TIBIAL COMPRESSION DURING ACTIVITIES OF DAILY LIVING IN YOUNG AND OLDER ADULTS

by

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DEDICATION

I would like to dedicate this thesis to my family, including my mother Shannon Walker, father Mark Walker, and my sister Brynn Walker, for their continuous guidance and support in my personal, academic, and professional endeavors. I would also like to dedicate this moment to my friend, Nic Hunt, for providing me with assistance and support throughout my time working on this thesis.

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ABSTRACT

Introduction: Stress fracture, particularly in the tibia, is a growing concern among older adults (greater than 65 years). Older adults may have inherent stress fracture risk from ageing-related changes to their musculoskeletal system. Specifically, older adults reduced ankle neuromuscular function may impair their ability to attenuate repetitive compressive forces experienced during daily locomotor tasks and increase the likelihood of suffering bone damage from decreased bone tissue elasticity. Yet, it is currently unknown if older adults exhibit greater tibial compression than their younger counterparts during locomotor tasks. **Purpose:** This study sought to quantify tibial compression for older and younger adults when walking and negotiating stairs and determine whether tibial compression is related to specific ankle biomechanics. **Methods:** 13 young (ages 18-25 years) and 13 older (greater than 65 years) adults had tibial compression, and ankle joint stiffness and biomechanics (peak joint angle and moment) quantified during an overground walk, and stair ascent and descent tasks. **Statistical Analysis:** Maximum and impulse of tibial compression, ankle joint stiffness, and peak of stance (0-100%) ankle flexion joint angle and moment were submitted to an independent t-test to assess the difference between young and older adults during each task. Then, correlation analysis determined the relation between tibial compression and ankle biomechanics for all participants, as well as the young and older adults. **Results:** Neither tibial compression (maximum and impulse), nor ankle biomechanics (joint stiffness, moment, and angle) differed between young and older adults (all: $p > 0.05$)

during the walk and stair ascent tasks. However, older adults exhibited ~15% smaller maximum tibial compression ($p = 0.004$) and ~10% peak ankle joint moment ($p = 0.037$) compared to young adults during the stair descent. Peak ankle flexion moment exhibited a moderate to strong relation with maximum tibial compression during each task (overground walk: $r = -0.69 \pm 0.26$; stair descent: $r = -0.48 \pm 0.32$; stair ascent: $r = -0.72 \pm 0.25$, respectively). Yet, older adults typically exhibited stronger relation between ankle biomechanics and tibial compression than their younger counterparts. Specifically, older adults exhibited a moderate linear relation between ankle joint stiffness and peak ankle joint moment with impulse of tibial compression during the walk ($r = 0.44 \pm 0.48$ and $r = -0.47 \pm 0.47$), and peak ankle joint moment with maximum tibial compression ($r = -0.48 \pm 0.47$) during stair descent task; whereas young adults exhibited a weak relation between the same ankle biomechanical and tibial compression measures ($r = 0.23$, -0.20, and -0.27, respectively) during the walk and stair descent tasks. **Conclusion:** Older adults exhibited a substantial, albeit statistically insignificant, 3% to 10% increase in impulse of tibial compression compared to young adults. The elevated compression impulse may place larger compressive forces on older adult's tibia, increasing likelihood of bone microdamage accumulation and stress fracture development. Yet, despite exhibiting a stronger relation between ankle biomechanics and tibial compression than their younger counterparts, there was not a specific alteration in older adults' ankle biomechanics that may predict the substantial change in their tibial compression.

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ADLs	Activities of Daily Living
vGRF(s)	Vertical Ground Reaction Forces
GRFs	Ground Reaction Forces
BMI	Body Mass Index
CM	Center of Mass
3D	Three-dimensional
AJC	Ankle Joint Center
KJC	Knee Joint Center
AJRF	Ankle Joint Reaction Force
SAM	Sagittal Ankle Moment

CHAPTER ONE: INTRODUCTION

Overuse musculoskeletal injury, such as tibial stress fracture, is a growing concern among ageing individuals in the United States. Older adults (over 65 years of age) may have inherent stress fracture risk, as age is purportedly a factor in fracture development and may be attributed to differences in cortical bone composition, fatigue life, and accumulation of microdamage compared with young adults (between 18 and 25 years of age).[24] The fatigue life of cortical bone is reportedly four times longer in young adults, and may stem from age-related decreases in bone tissue elasticity that increase the likelihood an older adult suffers the microdamage that characterize stress fracture.[26,27] In fact, compressive stress placed on older adults generated approximately 115% more cortical bone microdamage than young adults.[8,9] Upwards of 90% of stress fractures in older adults occur in the lower limb, and may result from repetitive compressive stress being placed on the lower limb musculoskeletal system during activities of daily living (ADLs), such as walking and stair negotiation.[25] Although older adults are expected to comprise nearly 19% of the US population, exceeding 72 million individuals by the year 2030, it is currently unknown whether older adults exhibit lower limb biomechanics that increase tibial compression and the likelihood of tibial stress fracture development during ADLs, which is a critical gap for an ageing population.[1]

Elevated stress fracture risk in older adults is reportedly attributed to ageing-related changes to the musculoskeletal system. In addition to the alterations in cortical bone, older adults exhibit a significant decrement in lower limb neuromuscular function.

In particular, older adults reportedly exhibit a 40% reduction in maximal lower limb musculature strength.[10] The age-related loss in lower limb strength may compromise an older adult's ability to attenuate impact forces (i.e., vertical ground reaction force (vGRF)) of walking, leading to larger, faster compressive stresses placed on the musculoskeletal system and an accompanying increase in stress fracture injury risk.[11-13,78] In fact, placing larger compressive stresses on a bone, such as the tibia, reduces the number of cycles (i.e., steps) before fatigue failure of the bone occurs.[34] Ageing-related changes to the musculoskeletal system may amplify the speed and magnitude of tibial compressive stresses, and subsequent injury risk. Both individuals with a history of tibial stress fracture and older adults exhibit significant increases in the impact peak (the first, rapid vGRF peak after heel strike) and vertical loading rate (vGRF transmission speed following heel strike) compared with healthy controls and young adults during ADLs.[14,17,18] In addition, older adults compromised neuromuscular function leads them to produce smaller ankle joint torques compared to healthy, young adults during ADLs.[15,16] This muscular decrement may further compromise their ability to safely attenuate compressive stresses during ADLs, and elevate their tibial stress fracture risk.[7]

Stair negotiation (both ascent and descent) elicits different physical demands, resulting in distinct ankle biomechanics, compared with level overground walking, particularly in older adults.[35-37] To ascend or descend stairs, individuals must either generate or dissipate sufficient work by ankle musculature to safely lift or lower the center of mass on the next step. During a stair ascent and descent, however, older adults exhibit an approximately 9% and 22% reduction in the peak ankle plantarflexion and dorsiflexion joint moment necessary to lift or lower the center of mass, respectively.[22,37]

In addition, older adults tend to have an earlier peak ankle joint moment and use up to 41% less ankle range of motion, particularly during the stair descent, compared with their younger counterparts.[22,36] These modifications in ankle biomechanics may be adopted by the older adults to alter joint stiffness, the ratio of ankle joint moment to angle between initial contact and peak ankle flexion angle, to safely perform the task, but may lead to increases in injury risk.[45] Alterations in ankle joint stiffness may influence tibial compression and elevate stress fracture risk. In addition to alterations in ankle biomechanics, stair ascent purportedly results in minimal increases in vGRFs, while stair descent results in large, significant increases in impact peak vGRFs compared to level walking.[35] Although older adults reportedly alter ankle biomechanics to safely negotiate stairs, it is unclear whether these alterations increase joint stiffness and tibial compression that may elevate their tibial stress fracture risk.

Tibial compression, a quantification of the compressive forces on the tibia, can be determined through the summation of muscle (internal) and ground reaction forces (GRFs) (external) placed on the lower limb musculoskeletal system. Specifically, quantifying tibial compression requires determination of the ankle flexion moment to predict the internal muscle force contributions, and projection of the ground reaction along the long axis of the tibia to estimate external force contributions.[19] Older adults exhibit changes in gait biomechanics, such as a substantial reduction in peak vGRF and maximum ankle joint torque during stair negotiation and level walking, that may theoretically decrease tibial compression compared with younger adults.[14,15] Yet, recent experimental evidence reported only a negligible to moderate correlation between vGRFs and tibial compression for individuals running and walking with and without body borne load.[19] It may be that specific ankle biomechanics, such as changes to joint stiffness or joint moments that stem from neuromuscular deficiencies commonly exhibited by older adults relate to hazardous increases in tibial compression and subsequent tibial stress fracture risk. However, it is currently unknown if tibial compression differs with age or whether tibial compression relates to specific ankle biomechanics (such as joint stiffness and associated components). This study seeks to fill that critical gap and quantify tibial compression for older and younger adults when walking and negotiating stairs and determine whether tibial compression is related to specific changes in ankle biomechanics.

