the role of knee joint stiffness and its association with shock attenuation in injury development, longitudinal studies are necessary.

An increase in peak knee flexion during stance was associated with an increase in shock attenuation within 3-35 Hz for the resultant and 36-50 Hz for the axial component. These findings corroborate prior observations by Lafortune et al. (1996) and Edwards et al. (2012) who suggested that the knee acts as a low-pass filter of energy >5 Hz, potentially reducing the demand placed on passive tissues to attenuate shock; Peak knee flexion magnitude may be dictated by eccentric knee extensor muscle action and the elongation of the elastic components of muscle-tendon units that are previously suggested to allow for mechanical energy absorption (Edwards et al., 2012; Garrett et al., 1987). Peak knee flexion also tends to be positively associated with knee flexion angle at foot-ground contact (Derrick, 2004). The present findings revealed that as knee flexion at contact increased, shock attenuation within 9-20 Hz and 36-50 Hz along the resultant component increased, supporting previous suggestions (Derrick, 2004). Thus, both peak knee flexion and knee flexion at contact may contribute to attenuating frequencies across the spectrum. This finding has implications for gait retaining strategies like a switch from rearfoot to non-rearfoot footstrike pattern which may increase knee flexion magnitude at ground contact (Doyle et al., 2022), thereby compelling runners to actively rather than passively attenuate shock.

An increase in ankle joint stiffness was associated with decreased attenuation of the resultant within 3-8 Hz, 9-20 Hz, and 36-50 Hz. Ankle stiffness is inversely related to mechanical energy absorption that occurs via eccentric muscle actions (Hamill et al., 2014). In the same way, the signal energy associated with shock may also be absorbed by eccentric muscle actions (Edwards et al., 2012), the contributions of which, to shock attenuation, may decrease with ankle stiffness.

A more compliant ankle is often observed when employing a plantarflexed rather than dorsiflexed ankle angle at initial contact which results in greater mechanical energy absorption at the ankle (Hamill et al., 2014). The present findings indicate increased dorsiflexion rather than plantarflexion at contact increased axial shock attenuation in 3-20 Hz. Two potential reasons for why shock attenuation increased along the axial with ankle dorsiflexion at contact could be greater energy contained across frequencies at the tibia thereby necessitating more shock attenuation (Gruber et al., 2014), or a more compliant knee often observed among runners who are more dorsiflexed than plantarflexed at contact (Hamill et al., 2014). On the contrary, a smaller anterior-posterior impact shock experienced at larger dorsiflexion than plantarflexion angles at contact (Van den Berghe et al., 2019a) (i.e., at larger contact angles) may require a smaller attenuation response potentially explaining why resultant attenuation in 3-20 Hz decreased with an increase in ankle contact angles. Importantly, the difference in associations between shock attenuation and ankle contact angle along the axial versus the resultant highlight the need to examine multi-directional shock attenuation.

Frontal plane ankle motion, both range of motion and peak ankle eversion, contributed to shock attenuation in the present study. Like the associations found in the sagittal plane, frontal plane ROM contributed to the attenuation of energy within 3-8 Hz and 36-50 Hz along the resultant component, possibly due to mechanical energy absorption via the ankle invertor muscles that control eversion during running stance. The rapid eversion of the ankle following foot-ground contact is resisted via elongation of the foot invertor muscles which may increasingly contribute to shock attenuation over a large range of motion; During running, 58% more mechanical energy is absorbed by the foot invertors when wearing pronation-augmenting versus neutral footwear (O'Connor and Hamill, 2004). Thus, frontal plane ankle motion may actively contribute to both low and high frequency resultant shock attenuation.

5.6 Conclusions

Overall, the present results suggest that sagittal knee and ankle joint stiffness, knee and ankle joint excursions, and peak knee flexion are associated with the degree of shock attenuation across 3-50 Hz while peak ankle eversion may only contribute to attenuation within 3-8 Hz. Therefore, attenuation of frequencies >8 Hz may be achieved not only via passive deformation of rigid and soft musculoskeletal tissues per previous suggestions, but also through active contributions of joint kinematics as revealed in this study. However, the degree to which joint kinematic variables contribute to shock attenuation depends on the direction of acceleration considered, emphasizing the importance of analyzing multi-direction shock attenuation.

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6 CHAPTER 6: Study 3

Manuscript Title: Age and Running Experience Influence Shock Attenuation During Human Running

Intended Journal: Medicine and Science in Sport and Exercise

6.1 Abstract

***Background*.** Older and novice runners may display heightened peak impact shock and shock attenuation than younger and recreational runners due to age and running experience-related alterations to gait and body composition. This study aimed to examine the independent associations of age and running experience to impact shock and shock attenuation in time and frequency-domains to understand the reliance on passive tissues and active gait mechanics for attenuation within each runner sub-population.

***Methods*.** Enrolled healthy runners were classified as middle-aged (n=20, age: 52.95±5.83 years) or young (n=24, age: 22.37±4.09) and novice (n=14, <1 year of experience, 3-13 mi·wk^{-1}) or recreational (n=24, >1 year of experience, 9-30 mi·wk^{-1}). Subjects ran overground at 3.35 m·s^{-1} while impact shock was measured using triaxial accelerometers on the tibia and low-back. Time-domain shock attenuation, peak tibial, and low-back shock were calculated along the axial and resultant components. Power spectral density was computed for a frequency-domain transformation of each signal. Signal power magnitude (SPM) at the tibia and low-back was calculated and shock attenuation was quantified using a transfer function within 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4). Associations of age and running experience to each time-domain and frequency-domain dependent variable were estimated with linear regression.

***Results*.** Middle-aged runners attenuated less resultant shock in the frequency-domain within FB1 (p=0.007, 95% confidence interval (CI): 3.20–20.03 β = 11.62) and less axial shock within FB2 (p=0.016, CI: 4.29–40.12, β = 22.21) and FB3 (p=0.002, CI: 22.58–96.69, β = 59.64) than younger runners. Novice runners attenuated less axial shock within FB3 than recreational runners (p=0.024, CI: 6.72–92.13, β =

49.43). Middle-aged runners attenuated more shock in the time-domain along both resultant (p=0.046, CI: -16.89 – -0.15, β = -8.52) and axial components (p=0.038, CI: 27.94 – -0.83, β = -14.38) compared with younger runners. Novices attenuated more shock in the time-domain along the axial (p-value: 0.010, CI: -35.85 – -5.14, β = -20.50) and resultant (p-value: 0.040, CI: -14.99 – -0.34, β = -7.67) components than recreational runners.

***Conclusion*.** Less high-frequency shock attenuation by middle-aged and novice runners than younger and recreational runners, respectively, may be due to differences in gait mechanics. Smaller peak knee flexion, an active shock attenuation mechanism, by middle-aged and novice runners may explain less high-frequency attenuation by these subpopulations, which may increase the attenuation demands on passive tissues. Age and running experience may interactively influence shock attenuation which may modulate injury risk for older adults adopting running for physical activity.

6.2 Introduction

Global engagement in running is on the rise given its numerous health benefits (Lavie et al., 2015; Lee et al., 2014; Oswald et al., 2020a). Unfortunately, running-related injury (RRI) rates are high in the general running population (Van Gent et al., 2007; Videbæk et al., 2015). Notably, novice (i.e., less-experienced) and older runners develop significantly more RRIs than recreational (i.e., more-experienced) and younger runners, respectively (Damsted et al., 2019; Linton and Valentin, 2018; McKean et al., 2006; Videbæk et al., 2015). Novice runners may cease running due to RRI development (Fokkema et al., 2019). The inhibition of physical activity levels post-injury (Davis and Gruber, 2019) may accelerate decrements in musculoskeletal function that naturally occur with age (Kirkendall and Garrett, 1998) among older runners. Therefore, to minimize these adverse effects of RRI development among older and novice runners, exploring factors that may influence RRI risk among these sub-populations is necessary.

Peak impact shock may be a risk factor for RRI among young recreational runners (Davis et al., 2004; Milner et al., 2006) and must be examined in novice and older runners as it may contribute to their particularly high risk for RRI. While peak impact shock describes the maximum magnitude of shock experienced at the tibia after foot-ground contact, analyzing shock attenuation among novice and older runners may also be insightful because it quantifies how much shock is absorbed by musculoskeletal structures as the shock propagates from the tibia to the low-back, and subsequently the head (Voloshin and Wosk, 1982; Voloshin et al., 1981; Wosk and Voloshin, 1981). Such a time-domain analysis may reveal if shock attenuation demands placed on musculoskeletal structures differ between runner sub-populations. Still, a frequency-domain analysis is necessary to understand the mechanisms relied upon by each group to attenuate shock.

In the frequency-domain, the impact shock generated soon after foot-ground contact consists of vibrations up to 50 Hz (Shorten and Winslow, 1992). Active mechanisms like joint compliance may attenuate energy between 3-8 Hz while energy >9 Hz may be attenuated by passive mechanisms like soft and rigid tissue deformation (Derrick et al., 1998; Edwards et al., 2012; Hamill et al., 1995; Lafortune et al., 1996; Nigg et al., 1995; Paul et al., 1978; Shorten and Winslow, 1992). Age-related and running experience-related changes to gait may influence tibial impact shock energy while age-related and running experience-related changes to both gait mechanics and passive tissues may modulate the ability to attenuate it. For instance, older runners display larger magnitudes of peak impact force and loading rates of the vertical ground reaction force than younger runners (Bus, 2003a), both positively correlated with peak impact shock (Hennig and Lafortune, 1991). Thus, relative to younger runners, older runners may experience more impact-related higher frequency content that may be passively attenuated (Paul et al., 1978). At the same time, older compared to younger runners display less knee flexion during stance, a smaller knee flexion-extension range of motion, and perform less negative mechanical work (Devita et al., 2016; Powell and Williams, 2018), all of which indicate a stiffer knee joint that may absorb less energy via eccentric muscle contractions. Consequently, active shock attenuation may diminish with age, compelling older runners to rely on passive tissues for attenuation. However, because rigid and soft tissues decrease mass and resiliency with age (Goodpaster et al., 2006; Warming et al., 2002), a heightened reliance on passive tissues for attenuation among older runners could potentially overload these tissues and result in RRI development.

Like with aging, injury risk to passive tissues may be heightened among novice runners as they tend to display smaller magnitudes of knee flexion throughout running stance than recreational runners (Clermont et al., 2017; Clermont et al., 2019; Moore, 2016), thereby increasing their reliance on passive attenuation mechanisms. Novice runners also run with less ankle eversion throughout stance than recreational runners (Harrison et al., 2021), so the tibialis posterior muscle may perform less negative mechanical work to resist eversion resulting in less actively shock attenuated than recreation runners experiencing greater eversion. Considering that the mass and resiliency of both rigid and soft tissues improve with long-term exposure to weight-bearing exercise (Luden et al., 2012; Ravnholt et al., 2018; Yu et al., 2022), novice runners' tissues may be inadequately adapted to meet large shock attenuation demands. Thus, a potential shift in attenuation responsibilities from active gait to passive tissue-related mechanisms with increasing age and less running experience may increase injury risk to passive tissues among older and novice runners.

Novice and older runners may be particularly susceptible to impact shock as an RRI mechanism; They may experience larger magnitudes of impact shock and high-frequency content due to their gait patterns than recreational and younger runners. Concurrently, their tissues may not be capable of optimally attenuating impact-related shock (Beck et al., 2016; Boyer et al., 2017; Devita et al., 2016; Warming et al., 2002). Assessing if impact shock and degree of shock attenuation are age and running experience-dependent is necessary to identify if older and novice runners' tissues are loaded differently than younger and more experienced runners. This information may provide insight into whether older and novice runners may incur greater injury risk compared to younger and more experienced runners. Therefore, this study aimed to estimate the independent effects of age and running experience on (1) tibial peak impact shock in the time-domain and in the frequency-domain by assessing signal power magnitude within 3-8 Hz (FB1), 9-20 Hz (FB2), 21-35 Hz (FB3), and 36-50 Hz (FB4), and (2) estimate the independent effects of age and running experience on the degree of shock attenuated in the time-domain and the frequency-domain within the same frequency bands. It was hypothesized that middle-aged runners will have:

(1) greater impact shock and shock attenuation within FB2-FB4 (e.g., time-domain peak tibial impact shock, tibial signal power magnitude within FB2-FB4, time-domain shock attenuation, and frequency-domain shock attenuation within FB2-FB4) compared with younger runners.
(2) less shock attenuation within FB1.

Additionally, it was hypothesized that novice runners will have:

(3) greater impact shock and shock attenuation within FB2-FB4 (e.g., time-domain peak tibial impact shock, tibial signal power magnitude within FB2-FB4, time-domain shock attenuation, and frequency-domain sock attenuation within FB2-FB4) than recreational runners.
(4) less shock attenuation within FB1.

6.3 Methods

6.3.1 Sample Characteristics

Forty-four total subjects from the Bloomington and Indianapolis (IN, USA) regions were recruited based on a-priori power calculations for age ($f^2 = 0.18$, α error probability = 0.05, desired power = 0.80, required sample size = 40) and running experience ($f^2 = 0.46$, α error probability = 0.05, desired power = 0.80, required sample size = 19) using the vertical ground reaction force impact peak effect size from Bus (2003a) and peak tibial shock data from Camelio et al. (2020), respectively.

Three sets of independent groups were recruited. The young and recreational group was used as the reference group for the independent effects of age and experience. This group was required to be 18-

30 years old with >1 year of running 9-30 miles·week^{-1}. The middle-aged group was required to be 45-65 years old with >1 year of running 9-30 miles·week^{-1}. The novice group was required to be 18-30 years old with <1 year of running 3-13 miles·week^{-1}. The running experience criteria were based on Honert et al. (2020) but, to ensure 80% power, the following adjustments were made for runners who ran for one year precisely or those who met either criteria for running-years or weekly mileage, but not both: Individuals who ran for exactly one year were included as either novice (n=3) or recreational (n=1) if their weekly mileage values were within ±1 standard deviation of the mean for the other subjects in those groups. Additionally, eight runners qualified as recreational per the running-years criteria but not weekly mileage criteria, while the opposite was true for one runner. These nine runners were classified as recreational if they had been running for >1 year (n=8) and novice if they ran for <1 year (n=1).

Subjects completed questionnaires to report their age, running injury history, total years spent running, current weekly mileage, and readiness for physical activity to determine eligibility for the study. Additional inclusion criteria required subjects to be free of chronic disease or illnesses that may disallow engagement in ≥15 minutes of low to moderate-intensity physical activity, have no history of low-back or lower limb orthopedic surgery in the past ten years, and no musculoskeletal injury within the past six months or resumed normal running training volume for ≥8 weeks before enrollment if sustaining an within the past six months. Tobacco users and women pregnant at the time of the study were excluded. All testing procedures were approved by the Indiana University Institutional Review Board (IRB Protocol 10585), and all subjects gave their written informed consent prior to participation.

6.3.2 Experimental Protocol

After completing the questionnaires, subjects changed into form-fitting clothes and lab-standard neutral running shoes (Reebok; One X CrossFit Cushion 3.0) to control for any potential effects of footwear on shock attenuation. Each subject's height and weight were measured using a stadiometer and weighing scale. Subjects then performed an ~5-8 minute warmup at their preferred running speed equating to light exertion (i.e., <11 on Borg's ratings of perceived exertion 6-20 scale [RPE]) (Borg, 1998).

Lightweight (12 grams) inertial measurement units (IMUs) (Blue Trident IMU, IMeasureU, Auckland, New Zealand) containing triaxial accelerometers were affixed bilaterally to the boniest aspect of the anteromedial distal tibia and the low-back (i.e., at the lumbosacral joint location) to quantify segmental accelerations and shock attenuation between segments. The IMUs were secured using elastic straps and self-adhesive tape tightened as firmly as possible yet to the subject's comfort. The sampling frequency of the accelerometers was 1600 Hz with a capture range of ±20g. The orientation of the IMU axes was as follows for all anatomical locations: X-axis as axial (upward = positive), Y-axis as mediolateral (right = positive), and Z-axis as anterior-posterior (anterior = positive). All acceleration data were collected with the Capture.U mobile application and stored onboard the sensors.

Subjects performed a jump at the start of each running trial which was manually flagged in the software (Qualisys Track Manager [QTM], Qualisys AB, Göthenburg, Sweden) capturing force plate data. This manual flag was used later to temporally align the force plate and IMU data and identify the stance phase acceleration signals retained for analysis. Immediately following the jump, subjects ran overground at a criterion speed of 3.35 m/s ±5% along an 18-meter runway embedded with three force plates (OR-6- 2000; AMTI Inc, Watertown, MA, USA) sampling at 1200 Hz. A criterion running speed was employed across subjects to minimize the known effects of running speed on impact shock and attenuation (Mercer et al., 2002; Sheerin et al., 2019). Running speed was monitored using two optoelectrical timing gates (TC-Gait, Brower Timing System, Draper, UT, USA) placed six meters apart in the center of the runway. Each subject completed a minimum of eight successful running trials for each limb. A trial was deemed successful if subjects ran within ±5% of the criterion speed, maintained a constant running speed between timing gates, did not noticeably modify their gait to target the force platforms, and contacted the force platforms with their entire foot. Subjects indicated their RPE every ten attempts to ensure they were not fatigued (i.e., RPE <13; (Borg, 1998)).

6.3.3 Data Processing

6.3.3.1 Age and Running Experience

Both age and running experience were treated as categorical variables. The independent effects of age were assessed among only the subjects who had been running for >1 year and ran between 9-30 miles per week on average. The young group (18-30 years) was treated as the reference group and coded "0", while the middle-aged group (45-65 years) was coded "1". Similarly, the independent effects of running experience were examined only within the young group wherein the recreational group was treated as the reference group and coded "0" while the novice group (45-65 years) was coded "1".

6.3.3.2 IMU Data

Acceleration data from the tibia and low-back IMUs along the axial (x-axis), anterior-posterior (z-axis), and mediolateral (y-axis) axes were extracted and exported to MATLAB for processing. Each signal was converted from m/s^2 to units of gravity (g) and detrended by subtracting a least-squares best-fit line to remove any existing linear trend in the data. Signal bias was removed by subtracting the signal's mean from each data point such that the signal oscillated about zero. The signals were not filtered and the resultant acceleration signal was calculated as $\sqrt{x^2 + y^2 + z^2}$.