Specific Aims

Specific Aim 1

To quantify tibial compression while young and older adults walk and negotiate stairs. Specifically, this study will quantify maximum and impulse of tibial compression during the stance phase as young (between 18 and 25 years) and older adults (over 65 years) walk at a self-selected speed overground and ascend and descend an 18.5 cm stair.

Hypothesis 1

Older adults will exhibit greater maximum and impulse of tibial compression compared with the young adults during overground walk, and stair ascent and descent tasks.

Significance 1

Determining whether tibial compression is greater for older adults during ADLs, particularly when walking overground and negotiating stairs, may provide valuable information regarding older adult's tibial stress fracture risk and can be used by clinicians to implement injury prevention strategies to reduce older adults tibial stress fracture risk.

Specific Aim 2

To examine ankle joint stiffness and associated biomechanics while young and older adults walk and negotiate stairs. Specifically, this study will quantify stance phase ankle joint stiffness, and peak of stance ankle flexion joint angle and moment as young (between 18 and 25 years) and older adults (over 65 years) walk at a self-selected speed overground and ascend and descend an 18.5 cm stair.

Hypothesis 2.1

Older adult participants will exhibit a significantly stiffer ankle during the overground walk, and stair ascent and descent tasks compared to young adults.

Hypothesis 2.2

Older adults will exhibit a significant reduction in peak ankle flexion joint angle and moment during the overground walk, and stair ascent and descent tasks compared to young adults.

Significance 2

Determining whether older adults modify ankle biomechanics, in particular ankle joint stiffness and associated measures during overground walking and stair negotiation may contribute valuable information to understand the underlying biomechanical factors that increase musculoskeletal injury risk with age. This information can be used by clinicians to reduce incidence of musculoskeletal injury for older adults during ADLs.

Specific Aim 3

To determine the relation between tibial compression and ankle biomechanics, and whether it differs between young and older adults during ADLs. Specifically, this study will assess the linear relation between tibial compression (both maximum and impulse) and ankle biomechanics (ankle joint stiffness, and ankle flexion angle and moment) while young (between 18 and 25 years) and older adults (over 65 years) walk at a self-selected speed overground, and ascend and descend an 18.5 cm stair.

Hypothesis 3.1

During both the overground walk and stair negotiation tasks, ankle joint stiffness and peak flexion angle and moment will exhibit a linear relation to both maximum and impulse of tibial compression.

Hypothesis 3.2

The strength of linear relations between ankle biomechanics (ankle joint stiffness, and peak of stance ankle joint moment and angle) and tibial compression metrics (maximum and impulse) will not differ between the young and older adults during the overground walk and stair negotiation tasks.

Significance 3

Determining whether ankle biomechanics metrics are related to increases in tibial compression would provide valuable knowledge for clinicians to target a reduction in tibial compression within injury prevention protocols in general, with a greater focus on explicit ankle biomechanics adaptations necessary for a meaningful decrease in tibial stress fracture risk specifically.

CHAPTER TWO: LITERATURE REVIEW

This following section seeks to detail ageing-related changes to lower limb biomechanics and tibial stress fracture risk by providing an overview of 1) older adults and ageing-related injury risk and costs, 2) musculoskeletal injury with a focus on musculoskeletal changes with age, and 3) lower limb biomechanics in general, with a particular emphasis on 4) ankle joint biomechanics and 5) tibial compression and stress fracture.

Older Adults

Older Adult Population

The older adult population has seen rapid growth in recent years, as modern advances in science and medicine lead to longer lifespans. During the 20th century, the older adult population in the United States went from 3.1 million to 35 million, a shift that is expected to continue.[48] An American Community Survey conducted in 2016 reported that adults over the age of 65 years old accounted for approximately 13% of the US population at 49 million, and experts predict that this number will reach as much as 19%, or more than 72 million individuals, by the year 2030.[1] One important side-effect of an ageing population is the accompanying financial burden on the healthcare system and individual alike. The average medical expenses for older adults are reportedly 2.6 times greater than that of the national average, accounting for one-third of medical spending in the US.[49] According to data from 2015, a large portion of these costs are covered through government subsidies and private insurance (about 65% and 13%, respectively), but the remaining costs of over 20% are paid for out-of-pocket by the individual for an average of over $5,700 per person annually, an increase of nearly 40% since 2005.[50] Compared with the general population ($4,342), older adult's out-of-pocket medical spending is 75% higher, accounting for 13% of their total spending compared to 8% for the general population.[50] Understanding both the physical and monetary costs associated with ageing provides necessary context to explore ageing-related changes in general, or the lower limb musculoskeletal system and associated biomechanics specifically.

Injury Risk and Cost

As the older adult population continues to grow, more than 25% of these older adults will experience a fall in a given year according to recent trends, with as much as 20% of those falls resulting in serious, even fatal, injuries.[1-3] This combination of a growing older adult population and a high rate of falls is expected to increase medical expenses, placing a substantial burden on the health care system and individual alike. In fact, approximately $50 billion in medical spending was attributed to falls among older adults in the United States in 2015, which represents an increase greater than twofold since the estimated $19.2 billion spent in 2000.[4,5] Considering the fact that falls have long been known as the leading cause of fractures among older adults, it is crucial to understand the age-related biomechanical factors that influence musculoskeletal injury in older adults.[6] Despite the fact that older adult fall-related fractures are most common at the hip, wrist and vertebrae, the age-related biomechanical changes to the musculoskeletal system indicate that older adults may be at an elevated risk for other musculoskeletal injuries, including tibial stress fracture.[7]

Musculoskeletal Injury

Overview

Overuse musculoskeletal injuries occur when the components of the musculoskeletal system, such as bone and skeletal muscle, fail to properly adapt to increasing mechanical stress caused by an increase in applied load, repetitive loading beyond functional capacity, sudden changes in the administration of loading, structural musculoskeletal changes, or a combination of these factors.[78-81] Despite fluctuating incidence rates based on level and intensity of physical activity and occupation, musculoskeletal injury can occur in any population owing to the broad nature of the injuries.[80-91] Thus, injuries to the musculoskeletal system are considered a major global public health problem, as they are a prominent source of death and disability.[92] Prominent risk factors associated with musculoskeletal injury include elevated GRFs, decrements in muscular strength, body mass index (BMI), the density of relevant minerals in bone, lifestyle, environment, sex, age, ethnicity, history of traumatic injury (such as stress fracture), and musculoskeletal changes in the lower limbs (capacity and structural).[81,82,87,93-97] Primary categories of musculoskeletal injury linked with overuse are stress fractures (particularly in the lower limb), musculoskeletal stress-related injuries, and muscular fatigue induced by overload of skeletal muscle.[82,83,93,98-100]

Ageing-Related Musculoskeletal Changes

One group at high risk of developing overuse musculoskeletal injuries as a result of changes to that system are older adults (65+ years old). Ageing has been shown to reduce muscle and bone mass, change the distribution minerals (shape) of bones, and alter the movement capabilities of the musculoskeletal system, including coordination decline, strength and speed decrements in muscle forces, and a reduced capacity to safely attenuate loads.[78] Additionally, ageing-related changes reduce the protection for internal organs through a combination of increased bone fracture risk and a decreased capability to properly heal.[78,101,103] Musculoskeletal changes to cortical bone strength and composition, as well as decrements in skeletal muscle capacity may be the primary sources for overuse musculoskeletal injuries during ADLs in older adults.