To calculate impact shock and attenuation in the frequency-domain, individual stance phases were detected in both the tibia and low-back accelerations such that the chosen stance phases were completed at a constant speed matching the criterion speed. Since the force plates were located mid-way between the optoelectrical timing gates that monitored running speed, the step coinciding with force-plate contact was detected in the acceleration signals ensuring it met the constant speed requirement. Further, it was necessary to temporally align the tibia and low-back signals to accurately quantify shock attenuation. Thus, the following steps were followed to ensure temporal alignment between IMU sensors and the force plates such that the data from same step and at a constant running speed were analyzed for all trials and participants:

a. Temporal alignment of the tibia and low-back acceleration signals

IMU data were first aligned based on Unix time stamps to account for observed lags in start times between the low-back and tibia sensors and the slightly different sampling frequency of each IMU (right leg: 1598.151 Hz, left leg: 1604.322, low-back: 1587.432 Hz). To algin the data, the location of the first common Unix timestamp in both signals was identified and designated as the beginning of both signals. Then each signal was linearly interpolated to 1600 Hz via the *resample* function in MATLAB.

b. Temporal alignment of the acceleration data with force plate data

The following procedures are described schematically in Figure 1. To align the force plate recorded by QTM and acceleration data measured by the IMU, the time elapsed (QTM period) between the manually flagged jump-landing in QTM (Q_{t1}) and foot-contact on a force plate (Q_{t2}) (i.e., when the ground reaction force surpassed 20 newtons) was calculated. The location of the jump-landing (A_{t1}) was also identified in the tibial acceleration signal as the location of the first peak ≥3g. Q_{t1} and A_{t1} were expected to occur at the same time, so the start of the QTM period was aligned with the tibial acceleration signal such that Q_{t1} and A_{t1} occurred as the first data point for both signals. Therefore, the time stamp of Q_{t2} corresponded to the time stamp in the tibial acceleration immediately that was succeeded by the step run at constant speed (i.e., the step coinciding with force plate contact).

Since the jump landing in QTM was manually flagged, an exploratory analysis (Appendix 7) was performed to identify the time lag or lead relative between to Q_{t1} and $A_{t1.}$ The investigation revealed that the manual flag (Q_{t1}) preceded A_{t1} by ~0.16 seconds. However, 0.16 seconds was small enough such that that the step in the acceleration signal preceding the one coincident with force-plate contact was excluded ensuring the same step was identified in both force plate and acceleration data.

Once the step run at constant speed was identified, it was necessary to detect the gait events of foot-contact and toe-off in the tibial acceleration signal to isolate individual stance phases. The force plate data could not be used for this purpose because there was a lag in start times between QTM and the IMU sensors. A validated foot-contact detection algorithm adapted from Aubol and Milner (2020) was utilized to identify foot-contact and toe-off for the step immediately following the QTM period. Data were filtered using a zero-lag lowpass 4th order Butterworth filter with a cutoff frequency of 70 Hz for step detection. The axial data from the tibial sensor was used for step detection. The algorithm first identified peaks in the jerk (i.e., derivative of acceleration) of the axial acceleration signal within 150 milliseconds of an acceleration peak shortly after foot-ground contact. Then, local minima in the axial acceleration signal that occurred within 75 milliseconds of each jerk peak were detected, while the minima that occurred after an acceleration peak less than 20% in magnitude of the third highest acceleration peak were removed. Finally, the location of the earliest local minima occurring before the axial acceleration peak was identified and designated as foot-contact. Toe-off was identified using contact time, calculated via force-plate data as the time between foot-contact and toe-off. The application of contact time to the instant of foot-contact identified via step detection yielded the location for toe-off. The locations of foot-contact and toe-off thus identified were subsequently applied to all axes of the unfiltered tibial and low-back sensor data, thereby detecting individual stance phases.

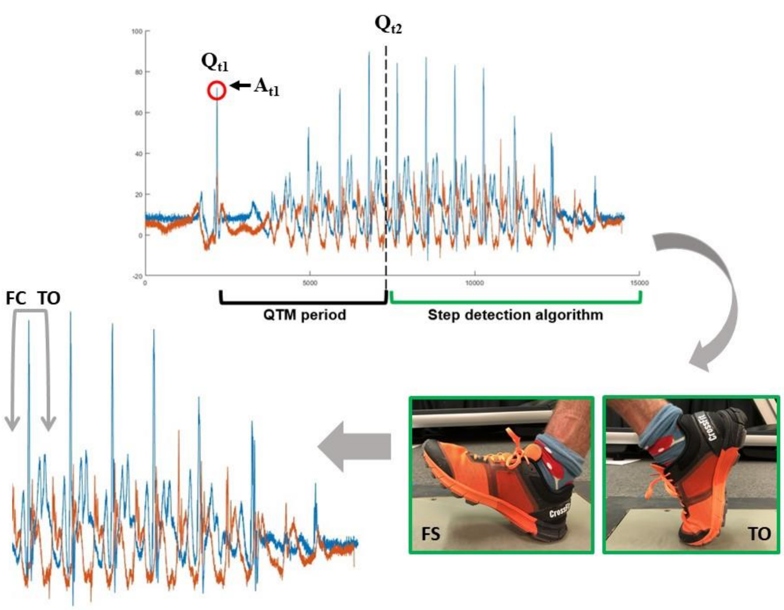


Figure 1. Method used to align the motion capture data with the IMU data so that the same step was identified in both data types. The QTM period was the time between the jump landing at Q_{t1} and foot-contact on a force plate at Q_{t2}. The black vertical dashed line in the top panel indicates the end of QTM period at Q_{t2}. The signal before Q_{t2} was discarded while the signal after it was entered into the step detection algorithm to detect the gait events of foot-contact (FC) and toe off (TO). These events were subsequently applied to all acceleration components. Q_{t1}: location of the jump landing manually flagged in QTM and A_{t1} was the same jump landing observed in the acceleration signal.

To analyze impact shock and attenuation in the frequency-domain, the unfiltered, detrended, and offset-removed stance phases of the tibia and low-back acceleration signals were zero padded at both ends such that the signal length was a power of two, ensuring periodicity (Shorten and Winslow, 1992). Signal power (i.e., energy) of the tibia and low-back accelerations was determined by computing the power spectral density (PSD) of each signal using a Hanning window. The PSD was normalized to 1Hz bins (Edwards et al., 2012; Gruber et al., 2014; Hamill et al., 1995). Signal power magnitude of FB1, FB2, FB3, and FB4 for the tibia and low-back accelerations were computed as the integral of the signal power contained within each bin. The degree of shock attenuated between the tibia and low-back was calculated using the following transfer function (Hamill et al., 1995):

$$\text{Shock attenuation} = 10\ \log_{10}\left(\frac{PSD_{lower\ back}}{PSD_{tibia}}\right)$$

The integral of the transfer function within FB1, FB2, FB3, and FB4 was calculated to assess the magnitude of shock attenuated or gained, with negative values (in decibels) indicating attenuation and positive values indicating a gain (Figure 2).

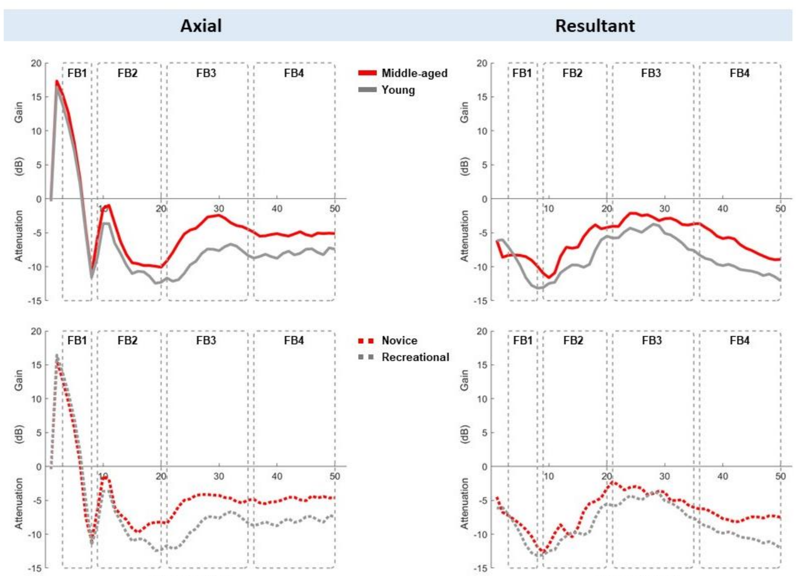


Figure 2. Frequency-domain transfer functions along the axial and resultant acceleration components calculated for the middle-aged and young group (top panel) and the novice and recreational group (bottom panel). Positive values indicate a gain in the signal while negative values indicate attenuation. FB1: 3-8 Hz, FB2: 9-20 Hz, FB3: 21-35 Hz, FB4: 36-50 Hz.

For time-domain analyses, tibial and low-back acceleration signals were filtered using a 4th order, zero-lag Butterworth low-pass filter with a cut-off frequency of 70 Hz. Previous literature has used a cut-off frequency between 40 Hz-100 Hz (Sheerin et al., 2019), however, resonance arising from accelerometer mass and mounting techniques typically occurs between 60-90 Hz and can amplify the magnitude of acceleration to a degree uncharacteristic of human running (Shorten and Winslow, 1992). Hence, a power spectrum analysis was performed using the MATLAB's *'obw'* function to quantify the frequency bandwidth within which 95% of signal power resided. This power analysis was performed on all trials across subjects and limbs and revealed that 95% of total signal power at the tibia was contained below 65.88 Hz. Hence, a cut-off frequency of 70 Hz was deemed appropriate for both the tibia and low-back signals.

For each stance phase, peak impact shock at the tibia was identified as the local maxima in the acceleration signal occurring within the first 50 milliseconds of stance with a magnitude of >3g (Figure 3); 3g is a threshold used to detect foot-contact previously, suggesting peak impact shock must exceed this minimum threshold (Van den Berghe et al., 2019b). Similarly, peak impact shock at the low-back was identified as the local maxima in the acceleration signal occurring immediately following instant of peak tibial shock (Figure 3).

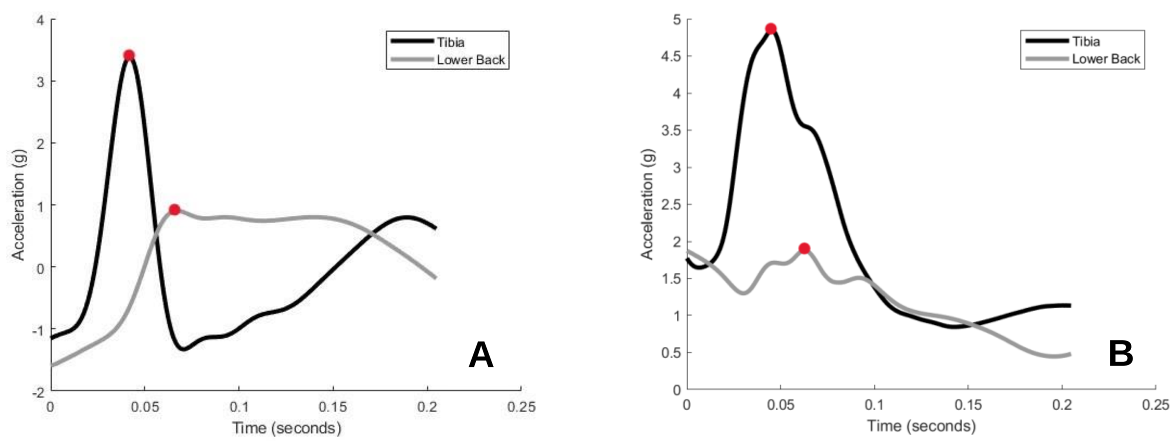


Figure 3. Detection of peak impact shock (red dot) in the time-domain along the axial (A) and resultant (B) acceleration components from a single trial of a representative subject.

In the time-domain, the degree of shock attenuation between the tibia and low-back was calculated using a transfer function as follows (Mercer et al., 2002; Reenalda et al., 2019):

$$Shock\ attenuation = \left(1 - \frac{Peak\ impact\ shock_{low-back}}{Peak\ impact\ shock_{tibia}} * 100\right)$$

For each subject, time and frequency-domain dependent variables were calculated for each limb as an average across eight trials.

6.3.3.3 Statistical Analysis

6.3.3.3.1 Bilateral Differences

Dependent variable data were collapsed across the right and left limbs as there is no evidence for bilateral differences in impact shock and shock attenuation in both time and frequency domains during running (Zifchock et al., 2006). However, before collapsing data across the right and left limbs, bilateral differences in these measures were assessed for the entire sample to ensure that these differences were accounted for if they existed in the current sample. Bilateral differences were examined using repeated measures t-tests for parametric data and Wilcoxon signed-rank tests for non-parametric data, with normality assessed via Shapiro-wilk tests ($\alpha \leq 0.05$). For the independent effects of age, only axial shock attenuation in the time-domain was significantly different between limbs (right mean: -35.00±20.95; left mean: -42.54±22.77, p=0.050, Cohen's d=0.34). For the independent effects of running experience, low-back signal power magnitude in FB1 (right mean: 0.14±0.10; left mean: 0.12±0.09, p=0.050, Cohen's d=0.15) and FB4 (right mean: 0.02±0.02; left mean: 0.01±0.01, p=0.050, Cohen's d=0.20) along the resultant was significantly different between limbs. Therefore, for regression analysis, data for these three variables were treated as independent subjects and data for the other variables were collapsed across limbs.

6.3.3.3.2 Regression Analyses

Unadjusted linear regression was used to assess the independent effects of age and running experience for each dependent variable (i.e., impact shock and attenuation) in both time and frequency-domains along the axial and resultant components. For the independent effects of age and running experience, the young and the recreational groups were treated as the reference groups, respectively. One regression model for each independent-dependent variable pairing was implemented. Linear regression assumption violations were assessed. The dependent variable was log-transformed for models that violated the assumed normality of residuals as assessed using Q-Q plots.

6.4 Results

Anthropometric and demographic data are presented in Table 1. Current weekly mileage and years spent running were significantly greater in the middle-aged recreational versus younger recreational group, while recreational young runners had been running for significantly more years than novice young runners.

Table 1. Anthropometrics, demographics, and running training features (mean±1 standard deviation) for each runner sub-population.

Age Effects	Young (n=24)	Middle-aged (n=20)	p-value	Cohen's d
Age (years)	22.37±4.09	52.95±5.83	<0.001	6.16
Height (m)	1.71±0.08	1.70±0.07	0.65	0.13
Body mass (kg)	68.48±10.03	68.64±10.35	0.95	0.01
Sex distribution (n)	M=15, F=9	M=8, F=12	0.13	Chi sq: 2.21
Current miles·week^{-1}	12.08±6.66	16.92±7.39	0.02*	0.69
Years running	7.42±4.98	22.09±13.40	<0.001*	1.50
Experience Effects	Recreational (n=24)	Novice (n=14)	p-value	Cohen's d
Age (years)	22.37±4.09	22.50±3.52	0.77	0.03
Height (m)	1.71±0.08	1.75±0.06	0.08	0.55
Body mass (kg)	68.48±10.03	72.32±8.35	0.21	0.40
Sex distribution (n)	M=15, F=9	M=10, F=4	0.57	Chi sq: 0.31
Current miles·week^{-1}	12.08±6.66	9.42±3.12	0.10	0.47
Years running	7.42±4.98	0.59±0.36	<0.001*	1.70

Note: The independent effects of age were examined within subjects considered to be recreational runners (per the Honert et al. (2020) definition) only, while the independent effects of running experience were examined within the young group (i.e., subjects aged 18-30 years) only. Therefore, the subject data used for the young group was the same as that for the recreational group; *statistically significant at the 0.050 level; M: male, F: female.

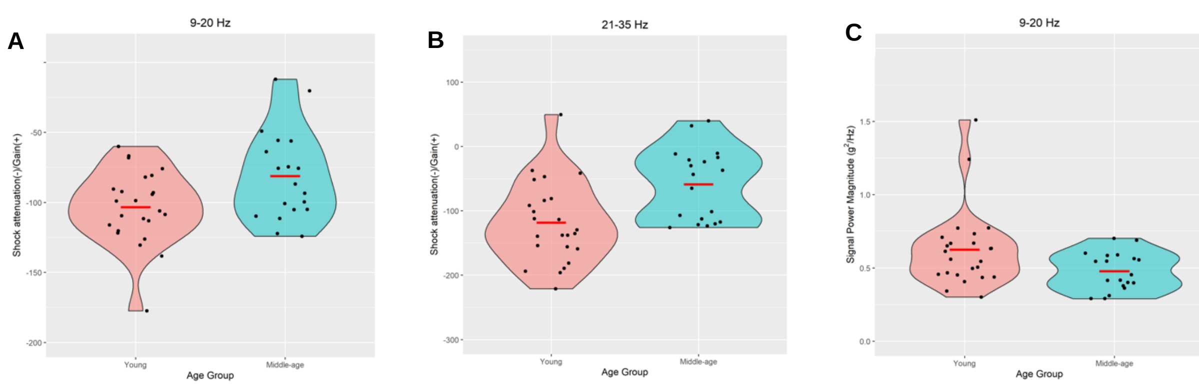
6.4.1 Age Effects

6.4.1.1 Axial Component

Results are displayed in Table 2 and visualized in Figure 4. In the frequency-domain, middle-aged runners attenuated less shock within FB2 (Figure 4A) and FB3 (Figure 4B) compared with young runners. Compared with the young runners, middle-aged runners displayed lower SPM at the tibia in FB2 (Figure 4C) and greater SPM at the low-back in FB3 (Figure 4D) and FB4 (Figure 4E). In the time-domain, middle-aged runners attenuated more shock than young runners in their left limb (Figure 4F). No other significant associations were found.

6.4.1.2 Resultant Component

Results are displayed in Table 2 and visualized in Figure 4. Middle-aged runners attenuated less shock in FB1 (Figure 4G) and displayed greater SPM at the back in FB1 (Figure 4H), FB2 (Figure 4I), FB3 (Figure 4J), and FB4 (Figure 4K) compared with young runners. In the time-domain, middle-aged runners attenuated more shock than young runners (Figure 4L). No other significant associations were found.