During the ageing process, there is a reduction in bone mass and mineral content, which may cause changes to the fatigue life and injury risk of cortical bone.[78] Healthy bone tissue maintains strength and volume through a balanced remodeling process, which is negatively affected during ageing, resulting in an inability to maintain consistent remodeling capabilities.[101] As fatigue life is reportedly four times longer in young adults when compared with older adults, these morphological and remodeling changes may account for the approximately 115% more cortical bone microdamage reported in older adults, increasing the likelihood of stress fracture.[8,9,26,27] Additionally, compressive forces placed on the bones of older adults reduce the number of cycles (i.e., steps) before this fatigue failure occurs.[34] Older adults also exhibit a significant decrement in neuromuscular function, with empirical evidence demonstrating a 40% reduction in maximal lower limb musculature strength.[10] Thus, the ageing process is thought to result in sarcopenia, or the ageing-related loss of both muscle mass and function.[78,103] Ageing-related changes to the neuromuscular system include both muscle composition and functionality differences.[78] Reportedly, there are ageing-related changes to muscular size and fiber type, ability to properly contract, and endurance, each of which contribute to the performance decrements observed in ageing skeletal muscle.[78,102]

Lower Limb

Ageing-Related Biomechanics

Musculoskeletal changes that occur with ageing not only produce structural and compositional differences, but also alter lower limb gait biomechanics during ADLs. During locomotion across different ADLs, older adults exhibit significant increases in the impact peak and vertical loading rate when compared against young adults, which may account for the reported decrease in their ability to safely attenuate impact forces resulting in larger and faster compressive stresses being transmitted to the lower limb musculoskeletal system.[11-14,17,18,78] Despite this, empirical evidence also shows that older adults exhibit significantly lower peak vGRFs and maximum ankle joint torques during both stair negotiation and level overground walking, highlighting the substantial gait strategy differences adopted by older adults when compared with their younger counterparts.[14,15] Modifications at the ankle joint reflect these biomechanical changes as a result of musculoskeletal changes particularly well, as older adults have reduced range of motion while also being more plantarflexed and generating 30% and 23% less work and angular impulse when compared with their younger counterparts, thus producing smaller joint torques overall.[15,16,105,119,120] Understanding the wider implications of these biomechanical changes requires analysis of these changes with regards to two common ADLs: walking and stair negotiation.

ADLs

Older adults tend to adopt slower walking speeds as a result of their decreases in stride lengths coupled with increases in both overall stance time and double-support stance phase when compared with younger adults.[104,111] Additionally, older adults also exhibit significant increases in gait variability during walking, including changes to cadence, stride length and stride width compared to their younger counterparts, all of which may increase the risk of frailty, falls, and musculoskeletal injury.[112] Despite these gait changes during walking, older adults are prone to operating approximately 22% closer to their individual relative lower limb muscular strength, an indication of an increased mobility cost that may prove unsustainable for long periods of overground walking.[106,107] Older adults also have between a 15% and 25% increase in the energy cost regardless of walking speed compared to younger adults, reflecting the U-shaped relationship that exists between speed and energy cost.[108-110] Reduced walking speeds are known to produce lower GRFs across all populations, meaning that older adults exhibit significantly smaller vGRFs while also exhibiting lower horizontal GRFs during the propulsion phase of gait owing to their slower reported walking speeds.[113,114] Muscular strength disparities may also influence both walk speed and GRFs, as lower strength values in older adults are linked with slower walk speeds and smaller vGRF values, particularly during the weight acceptance phase of stance.[115] Ageing-related changes to overground walking gait and GRFs reflect the mobility limitations, disabilities, and loss of independence that older adults are predisposed to and highlight the compensatory gait strategies adopted by older adults that alters lower limb biomechanics and loading of the lower limb joints.[116-118]

Stair negotiation produces different physical demands during both stair ascent and stair descent, resulting in lower limb biomechanics that are distinct from those observed during overground walking, particularly at the ankle joint.[35-37] To safely ascend or descend stairs it is necessary to either generate or dissipate sufficient work within the associated musculature at the ankle joint to move the center of mass (CM) between steps. However, older adults tend to exhibit a 9% and 22% reduction in the plantarflexion and dorsiflexion joint moment, respectively, which may diminish their ability to safely lift and lower their CM between the stairs during stair negotiation tasks.[22,37] These muscular deficiencies and ankle biomechanics changes lead older adults to produce peak ankle joint moments earlier during stance with up to a 41% reduction in the range of motion at the ankle compared with younger adults, which is particularly evident during stair descent tasks.[22,36] Additionally, empirical evidence shows that stair descent results in significant increases in impact peak vGRFs, compared with the both the minimal increases in stair ascent values and level walking.[35] This combination of ankle biomechanics and GRF changes influence the tibial compression experienced by older adults, and may place them at greater risk for stress fracture development.

Ankle Biomechanics

Ankle Joint Biomechanics

Ankle biomechanics, and more specifically ankle joint moment and angle, and ankle joint stiffness, are each critical to understanding and assessing lower limb injury risk, particularly with regards to tibial compression and tibial stress fracture. Quantification of tibial compression includes an internal component, determined using sagittal ankle joint moment, providing justification for the inherent relationship between ankle biomechanics and tibial compression and stress fracture risk.[19,26,51] On average, maximum moments (torques) at the ankle are substantially larger than the other lower limb joints, owing largely to the comparatively greater flexion-extension cycle at the ankle during locomotion.[70] In fact, the ankle is typically flexed more than twice the magnitude of the knee, which is considered necessary to maintain posture and balance during movement.[21,70] These large changes in ankle angle values during stance demonstrate why greater displacements in CM are associated with higher ankle moments.[21] Due to reported muscle strength deficiencies, older adults cannot develop ankle moments at the same rate as younger adults, which is thought to be vital for balance recovery and increase injury risk.[22,75-77] These biomechanical deficiencies account for the smaller overall maximum ankle moments that occur at greater dorsiflexion angles in older adults, despite the reported 41% less ankle dorsiflexion when compared with younger adults.[22]

Ankle joint stiffness incorporates each of these ankle biomechanics, as it is a measure of the change in ankle joint flexion moment divided by the corresponding change in ankle joint flexion angle from heel strike to peak ankle flexion angle, with the magnitude determined as the slope of the moment-angle relationship over that period.[45] Stiffness value magnitudes at the ankle are subject to a number of factors, including posture (shown to increase during periods of dorsiflexion), the forces applied to the ankle, the magnitude of the muscle activation from associated musculature, movement velocity, and the activity in question.[21,69,71-74] Empirical evidence shows a substantial variation in ankle joint stiffness measures during different activities of daily living, including walking and stair negotiation.[23,71] This is particularly relevant for older adults, as these individuals develop different torque patterns that account for differences in ankle stiffness measures, which provides evidence of altered strategies used by older adults during ADLs.[22] Understanding these changes in ankle biomechanics and the influence this has on injury risk is essential for reducing musculoskeletal injury risk in older adults, including risk of tibial stress fracture.

Tibial Compression

Formulation

Tibial compression metrics can negatively impact the lower limb musculature through a combination of bone loading and muscle force generation. Quantification of tibial compression requires the summation of both an external (ground reaction) and internal (muscle) force, which combine to compress the tibia.[19,26,51] Evidence shows that increases in GRFs, in particular vGRFs, may be accompanied with greater force transmission to lower limb musculoskeletal system, which would increase tibial compression.[20] The forces experienced by internal structures, such as the tibia, are influenced by GRFs, but are subject to other factors.[26,52-56] Despite increasing concurrently, recent evidence suggests that increases in tibial compression are not strongly correlated with increases in GRFs, indicating that the internal muscle component is the primary influence.[19,51] As such, GRFs may not provide an accurate indicator of the repetitive loads indicative of tibial stress fracture as a result of tibial compression.[19,51] Empirical evidence supports this notion, as the loads experienced by the tibia are typically much greater than measured GRF values by as much as 3 to 5 times.[26,52,53,55,56] Additionally, tibial bone loads can increase without a concurrent increase in GRFs, suggesting that other factors have a greater influence.[57]

Tibial Compression & Stress Fracture

Repetitive loading of the tibia may result in tibial stress fracture, a common overuse injury associated with compression of the tibia, brought about by the accumulation of microdamage.[58,59,62,63] Increases in the tibial compression forces experienced by the lower limb musculoskeletal system are believed to initiate and even accelerate this microdamage and remodeling of the tibia that are critical components of stress fracture development.[63] Several factors are linked with the tibial bone stress fracture and injury risk, including the intensity of the bone loading force, the rate at which the remodeling of the osseous tissue occurs, bone composition, and age.[60] In fact, it is thought that microdamage density is primarily responsible for rising stress fracture injury risk in the elderly.[62]

Biomechanical factors associated with tibial compression also play a role in stress fracture development. Many of these factors associated with tibial stress fracture are modified with fatigue, which may limit the ability of the lower limb musculoskeletal system to safely attenuate impact forces, thereby increasing stress fracture injury risk.[61] Although these factors are not strongly correlated, increases in the vGRFs experienced by the lower limb musculoskeletal system increase both the magnitude and the speed at which forces are transmitted to the lower limb and tibia specifically, which may be an indication of heightened stress fracture injury risk.[64] Increases in the rate at which forces are applied to the tibia have previously been linked with greater microdamage, as strain rate increases decrease the fatigue life of bone, accelerating microdamage accumulation.[63] Loading rate and impact peak increases have also been linked with elevated tibial stress fracture injury risk, as both lead to increases in the magnitude and rate of tibial compression.[63] Additionally, as empirical evidence shows that the internal structures such as muscles and bones experience substantially greater tibial compression forces and tibial compression impulse during walking tasks when compared with vGRFs and vGRF impulse, there is a discrepancy that can be accounted for by the muscle forces required for locomotion.[53] Previous work has highlighted an increase in peak ankle joint moment, which is necessary to attenuate impact forces and propel the body forward during locomotion.[66-68] As the tibial compression calculation includes sagittal ankle joint moment, there may be an increase in gastrocnemius force and tibial compression, elevating tibial stress fracture risk.[46,65]

Summary

The older adult population is growing in size, and this trajectory is projected to continue for several decades to come. Accompanying this rise are increased medical costs owing in part to a greater number of traumatic and overuse musculoskeletal injuries, which place a substantial burden in the individual and healthcare system alike. Older adults may be predisposed to some overuse musculoskeletal injuries due to neuromuscular and bone tissue changes, which show decrements in muscle strength and increased accumulation of microdamage indicative of stress fracture. Changes to the lower limb biomechanics, particularly at the ankle joint, lead older adults to adopt alternative gait strategies during walking and stair negotiation that may further increase injury risk. Increased focus on the changes to ankle joint moments, angles and stiffness that occur with ageing may increase understanding of stress fracture development, particularly in the tibia. Despite knowing many of these important changes to lower limb biomechanics, and the influence these can have on tibial compressive forces and subsequent tibial stress fracture risk, it is currently unknown how tibial compression differs with age, and whether a significant relationship exists between tibial compression and ankle biomechanics. This work seeks to further examine this interaction by quantifying tibial compression and determine its relation to specific changes in ankle biomechanics during ADLs (walking and stair negotiation).

CHAPTER THREE: MANUSCRIPT

Introduction

Overuse musculoskeletal injury, such as tibial stress fracture, is a growing concern among ageing individuals in the United States. Older adults (over 65 years of age) may have inherent stress fracture risk, as age is purportedly a factor in fracture development and may be attributed to differences in cortical bone composition, fatigue life, and accumulation of microdamage compared with young adults (between 18 and 25 years of age).[24] The fatigue life of cortical bone is reportedly four times longer in young adults, and may stem from age-related decreases in bone tissue elasticity that increase the likelihood an older adult suffers the microdamage that characterize stress fracture.[26,27] In fact, compressive stress placed on older adults generated approximately 115% more cortical bone microdamage than young adults.[8,9] Upwards of 90% of stress fractures in older adults occur in the lower limb, and may result from repetitive compressive stress being placed on the lower limb musculoskeletal system during activities of daily living (ADLs), such as walking and stair negotiation.[25] Although older adults are expected to comprise nearly 19% of the US population, exceeding 72 million individuals by the year 2030, it is currently unknown whether older adults exhibit lower limb biomechanics that increase tibial compression and the likelihood of tibial stress fracture development during ADLs, which is a critical gap for an ageing population.[1]

Elevated stress fracture risk in older adults is reportedly attributed to ageing-related changes to the musculoskeletal system. In addition to the alterations in cortical bone, older adults exhibit a significant decrement in lower limb neuromuscular function. In particular, older adults reportedly exhibit a 40% reduction in maximal lower limb musculature strength.[10] The age-related loss in lower limb strength may compromise an older adult's ability to attenuate impact forces (i.e., vertical ground reaction force (vGRF)) of walking, leading to larger, faster compressive stresses placed on the musculoskeletal system and an accompanying increase in stress fracture injury risk.[11-13,78] In fact, placing larger compressive stresses on a bone, such as the tibia, reduces the number of cycles (i.e., steps) before fatigue failure of the bone occurs.[34] Ageing-related changes to the musculoskeletal system may amplify the speed and magnitude of tibial compressive stresses, and subsequent injury risk. Both individuals with a history of tibial stress fracture and older adults exhibit significant increases in the impact peak (the first, rapid vGRF peak after heel strike) and vertical loading rate (vGRF transmission speed following heel strike) compared with healthy controls and young adults during ADLs.[14,17,18] In addition, older adults compromised neuromuscular function leads them to produce smaller ankle joint torques compared to healthy, young adults during ADLs.[15,16] This muscular decrement may further compromise their ability to safely attenuate compressive stresses during ADLs, and elevate their tibial stress fracture risk.[7]

Stair negotiation (both ascent and descent) elicits different physical demands, resulting in distinct ankle biomechanics, compared with level overground walking, particularly in older adults.[35-37] To ascend or descend stairs, individuals must either generate or dissipate sufficient work by ankle musculature to safely lift or lower the center of mass on the next step. During a stair ascent and descent, however, older adults exhibit an approximately 9% and 22% reduction in the peak ankle plantarflexion and dorsiflexion joint moment necessary to lift or lower the center of mass, respectively.[22,37] In addition, older adults tend to have an earlier peak ankle joint moment and use up to 41% less ankle range of motion, particularly during the stair descent, compared with their younger counterparts.[22,36] These modifications in ankle biomechanics may be adopted by the older adults to alter joint stiffness, the ratio of ankle joint moment to angle between initial contact and peak ankle flexion angle, to safely perform the task, but may lead to increases in injury risk.[45] Alterations in ankle joint stiffness may influence tibial compression and elevate stress fracture risk. In addition to alterations in ankle biomechanics, stair ascent purportedly results in minimal increases in vGRFs, while stair descent results in large, significant increases in impact peak vGRFs compared to level walking.[35] Although older adults reportedly alter ankle biomechanics to safely negotiate stairs, it is unclear whether these alterations increase joint stiffness and tibial compression that may elevate their tibial stress fracture risk.

Tibial compression, a quantification of the compressive forces on the tibia, can be determined through the summation of muscle (internal) and ground reaction forces (GRFs) (external) placed on the lower limb musculoskeletal system. Specifically, quantifying tibial compression requires determination of the ankle flexion moment to predict the internal muscle force contributions, and projection of the ground reaction along the long axis of the tibia to estimate external force contributions.[19] Older adults exhibit changes in gait biomechanics, such as a substantial reduction in peak vGRF and maximum ankle joint torque during stair negotiation and level walking, that may theoretically decrease tibial compression compared with younger adults.[14,15] Yet, recent experimental evidence reported only a negligible to moderate correlation between vGRFs and tibial compression for individuals running and walking with and without body borne load.[19] It may be that specific ankle biomechanics, such as changes to joint stiffness or joint moments that stem from neuromuscular deficiencies commonly exhibited by older adults relate to hazardous increases in tibial compression and subsequent tibial stress fracture risk. However, it is currently unknown if tibial compression differs with age or whether tibial compression relates to specific ankle biomechanics (such as joint stiffness and associated components). This study seeks to fill that critical gap and quantify tibial compression for older and younger adults when walking and negotiating stairs and determine whether tibial compression is related to specific changes in ankle biomechanics. We hypothesize that older adults will exhibit significantly greater tibial compression metrics, and a significant reduction in peak ankle flexion angle and moment that increases ankle joint stiffness compared with their younger counterparts, and these tibial compression metrics will have a strong, linear relationship with ankle joint biomechanics that does not differ based on age grouping.

Methods

Participants

Two cohorts of 13 individuals were recruited. Cohort one consisted of young, healthy adults (between 18 and 25 years of age) who self-report no history of musculoskeletal injury or disease. Cohort two consisted of older adults (over 65 years of age) who self-reported at least one accidental fall within the past 12 months. Any potential participant was excluded if they self-reported: (1) previous injury or surgery of the lower extremity and/or back, (2) current (within the past 6 months) pain or injury to the lower extremity and/or back, and (3) any known neurological disorder. Every attempt was made to match participants in each cohort according to sex, height, and BMI (Table 3.1). Prior to testing, research approval was obtained from the local Institutional Review Board, and each participant provided written consent to participate.

Table 3.1 Mean (SD) subject demographics for each cohort (young and older adults)

	N (M/F)	Age (yrs)	Height (m)	Weight (kg)	Walk Speed (m/s)
Young	13 (7 M, 6 F)	21.92 (2.22)	1.76 (0.10)	72.27 (16.96)	1.08 (0.06)
Older	13 (7 M, 6 F)	70.46 (2.82)	1.72 (0.12)	75.35 (18.11)	1.02 (0.17)
p-value	-	< 0.001	0.396	0.659	0.264

Experimental Protocol

Each participant was required to complete one orientation and one test session. The orientation session lasted approximately 30 minutes. The test session lasted approximately 4 hours. The orientation and test sessions were separated by a minimum of 24 hours.