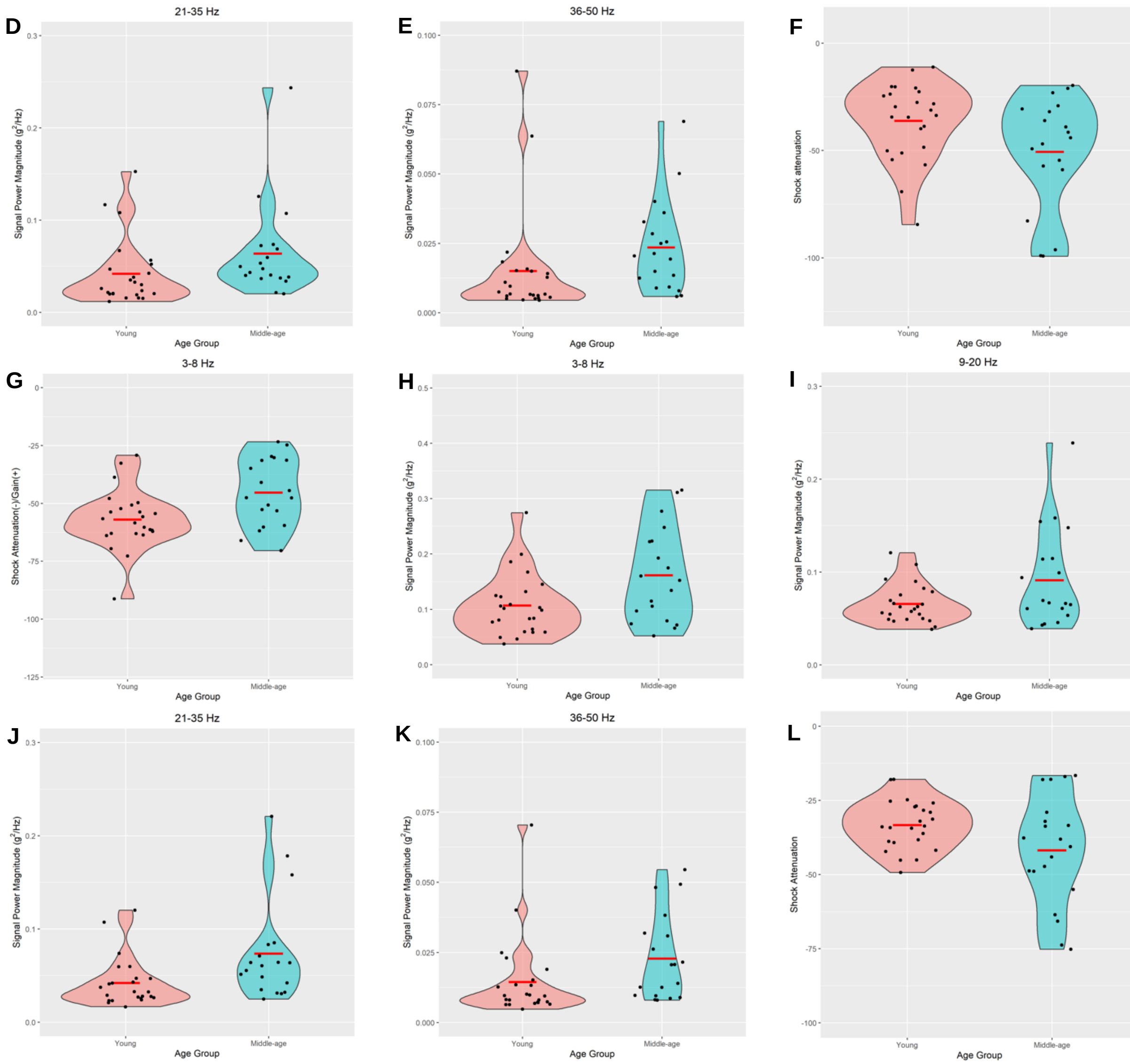


Figure 4. Violin plots depicting the mean (red crossbars) and density distribution for each dependent variable for the young (pink) and middle-aged (blue) groups. Each black dot represents an individual subject. The width of each violin plot indicates the density of data concentration in that region. Plots A-G repesent data along the axial component, while plots H-L represent data along the resultant component. A: shock attenuation in FB2 (9-20 Hz); B: shock attenuation in FB3 (21-35 Hz); C: shock attenuation in FB4 (36-50 Hz); D: signal power magnitude, tibia in FB2 (9-20 Hz); E: signal power magnitude, low-back in FB3 (21-35 Hz); F: signal power magnitude, low-back in FB4 (36-50 Hz); G: time-domain shock attenuation; H: shock attenuation in FB1 (3-8 Hz); signal power magnitude at the low-back in FB1 (I; 3-8 Hz), FB2 (J; 9-20 Hz), FB3 (K; 21-35 Hz), FB4 (L; 36-50 Hz); M: time-domain shock attenuation.

Table 2. Independent effects of age tested using unadjusted linear regression models for each independent-dependent variable pairing.

Frequency Domain			
	β	p-value	95% CI
Resultant Acceleration			
FB1 SPM – tibia^T	-0.13	0.439	-0.47 – 0.20
FB2 SPM – tibia^T	-0.19	0.462	-0.73 – 0.33
FB3 SPM – tibia^T	-0.05	0.781	-0.43 – 0.32
FB4 SPM – tibia	-0.002	0.931	-0.05 – 0.04
FB1 SPM – back	0.05	0.015*	0.01 – 0.09
FB2 SPM – back	0.02	0.035*	0.001 – 0.04
FB3 SPM – back	0.50	0.005*	0.15 – 0.84
FB4 SPM – back^T	0.49	0.017*	0.09 – 0.90
FB1 SA	11.62	0.007*	3.20 – 20.03
FB2 SA	21.32	0.095	-3.94 – 46.60
FB3 SA	24.47	0.125	-7.10 – 56.03
FB4 SA	31.37	0.106	-6.95 – 69.68
Axial Acceleration			
FB1 SPM – tibia	-0.03	0.202	-0.09 – 0.02
FB2 SPM – tibia	-0.14	0.034*	-0.28 – -0.01
FB3 SPM – tibia^T	-0.43	0.063	-0.90 – 0.02
FB4 SPM – tibia^T	-0.08	0.752	-0.59 – 0.43
FB1 SPM – back	-0.008	0.678	-0.04 – 0.03
FB2 SPM – back	0.02	0.178	-0.01 – 0.06
FB3 SPM – back^T	0.48	0.020*	0.07 – 0.88
FB4 SPM – back^T	0.61	0.010*	0.15 – 1.08
FB1 SA	4.12	0.476	-7.45 – 15.69
FB2 SA	22.21	0.016*	4.29 – 40.12
FB3 SA	59.64	0.002*	22.58 – 96.69
FB4 SA	39.67	0.062	-2.22 – 81.56
Time Domain			
Resultant Acceleration			
Peak shock – tibia	-0.03	0.946	-1.06 – 0.99
Peak shock – back	0.29	0.117	-0.07 – 0.66
SA	-8.52	0.046*	-16.89 – -0.15
Axial Acceleration			
Peak shock – tibia	-0.43	0.137	-1.00 – 0.14
Peak shock - back	0.05	0.825	-0.44 – 0.55
SA, R	-2.44	0.704	-15.38 – 10.48
SA, L	-14.38	0.038*	-27.94 – -0.83

*statistically significant at the p<0.050 level; β: beta coefficient; CI: 95% confidence intervals for the beta coefficients. FB1: 3-8 Hz, FB2: 9-20 Hz, FB3: 21-35 Hz, FB4: 36-50 Hz; SPM: signal power magnitude; ^{T}Transformed dependent variable.

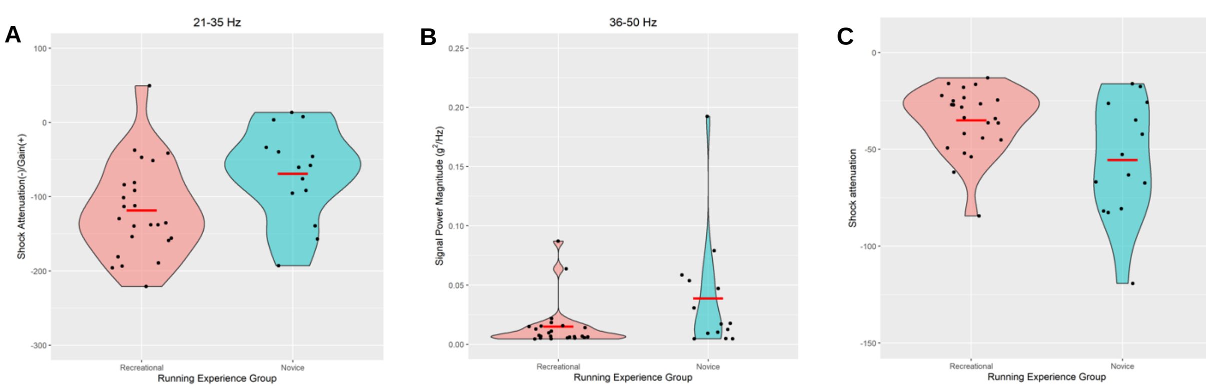
6.4.2 Running Experience Effects

6.4.2.1 Axial Component

Results are displayed in Table 3 and visualized in Figure 5. In the frequency-domain, FB3 shock attenuation was less in novice compared with recreational runners (Figure 5A). Novice runners displayed greater signal power magnitude at the low-back in FB4 than recreational runners (Figure 5B). In the time-domain, shock attenuation (Figure 5C) and peak impact shock at the low-back (Figure 5D) were greater among novice than recreational runners. No other significant associations were found.

6.4.2.2 Resultant Component

Results are displayed in Table 3 and visualized in Figure 5. In the frequency-domain, signal power magnitude at the low-back was greater for the novice compared with recreational runners in FB1 (right limb: Figure 5E; left limb: Figure 5F), FB2 (Figure 5G), FB3 (Figure 5H), and FB4 (right limb: Figure 5I, left limb: Figure 5J). In the time-domain, novice runners attenuated more shock (Figure 5K) and generated a greater peak impact shock at the low-back (Figure 5L) than recreational runners. No other significant associations were found.



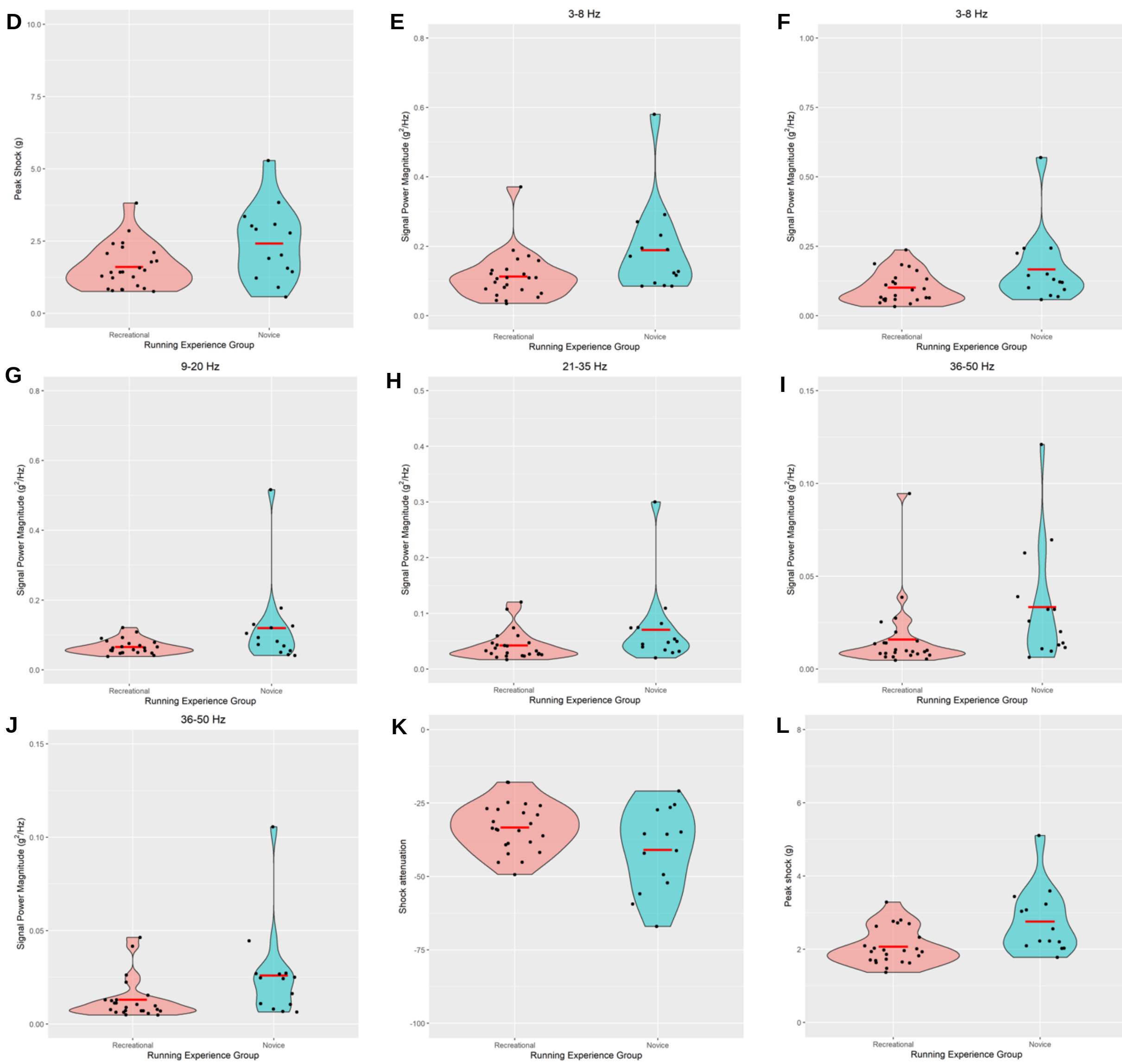


Figure 5. Violin plots depicting the mean (red crossbars) and density distribution for each dependent variable for the recreational (pink) and novice (blue) groups. Each black dot represents an individual subject. The width of each violin plot indicates the density of data concentration in that region. Plots A-D represent data along the vertical component, while plots H-L represent data along the resultant component. A: shock attenuation in FB3 (21-35 Hz); B: signal power magnitude, low-back (36-50 Hz); C: time-domain shock attenuation; D: peak impact shock, low-back; E: signal power magnitude, low-back in FB1 for the right limb (3-8 Hz); F: signal power magnitude, low-back in FB1 for the left limb (3-8 Hz); G: signal power magnitude, low-back in FB2 (9-20 Hz); H: signal power magnitude in FB3 (21-35 Hz); signal power magnitude at the low-back in FB4 (36-50 Hz) for the right limb (I) and left limb (J); K: time-domain shock attenuation; L: peak impact shock, low-back.

Table 3. Independent effects of running experience tested using unadjusted linear regression models for each independent -dependent variable pairing.

	β	p-value	95% CI
Frequency Domain			
Resultant Acceleration			
FB1 SPM – tibia	0.13	0.452	-0.23 – 0.51
FB2 SPM – tibia	0.02	0.818	-0.15 – 0.19
FB3 SPM – tibia^{T}	0.01	0.618	-0.03 – 0.06
FB4 SPM – tibia	0.41	0.129	-0.12 – 0.95
FB1 SPM – back, R^{T}	0.49	0.011*	0.11 – 0.87
FB1 SPM – back, L^{T}	0.44	0.026*	0.05 – 0.83
FB2 SPM – back	0.05	0.037*	0.003 – 0.104
FB3 SPM – back^{T}	0.39	0.047*	0.004 – 0.781
FB4 SPM – back, R^{T}	0.68	0.009*	0.17 – 1.19
FB4 SPM – back, L^{T}	0.60	0.012*	0.13 – 1.06
FB1 SA	6.02	0.212	-3.59 – 15.65
FB2 SA	19.63	0.166	-8.55 – 47.82
FB3 SA	16.705	0.316	-16.63 – 50.04
FB4 SA	21.32	0.375	-26.83 – 69.47
Axial Acceleration			
FB1 SPM – tibia	0.04	0.213	-0.02 – 0.10
FB2 SPM – tibia	-0.01	0.886	-0.19 – 0.16
FB3 SPM – tibia	-0.07	0.233	-0.20 – 0.05
FB4 SPM – tibia	0.002	0.931	-0.04 – 0.05
FB1 SPM – back	-0.01	0.371	-0.05 – 0.02
FB2 SPM – back	0.33	0.087	-0.05 – 0.71
FB3 SPM – back	0.52	0.058	-0.02 – 1.07
FB4 SPM – back	0.71	0.030*	0.07 – 1.35
FB1 SA	-4.67	0.435	-16.67 – 7.33
FB2 SA	21.85	0.065	-1.42 – 45.14
FB3 SA	49.43	0.024*	6.72 – 92.13
FB4 SA	44.36	0.116	-11.53 – 100.26
Time Domain			
Resultant Acceleration			
Peak shock – tibia	0.47	0.295	-0.43 – 1.38
Peak shock – back	0.68	0.004*	0.22 – 1.14
SA	-7.67	0.040*	-14.99 – -0.34
Axial Acceleration			
Peak shock – tibia	-0.07	0.837	-0.77 – 0.630
Peak shock - back	0.81	0.019*	0.14 – 1.48
SA	-20.50	0.010*	-35.85 – -5.14

*statistically significant at the p<0.050 level; β: beta coefficient; CI: 95% confidence intervals for the beta coefficients. FB1: 3-8 Hz, FB2: 9-20 Hz, FB3: 21-35 Hz, FB4: 36-50 Hz; SPM: signal power magnitude; ^{T}Transformed dependent variable.

6.5 Discussion

This study aimed to identify the associations of age and running experience to impact shock and shock attenuation in both the time and frequency-domains during overground running. Axial shock attenuation was substantially greater in the time-domain among middle-aged and novice than young and recreational runners, respectively. Time-domain findings suggest a greater demand on musculoskeletal tissues to attenuate shock with increasing age and less running experience. Such heightened shock attenuation demands with age may be achieved actively rather than passively because in the frequency-domain, middle-aged runners attenuated less shock between 9-35 Hz than young runners. Similarly, novices attenuated less shock between 21-35 Hz than recreational runners. These results may suggest a lesser attenuation demand on passive tissues among middle-aged recreational and young novice runners. However, chapter 5 findings connote that not all higher-frequency shock is attenuated passively; Active gait mechanics are also associated with high-frequency attenuation. Therefore, these results may reflect the diminished ability of middle-aged and novice runners to attenuate higher-frequency shock via active muscle actions thereby modulating injury risk to passive tissues.