Orientation Session

During the orientation session, we collected participant demographic and strength data, and participants were afforded the opportunity to practice the testing procedures. Each participant had height (m), weight (kg), age (years), and foot dominance recorded. In order to determine foot dominance, participants were asked which foot they prefer to kick a ball with.[30] Each participant also had dominant limb strength recorded on an isokinetic dynamometer (HUMAC NORM, CSMI, Stoughton, MA, USA). Specifically, maximal isometric hip and knee flexion and extension, and ankle plantar- and dorsi-flexion were recorded. To record hip strength, participants stood with their dominant hip flexed at 15 degrees, and either maximally flexed or extended their leg. To record knee strength, participants were seated with the hip and knee secured at 90 and 60 degrees, respectively, and maximally flexed and extended their knee.[31] To record ankle strength, participants were prone with a neutral ankle (i.e., 0 degrees of plantarflexion), and either maximally plantar- or dorsi-flexed their foot.[32] Participants performed three maximal 5 second isometric contractions for each movement. Participants were provided 15 seconds of rest between each maximal contraction, and a minimum of 40 seconds of rest between each movement.[33] Maximum torque produced during each contraction was recorded for analysis. During orientation, each participant was also required to practice the study tasks and provide a verbal confirmation that they could comfortably perform each task prior to testing.

Biomechanical Testing

During the biomechanical test session, participants wore black spandex shorts and shirt, and broken-in tennis shoes. For safety, all participants wore a safety harness attached to an overhead gantry to prevent an accidental fall (Figure 3.1). Prior to data collection, a 4 x 4 Latin Square Design was used to randomize the test order of locomotor activities (Table 3.2), and a 3 x 3 Latin Square Design was used to randomize the test order of terrain (Table 3.3) for each participant.

Figure 3.1 Overhead gantry and safety harness used during stair negotiation

Table 3.2 4 x 4 Latin Square Design to randomize locomotor activity test order

	Task 1	Task 2	Task 3	Task 4
Order 1	Pivot	Stair Descent	Walk	Stair Ascent
Order 2	Stair Ascent	Pivot	Stair Descent	Walk
Order 3	Walk	Stair Ascent	Pivot	Stair Descent
Order 4	Stair Descent	Walk	Stair Ascent	Pivot

Table 3.3 3 x 3 Latin Square Design to randomize terrain test order

	Surface 1	Surface 2	Surface 3
Order 1	Normal	Uneven	Slick
Order 2	Slick	Normal	Uneven
Order 3	Uneven	Slick	Normal

During each locomotor activity, participants had synchronous three-dimensional (3D) lower limb (hip, knee and ankle) biomechanical data recorded. Ten high-speed (240 Hz) optical cameras (Vantage, Vicon Motion Systems, LTD, Oxford, UK) recorded lower limb motion data and a single in-ground force platform (2400 Hz, AMTI OR6 Series, Advanced Mechanical Technology Inc., Watertown, MA) recorded ground reaction force data.

For the purposes of this study, only the walk, and stair ascent and descent tasks on the normal surface were analyzed and described hereafter. For the walk task, participants walked at a self-selected speed through the motion capture space (about 10 meters) and contacted the force platform with their dominant limb. For the stair negotiation tasks, participants walked at a self-selected speed to either ascend or descend two stairs (18.5 cm rise) fixated atop the force platform (Figure 3.2). Stair height was set according to the requirements of the 2021 International Residential Code.[38] During each stair negotiation task, a flat, painted wood panel was used to simulate the normal surface. More specifically, the stair ascent task required participants to walk through the level capture space, place their dominant limb on the first stair, and ascend to the second step; while the stair descent task required participants start atop the second step and descend the stairs to the floor by placing their dominant limb on the first step and then walk through the motion capture space at their self-selected speed. To determine their self-selected speed, each participant was instructed to “walk at a comfortable speed” through the motion capture space five times. During each trial two sets of infrared timing gates (TracTronix TF100, TracTronix Wireless Timing Systems, Lenexa, KS), placed 1.8 meters apart, recorded walking speed. The self-selected speed calculated as an average of those five trials. For testing, participants performed three successful trials of each task (overground walk, and stair ascent and descent). A trial was deemed successful if the participant walked at the target speed (± 5%), only contacted the force platform with their dominant leg, and did not slop or fall during the movement trial.

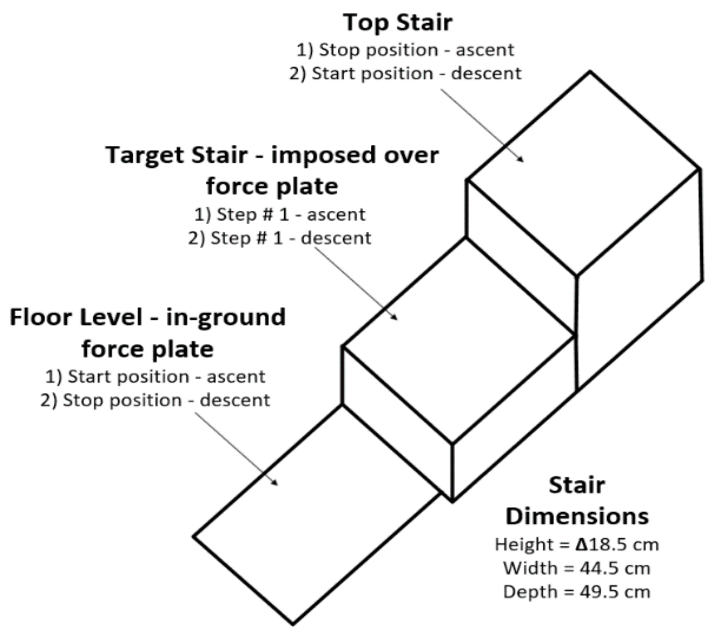

Figure 3.2 Staircase model in the motion capture space for stair negotiation tasks

Biomechanical Analysis

During each walk or stair negotiation trial, lower limb biomechanical data was calculated from the 3D coordinates of 50 retro-reflective markers and four virtual markers (Table 3.4). All reflective markers were placed over specific anatomical landmarks and secured using double sided and elastic tape (Cover-Roll Stretch, BSN Medical GmbH, Hamburg, Germany). Virtual markers were digitized at specific anatomical landmarks in the global coordinate system using a Davis Digitizing Pointer (C-Motion Inc., Rockville, MD). Following maker placement, a participant-specific kinematic model (comprised of trunk, pelvis, and bilateral thigh, shank and foot segments with 27 degrees of freedom) was created in Visual 3D (v6, C-Motion, Inc., Germantown, MD, USA) using a static recording of the participant in anatomical position. Each segment was assigned a local coordinate system and three orthogonal axes (x, y and z).[39] The trunk segment had an origin at the midpoint between the acromion process and the seventh cervical vertebrae and jugular notch on the sternum, and a local coordinate system with three degrees of freedom. The pelvis segment had an origin at the midpoint between the right and left superior iliac spines, and a local coordinate system with three rotational and three translational degrees of freedom. For the thigh segment, the origin was the hip joint center, determined in accordance with Rozumalski and Schwartz, and the segment assigned a local coordinate system with three degrees of freedom.[40] For the shank, the origin was the knee joint center, determined as the midpoint between the medial and lateral femoral epicondyles and segment assigned a local coordinate system with three degrees of freedom.[41] The origin of the foot segment was the ankle joint center, determined as the midpoint between the medial and lateral malleoli and segment assigned a local coordinate system with three degrees of freedom.[42]

Table 3.4 Marker set for whole-body retro-reflective and virtual markers

	Markers
Trunk	*Acromion process,* **jugular notch, xiphoid process, V7 vertebrae, T12 vertebrae**
Pelvis	*Anterior-superior iliac spines, posterior-superior iliac spines,* and iliac crests
Thigh	Greater trochanter, distal thigh, *medial* and *lateral femoral epicondyles*
Shank	Tibial tuberosity, lateral fibula, distal tibia, *medial* and *lateral malleoli*
Foot	Posterior heel, *first* and *fifth metatarsal heads*

Note: *Italic* indicates calibration markers. **Bold** indicates virtual markers.

For each trial, the synchronous marker and ground reaction force data was low pass filtered with a fourth-order Butterworth filter (cutoff frequency of 12 Hz). Filtered marker trajectories were then processed in Visual 3D to calculate ankle joint rotations using a joint coordinate systems approach.[41] Filtered kinematic and ground reaction force data was submitted to standard inverse-dynamics analysis in Visual 3D to obtain the 3D ankle joint moments. Segmental inertial properties were defined according to Dempster.[43,44] Joint moment data was normalized to participant's body mass and height and expressed as external moments. Biomechanical data was normalized from 0% to 100% of stance phase and resampled at 1% increments (n = 101). Stance phase (0% to 100%) was defined as heel strike to toe-off (the instance ground reaction force values exceed or fall below 10 N).

Custom MATLAB (r2018a, MathWorks, Natick, MA) code was used to quantify tibial compression according to Matijevich et al. (2019), and ankle joint stiffness according to Brown et al. (2020).[19,45] Specifically, tibial compression was defined as the summation of the external net force on the ankle (estimated as the 3D ankle joint reaction force projected along the 3D axis connecting the ankle joint and knee joint centers) and internal muscle force contributions (estimated as sagittal ankle moment divided by Achilles tendon moment arm). Achilles tendon moment arm was considered a constant 0.05 meters.[46,47] First, to calculate tibial compression, the 3D ankle joint reaction force projected along the 3D axis connecting the ankle joint and knee joint centers ($Projection_{\alpha}\beta$) was determined using Equation 1:

Equation 1 Vector projection

$$Projection_{\alpha}\beta = \left(\frac{\beta \cdot \alpha}{\|\alpha\|^2}\right)\alpha$$

where: α is a vector connecting ankle (AJC) and knee joint centers (KJC), and β is ankle joint reaction force (AJRF).

Next, the external net force on the ankle (F_{ext}) and the internal muscle force contributions (F_{int}) were calculated using the vector components of the $Projection_{\alpha}\beta$ and the ankle joint flexion moment using Equations 2 and 3:

Equation 2 External force contribution

$$F_{ext} = \sqrt{x^2 + y^2 + z^2}$$

where: x, y and z are vector components of $Projection_{\alpha}\beta$.

Equation 3 Internal force contribution

$$F_{int} = \frac{SAM}{0.05}$$

where: SAM is the ankle joint flexion moment ($^{N*m}/_{kg*m}$).

Finally, tibial compression (Tib_{comp}) was quantified as the summation of F_{ext} and F_{int} using Equation 4:

Equation 4 Tibial compression

$$Tib_{comp} = F_{ext} + F_{int}$$

For analysis, the maximum value and impulse of tibial compression across stance phase was calculated. Ankle joint stiffness was calculated as the change in ankle joint flexion moment divided by the corresponding change in ankle joint flexion angle from heel strike to peak ankle flexion angle, and magnitude quantified as the slope of the moment-angle relationship during that time period.[45]

Statistical Analysis

Predefined ankle biomechanics related to tibial compression was submitted to statistical analysis. Specifically, the dependent variables included maximum and impulse of tibial compression, and ankle joint stiffness and peak of stance ankle flexion joint angle and moment. Each dependent variable was averaged across the three successful trials to create a participant-based mean. Each participant-based mean was submitted to an independent t-test to assess the difference between young and older adults for each task. Pearson correlation coefficients (r) were calculated to determine the relation between tibial compression (maximum and impulse) and ankle biomechanics (ankle joint stiffness, and peak ankle flexion joint angle and moment). Each correlation coefficient was transformed to a z score using Fisher's r to z transformation for correlation comparisons. All statistical analysis was performed using SPSS (v26, IBM, Armonk, NY), with alpha level set to *a priori* at $p < 0.05$.

Results

Age ($p < 0.001$), but not height ($p = 0.396$), weight ($p = 0.659$), or walk speed ($p = 0.264$) differed between the young and older adults (Table 3.1).

Overground Walk

Neither tibial compression (maximum and impulse), nor ankle biomechanics (joint stiffness, moment, and angle) differed between the young and older adults (all: $p > 0.05$) during overground walk task (Appendix B: Figures B.1 and B.3, and Table 3.5).

Table 3.5 Mean (SD) for each tibial compression and ankle biomechanics measure for young and older adults during all tasks

		Maximum Tibial Compression (BW)	**Impulse of Tibial Compression** (BW*s)	**Peak Ankle Flexion Angle** (Deg)	**Ankle Joint Stiffness** (N*m/Deg)	**Peak Ankle Flexion Moment** (N*m/kg*m)
Walk	Young	6.13 (0.64)	3.32 (0.23)	12.66 (8.95)	0.040 (0.019)	-0.66 (0.09)
	Older	6.09 (0.62)	3.64 (0.52)	7.84 (7.90)	0.044 (0.013)	-0.64 (0.08)
Ascent	Young	6.88 (0.72)	3.53 (0.30)	10.05 (7.66)	0.088 (0.189)	-0.76 (0.09)
	Older	6.58 (0.70)	3.62 (0.57)	5.72 (8.65)	-0.456 (1.227)	-0.71 (0.10)
Descent	Young	10.76 (1.26)*	4.74 (0.28)	33.17 (10.29)	0.012 (0.004)	-0.78 (0.07)*
	Older	9.38 (0.95)*	4.95 (0.64)	27.68 (8.18)	0.011 (0.002)	-0.71 (0.08)*

*Denotes a significant effect of age.

During the overground walk, ankle biomechanics exhibited a weak to moderate correlation with tibial compression for all participants (Tables 3.6 and 3.7). Specifically, ankle joint stiffness exhibited a moderate, positive relation to maximum (0.30 ± 0.36) and impulse of tibial compression (0.33 ± 0.35) (Figure B.5), while peak ankle joint angle and moment exhibited a weak, positive and moderate, negative relation to maximum (0.25 ± 0.37; -0.69 ± 0.26) and impulse of tibial compression (0.11 ± 0.38; -0.30 ± 0.36), respectively (Figures 3.3 and B.6).

Table 3.6 Mean ± SD for the correlation coefficient between maximum tibial compression and each ankle biomechanics measure (peak ankle flexion joint angle and moment, and ankle joint stiffness) during each activity for all data and based on age grouping

		Tibial Compression		
		Angle	**Moment**	**Stiffness**
	All	0.25 ± 0.37	-0.69 ± 0.26	0.30 ± 0.36
Walk	Young	0.40 ± 0.49	-0.69 ± 0.40	0.17 ±0.54
	Older	0.06 ± 0.55	-0.69 ± 0.40	0.51 ± 0.46
	All	0.15 ± 0.38	-0.72 ± 0.25	0.03 ± 0.39
Ascent	Young	0.47 ± 0.47	-0.72 ± 0.39	0.30 ± 0.52
	Older	-0.27 ± 0.52	-0.72 ± 0.39	-0.10 ± 0.55
	All	0.23 ± 0.37	-0.48 ± 0.32	0.06 ± 0.39
Descent	Young	0.33 ± 0.51	-0.27 ± 0.52	0.06 ± 0.55
	Older	-0.30 ± 0.52	-0.48 ± 0.47	-0.19 ± 0.54

Table 3.7 Mean ± SD for the correlation coefficient between the impulse of tibial compression and each ankle biomechanics measure (peak ankle flexion joint angle and moment, and ankle joint stiffness) during each activity for all data and based on age grouping

		Impulse of Tibial Compression		
		Angle	**Moment**	**Stiffness**
	All	0.11 ± 0.38	-0.30 ± 0.36	0.33 ± 0.35
Walk	Young	0.01 ± 0.55	-0.20 ± 0.54	0.23 ± 0.53
	Older	0.38 ± 0.50	-0.47 ± 0.47	0.44 ± 0.48
	All	0.44 ± 0.33	0.08 ± 0.39	-0.17 ± 0.38
Ascent	Young	0.32 ± 0.51	-0.27 ± 0.52	0.31 ± 0.51
	Older	0.56 ± 0.44	0.23 ± 0.53	-0.19 ± 0.54
	All	0.14 ± 0.38	0.10 ± 0.38	-0.13 ± 0.38
Descent	Young	-0.27 ± 0.52	0.14 ± 0.54	-0.18 ± 0.54
	Older	0.49 ± 0.47	-0.05 ± 0.55	-0.12 ± 0.55

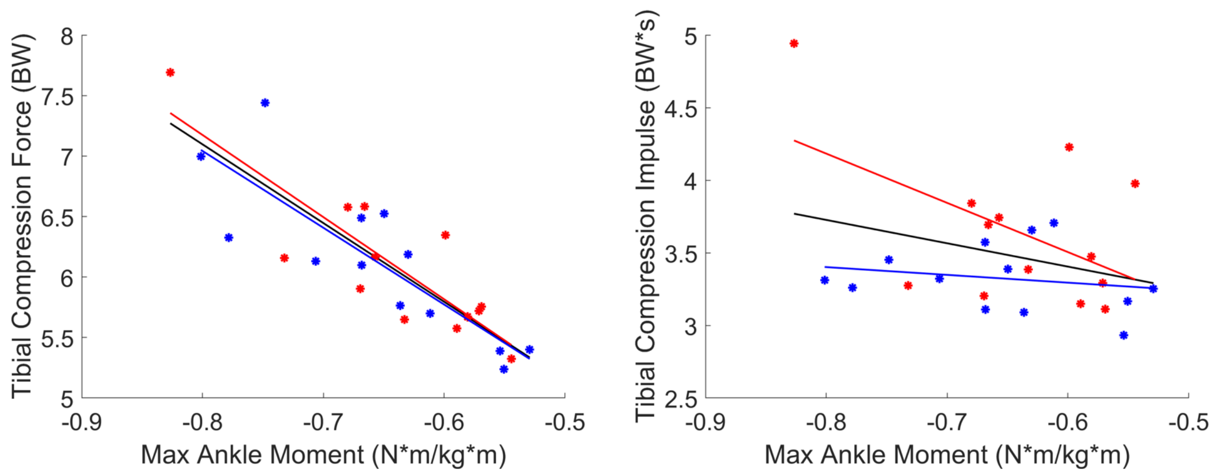

Figure 3.3 Scatterplot of maximum ankle moment (x-axis) and tibial compression metrics (y-axis), maximum tibial compression (right) and impulse of tibial compression (left), during the overground walk task for all participants (black), older adults (red) and younger adults (blue)