Demands placed on and injury risk to passive tissues may be dictated by active gait mechanics like peak knee flexion sagittal knee range of motion. For example, peak knee flexion was positively associated with the attenuation of 5-45 Hz previously (Lafortune et al., 1996), and sagittal knee ROM increased attenuation within 9-20 Hz in chapter 5 of this dissertation. Runners with a more flexed knee during stance may shift attenuation responsibilities away from passive tissues, as suggested earlier (Edwards et al., 2012). Middle-aged runners, however, may not be able to shift attenuation responsibilities from passive to active mechanisms because they typically run with smaller peak knee flexion magnitudes and ranges of motion during stance than young runners (Bus, 2003a; Devita et al., 2016; Fukuchi and Duarte, 2008; Powell and Williams, 2018), potentially addressing why they attenuated less shock within 9-35 Hz. Thus, since peak knee flexion and sagittal knee range of motion contributes to high-frequency attenuation, and middle-aged runners display smaller magnitudes of both these variables, their reduced ability to attenuate 9-35 Hz shock actively may overload passive tissues. Attenuation demands placed on passive tissues may, however, be frequency-band dependent; the lesser degree of attenuation within 9-20 Hz by middle-aged than young runners may not have been due to a reduced ability to actively attenuate higher frequencies but rather due to a smaller input signal, i.e., less signal power at the tibia. Therefore, less attenuation within this band may indicate a smaller passive tissue response was required in middle-aged than young runners. Lower power at the tibia among middle-aged runners may be because they tend to run with short stride lengths (Devita et al., 2016) associated with less impact-related energy than longer stride lengths (Mercer et al., 2003b). Thus, it may be that runners optimize their gait to reduce impact energy as they get older.

Middle-aged and novice runners may rely on similar mechanisms to attenuate axial shock within 21-35 Hz as they both attenuated less shock compared with their counterparts within this frequency band. The physical properties of passive tissues likely contribute to higher-frequency attenuation change with both age and physical activity exposure. For instance, fat mass and fat percentage increase with age among both non-athletes and athletes (Alvero-Cruz et al., 2021; Kyle et al., 2001a) and may predict a decline in leg lean mass (Koster et al., 2011) and skeletal muscle relative to adipose tissue (Schautz et al., 2012). Fat mass is also inversely associated with physical activity (Kyle et al., 2001b), and novice runners who increasingly engage in running along with diet modifications decrease their fat mass (Nielsen et al., 2015). Previously, adipose tissue was evidenced to dissipate energy (Alkhouli et al., 2013). However, data from chapter 4 suggests that shock attenuation of higher frequencies, specifically, decreases with an increase in fat mass. Therefore, increased fat mass in middle-aged and novice runners may explain their diminished attenuation of 21-35 Hz shock. These potential similarities in attenuation mechanisms between middle-age and novice runners suggest future studies should recruit middle-aged runners who are novice to examine an interaction between age and running experience as it may have implications for middle-aged individuals seeking to adopt running as a means of physical activity.

While middle-aged and novice runners attenuated less 21-35 Hz axial shock, they ran with more signal power at the low-back than young and recreational runners for both axial and resultant shock components. Substantial signal power at the low-back could increase attenuation demands within the spinal column which may be insufficiently met by middle-aged runners specifically, as they may display an age-related decline in lumbar lordosis (Arshad et al., 2019) – a structural component that contributes to shock attenuation (Castillo and Lieberman, 2018). However, whether high-frequency energy amplification at the low-back increases running related injury risk along the spinal column is yet unknown.

An amplification of signal power at the low-back can result from the horizontal movement of the trunk, arms, and head which add to low-back energy during deceleration (Chu and Caldwell, 2004; Shorten and Winslow, 1992). Thus, middle-aged and novice runners may have had different upper body kinematics than young and recreational runners which augmented their low-back energy. This hypothesis, however, contrasts previous evidence that found smaller trunk sagittal and transverse plane ranges of motion among older than young runners during stance (Fukuchi et al., 2014) that would lessen and not augment signal power at the low-back. Similarly, novice runners reduced anterior-posterior deceleration of the trunk (Carter et al., 2022) also suggests a smaller signal power magnitude than recreational runners. Therefore, instead of their kinematics, data dispersion may be driving observed group differences in low-back signal power magnitude. For instance, less experienced runners display greater variability in the vertical oscillation of their center of mass (Fadillioglu et al., 2022) suggesting the signal power magnitude data in the current study may also be more variable among novice than recreational runners. An exploratory analysis revealed low-back signal power magnitude was indeed more dispersed about the mean in the novice compared to recreational group for FB1 (novice right limb: 0.69, recreational right limb: 0.61; novice left limb: 0.78, recreational left limb: 0.54), FB2 (novice: 1.00, recreational: 0.31), and FB3 (novice: 0.99, recreational: 0.61), and FB4 (novice left limb: 0.97; recreational left limb: 0.83).

Therefore, the group differences in lower back signal power magnitude should be interpreted with caution.

The axial acceleration component has been typically studied during running but the results of the present study support past recommendations of examining impact shock and attenuation in all three directions (LaFortune, 1991). Middle-aged and novice compared with young and recreational runners attenuated less high-frequency shock axially, though these differences were not observed along the resultant component. Therefore, the different runner sub-populations may experience similar tissue loads due to higher-frequency shock vibrations when accounting for the anteroposterior and medio-lateral shock components. Another dissimilarity between the results obtained along the axial and resultant components was that attenuation within 3-8 Hz reduced with age along the resultant but not the axial component. While this dissimilarity highlights the need to examine both axial and resultant shock, it also reiterates the declining capacity to actively attenuate shock across the frequency spectrum with age – specifically within lower frequencies along the resultant and higher frequencies along the vertical.

Limitations

Potential sex differences in running biomechanics like reduced peak rearfoot eversion among female than male runners (Vannatta et al., 2020) may have influenced the associations of age and running experience to impact shock and attenuation in the current sample. However, sex distribution was statistically similar between running experience groups (chi squared statistic: 0.31, p-value: 0.57) and age groups (chi squared statistic: 2.21, p-value: 0.13) in this study, minimizing the effects on study findings. Nonetheless, female runners are more prone to bone stress injuries (Hollander et al., 2021) associated with high tibial shock (Davis et al., 2004; Milner et al., 2006), therefore sex specificities should be examined in future work.

Middle-aged recreational runners ran significantly more miles per week and had engaged in running for several more years than young recreational runners, suggesting running experience may have modulated the effects of age on impact shock and attenuation. While this study was insufficiently powered to test for an interaction between age and experience, a supplemental analysis (Supplemental Content 1) was performed to identify if the adjustment for current weekly mileage and years running altered the associations between age and each dependent variable. While age was not associated with shock attenuation in FB2 and FB3 along the resultant originally, shock attenuation significantly decreased with age when adjusting for weekly mileage and years spent running. This finding provides further support for the hypothesis that middle-aged runners may possess a diminished ability to attenuate higher frequency shock via active actions like peak knee flexion. Additionally, the previously significant increase in time-domain shock attenuation along the resultant with age was no longer significant when implementing the adjusted model. Along the axial component, shock attenuation in FB2, signal power magnitude at the tibia in FB2 and at the low-back in FB3, and left limb shock attenuation in the time-domain were no longer associated with age upon adjusting for weekly mileage and years of running engagement. Together, these results suggest that the effects of age on impact shock and attenuation in both time and frequency-domains are influenced by running experience. Thus, the combined effects of age and experience must be explored in future work with an adequate sample size.

Due to recruitment challenges, the present study was unable to recruit middle-aged runners who were novice. Therefore, the results of this study for the independent effects of age are limited to recreational runners, while those for the independent effects of running experience are limited to younger runners.

6.6 Conclusions

Middle-aged and novice runners attenuate more axial shock in the time-domain but attenuate less shock between 9-35 Hz and 21-35 Hz, respectively. These findings suggest shock attenuation demand increases with age and less running experience in the time-domain which could explain the greater injury likelihood previously observed among older and less experienced runners (Damsted et al., 2019; Linton and Valentin, 2018; McKean et al., 2006; Videbæk et al., 2015). Age and physical activity-related alterations to body composition may explain why both middle-aged and novice runners attenuated less shock within 21-35 Hz. The similar results observed among middle-aged and novice runners suggests an interaction between age and running experience should be explored in future work. Altered associations between age and certain dependent variables upon adjusting for weekly running mileage and years spent running further highlight the need to examine the combined effects of age and experience on running biomechanics. Additionally, middle-aged runners may be unable to significantly attenuate axial 9-35 Hz shock and resultant 3-8 Hz shock due to a diminished contribution from their typically smaller peak knee flexion magnitudes compared to young runners. However, future work is needed to identify the degree to which joint kinematics and kinetics enable shock attenuation within middle-aged versus young runners. Finally, differences in results when considering the axial versus resultant components of acceleration suggest both should be examined, as previously suggested (LaFortune, 1991), in future running injury studies as they may differently relate to the demands placed on both active and passive mechanisms of shock attenuation.

6.7 Supplemental Content

Supplemental Content 1.

Independent effects of age were tested using unadjusted linear regression models for each independent-dependent variable pairing while adjusting for current weekly mileage and years spent running. The values in bold text indicate results that differed from the unadjusted models.

Frequency Domain			
	β	p-value	95% CI
Resultant Acceleration			
FB1 SPM – tibia^{T}	-0.18	0.437	-0.66 – 0.29
FB2 SPM – tibia^{T}	-0.36	0.327	-1.11 – 0.38
FB3 SPM – tibia^{T}	-0.17	0.499	-0.70 – 0.34
FB4 SPM – tibia	0.0005	0.998	-0.06 – 0.06
FB1 SPM – back	0.07	0.018*	0.01 – 0.13
FB2 SPM – back	0.003	0.041*	0.001 – 0.06
FB3 SPM – back	0.05	0.002*	0.01 – 0.08
FB4 SPM – back^{T}	0.74	0.010*	0.18 – 1.29

FB1 SA	15.11	0.011*	3.59 – 26.63
FB2 SA	**37.17**	**0.035***	**2.72 – 71.62**
FB3 SA	**50.44**	**0.019***	**8.66 – 92.21**
FB4 SA	34.28	0.206	-19.65 – 88.22
Axial Acceleration			
FB1 SPM – tibia	-0.01	0.688	-0.08 – 0.05
FB2 SPM – tibia	**-0.09**	**0.309**	**-0.28 – 0.09**
FB3 SPM – tibia^T	-0.31	0.337	-0.96 – 0.33
FB4 SPM – tibia^T	-0.12	0.720	-0.85 – 0.59
FB1 SPM – back	-0.01	0.530	-0.07 – 0.03
FB2 SPM – back	0.04	0.131	-0.01 – 0.09
FB3 SPM – back^T	**0.51**	**0.069**	**-0.04 – 1.06**
FB4 SPM – back^T	0.63	0.050*	-0.0007 – 1.28
FB1 SA	2.19	0.777	-13.43 – 17.82
FB2 SA	**24.78**	**0.053**	**-0.53 – 49.93**
FB3 SA	58.43	0.027*	6.84 – 110.02
FB4 SA	48.55	0.094	-8.80 – 105.92
Time Domain			
Resultant Acceleration			
Peak shock – tibia	0.03	0.960	-1.33 – 1.40
Peak shock – back	0.43	0.078	-0.05 – 0.92
SA	**-10.27**	**0.069**	**-21.39 – 0.84**
Axial Acceleration			
Peak shock – tibia	-0.09	0.792	-0.83 – 0.63
Peak shock - back	-0.02	0.949	-0.69 – 0.64
SA, R	0.18	0.982	-16.91 – 17.27
SA, L	**-16.74**	**0.082**	**-35.73 – 2.25**

*statistically significant at the p<0.050 level; β: beta coefficient; CI: 95% confidence intervals for the beta coefficients. FB1: 3-8 Hz, FB2: 9-20 Hz, FB3: 21-35 Hz, FB4: 36-50 Hz; SPM: signal power magnitude; ^{T}Transformed dependent variable.

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7 CHAPTER 7: Executive Summary

7.1 Overview

Body segment vibrations that occur during running motion after foot-ground contact range from 3-50 Hz (Gruber et al., 2014; Shorten and Winslow, 1992). Frequencies of 8 Hz and lower may be absorbed by active eccentric muscle actions while frequencies of 9 Hz and higher may be attenuated by passive biological tissue deformation (Shorten and Winslow, 1992). However, it is important to determine the degree of attenuation and reliance on these active and passive shock attenuation mechanisms during human running to assess their potential running-related injury risk (RRI).

Study 1 examined the relationship between body composition measures (lower extremity and pelvis bone mineral density [BMD] and content [BMC], lower extremity and gynoid fat mass, and lower extremity and gynoid lean mass) and the degree of shock attenuation between the tibia and lower back within specific frequency bands (3-8 Hz, 9-20 Hz, 21-35 Hz, and 36-50 Hz) along both axial and resultant acceleration components among runners. Given that bone and soft tissue in animal models effectively attenuated high-frequency shock (Paul et al., 1978), it was hypothesized that an increase in body composition measures would increase shock attenuation within 9-50 Hz, but not within 3-8 Hz.

Likewise, Study 2 aimed to identify the associations between previously proposed gait-related shock attenuation mechanisms and the degree of shock attenuation within 3-8 HZ-36-50 HZ along both axial and resultant acceleration components among runners. It was expected that peak joint angles, ranges of motion, negative mechanical joint work, and power at the knee and ankle would be positively associated with shock attenuation within 3-8 Hz but not 9-50 Hz, while ankle and knee joint stiffness would be negatively associated with shock attenuation within 3-8 Hz but not 9-50 Hz because active mechanisms are proposed to attenuate low but not high-frequency shock.

The general premise of study 1 was that body composition influences the degree of shock attenuation that occurs during running. Running improves musculoskeletal tissue health and resiliency and decreases body fat (Nielsen et al., 2015) whereas aging increases body fat (Alvero-Cruz et al., 2021; Kyle et al., 2001a) and decreases musculoskeletal tissue health (Warming et al., 2002). Consequently, these changes in body composition due to age and running experience suggest that middle-aged and novice runners may exhibit different degrees of shock attenuation via passive tissue deformation compared with younger and recreational runners. Determining the association between age- and experience-related differences in tissue health may provide insights into the differing injury rates observed between these specific runner sub-populations (Damsted et al., 2019; Linton and Valentin, 2018; McKean et al., 2006; Videbæk et al., 2015). Study 3 therefore aimed to investigate the associations between age, running experience, impact shock, and shock attenuation in both the frequency and time domains along axial and resultant acceleration components. The hypothesis was that age would be positively associated with peak impact shock, time-domain shock attenuation, signal power magnitude, and shock attenuation in 9-20 Hz, 21-35 Hz, and 36-50 Hz, but negatively associated with signal power magnitude and shock attenuation in 3-8 Hz. Conversely, running experience would be negatively associated with peak impact shock, time-domain shock attenuation, signal power magnitude, and shock attenuation in 9-20 Hz, 21-35 Hz, and 36-50 Hz, but positively associated with signal power magnitude and shock attenuation in 3-8 Hz.

7.2 Studies 1 and 2: Passive and Active Shock Attenuation Mechanisms

The frequencies between 3-8 Hz are hypothesized to be attenuated by active mechanisms rather than passive deformation of biological tissues (Paul et al., 1978). Study 1 confirmed this result given that, within 3-8 Hz, there was no significant association between axial and resultant shock attenuation and body composition within the lower extremity. Instead, Study 2 revealed that axial shock attenuation within 3-8 Hz is achieved through an active process of increasing knee joint compliance during deceleration, dorsiflexion angle at foot-ground contact, and peak ankle eversion during stance. Similarly, increasing peak knee flexion during stance, sagittal and frontal ankle ranges of motion during deceleration, and reducing sagittal plane ankle joint stiffness increase resultant shock attenuation within

3-8 Hz. Thus, axial and resultant shock within the range of 3-8 Hz is actively rather than passively attenuated, while high-frequency shock attenuation may be attributed to biological tissues.

Contrary to expectations based on the role of fat mass in attenuating shock within the tibia during a pendulum experiment (Schinkel-Ivy et al., 2012), the results of Study 1 indicate that an increase in leg and gynoid fat mass significantly reduces axial shock attenuation within 9-50 Hz, and within 9-20 Hz and 36-50 Hz after adjusting for the effects of age. In the gynoid region, these findings may be attributed to the negative correlation found between gynoid fat mass and lean mass, which indicates that as fat mass increases, there is less connective and muscle tissue available to attenuate high-frequency shock; muscle tissue like the rectus femoris muscle has been found to effectively absorb shock frequencies exceeding 25 Hz in running humans (Boyer and Nigg, 2006).

Joint kinematics like sagittal knee and ankle ranges of motion that are dictated by active muscle actions can also contribute to high-frequency shock attenuation, as suggested by study 2 results. Such active contributions to the attenuation of shock >9 Hz may have masked the associations between BMD, BMC, and lean mass to high-frequency shock attenuation in study 1. These associations may be revealed in future work by controlling for muscle activation given that Schinkel-Ivy et al. (2012) found significant contributions of bone mass and wobbling mass to shock attenuation within the tibia when muscle activation was accounted for. Further, it may be important to explore the structural properties of bone, fat mass, and lean mass like material stiffness that contribute to their deformation and thereby shock attenuation in addition to the mass-related properties explored presently.

It is important to acknowledge the potential influence of selection bias on the results of study 1. Out of the 57 subjects who consented for study 2, only 38 agreed to undergo the dual x-ray absorptiometry (DXA) scan for study 1. Among the remaining 19 subjects, five were unable to undergo the scan due to scheduling challenges, while the other 16 subjects chose not to participate in the scan procedures due to concerns about radiation exposure. There is a possibility that the subjects who declined the scan were older and more cautious about radiation, or they may have been less experienced runners who were not as motivated to complete the body composition procedures. This potential bias could lead to a sample that is skewed towards a younger and more experienced running group. However, Table 1 demonstrates that the primary sample from study 2 and the subset of subjects who underwent the DXA scan in study 1 did not significantly differ in terms of their anthropometric and demographic characteristics. This finding indicates that the risk of selection bias in study 1 is minimized, as the two groups are comparable. Overall, while there is a potential for selection bias in study 1, the lack of significant differences between the primary sample and the subset of subjects who underwent the DXA scan provides some assurance regarding the generalizability of the findings.

Table 1. Anthropometric and demographic variables for subjects from study 1 compared with those from study 2.

	Mean ± 1 SD, Aim 1	Mean ± 1 SD, Aim 2	p-value
Number of participants (n)	38	57	-
Age (years	31.5 ± 14.9	32.66 ± 15.32	0.74
Sex distribution	17 female, 21 male	24 female, 33 male	0.79
Height (m)	1.70 ± 0.08	1.72 ± 0.07	0.53
Body mass (kg)	68.3 ± 10.3	69.52 ± 9.82	0.58
Current weekly mileage (miles·wk^{-1})	12.5 ± 5.9	12.66 ± 6.82	0.42
Years spent running (years)	10.75 ± 12.32	10.95 ± 12.11	0.89

SD: standard deviation; Current weekly mileage and years spent running overall were self-reported

7.3 Study 3: Age and Experience Effects on Impact Shock and Shock Attenuation

7.3.1 Frequency domain findings within the axial component

Study 2 found that joint kinematics contribute to the attenuation of axial low and high-frequency shock. These findings may explain some of the associations observed in Study 3, which investigated the independent effects of age and running experience on impact shock and shock attenuation in both time and frequency domains during running. Specifically, in Study 3, middle-aged runners attenuated less axial shock between 9-50 Hz than younger runners. As older runners typically display less peak knee flexion throughout stance and a smaller sagittal knee range of motion (Bus, 2003a; Devita et al., 2016; Fukuchi and Duarte, 2008; Powell and Williams, 2018), reduced shock attenuation by middle-age runners aligns with the Study 2 finding that axial shock attenuation within 9-20 Hz and 36-50 Hz increases with an increase in sagittal plane knee range of motion during deceleration and peak knee flexion during stance, respectively. The reduced active shock attenuation observed in high frequency bands due to a less complaint lower extremity may lead to an increase in shock attenuation demands on the passive tissues of middle-aged and novice runners, potentially increasing their risk of RRI.