When separated by age, the older adults typically exhibited a stronger relation between ankle biomechanics and tibial compression during the overground walk (Tables 3.6 and 3.7). The older adults exhibited a moderate, positive relation between ankle joint stiffness and maximum (0.51 ± 0.46) and impulse of tibial compression (0.44 ± 0.48), while young adults joint stiffness exhibited a weak, positive relation with maximum (0.17 ± 0.54) and impulse of tibial compression (0.23 ± 0.53). Both young and older adults exhibited negligible to moderate, positive relations between peak ankle joint angle and maximum (0.40 ± 0.49; 0.06 ± 0.55) and impulse of tibial compression (0.01 ± 0.38; 0.38 ± 0.50). Yet, older adults exhibited a moderate, negative relation between peak ankle joint moment and tibial compression (maximum: -0.69 ± 0.40; impulse: -0.47 ± 0.47); whereas young adults exhibited a weak to moderate, negative relation between ankle moment and tibial compression (maximum: -0.69 ± 0.40; impulse: -0.20 ± 0.54).

Stair Ascent

Age did not impact tibial compression (maximum and impulse) or ankle biomechanics (joint stiffness, moment, and angle) (all: p > 0.05) during the stair ascent task (Figures B.2 and B.4, and Table 3.5).

During the stair ascent task, ankle biomechanics exhibited a negligible to strong correlation with tibial compression for all participants (Tables 3.6 and 3.7). Specifically, ankle joint stiffness exhibited a negligible, positive relation to maximum tibial compression (0.03 ± 0.39) and a weak, negative relation to impulse of tibial compression (-0.17 ± 0.38) (Figure B.7), while peak ankle joint moment exhibited a strong, negative relation to maximum tibial compression (-0.72 ± 0.25) and a negligible, positive relation to impulse of tibial compression (0.08 ± 0.39) (Figure 3.4). Finally, peak ankle joint angle exhibited a weak, positive relation to maximum (0.15 ± 0.38) and impulse of tibial compression (0.44 ± 0.33) (Figure B.8).

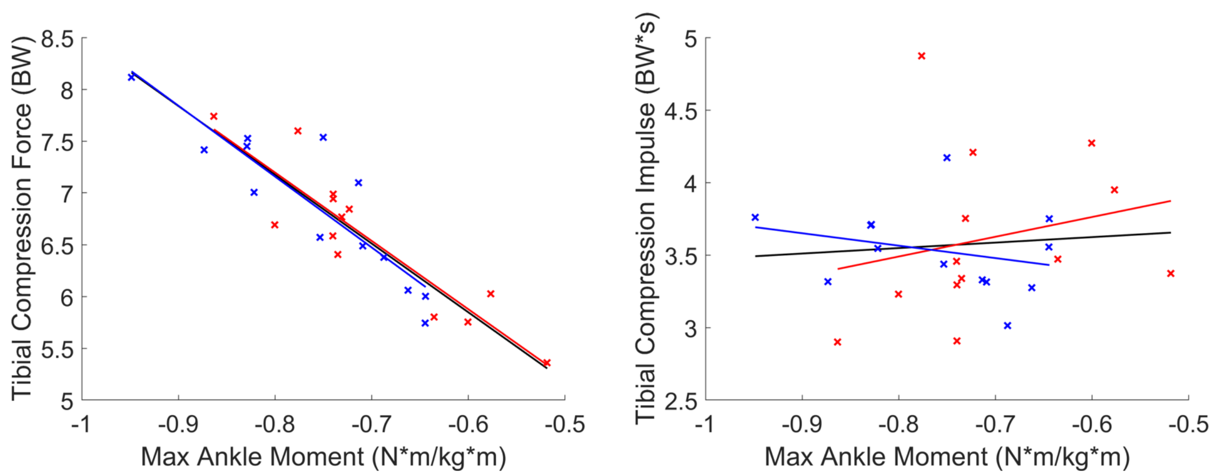

Figure 3.4 Scatterplot of maximum ankle moment (x-axis) and tibial compression metrics (y-axis), maximum tibial compression (right) and impulse of tibial compression (left), during the stair ascent task for all participants (black), older adults (red) and younger adults (blue)

When separated by age, the older adults typically exhibited weaker relations between ankle biomechanics and tibial compression during the stair ascent (Tables 3.6 and 3.7). The older adults exhibited a weak, negative relation between ankle joint stiffness and maximum (-0.10 ± 0.55) and impulse of tibial compression (-0.19 ± 0.54), while young adults exhibited a moderate, positive relation between ankle joint stiffness and maximum (0.30 ± 0.52) and impulse of tibial compression (0.31 ± 0.51). Both older and young adults exhibited a strong, negative relation between peak ankle joint moment and maximum tibial compression (-0.72 ± 0.39; -0.72 ± 0.39) and a weak relation between peak ankle joint moment and impulse of tibial compression (older adults: 0.23 ± 0.53; young adults: -0.27 ± 0.52). Yet, older adults exhibited a weak to moderate relation between peak ankle joint angle and maximum (-0.27 ± 0.52) and impulse of tibial compression (0.56 ± 0.44); whereas young adults exhibited a moderate, positive relation between peak ankle joint angle and maximum (0.47 ± 0.47) and impulse of tibial compression (0.32 ± 0.51).

Stair Descent

During the stair descent, older adults exhibited a smaller maximum tibial compression ($p = 0.004$) (Figure 3.5) and peak ankle joint moment ($p = 0.037$) compared to young adults (Figure 3.6). Age did not impact any other tibial compression or ankle biomechanics measure (all: $p > 0.05$) during the stair descent task (Table 3.5).

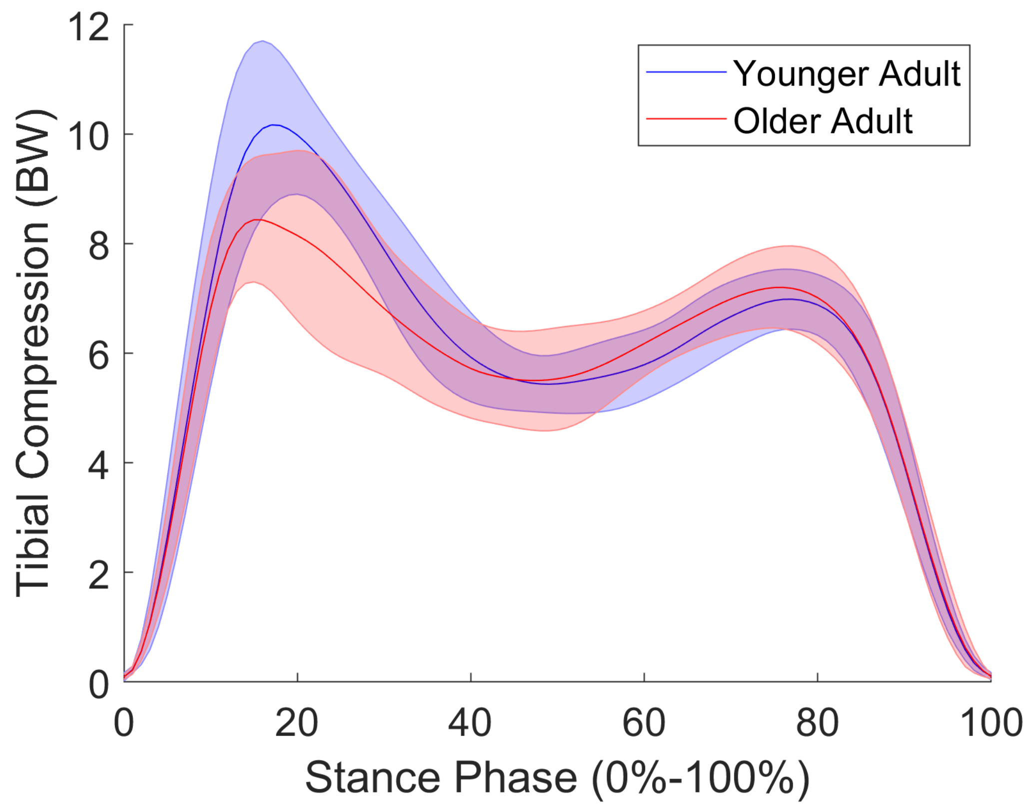

Figure 3.5 Mean ± SD stance phase (0% - 100%) tibial compression for younger and older adults during the stair descent task