Middle-aged runners may have attenuated less axial shock within 9-20 Hz not only due to a less compliant knee joint, but also because of the smaller input signal (i.e., signal power magnitude) found at the tibia compared with the young group. This smaller input signal could be a consequence of a less extended knee orientation (i.e., more compliant) at foot-ground contact, which is observed with shorter stride lengths (Heiderscheit et al., 2011) typically associated with age (Devita et al., 2016; Willy and Paquette, 2019). Studies have shown that shorter stride lengths are linked to decreased signal power at the tibia (Mercer et al., 2003b). Thus, the need for a smaller attenuation response among middle-aged runners than younger runners may be attributed to a combination of a smaller input signal resulting from a less-extended knee at foot-ground contact characteristic of older runners' gait (Bus, 2003a), and a less compliant lower extremity during stance (Bus, 2003a; Devita et al., 2016; Fukuchi and Duarte, 2008; Powell and Williams, 2018). It is challenging to determine which factor, the smaller input signal, or the less compliant lower extremity, primarily contributed to the lower attenuation observed in middle-aged runners compared to younger runners. To parse out the shock attenuation effects of a smaller input signal versus a less compliant lower extremity would require one of these variables to be controlled while the other is varied.

Study 3 found that both middle-aged runners and novice runners attenuated less axial shock within 21-35 Hz compared to younger and recreational runners, respectively. Given that gait mechanics did not contribute to axial attenuation within 21-35 Hz in study 2, the reduction in attenuation with age and less running experience may be explained by other factors like fat mass, which was found to decrease axial attenuation within 21-35 Hz in study 1. Specifically, fat mass increases with age (Alvero-Cruz et al., 2021; Kyle et al., 2001a; Ponti et al., 2020) and less activity exposure (Kyle et al., 2001b; Nielsen et al., 2015). Therefore, middle-aged runners and novice runners who may have more fat mass than their younger and more experienced counterparts may thereby attenuate less shock due to this body composition characteristic.

7.3.2 Frequency domain findings within the resultant component

The axial component of segmental acceleration during running is the dominant contributor to the magnitude of the resultant. Therefore, several of the shock attenuation mechanisms discussed along the axial component also apply to the resultant component. However, Van den Berghe et al. (2019a) demonstrated that the anterior-posterior and mediolateral components are insightful in addition to the axial direction. For example, a non-rearfoot strike was found to elicit a greater resultant but not axial impact shock than a rearfoot strike. This suggests that distinct mechanisms may be employed to attenuate the varying magnitudes of shock experienced along each component.

Indeed, the findings from studies 1 and 2 indicate distinct mechanisms for shock attenuation along the axial and resultant components. In study 1, increased lower extremity fat mass reduced axial but not resultant shock attenuation within 9-50 Hz. This discrepancy may explain why middle-aged and younger runners in study 3, who tend to differ in fat mass concentrations, showed different axial but similar resultant attenuation in 9-50 Hz. Furthermore, study 2 revealed that increased sagittal knee range of motion was associated with higher axial but lower resultant attenuation within 3-20 Hz. Similarly, knee joint stiffness decreased axial but increased resultant attenuation within the 9-20 Hz range. These findings underscore the role of knee joint compliance in axial attenuation within 9-20 Hz, while ankle range of motion appears to be a significant factor in resultant attenuation within this frequency band, as demonstrated in study 2. These differences highlight the importance of studying multi-directional shock attenuation.

Study 3 further highlights distinct differences in shock attenuation along the axial and resultant components. Middle-aged runners showed lower resultant shock attenuation than younger runners in the

3-8 Hz range, while axial attenuation did not significantly differ between the groups within this range. These findings suggest that age-related reductions in knee and ankle joint compliance, commonly observed (Devita et al., 2016; Fukuchi et al., 2014; Powell and Williams, 2018), may contribute to decreased resultant attenuation among middle-aged runners in the 3-8 Hz range. However, these same mechanisms may explain their decreased axial attenuation within the 9-50 Hz range compared with younger runners. Thus, the degree of contribution by ankle and knee joint compliance to shock attenuation within specific frequency bands may depend on the direction of acceleration.

The dominance of the axial component in resultant acceleration may explain the similarities observed in the results between the axial and resultant directions. For instance, an increase in peak knee flexion increased axial and resultant attenuation within 36-50 Hz, and 3-35 Hz respectively. As suggested previously (Edwards et al., 2012), increasing knee flexion at contact and peak knee flexion during ground contact may increase eccentric muscle action contribution to shock attenuation, potentially explaining the associations of peak knee flexion to both axial and resultant attenuation.

7.3.3 Time domain findings

In the time domain, novice and middle-aged runners demonstrated more axial and resultant attenuation compared with younger and recreational runners, respectively. However, within the 9-50 Hz frequency range, which was found to be attenuated through active mechanisms in study 2, both novice and middle-aged runners exhibited lower shock attenuation than their younger and more experienced counterparts. This suggests that the increased demand for attenuation may have been met passively by these groups.

7.4 General Summary

This dissertation provides empirical evidence that emphasizes the significance of gait and body composition-related mechanisms in shock attenuation during running. Importantly, the results of study 3 reveal that runners of different ages and exposures to running possess distinct shock attenuation capacities possibly due to the employment of shock attenuation mechanisms to different degrees. Although the relationship between frequency-domain shock attenuation and prospective running injury risk has not been established yet, it is possible that the reduced active contribution to high-frequency shock attenuation observed in middle-aged and novice runners could increase their injury risk by overloading their passive tissues that may be less adapted to bear such loads compared to younger and recreational runners. Further, the results of studies 1 and 2 suggest that body composition and gait mechanics influence shock attenuation across the frequency spectrum, therefore alterations to these variables with age and exposure to physical activity likely alter a runner's shock attenuation response.

7.5 Future Directions

7.5.1 Combined effects of age and running experience on shock attenuation

This study demonstrated that both age and running experience can affect impact shock and attenuation in the time and frequency domains. Consequently, tailoring training or footwear interventions may be necessary for older runners based on their running experience, as an older novice runner may attenuate shock differently than an older recreational runner. Therefore, future studies should investigate the interaction between age and running experience with a sufficiently large sample size and a wide range of experience levels in both older and younger age groups.

7.5.2 Potential pathways for the associations between age, running experience, and shock attenuation

This dissertation identified differences in shock attenuation between runner sub-populations in study 3 and revealed body composition and gait mechanisms of shock attenuation in studies 1 and 2. Therefore, a natural follow-up would be to examine if the identified active and passive mechanisms uncovered in studies 1 and 2 serve as pathways for the associations between age, running experience, and shock attenuation. For instance, if the association between age and shock attenuation is mediated by sagittal knee range of motion which was a shock attenuation mechanism in study 2, then it will mean that differences in shock attenuation between middle-aged and young runners are due to their varying degrees of sagittal knee range of motion specifically. Thus, a next step to this dissertation would be a mediation analysis to explain the significant associations found between age and shock attenuation.

7.5.3 Gait retraining interventions

Study 2 indicates that resultant shock attenuation within 9-20 Hz and 36-50 Hz could be increased by increasing knee flexion at foot-ground contact, leading to a shift in the responsibility of shock attenuation away from passive tissues. Retraining gait through interventions such as switching from a rearfoot to a non-rearfoot strike pattern, which has been found to increase knee flexion at contact (Doyle et al., 2022), may be beneficial for older and novice runners who may not have well-adapted tissues. Additionally, adopting a non-rearfoot strike pattern could lead to a reduction in signal power magnitude at the tibia across frequencies (Gruber et al., 2014), which may result in the need for less shock attenuation among middle-aged and novice runners. To further investigate the findings of Study 3, the effects of foot-strike pattern retraining on shock attenuation characteristics should be examined in each runner sub-population and its potential impact on prospective injury risk should be assessed.

7.5.4 Body composition measurements

Dual x-ray absorptiometry was used in study 1 to obtain measures of bone mass because bone mass was associated with shock attenuation in the past (Schinkel-Ivy et al., 2012). Future studies can utilize techniques such as computed tomography, magnetic resonance imaging, and finite element modeling to assess tissue-level material properties of biological tissue, in addition to bone mass considered in Study 1. Material properties will supplement the properties of mass as bone architecture and structure, including bone cross-sectional area, width, thickness, porosity, volume, shape, and trabeculae number and connectivity, also play a role in determining the mechanical properties of bone (Carbonare and Giannini, 2004; van der Linden and Weinans, 2007), which are important in energy dissipation and potentially shock attenuation. Therefore, such assessments may complement iDXA measurements and provide insights into a runner's ability to attenuate shock.

7.6 References

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8 Appendix 1

Table 1. Summary chart of previous studies exploring the determinants of impact shock and its correlates, namely VGRF-IP and loading rates.

Determinant	Author (years)	Study Design	Activity	Participants	Notable exposure characteristics	Impact-related outcome variables	Directionality of results
Vertical touchdown velocity	Gerritsen et al. (1995)	Observational	Running simulation	None: Model parameters adopted from previous studies	Vertical touchdown velocity of the heel was simulated	Peak impact force magnitude and vertical loading rate	As vertical touchdown velocity of the heel increased: Peak impact force ↑ Vertical loading rate: not reported
	Lafortune et al. (1996)	Observational	Impacts administered via a human pendulum while lying supine	17 active men	Impact velocity of the human pendulum device was varied such that touchdown velocity changes	Peak ,time to peak, and power spectra of shank and head impact shock	As impact velocity increased: Peak shank shock ↑ Peak head shock ↑ Time to peak for both head and shank shock ↓ Power or energy in lower and higher frequencies at both head and shank ↑

Sagittal plane knee contact angle	Bobbert et al. (1991)	Observational	Running	3 males	Knee flexion not directly measured but Groucho running analyzed which includes significantly more knee flexion compared with normal running	Contributions of support leg segments to the impact peak	As running style changed to Groucho running: Contributions of support leg segments to the impact peak ↑
	Derrick (2004)	Review Article	Running, in-line skating, and mass-spring-damper model of running	N/A	0-60 degrees knee flexion during in-line skating and up to 20 degrees knee flexion at contact during running across studies	Peak impact force and peak impact shock	Reviewed literature suggests increased knee flexion may result in the following if the effective mass effect dominates: Peak impact force ↓ Peak impact shock ↑

Sagittal plane knee contact angle	Lafortune et al. (1996)	Observational	Impacts administered via a human pendulum while lying supine	21 active men	Knee flexion from 0 (full extension) to 40 degrees of flexion was achieved via suspension system	Peak ,time to peak, and power spectra of shank and head impact shock	As knee flexion at contact increased: Peak shank shock ↑ Peak head shock ↓ Time to peak for both head and shank shock ↓ Power or energy in frequencies between 5-45 Hz at the head ↓ Power or energy in frequencies >30 Hz at the shank ↑
	Edwards et al. (2012)	Observational	Impacts delivered via bumps on a treadmill while wearing in-line skates	4 females and 6 males	0-30 degrees of knee flexion assessed	Power spectral density at the head and lower leg	As knee flexion at contact increased: The energy contained in the higher impact-related frequencies (15-35 Hz) ↑

Sagittal plane foot and/or ankle contact angle	Gerritsen et al. (1995)	Observational	Running simulation	None: Model parameters adopted from previous studies	Foot angle relative to the ground was simulated. Increase in foot angle indicates more dorsiflexion.	Peak impact force magnitude and vertical loading rate	As dorsiflexion at contact increased: Peak impact force ↑ Vertical loading rate ↑
	Gruber et al. (2014)	Cross-sectional	Treadmill running at 3.5 m/s	19 rearfoot (12 males) and 19 forefoot (14 males) runners	Greater ankle dorsiflexion observed among rearfoot runners, greater plantarflexion observed among forefoot runners	Frequency at which peak power occurs at the tibia and head in the lower (3-8 Hz) and higher (9-20 Hz) frequency range, energy contained in lower and higher frequencies at the tibia and head	Among rearfoot runners with greater dorsiflexion compared to forefoot runners with greater plantarflexion: Frequency at which peak power occurred in the lower range at the head ↓ Frequency at which peak power occurred in the higher range at the tibia ↑ Energy contained in both lower and higher frequencies at the tibia ↑

Sagittal plane foot and/or ankle contact angle	Gruber et al. (2017)	Cross-sectional	Overground running at 3.5 m/s	20 habitual rearfoot (13 males) and 20 non-rearfoot (15 males) runners	Habitual rearfoot runners were those with greater than 8 degrees of initial dorsiflexion at ground contact, while non-rearfoot runners were those with less than 8 degrees of initial dorsiflexion at ground contact	Time-frequency space waveform components of the vertical and resultant ground reaction forces during the impact phase of running. Specifically the maximum signal power of impact-related higher frequencies	Among the rearfoot (i.e. more dorsiflexion at contact) compared to non-rearfoot group (i.e. less dorsiflexion at contact): Frequencies at which maximum signal power occured ↑ , while also occuring 12% sooner in stance for both the resultant and vertical GRFs
Frontal plane rearfoot and/or ankle contact angle	Nigg & Morlock. (1987)	Repeated measures design	Overground running at 4 m/s	14 male runners	Rearfoot angle as a function of positive, none, and negative lateral heel flare. Initial pronation at touchdown decreases as heel flare goes from positive to negative.	Vertical ground reaction force impact peak, timing of the impact peak, and vertical loading rate.	As pronation increases: Vertical ground reaction force impact peak ≅ Vertical loading rate ≅ Timing of impact peak ↓

Frontal plane rearfoot and/or ankle contact angle	Perry & Lafortune (1995)	Repeated measures design	Overground running at 3.8 m/s	10 males	Rearfoot motion	Resultant ground reaction force impact peak, vertical loading rate, and peak tibial shock	As pronation increased: Resultant ground reaction force impact peak ↓ Vertical loading rate ↓ Peak tibial shock ↓
	Derrick et al. (2002)	Repeated measures design	Fatiguing run performed on a treadmill	10 recreational runners	Rearfoot motion	Peak impact shock at the tibia and head in the time domain	As the rearfoot angle became more inverted towards the end of the run: Peak impact shock at the tibia ↑ Peak impact shock at the head ≅
Stride length	Mercer et al. (2002)	Repeated measures design	Treadmill running at incremental speeds ranging from 50%-100% of their maximal speed.	8 male runners	Stride length freely varied with running speed, exhibiting a positive relationship	Peak impact shock at the tibia and head in the time domain	As stride length increased with running speed: Peak impact shock at the tibia ↑ Peak impact shock at the head ↑ but to a much smaller extent compared to the tibia

Stride length	Mercer et al. (2005)	Repeated measures design	Treadmill running at different speeds and both self selected and constrained stride lengths	8 volunteers	Stride length was both self-selected and constrained to 2.5m at different running speeds.	Peak impact shock at the tibia and head in the time domain, and vertical ground reaction force impact peak	When freely chosen stride length increased with running speed: Vertical ground reaction force impact peak ↑ Peak impact shock at the tibia ↑
	Schubert et al. (2014)	Review Article	Treadmill and overground running at constant speed	10 articles included	Stride length, preferred and constrained at a higher and lower length compared to preferred	Ground reaction force impact peak, peak impact shock at the tibia and head	As stride length increases, Peak impact shock at the tibia ↑ Ground reaction force impact peak ↑
Stride frequency	Schubert et al. (2014)	Review Article	Treadmill and overground running at constant speed	10 articles included	Stride frequency, preferred and constrained at a higher and lower frequency compared to preferred	Ground reaction force impact peak, peak impact shock at the tibia and head	As step frequency increases: Peak impact shock at the tibia ↓ Ground reaction force impact peak ↓

Stride frequency	Hamill et al (1995)	Repeated measures design	Treadmill running at preferred running speed	10 males	Stride frequency, preferred and constrained at a higher and lower frequency compared to preferred	Power spectrum at the tibia and head	As stride frequency increased: Energy contained in higher and lower frequencies at the tibia ↓ Energy contained in higher and lower frequencies at the head ≅

↑ increased, ↓ decreased, ≅ remained unchanged/was similar

9 Appendix 2

Table 1. Overview of the studies exploring the determinants of active and passive shock attenuation mechanisms.

Mechanisms	**Author (years)**	**Study Design**	**Activity**	**Participants**	**Notable exposure characteristics**	**Attenuation-related outcome variables**	**Directionality of results**
Potential Active Mechanisms - Gait related							
Negative mechanical joint work	Derrick et al., (1998)	Observational, Repeated measures design	Overground running at 3.83 m/s using preferred stride length, +20% of preferred and -20% of preferred stride length	10 male subjects	Negative mechanical work at the hip, knee and ankle joint	Attenuation within each frequency bin between the tibia and head power spectral densities	As stride length increased, negative mechanical work increased, resulting in: Attenuation between the tibia and head ↑
Knee flexion magnitudes	Lafortune et al. (1996)	Observational	Impacts administered via a human pendulum while lying supine	21 active men	Knee flexion from 0 (full extension) to 40 degrees of flexion was achieved via suspension system	Impact shock peak ratio (head/tibia), Attenuation between the tibia and head power spectrums	As initial knee flexion angle increased: Impact shock peak ratio ↓ Attenuation of frequencies >5 Hz ↑

Knee flexion magnitudes	Edwards et al. (2012)	Observational	Impacts delivered via bumps on a treadmill while wearing in-line skates	4 females and 6 males	0-30 degrees of knee flexion assessed	Attenuation within each frequency bin between the tibia and head power spectral densities Cutoff frequency that minimized the difference in peak tibial and head acceleration - termed the low-pass filter cutoff frequency of the body	As knee flexion increased: Shock attenuation between the tibia and head ↑ Low-pass filter cut-off frequency ↓ meaning higher frequencies were attenuated while lower frequencies were allowed to transmit to the head
Sagittal plane ankle motion	Gruber et al. (2014)	Cross-sectional	Treadmill running at 3.5 m/s	19 rearfoot (12 males) and 19 forefoot (14 males) runners	Greater ankle dorsiflexion observed among rearfoot runners, greater plantarflexion observed among forefoot runners	Attenuation within each frequency bin between the tibia and head power spectral densities	In rearfoot runners (greater dorsiflexion) than forefoot runners (greater plantarflexion): Shock attenuation between the tibia and head for both lower (4-8 Hz) and higher (9-20 Hz) frequencies ↑

Frontal plane ankle and/or rearfoot motion	Perry & Lafortune, 1995	Repeated measures design	Overground running at 3.8 m/s	10 males	Rearfoot motion	Resultant ground reaction force impact peak, vertical loading rate, and peak tibial shock	As pronation increased: Resultant ground reaction force impact peak ↓ Vertical loading rate ↓ Peak tibial shock ↓
	Nigg & Morlock, 1987	Repeated measures design	Overground running at 4 m/s	14 male runners	Rearfoot angle as a function of positive, none, and negative lateral heel flare. Initial pronation at touchdown decreases as heel flare goes from positive to negative.	Vertical ground reaction force impact peak, timing of the impact peak, and vertical loading rate.	As pronation increases: Vertical ground reaction force impact peak ≅ Vertical loading rate ≅ Timing of impact peak ↓

Joint/lower extremity stiffness	Chu & Caldwell (2004)	Cross-sectional study design and modeling	Treadmill running at 4.17 m/s. Treadmill grade was varied between 0%-12% downhill. A spring-mass-damper model was used to assess lower extremity stiffness and damping.	10 experienced male runners with low and high shock attenuation	Lower extremity stiffness was simulated using a mass-spring-damper model with initial conditions such as touchdown velocity and mass taken from the treadmill running data	Shock transmission and damping as a measure absorption	As downhill grade increased, in the high shock attenuation group: Lower extremity stiffness ↑ Shock transmission ↓ Damping ↑ In the low shock attenuation group: lower extremity stiffness ≅ Damping ↓
	Hamill et al. (2014)	Cross-sectional	Overground running at 3.5 m/s	20 habitual rearfoot and 20 habitual forefoot, young runners	Ankle and knee joint stiffness calculated using joint moment-angle curves	Energy absorption at the ankle and knee joint represented as negative joint work performed during the shock absorption phase of stance	Forefoot running than rearfoot running in both groups resulted in less ankle but higher knee stiffness, with the following differences in energy absorption: Forefoot than rearfoot running displayed: Energy absorption at the ankle ↑ Energy absorption at the knee ↓

Active muscle actions	Mercer et al. (2003a)	Repeated measures design	Fatiguing run performed on a treadmill	10 healthy males	Maximal graded exercise test	Shock attenuation between the distal tibia and head power spectral densities across a frequency range of 11-18 Hz.	Post the fatiguing run: Shock attenuation ↓
	Derrick et al. (2002)	Repeated measures design	Fatiguing run performed on a treadmill	10 recreational runners	Volitional run to exhaustion	Peak impact shock at the tibia and head in the time domain	As the rearfoot angle became more inverted towards the end of the run: Peak impact shock at the tibia ↑ Peak impact shock at the head ≅

Potential Passive Mechanisms - Body composition related							
Body composition	Schinkel-Ivy et al. (2012)	Observational study	Human pendulum method	36 healthy participants divided into low, medium and high body fat percentage groups	Body fat percent, lean mass, wobbling mass, bone mineral content	Peak impact shock, slope of peak impact shock, and time to peak impact shock. Note: the accelerometer was attached to the proximal tibia meaning all impact shock parameters represent the degree of shock attenuation as well.	As lean mass, wobbling mass, and bone mineral content increased: Peak impact shock ↓ Peak impact shock slope ↓ As wobbling mass increased: Time to peak impact shock ↑

Bone	Radin et al. (1970)	Interventional study	Longitudinally applied external impulsive force to adult bovine bone	Adult bovine's proximal interphalangeal joints	Longitudinally applied rapid forces to interphalangeal joints	Force transmitted through the joint as a function of the presence of (1) All joint structures (2) periarticular soft tissues were extracted while synovial joint fluid, cartilage, and bone were left intact, (3) periarticular soft tissues and synovial joint fluid was extracted while cartilage and bone remained intact, (4) periarticular soft tissues, synovial joint fluid, and the articular cartilage were completely removed leaving the subchondral bone to be the primary shock absorber.	When only subchondral bone was left intact: peak forces experienced at the joint ↓

Bone	Chu et al. (1986)	Interventional study	19g impact shocks administered with the knee suspended in full extension	6 cadaveric lower extremities with all soft tissue removed except at the knee	Impulse applied longitudinally with a magnitude of 19g and pulse rate of 0.5 Hz	Impulse wave transmissibility through (1) excised knee with soft tissues and bone intact, (2) menisci and articular cartilage removed, (3) total prosthetic cadaveric joint with the absence of subchondral bone and soft tissue	Shock transmission increased, i.e., shock attenuation decreased successively from condition 1-3. However, the largest decrease was found in condition 3 where subchondral bone in addition to soft tissue was absent. Thus, with the absence of bone as a potential shock absorber: Impact shock attenuation at the knee ↓
	Paul et al. (1978)	Observational	External impacts applied to deceased and alive rabbit tibia	Alive and deceased rabbit tibia	Externally applied impacts shocks at 3Hz, between 60-360 Hz and above 500 Hz	Shock transmission measured using force transducers implanted into the proximal tibia	As impact frequency to the bone increased: Shock attenuation ↑ Indicating bone attenuates higher-frequency shock better

Soft tissue (fat and lean mass)	Pain & Challis (2001)	Simulation study	Two-dimensional modeling of the heel pad and tibia soft tissues simulating a pendulum impact	N/A	Simulated pendulum impacts to the heel pad and tibia modeled as rigid bodies or wobbling masses	Energy dissipation at impact	As wobbling mass was introduced into the model: Energy dissipation at both the tibia and heel pad ↑
	Pain & Challis (2002)	Interventional study	Repetitive striking task with the right forearm while (1) forearm muscles were completely relaxed, (2) forearm muscles stiffened, (3) forearm muscles are stiff as possible	1 male participant	Muscle tension pre-impact and soft tissue deformation post impact as assessed via the deformation of an array of retroreflective markers	Time to impact peak, energy transmission along the length of the forearm	As muscle tension increased: Time to peak impact ↓ Soft tissue deformation accounted for ~70% of the total energy dissipated by the forearm during impact
	Boyer & Nigg (2006)	Observational study	Overground running at 3.0 m/s	5 male recreational athletes	Accelerometer location at the distal and proximal rectus femoris muscles	Time to peak acceleration, power spectrum and attenuation of power for lower (10-25 Hz) and middle-frequency (25-50 Hz) ranges	For the distal than proximal location, higher frequency power ↑; Attenuation between distal and proximal rectus femoris for frequencies >25 Hz ↑

Negative mechanical work performed by soft tissues	Zelik & Kuo (2010)	Randomized experimental trial	Treadmill walking at 0.7-2.0 m/s	10 healthy adults	Walking velocity ranging from 0.7-2.0 m/s	Negative work performed by soft tissue	Soft tissues performed negative work across all walking speeds. As walking speed increased: Negative work. i.e., energy absorption by soft tissues ↑
	Riddick & Kuo, (2016)	Randomized experimental trial	Treadmill running at 2.2-5 m/s	8 healthy adults	Running velocity ranging from 2.2-5 m/s	Ngeative work performed by soft tissues	Soft tissues performed negative work across all running speeds. As running speed increased: Negative work. i.e., energy absorption by soft tissues ↑
	van der Zee & Kuo, 2021	Randomized experimental trial	Treadmill walking at different combinations of walking speed and step length	9 healthy adults	Different combinations of walking speed and step length	Negative work performed by soft tissues	As walking speed and step length increased: Negative work. i.e., energy absorption by soft tissues ↑

↑ increased, ↓ decreased, ≅ remained unchanged/was similar

10 Appendix 3

Modified Physical Activity Readiness Questionnaire & Health History

The American College of Sports Medicine Exercise Pre-participation Health Screening

Date: ________________________

Please answer the following questions to the best of your knowledge (circle YES or NO)

1.	YES	NO	Do you participate in regular exercise?
2.	YES	NO	Have you ever been diagnosed with a heart condition, cardiovascular disease (e.g. high blood pressure (hypertension), coronary heart disease, coronary artery disease, high cholesterol), metabolic disease (e.g. Type I or Type II diabetes), renal disease, or pulmonary disease (excluding regular asthma)?
	YES	NO	2a. If yes, have you experienced any of the following in the past 12 months: chest pain, shortness of breath, pain or numbness in the arms or legs, fainting, swelling of the ankles, abnormally quick heartbeat, cramping in the leg, or pain in the neck, back, or upper abdomen?
	YES	NO	2b. If you answered yes to the above question (2a), has a doctor cleared you to continue with regular exercise <u>including</u> vigorous intensity exercise?
3.	YES	NO	Has a doctor ever said that you should only perform medically supervised physical activity?
4.	YES	NO	In the past month, have you developed chest pain when you are not performing any physical activity?
	YES	NO	4a. If yes, has a doctor told you it is okay to continue exercise <u>including</u> vigorous intensity exercise? Please provide reasoning: ______________________ ____
5.	YES	NO	Do you ever feel faint, pass out, or have spells of severe dizziness, palpitations, or excessively rapid heartbeat at rest or during exercise?
	YES	NO	5a. If yes, has a doctor told you it is okay to continue exercise <u>including</u> vigorous intensity exercise? Please provide reasoning: _______________________ ____
6.	YES	NO	Have you ever been diagnosed with a neurological problem (e.g. stroke, MS, ALS, neuropathy, etc.)?

7. YES NO Have you injured bones, joints, or muscles in your back or legs in the last 3 months?

YES NO 7a. If yes, have you returned to your normal running training and mileage?

7b. If yes to question 7a above, how long have been performing your normal running training and mileage?: ______________________________________

8. YES NO Have you had any surgery to your legs or back in the past 10 years?

9. YES NO Is there any physical reason not mentioned here why you should not engage in physical activity including vigorous intensity exercise? If yes, please explain:

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11 Appendix 4

INDIANA UNIVERSITY INFORMED CONSENT STATEMENT FOR

The Effects of Age and Running Experience on Running Biomechanics

You are invited to participate in a research study that will investigate the relationship between your age, running experience and running biomechanics. We will determine this relationship by inquiring about your age and running experience via a short questionnaire after which you would be required to visit the Indiana University-Bloomington Biomechanics Laboratory only once for a gait analysis. The question we seek to answer is, does a runner's age and running experience affect their running movement patterns.

You were selected as a possible participant in this study because you are either 1) a runner who has participated in consistent running of between 3-13 miles per week for less than one year, 18-30 years old or 45-65 years old, and are a non-smoker (the "novice" group for the purposes of this study), or 2) a runner who has participated in consistent running of between 9-30 miles per week for more than one year, between 18-30 years old or 45-65 years old, and are a non-smoker (the "recreational" group for the purposes of this study). We ask that you read this form carefully and ask any questions you may have before agreeing to be in the study.

The study is being conducted by Dr. Allison Gruber and Gauri Desai M.Sc. of Indiana University-Bloomington.

STUDY PURPOSE
The purpose of this study is to determine the relationship between age, running experience and running mechanics. Specifically, we seek to understand how a runner's age and running experience affects their running mechanics that may in turn influence their ability to respond to running related physical stresses with implications for running related injury risk.

NUMBER OF PEOPLE TAKING PART IN THE STUDY:
If you agree to participate, you will be one of 100 participants who will be included in this research study.

PROCEDURES FOR THE STUDY:
If you agree to be in the study, you will be required to do the following things:

After reading this document and if you agree to participate in this study, we will ask you to complete several questionnaires so we can better understand the following:

1) Your readiness to participate in physical activity (modified PARQ)
2) Your current running training and history, physical activity history within the past year, running injury history (Running injury Hx questionnaire) and sedentary behavior (Sedentary Behavior Questionnaire).
3) Lifetime physical activity levels (Lifetime PA Questionnaire)

If the analysis of your running form was scheduled for another day, your appointment today will be complete after the above surveys and study information have been presented. You will be required to schedule your running form analysis to be completed within two weeks of enrolling in the study. All participants will complete the running form analysis at the IU Biomechanics lab. If the analysis of your running form (gait analysis) was scheduled for today, we will first measure your height and weight. You will then be escorted to a room across the hallway from the lab where the bioelectric impedance analysis (BIA) machine is located to complete body composition procedures. Once in the room, you will be instructed to remove your shoes and socks and step onto the BIA machine. The BIA is a machine that sends a small electrical current through your body, which you will not be able to feel, for a duration of ~5-10 seconds. For this measure, you will be asked to stand without moving on a scale while holding two metal handles in your hands. From this procedure, we will obtain measures of your body fat and lean muscle mass.

Upon completion of BIA procedures, you will be escorted back to the biomechanics lab. The analysis of your running form will involve you first changing into form fitting clothes and lab running shoes. Clothing provided by the laboratory can be borrowed if necessary. Next, you will kick a soccer ball three times to establish limb dominance. Then, you will warm up on the treadmill for 5 minutes or until you feel ready. During the warm-up, we will increase the treadmill speed until you feel like you are running at the speed you typically run. You will run at that speed for about a minute then we will increase and decrease the treadmill speed in small increments (no more than ±2 mph) until you identify the same speed as your preferred speed three times. We will then place reflective markers on your legs, hips, and the shoes to measure how your body moves during running. We will also attach accelerometer sensors to both your lower legs (above your ankles), your lower back, and your forehead to study the movements of these body segments. These sensors and the markers will be secured with wraps and athletic tape. The wraps and tape need to be tight, but not uncomfortable. We will then have you run through the data collection space in the Biomechanics Laboratory at the speed you typically run identified during warm up as well as a standardized speed (3.35 m/s = 7.5 mph = 8 min-mile pace). Eleven motion capture cameras will capture the position of the markers as you run through the collection space that is approximately 18 m (~60 ft) long. The motion capture cameras cannot record any information except for the position of the markers. Therefore, you may be rest assured you will not be personally identified by these recordings. The data collection space also has three force platforms in line with one another in the center of the space. These force platforms are like fancy scales and will measure the forces you apply to the ground when you run. You will be asked to complete at least ten trials each at your previously established speed and the standardized speed of 3.35 m/s. You may be asked to complete more than ten trials if any of the trials do not result in successful recording of the markers, accelerometer, or force platform data. The starting point of your run will be adjusted so that the cameras can clearly record your running form. We will continually monitor how you are feeling via the Borg Rating of Perceived Exertion (RPE) scale to ensure you are not fatigued. This scale ranges from 6 to 20 (6 = no exertion at all; 20 = maximal exertion). We will ask you to rate your effort every five minutes using this scale. After the running form analysis is complete, the markers and accelerometers will be removed, and any borrowed lab clothing and shoes will be returned.

Completing the consent procedures and the additional questionnaires will take approximately 20 min. The running form analysis will take approximately 120-180 min. Your total time in the lab today, if the enrollment procedures and running form analysis were completed on the same day, will be approximately 2.5-3 hrs.

You will not receive the results of any of these tests or procedures because they are being done only for research purposes.

RISKS OF TAKING PART IN THE STUDY:

While on the study, the risks are:

Muscle soreness or fatigue

Possible discomfort from the adhesive and wraps used to secure the reflective markers
Exposure to an imperceptible electrical signal from the BIA.
A possible loss of confidentiality.

The testing procedures to measure how you run are considered safe and are identical to procedures currently being used in the Biomechanics Laboratory. By performing the gait analysis, you are at no further risk of injury than by engaging in running training on your own.

During the testing protocol, there is minimal risk that you may experience fatigue or muscle soreness due to performing the running form analysis. Risk of fatigue and muscle soreness will be decreased by allowing you to rest as you many times and for as long as you need between the over-ground trials. The running speeds that you will run at are your preferred speed and a standardized speed that has been deemed comfortable by runners across running experience levels. Therefore, risk of muscle soreness or fatigue is further minimized if not eliminated.

You may develop skin irritation and/or a slight rash due to the application and removal of the reflective markers or accelerometers. Please let the researchers know if your skin is sensitive, or if you experience skin discomfort. If necessary, there are sterile alcohol pads available in the laboratory to provide temporary skin relief. Any potential discomfort would be no greater than wearing a Band-Aid.

While use of the BIA is low risk, there are risks for individuals who have pacemakers. <u>There is a risk that the electrical signal from the machine could interfere with the functioning of your pacemaker device.</u> **YOU SHOULD NOTIFY THE RESEARCHER IF YOU HAVE A PACEMAKER.**

There also may be other side effects that we cannot predict. There is also a small possibility of loss of confidentiality. The likelihood of loss of confidentiality is very low because only members of the research staff will be in the lab when you are being tested and see your data. We also have a system to prevent anyone outside of the research staff from identifying you by your data. All questionnaires will be coded by your subject number only. Every effort will be made to ensure confidentiality.

BENEFITS OF TAKING PART IN THE STUDY:
There are no direct benefits to taking part in this study. However, by taking part in this study, you will be contributing to the knowledge of the field to help us better understand the relationships between age, running experience and biomechanical factors. This information will contribute to the enhancement of running related injury risk reduction protocol design, training prescription, and footwear design and selection.

CONFIDENTIALITY
Efforts will be made to keep your personal information confidential. We cannot guarantee absolute confidentiality. Your personal information may be disclosed if required by law. Your identity will be held in confidence in reports in which the study may be published. Data will be recorded with numerical identifiers, will be stored securely in a laboratory computer, and will be made available only to persons involved with the study unless you specifically give permission in writing to do otherwise. No reference will be made in verbal or written reports which could link you to this study.

Organizations that may inspect and/or copy your research records for quality assurance and data analysis include groups such as the study investigator and his/her research associates, the Indiana University Institutional Review Board or its designees.

The following individuals and organizations may receive or use your identifiable information:

- The researchers and research staff conducting the study
- The Institutional Review Boards (IRB) or its designees that review this study
- Indiana University

WILL MY INFORMATION BE USED FOR RESEARCH IN THE FUTURE?
Information for this research may be used for future research studies or shared with other researchers for future research. If this happens, information which could identify you will be removed before any information is shared. Since identifying information will be removed, we will not ask for your additional consent.

DOES MY PARTICIPATION DISQUALIFY ME FROM TAKING PART IN OTHER STUDIES?
Taking part in this study will not prevent you from participating in other studies unless the study requires you to stop running or exercising or if it increases your risk of injury.

COSTS
There are no costs associated with participation in this study.

PAYMENT
All participants will receive a $30 gift card and an analysis of their running gait (~$100+ value).

COMPENSATION FOR INJURY
In the event of physical injury resulting from your participation in this research when you are in the lab, lab personnel will assist you in obtaining any necessary medical treatment which will be billed as part of your medical expenses. Any medical treatment and costs not covered by your health care insurer will be your responsibility. Also, it is your responsibility to determine the extent of your health care coverage. There is no program in place for other monetary compensation for such injuries. However, you are not giving up any legal rights or benefits to which you are otherwise entitled. If you are participating in research which is not conducted at a medical facility, you will be responsible for seeking medical care and for the expenses associated with any care received.

CONTACTS FOR QUESTIONS OR PROBLEMS
For questions about the study or a research-related injury, contact the researcher Gauri Desai at (513) 546-9612 or gaudesai@iu.edu.

In the event of an emergency, you may contact Allison Gruber at (978) 697-2631. If the emergency is medical in nature, call 911 and contact Allison Gruber after the emergency has passed.

For questions about your rights as a research participant or to discuss problems, complaints or concerns about a research study, or to obtain information, or offer input, contact the IU Human Subjects Office at (800) 696-2949 or irb@iu.edu.

VOLUNTARY NATURE OF STUDY
Taking part in this study is voluntary. You may choose not to take part or may leave the study at any time. Leaving the study will not result in any penalty or loss of benefits to which you are entitled. Your decision whether or not to participate in this study will not affect your current or future relations with the investigators or Indiana University. We ask that you contact a member of the research team if you wish to withdraw from the study.

SUBJECT'S CONSENT

In consideration of all of the above, I give my consent to participate in this research study.
I will be given a copy of this informed consent document to keep for my records. I agree to take part in this study.

Subject's Printed Name: ____

Subject's Signature: ____ **Date:** _______
(must be dated by the subject)

Printed Name of Person Obtaining Consent: ____

Signature of Person Obtaining Consent: ___________________________________ **Date:** _______

Participant Contact Information

Cell phone number: ______________________________________

Work or home phone number: ______________________________________

Living Address: ______________________________________
____
____

Permanent Address: ______________________________________
____
____

12 Appendix 5

INDIANA UNIVERSITY INFORMED CONSENT STATEMENT FOR

Body Composition and Bone Health among Runners

You are invited to participate in a research study that will investigate the relationship between bone health, body composition and running biomechanics. We will determine this relationship by conducting a body scan for which you would be required to visit the Indiana University-Bloomington Biomechanics Laboratory only once. The question we seek to answer is, does a runner's bone health and body composition affect their ability to respond to running related physiological stresses.

You were selected as a possible participant in this study because you were enrolled in IRB study number 10585, titled "The Effects of Aging and Running Experience on Running Biomechanics". You qualified for that study because you are either 1) a runner who has participated in consistent running of between 3-13 miles per week (or 5-20 km per week per week) for less than one year, aged between 18-30 years or 45-65 years old, and are a non-smoker (the "novice" group for the purposes of this study), or 2) a runner who has participated in consistent running of between 9-30 miles per week (or 15-50 km per week) for more than one year, aged between 18-30 years or 45-65 years old, and are a non-smoker (the "recreational" group for the purposes of this study). Additionally, you qualify for the current study because you did not develop an injury or experience pain since your previous gait analysis visit that caused you to significantly change or stop your running training. This current study is an additional, secondary analysis to serve the purposes described below. Please read this form carefully and ask any questions you may have before agreeing to be in the study.

The study is being conducted by Dr. Allison Gruber and Gauri Desai, M.Sc. of Indiana University-Bloomington.

STUDY PURPOSE

This study accompanies another study (IRB study protocol #10585) you are or were enrolled in during which we measured your running biomechanics. The purpose of this additional study is to determine the relationship between bone health, body composition and running movement patterns. Specifically, we seek to understand how a runner's bone mineral density, fat mass and lean muscle mass affects their ability to respond to running related physiological stress with implications for running related injury risk. Your data from the previous study will be analyzed with the data from this current study to answer this question.

NUMBER OF PEOPLE TAKING PART IN THE STUDY:

If you agree to participate, you will be one of 80 participants who will be included in this research study.

PROCEDURES FOR THE STUDY:

If you agree to participate, you will do the following:

During your visit for this study, you will fill out brief questionnaires about your running training, running injury occurrence, and your non-running physical activity levels including sedentary behaviors for the past two weeks and any changes that may have occurred since your participation in study protocol #10585

- “The effects of aging and running experience on running biomechanics”. Your height and weight will then be recorded.

Next, you will be escorted to a room across the hallway from the lab where the bioelectric impedance analysis (BIA) and dual x-ray absorptiometry (iDXA) machines are located to complete the body composition and bone scan procedures. Once in the room, you will be instructed to remove your shoes and socks and step onto the BIA machine. The BIA is a machine that sends a small electrical current through your body, which you will not be able to feel, for a duration of ~5-10 seconds. For this measure, you will be asked to stand without moving on a scale while holding two metal handles in your hands. From this procedure, we will obtain measures of your body fat and lean muscle mass. You will only be included in further study procedures if your measured body composition is similar compared to that measured during your participation in study protocol #10585, if your overall activity levels (as reported in the questionnaires) are similar to those you reported during your participation in study protocol #10585, and you have not developed a running injury that caused you to significantly alter or stop your running training.

Women of childbearing potential will then be required to take a urine pregnancy test to ensure they are not pregnant at the time of the body scan. If you are found to be pregnant, your participation in the study will end. If you are determined to not be pregnant, you will undergo a body scan using the iDXA machine from which we will obtain measures of your bone mass in addition to fat mass and lean muscle mass in greater detail. Per the requirements of the iDXA test, you will then be required to remove any jewelry or clothing that might include metal in it. Sanitary scrubs will be made available to you if your clothing contains metal in the form of buttons, zips, suspenders, or bras with metal fastenings. You will then lie flat on your back on the iDXA machine platform with both arms by the side of your body. Elastic straps will be placed comfortably around your ankles and knees to prevent any movement during the scan as it may invalidate the scan results. The scan will then begin and last for approximately 7 minutes during which time the research personnel will leave the room but remain just outside in the hallway. Once the scan is complete, the research personnel will return to the room and you will be instructed to dismount from the machine.

Total time taken to complete these procedures will be approximately 30 minutes. This study will include only one visit if you choose to sign the informed consent form and complete study procedures on the same day. If you wish to complete these on different days, then you will need to make two visits to the laboratory.

You will not receive the results of any of these tests or procedures because they are being done only for research purposes.

RISKS OF TAKING PART IN THE STUDY:
While on the study, the risks are:

A possible loss of confidentiality.
Exposure to a small electrical signal from the body composition exam.
Exposure to a minimal and safe amount of radiation from the iDXA scan.
Discomfort answering questions regarding overall health and physical activity.

There is a small possibility of loss of confidentiality. The likelihood of loss of confidentiality is very low because only members of the research staff will be in the lab when you are being tested and see your data.

We also have a system to prevent anyone outside of the research staff from identifying you by your data. All data will be coded by your subject number only. Every effort will be made to ensure confidentiality.

While use of the BIA is low risk, there are risks for individuals who have pacemakers. There is a risk that the electrical signal from the machine could interfere with the functioning of your pacemaker device. **YOU SHOULD NOTIFY THE RESEARCHER IF YOU HAVE A PACEMAKER.**

The iDXA scan will expose you to low amounts of radiation. Every day, people are exposed to low levels of radiation that come from the natural environment and man-made radiation sources around them. This type of radiation is called "background radiation." The amount of radiation you will get from the iDXA scan in this study is approximately, or less than, 1 years' worth of background radiation. The probability of harm from participating in this study is low compared to other everyday risks. Certain diseases or conditions may affect your sensitivity to radiation. For more detailed information on the risks of radiation or if you wish to have a more detailed dose estimate, you should ask your personal physician before completing the DXA scan.
It is possible that you feel uncomfortable responding to the past and current running training, physical activity, sedentary behaviors and running injury history questionnaires. In such a case, you can choose to not answer the questions that make you feel uncomfortable.

BENEFITS OF TAKING PART IN THE STUDY:
There are no direct benefits to taking part in this study. However, by taking part in this study, you will be contributing to the knowledge of the field to help us better understand the relationships between bone health, body composition and the body's response to running related stress. This information will contribute to the enhancement of running related injury risk reduction protocol design, training prescription, and footwear design and selection.

CONFIDENTIALITY
Efforts will be made to keep your personal information confidential. We cannot guarantee absolute confidentiality. Your personal information may be disclosed if required by law. Your identity will be held in confidence in reports in which the study may be published. Data will be recorded with numerical identifiers, will be stored securely in a laboratory computer, and will be made available only to persons involved with the study unless you specifically give permission in writing to do otherwise. No reference will be made in verbal or written reports which could link you to this study. Your personal information may be shared outside the research study if required by law and/or to individuals or organizations that oversee the conduct of research studies

The following individuals and organizations may receive or use your identifiable information:

- The researchers and research staff conducting the study
- The Institutional Review Boards (IRB) or its designees that review this study
- Indiana University

WILL MY INFORMATION BE USED FOR RESEARCH IN THE FUTURE?
Information for this research may be used for future research studies or shared with other researchers for future research. If this happens, information which could identify you will be removed before any information is shared. Since identifying information will be removed, we will not ask for your additional consent.

DOES MY PARTICIPATION DISQUALIFY ME FROM TAKING PART IN OTHER STUDIES?
Taking part in this study will not prevent you from participating in other studies unless the study requires you to stop running or exercising or if it increases your risk of injury.

COSTS
There are no costs associated with participation in this study.

PAYMENT
There is no payment for participation in this current study.

COMPENSATION FOR INJURY
In the event of physical injury resulting from your participation in this research when you are in the lab, lab personnel will assist you in obtaining any necessary medical treatment which will be billed as part of your medical expenses. Any medical treatment and costs not covered by your health care insurer will be your responsibility. Also, it is your responsibility to determine the extent of your health care coverage. There is no program in place for other monetary compensation for such injuries. However, you are not giving up any legal rights or benefits to which you are otherwise entitled. If you are participating in research which is not conducted at a medical facility, you will be responsible for seeking medical care and for the expenses associated with any care received.

CONTACTS FOR QUESTIONS OR PROBLEMS
For questions about the study or a research-related injury, contact the researcher Gauri Desai at (513) 546-9612
or gaudesai@iu.edu.

In the event of an emergency, you may contact Allison Gruber at (978) 697-2631. If the emergency is medical in nature, call 911 and contact Allison Gruber after the emergency has passed.

For questions about your rights as a research participant or to discuss problems, complaints or concerns about a research study, or to obtain information, or offer input, contact the IU Human Subjects Office at (800) 696-2949 or irb@iu.edu.

VOLUNTARY NATURE OF STUDY
After reviewing this form and having your questions answered, you may decide to sign this form and participate in the study. Or, you may choose not to participate in the study. This decision is up to you. If you choose not to participate in this study or change your mind after signing this document, it will not affect your usual medical care or treatment or any relationship you may have with Indiana University and/or IU Health Bloomington Hospital.

If you change your mind and decide to leave the study in the future, the study team will help you withdraw from the study safely. If you decide to withdraw, your responses to all questionnaires will be discarded. Email thread exchanges between you and the researchers will also be deleted. The researchers will be available to answer any further questions you may have about your withdrawal. There are no risks associated with your withdrawal from this study.

SUBJECT'S CONSENT

In consideration of all of the above, I give my consent to participate in this research study.
I will be given a copy of this informed consent document to keep for my records. I agree to take part in this study.

Subject's Printed Name: ____

Subject's Signature: ____ **Date:** _______
(must be dated by the subject)

Printed Name of Person Obtaining Consent: ____________________________________

Signature of Person Obtaining Consent: _________________________________ **Date:** _______

Participant Contact Information

Cell phone number: ____________________________________

Work or home phone number: ____________________________________

Living Address: ____________________________________
____
____

Permanent Address: ____________________________________
____
____

13 Appendix 6

Survey will be administered electronically using Qualtrics (http://iu.qualtrics.com).

Running Training and Injury History Questionnaire

Instructions: In the following survey, you will be asked questions about your current running training, running training history, any non-running physical activity that you preform, about your sedentary behaviors, about your shoe and orthotic preferences, and about any orthopedic injuries that you have sustained. Please be as accurate and detailed as possible in your responses - the more accurate and detailed your responses, the better we will be able to identify the causes of running-related injuries.

Running Training History

1) What type of runner do you consider yourself?
Health/fitness Recreational Competitive Other

2) Why did you start running?
- □ To train for a race.
- □ For general health/fitness purposes.
- □ To be part of a group/the social aspect.
- □ Other. Please explain: _______________________

3) How long have you been a runner (or engaged in running activity consistently)?

Years	Months

4) How long have you been consistently running at your current weekly mileage?

Years	Months

5) What has been your average miles per week over that time?

6) Have you ever taken any long breaks or taken time off from running? YES NO
[If yes, proceed to the next question]
[if no, skip to the Current Training section]

7) How many long breaks or periods of time off from running have you taken since you started running at least 6 miles per week consistently? [drop down menu]
[the next two questions will be repeated for the number of breaks they indicate]

8) When did the [first/second/third/etc] time off from running begin?

9) How long was did you take time off from running the [first/second/third/etc] time?

Years: _________________ Months: _____________

10) Have you ever experienced periods of poor running performance that was not caused by an injury or illness?

Yes
No

Current Training – ***The following questions refer to the training you have done in last 3 months or the most recent change to your running habits within that time.***

1) What has your average weekly running mileage been for the last 3 months?

2) What is the typical distance you complete for each run (in miles)?

3) How many runs do you perform each week?

4) How many hours per week do you currently run?

5) How many months out of the year do you typically run?

6) What is your typical perceived exertion for typical run? (0 = no exertion at all; 10 = maximal exertion)

0 1 2 3 4 5 6 7 8 9 10
no exertion at all maximal exertion

7) What is your usual running pace or running speed for an 'easy run'?

Pace (min-mile)	Speed (mph)

8) What surfaces do you usually run on? For each surface, please tell us the time and/or the number of miles per week you run on this surface.

Surface	No. of runs per week	Time (in minutes)	Distance (in miles)
Pavement			
Dirt path/trail			
Track			
Grass			
Treadmill			
Other			

9) Do you typically complete a 'long' run each week? YES NO
[If yes, proceed to question 15]
[if no, skip to question 16]

10) How long is your 'long' run?
Miles: ________________
Minutes: ______________________

Special Running Training

1) Do you usually perform any special running training (such as sprinting, intervals, hills, tempo, etc.)?
□ Yes □ No

[If yes, participant will be directed to the next question]
[If no, participant will be directed to Running Shoe section]

2) For each type of special running training that you usually perform, please tell us the total distance in miles and/or the total time (in minutes), the intensity, how often you perform these activities, and any other additional information you would like to share.

	Do you perform?	Estimated total time per bout	times per week	weeks per month	#months per year	Perceived Intensity 0-10 scale 10=hardest	Surface (e.g. asphalt, grass, etc)	Additional info to share
Fartlek's (typical interval type)	YES/NO							
Sprint/track-interval workout	YES/NO							
Tempo runs	YES/NO							
Hill repeats	YES/NO							
Stair repeats	YES/NO							
Other	YES/NO							

3) In the space provided below, please provide any additional information that you would like us to know about your current running training.

Running Shoes and Orthotics

1) When do you wear your running shoes (including barefoot style shoes)?
   - □ I wear my running shoes for running only
   - □ I use the same pair of shoes for running and non-running exercise, but for only exercise (no general purpose)
   - □ I wear my running shoes for general purposes in addition to exercise
   - □ I run barefoot primarily (I do not wear shoes when I run)
   - □ Other. Please explain:

2) Do you run in a single pair of shoes or alternate between shoes?
   - □ Single pair
   - □ I Switch between two or more pairs throughout the week
   - □ I switch between barefoot running and running with shoes (proceed to question 4).
   - □ I run barefoot for all of my runs, regardless of speed, distance, surface, etc. (proceed to question 7)
   - □ Other. Please explain:

3) Do you wear your running shoes for all other types of physical activity or exercise? Or do you have different shoes for running than you wear for other forms of physical activity or exercise?
   - □ I wear my running shoes for running only.
   - □ I use the same pair of shoes for all types of physical activity or exercise
   - □ Other. Please explain:

4) What type of running shoe(s) do you wear?

Brand	Model	Style	Male of	Size	Primary use (easy runs, long runs,

			Female sizing?		trail running, non-running exercise, etc.)

5) How often do you replace your running shoes? Please report in months and/or miles.

Months	Miles

6) Do you complete any barefoot runs?

□ Yes □ No

[If yes, participant will be directed to the next question]
[if no, participant will be directed to question 8]

7) Please complete the following information about your barefoot runs?

Times per week	Times per month	Miles	Speed or Pace	Type of surface

8) Do you wear orthotics or shoe inserts?

□ Yes □ No

[If yes, participant will be directed to the next question]
[if no, participant will be directed to the last question in this section]

9) Do you were your orthotics all the time or only during certain activities?

□ I wear my orthotics all the time.
□ I wear my orthotics when I run only.
□ I wear my orthotics when I perform any physical activity or exercise.
□ I wear my orthotics for exercise, if I remember.
□ I have orthotics, but I usually don't wear them.

10) Were your orthotics recommended or prescribed by a doctor?

□ Yes, my doctor prescribed (required) me to wear orthotics.
□ Yes, my doctor recommended or suggested that I wear orthotics but it was up to me.
□ No, I decided to wear orthotics on my own.

11) Are your orthotics custom made or store bought?

□ Custom made by my doctor.
□ Semi-custom orthotics determined by the Dr. Scholls FootMapping® System or similar
□ Off the shelf

12) What are your orthotics designed to do or treat?

□ I'm not sure
□ increase cushioning
□ alignment
□ Arch support
□ over pronation

- □ over supination
- □ rearfoot control
- □ other. Please describe:

13) What is the reason you purchased the shoes that you wear most often for running? *[free text]*

14) In the space provided below, please provide any additional information that you would like us to know about your running shoe preferences? *[free text]*

Other Activity

1) Do you regularly participate in any non-running exercise or physical activity at least 10 or more times in the past year, besides running?
   - □ Yes, other types of exercise in addition to running for all of the past year
   - □ Yes, other types of exercise in addition to running for only all or part of the last three months
   - □ Yes, I did do other activity but I have not done it within the past three months.
   - □ No, only running

   [If yes, proceed to question 2]
   [if no, skip to question 3 on page 7]

2)

In the past 12 months, have you participated in any of the following activities 10 or more times? If yes, please check the box next to each activity. Then, please tell us which months out of the year you participated in the activity, the average number of days per week you participated, and the average number of hours per week that you participated in the activity:

	Have you participated in these activities within the past 12 months?		Months of that you participate in activity												Ave days per week of activity	Ave hours per week of activity	Perceived intensity 0-10 scale 10=hardest
	Yes	No	Jan 1	Feb 2	Mar 3	Apr 4	May 5	Jun 6	Jul 7	Aug 8	Sep 9	Oct 10	Nov 11	Dec 12	(# of days/wk)	(# of hrs/wk)	
Aerobics																	
Band/Drill Team																	
Baseball																	
Basketball																	
Bicycling																	
Bowling																	
Cheerleading																	
Dance Class																	
Elliptical																	
Football																	
Garden/Yard Work																	
Gymnastics																	
High Intensity Interval Training																	
Hiking																	
Heavy Household																	
Ice Skating																	
Pilates or Barre																	
Roller Skating																	
Running for Exercise																	
Skateboarding																	
Snow Skiing																	
Soccer																	
Softball																	
Stair master or climbing machine																	
Street Hockey																	
Swimming (Laps)																	
Tennis																	
Volleyball																	
Walking for Exercise																	
Water Skiing																	
Weight Training																	
Wrestling (competitive)																	
Yoga																	
Other, field sports																	
Other, stationary aerobic machines																	
Other, aerobics classes																	
Other, not listed																	

3) In the space provided below, please provide any additional information that you would like us to know about your other usual forms of physical activities or exercise.

Running Injury History

1) Have you ever experienced persistent pain or injury to your hips, legs, ankles, knees, feet, toes, or back due to exercise and/or running? These can be either acute/happened all of a sudden or chronic injuries caused by running or other exercise or physical activity.

□ Yes *[If yes, proceed to question 2]* □ No *[if no, skip to question 3]*

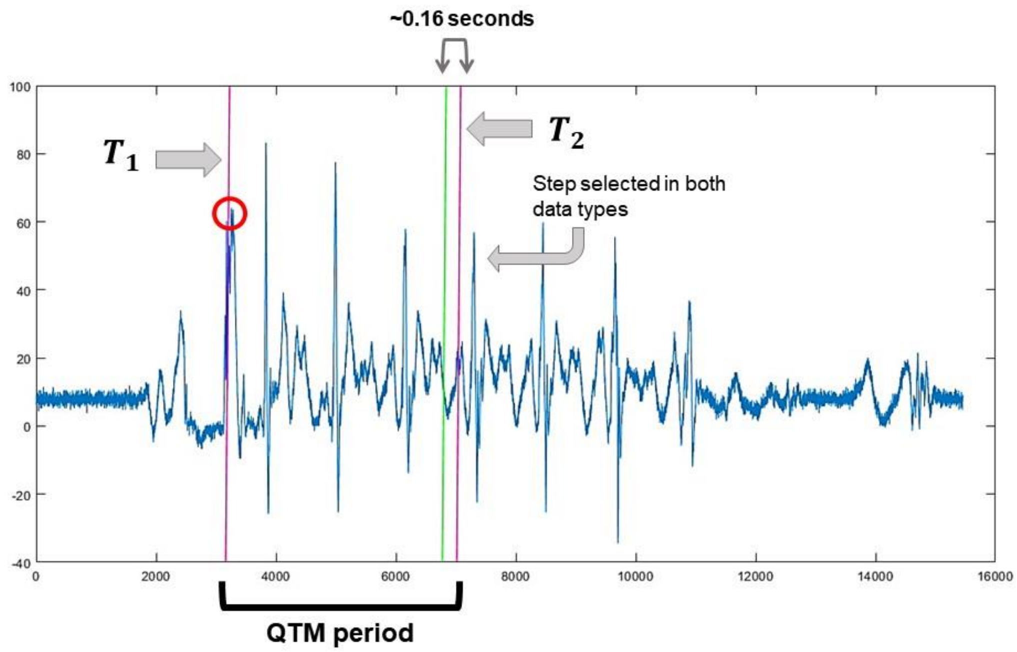
2) Please describe all of the persistent pain or injuries you have experienced due to running or other exercise by completing the table below, one row for each injury or location of persistent pain. **For the "Additional information to share with us" row, please tell us more about the injury if you wish. For example, how it happened, how long it affected you, if it bothered you only when you were running, etc.*

Injury number	**1**	**2**	**3**	**4**	**5**
Location (body part)					
Side of the body (R or L)					
Diagnosis or injury description					
How long ago? (months)					
Pain Level 0 (no pain) – 10 (horrible pain)					
Did it affect your running training?	□ reduce, □ stop, □ no change	□ reduce, □ stop, □ no change	□ reduce, □ stop, □ no change	□ reduce, □ stop, □ no change	□ reduce, □ stop, □ no change
Seek Medical treatment?	□ Yes □ No	□ Yes □ No	□ Yes □ No	□ Yes □ No	□ Yes □ No
Was it caused by running specifically?	□ Yes □ No	□ Yes □ No	□ Yes □ No	□ Yes □ No	□ Yes □ No
Additional info to share					

3) In the space provided below, please provide any additional information that you would like us to know about your running injury history?

14 Appendix 7

Analysis to identify the lag or lead in the manually flagged jumped landing in Qualisys Track Manager (QTM) relative to the jump-landing peak observed in the tibial acceleration signal to ensure the same step was selected in both the QTM and acceleration data when analyzing the kinematic and kinetic mechanisms of shock attenuation during running stance.



The figure displays the tibial acceleration signal for the right limb as an average across 100 trials. To select the same running step in both motion capture and acceleration data, it was necessary to align the two data types. To do so, each subject was instructed to perform a jump at the beginning of each trial. The landing of this jump was manually flagged in QTM (i.e. the motion capture system). The manual flag was created as soon as the researcher heard the impact associated with the jump landing during data collection. The jump landing was also detectable as the first peak >3g in magnitude in the tibial acceleration signal (red circle). The manual flag in QTM and the jump landing peak in the tibial acceleration signal were expected to occur in very close proximity of each other. Therefore, the time elapsed between the manual flag and the instant of foot strike on a force plate (i.e., instant when the ground reaction force >20 newtons; T_2) was calculated (QTM period) and applied to the tibial acceleration signal starting at the jump-landing acceleration peak (red circle, T_1). The acceleration before T_2 was discarded while that after the QTM period was entered into step detection algorithm to detect the first step immediately following T_2. This step coincided with the step registered on a force plate in the motion capture system that was then used to calculate the kinematic and kinetic variables associated with shock attenuation. Thus, the same step was identified for motion capture and IMU data ensuring both shock attenuation and the proposed mechanisms of attenuation were calculated using the same step.

It was possible, however, that the jump landing manually flagged in QTM either led or lagged the acceleration peak recorded at the tibia. If this lead or lag was too large, the wrong step could have been identified in the acceleration signal that did not coincide with the motion capture data. To ensure this error did not occur, the time lead/lag was identified using 100 trials. Lead/lag times were identified as the difference in Unix time stamps between the manual flag in QTM and T_1 after accounting for differences in the trial's start time between systems. The Unix time-stamp for the manual flag was obtained by adding its time in seconds to the Unix timestamp at the start of the trial while that for T_1 was identified as the unix time corresponding to the location of the jump-landing peak. This analysis revealed that on average, the manual flag always preceded T_1 by ~0.16 seconds. This means that if this lead time was corrected for, the QTM period would end slightly 0.16 seconds earlier than calculated; it would likely end at the solid green vertical line rather than T_2. However, as is evident, this small difference between the manual flag and the jump landing in the acceleration data was not large enough to include the impact shock peak occurring before the one currently being identified. Thus, the first step selected after the QTM period was the one common to both motion capture and acceleration data.

Curriculum Vitae

GAURI DESAI

EXPERIENCE

Aug 2023-Aug 2025 (expected)

University of Maryland, Neuromechanics Core

Postdoctoral Research Fellow

- Research focus: Effects of a traditional versus osseointegrated lower extremity prosthetic on gait function and oxygen utilization in patients with transtibial and transfemoral limb loss.

July 2022-Present

The Micheli Center for Sports Injury Prevention, Boston Children's Hospital, MA

Research Collaborator

- Collaborating with post-doctoral fellows and medical professionals to assess ground reaction forces during running, dual x-ray absorptiometry bone health metrics, and bone stress injuries among adolescent runners.

Jan 2021-May 2021

Women in Sports Tech Inc. (WiST)

Program Coordinator - WiST Next Gen Ambassador Program

- Mentoring and facilitating weekly webinars for high-school girls enrolled in the WiST Next Gen Ambassador Program with the aim of building a community for young women in the sports and tech world.
- We provide valuable resources related to sports and tech to these young minds to aid skill development including networking, communication, and leadership.

Aug 2018-May 2023

Indiana University-Bloomington, Bloomington, IN

Teaching Assistant

- Subject area: Biomechanics (Undergraduate)
- Topics covered: Linear and angular kinetic and kinematic concepts and clinical biomechanics - specifically, applying the concepts of torque and equilibrium to modify reconstructed anterior cruciate ligament rehabilitation protocols.
- Duties: creating coursework and teaching lab-based classes where students learn human movement-related biomechanical concepts through application.

May 2021-Oct 2021

IMeasureU, Bloomington, IN

Research Intern

- I aimed to answer the question "what is the relationship of age and running experience to running biomechanics, specifically impact shock and shock attenuation?".
- Running biomechanics data were collected using inertial measurement units, 3D motion capture, and force platforms.
- Kinetic and kinematic data were obtained via musculoskeletal modeling in Visual 3D software.
- Acceleration data obtained from the sensors is being processed and analyzed using MATLAB while statistical analyses are performed in R.

Sep 2017-Aug 2018

H.N. Reliance Hospital Sports Medicine Center, Mumbai, Maharashtra

Sport Science Intern

- Routine Vo2max testing for soccer players and recreational runners.
- Administration of graded rehabilitation programs for a wide range of musculoskeletal injuries as part of the physical therapy unit.
- Development of an 18-month graded ACL rehabilitation and Return to Sport protocol.

Apr 2018-May 2018

Varanashi Swimming Academy, Puttur, Karnataka

Sport Scientist and Assistant Coach

- Developed in-water and dryland testing protocols for performance testing among youth swimmers.
- Collected data using both protocols to study the effect of the camp on performance.
  in a competitive and non-competitive group and logged data for upcoming years on several parameters such as aerobic power and soundness of swimming technique.

Jun 2017-Jul 2017

Total Performance Swim Camps at Kenyon College, Gambier, OH

Camp Counselor

- Duties included administering daily swim practices and conducting daily performance/technique tests for adolescent competitive swimmers.

EDUCATION

August 2018-July 2023

Indiana University-Bloomington
Bloomington, IN, USA
Doctor of Philosophy (Ph.D.)

- **Major:** Human Performance
- **Track:** Biomechanics
- **Relevant Coursework:**
- Mechanical Analysis of Human Performance (Kinesiology- 550)
- Quantitative Mechanical Analysis of Human Movement (Kinesiology -631)
- Journal Club, Human Performance (Kinesiology-630)
- Physiological Basis of Human Performance (Kinesiology-535)
  Scientific Writing and Proposal Development (Epidemiology-790)
- Experimental Analysis and Design (Biostatistics-601)
- Semiparametric Regression with R (Biostatistics-650)
- Human Biological Signal Processing (Kinesiology-550)
  Advanced Theories of High-Level Performance (Kinesiology-533)

- **Extracurricular Activities:** IU Run Club Cross Country Team (August 2018 - December 2018)

Sep 2017

University of Exeter
Exeter, UK
Master of Science (M.Sc.) Sport and Health Sciences

- **Grade:** Merit
- **Relevant Coursework:** Statistical and Research Methods in Sport Science, Biomechanics of Human Movement, Exercise Physiology in Human Performance

May 2016

Sophia College for Women Mumbai, Maharashtra, India
Bachelor of Arts (B.A.) Psychology

- **GPA:** 6.74/7

- **Relevant Coursework:** Cognitive Psychology, Developmental Psychology, Abnormal Psychology, Social Psychology
- **Extracurricular Activities:** Sports Event Coordinator at the annual college festival

RESEARCH EXPERIENCE & PRESENTATIONS

May 2021-Jun 2023

Indiana University-Bloomington & IMeasureU

Investigator

- Investigating the active shock attenuation mechanisms during human running. We seek to assess whether joint kinematics and kinetics at the knee and ankle are associated with the degree of shock frequencies attenuated between the lower leg and tibia during overground running (Abstract presented at ACSM, Denver, USA, June 2023 for a thematic poster presentation).

May 2021-Jun 2023

Indiana University-Bloomington & IMeasureU

Investigator

- Investigating the passive shock attenuation mechanisms during human running. We seek to assess whether dual x-ray absorptiometry-derived body composition variables are associated with the degree of shock frequencies attenuated between the lower leg and tibia during overground running (Abstract presented at ACSM, Denver, USA, June 2023 for a thematic poster presentation).

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May 2021-Jun 2023

Indiana University-Bloomington & IMeasureU

Investigator

- Investigating the effects of age and running experience on running biomechanics, specifically impact shock and the ability to attenuate higher and lower frequency shock during running. We seek to assess whether age and running experience-related alterations to gait mechanics and body composition parameters affect one's ability to absorb impact shock of varying frequencies during running.

Feb 2022-Aug 2023 (expected)

Indiana University-Bloomington

Investigator

- Conducting a systematic review to be published in a peer reviewed journal focusing on bringing the scientific community up to date regarding the effects of running experience levels on running kinematics and kinetics. We hope to inform future research on the implications of running experience definitions in the interpretations of biomechanics research findings.

May 2020-Jun 2023 (expected)

Indiana University-Bloomington

Investigator

- Investigating the effects of moderate to vigorous non-running physical activity participation on the tendency to rely on hip, knee, and ankle musculature for force and power generation during overground running.

Sep 2021-Jan 2022

Indiana University-Bloomington

Investigator

- Investigated the combined effects of age and running experience on higher and lower frequency impact shock magnitudes. Neither age nor running experienced independently or combined were associated with higher and lower frequency impact shock (Abstract accepted to NACOB 2022, Ottawa, Canada).

Jan 2020-Aug 2022 (expected)

Indiana University-Bloomington

Investigator

- Investigating distal to proximal shifts in joint function of the lower extremity due to engagement in moderate to vigorous non-running physical activity among recreational runners. (Abstract presented at the ACSM Annual Conference, 2022, San Diego, California as a rapid-fire Oral Presentation).

Jan 2019-Dec 2021

Indiana University-Bloomington

Investigator

- Investigated Coordination Variability in prospectively injured and uninjured runners quantified using a modified vector coding method. Identified significant between- group differences in coordinative variability stimulating further research into a specific limb as a potential running related injury mechanism. (Poster presentation at ISB/ASB Conference, Calgary, May 2019 and Journal article published in Journal of Biomechanics, December 2021).

Sept 2018-Aug 2020

Indiana University-Bloomington

Investigator

- Investigated segment Coordination and Variability in prospective injured and uninjured runners quantified using a modified vector coding method. Identified coordinative variability and specific coordination patterns as potential running related injury mechanisms. (Abstract published in MSSE and Journal article published in Journal of Sports Sciences, August 2020)

Aug 2018-Feb 2019	**Indiana University-Bloomington** *Investigator* • Investigated lower limb joint coupling in prospectively injured and uninjured runners quantified using dynamical versus discrete measures of running gait. Identified support for the use of dynamical rather than discrete measures. (Presented at Mid-South Biomechanics Conference, Memphis, February 2019).
Dec 2017-Aug 2018	**University of Exeter, UK** *Investigator* • Investigated the effects of a Plyometric Jump Training program intervention on bone mineral density and content in adolescents engaged in high impact versus low impact sports (Soccer, Cycling, Swimming). Identified significant differences post the intervention among swimmers at the hip and ankle.

PUBLICATIONS

- **Desai, Gauri A**., and Allison H. Gruber. "Bilateral differences in coordination variability among injured and uninjured runners: A prospective study." *Journal of Biomechanics* 132 (2022): 110938.
- **Desai, Gauri A.**, and Allison H. Gruber. "Segment coordination and variability among prospectively injured and uninjured runners." *Journal of Sports Sciences* 39.1 (2021): 38-47.
- **Desai, Gauri A**., and Allison H. Gruber. "Assessing Between-limb Differences In Prospectively Injured And Uninjured Runners Using Dynamical Measures Of Gait: 2632 Board# 93 May 29 9: 30 AM-11: 00 AM." Medicine & Science in Sports & Exercise 52.7S (2020): 718.
- **Desai, Gauri A.,** and Allison H. Gruber. "Weekly Moderate-to-vigorous Physical Activity and Biomechanical Plasticity Among Active Adults: A Prospective Study: 778." *Medicine & Science in Sports & Exercise* 54.9S (2022): 186.

AWARDS AND HONORS

May 2022	• Biomechanics Interest Group Student Research Award, American College of Sports Medicine
May 2022	• Cooper Fellowship, Indiana University-Bloomington
May 2021	• Women in Sports Tech Fellowship
May 2020	• Biomechanics Interest Group Student Research Award, American College of Sports Medicine

May 2020	• Marian Godeke Miller Fellowship, Indiana University-Bloomington
August 2019	• Cooper Fellowship, Indiana University-Bloomington
May 2010	• Government of India (Ministry of Youth Affairs and Sports) Scholarship

SPORTING EXPEREINCE

- Gold medal in the 4x100 medley relay for the Indian Women's swim team at the 2010 South Asian Games, Dhaka (Bangladesh)
- Represented India at the World School Games, 2010, Doha (Qatar)
- Won over 30 national medals at the Sub Junior, Junior and Senior Indian Nationals in swimming over 10 consecutive years
- Set 6 national records at various age group national championships
- Was declared Individual Champion at the Indian School Nationals in 2009

PEER REVIEW EXPERIENCE

- Judge and Interviewer: Women in Sports Tech Fellowship Applications, 2022
- Judge and Interviewer: Women in Sports Tech Fellowship Applications, 2023

ADDITIONAL SKILLS

Programming in MATLAB | Programming and Statistical Analysis in Rstudio | Musculoskeletal Modelling in Visual 3D software | Biological Signal Processing | Peer Mentoring | Motion Capture with Qualisys Track Manager | AMTI force plates | Excel | Dual Energy X-ray Absorptiometry | Scientific Writing and Communication | Project Ideation, Implementation and Management | Resiliency and Adaptability | Data analysis | Data visualization | Collaboration | Presentation

CERTIFICATIONS

- Supervised Machine Learning, Stanford University via Coursera (expected completion: August 2023)
- Science of Training Young Athletes, Part II, Coursera (May 2018)