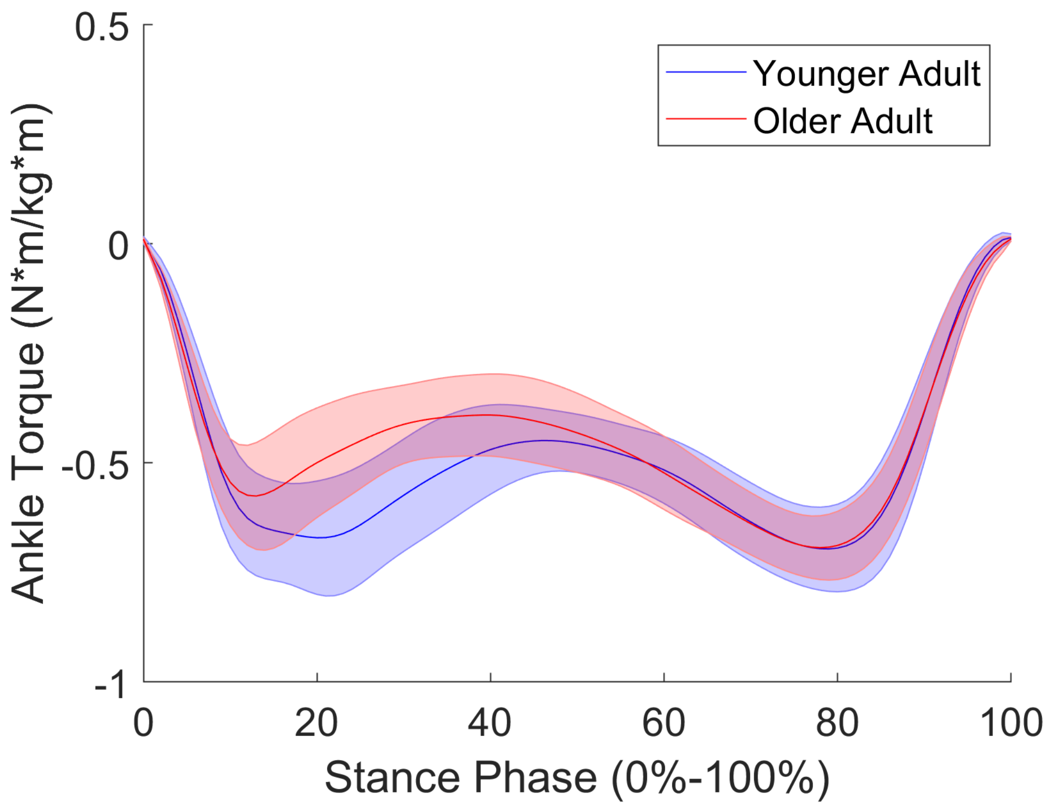

Figure 3.6 Mean ± SD stance phase (0% - 100%) ankle torque (moment) for younger and older adults during the stair descent task

During the stair descent task, ankle biomechanics exhibited a negligible to moderate correlation with tibial compression for all participants (Tables 3.6 and 3.7). Specifically, ankle joint stiffness exhibited a negligible to weak relation to maximum (0.06 ± 0.39) and impulse of tibial compression (-0.13 ± 0.38) (Figure B.9). Peak ankle joint moment exhibited a moderate, negative relation to maximum tibial compression (-0.48 ± 0.32) and a weak, positive relation to impulse of tibial compression (0.10 ± 0.38) (Figure 3.7), while peak ankle joint angle exhibited a weak, positive relation to maximum (0.23 ± 0.37) and impulse of tibial compression (0.14 ± 0.38) (Figure B.10), respectively.

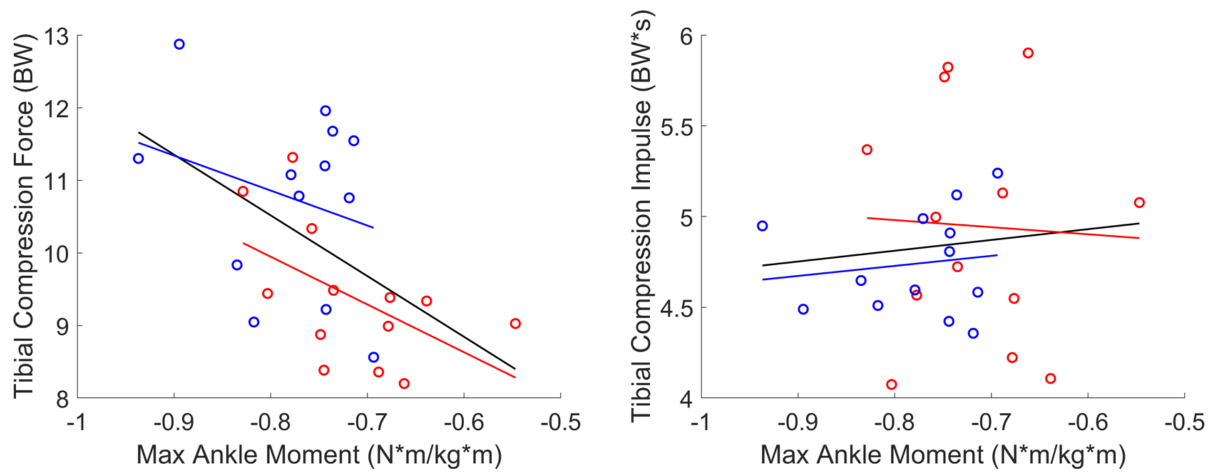

Figure 3.7 Scatterplot of maximum ankle moment (x-axis) and tibial compression metrics (y-axis), maximum tibial compression (right) and impulse of tibial compression (left), during the stair descent task for all participants (black), older adults (red) and younger adults (blue)

When separated by age, the older adults typically exhibited a stronger relation between ankle biomechanics and tibial compression during the stair descent (Tables 3.6 and 3.7). The older adults exhibited a weak, negative relation between ankle joint stiffness and maximum (-0.19 ± 0.54) and impulse of tibial compression (-0.12 ± 0.55); whereas young adults exhibited a negligible to weak relation between ankle joint stiffness and maximum (0.06 ± 0.55) and impulse of tibial compression (-0.18 ± 0.54). Older adults also exhibited a negligible to moderate, negative relation between peak ankle joint moment and maximum (-0.48 ± 0.47) and impulse of tibial compression (-0.05 ± 0.55); whereas young adults exhibited a weak relation between peak ankle joint moment and maximum (-0.27 ± 0.52) and impulse of tibial compression (0.14 ± 0.54). Both older and young adults exhibited a moderate relation between peak ankle joint angle and maximum tibial compression (older adults: -0.30 ± 0.52; young adults: 0.33 ± 0.51). However, the older adults exhibited a moderate, positive relation between peak ankle joint angle and impulse of tibial compression (0.49 ± 0.47), while young adults exhibited a weak, negative relation between peak ankle joint angle and impulse of tibial compression (-0.27 ± 0.52).

Discussion

This study quantified tibial compression and ankle biomechanics for young and older adults during an overground walk and stair negotiation tasks, and determined whether the relationship between tibial compression and ankle biomechanics differed by age. Contrary to our hypotheses, older adults did not exhibit consistent differences in tibial compression compared to young adults, and a strong linear relation between ankle biomechanics and tibial compression was not observed. Yet, older adults exhibited differences in ankle biomechanics, and a stronger relation between ankle biomechanics and tibial compression than young adults.

Contrary to our hypotheses, older adults exhibited smaller maximum tibial compression than their younger counterparts. Specifically, older adults' maximum tibial compression was ~ 1%, 5%, and 15% smaller than young adults during the overground walk and stair ascent and descent, respectively. Yet, despite the smaller maximum tibial compression, older adults may place greater load on the tibia as they exhibited substantial, albeit insignificant, 3% to 10% increases in impulse of tibial compression during the study walk and stair negotiation tasks. Large increases in time-dependent loading (i.e., impulse) reportedly leads to cyclic bone fatigue and damage accumulation.[58,121-127] This elevated tibial loading may be particularly problematic for older adults as they purportedly suffer approximately 115% more cortical bone microdamage and faster fatigue failure from compressive bone stresses than their younger counterparts.[8,9,26,27,34,101] Older adults are reported to exhibit other lower limb biomechanical adaptations, such as greater impact peak and loading rate of vGRF, that, in combination with their reduced muscular strength and bone tissue elasticity, may alter their capacity to attenuate ground impact forces. These alterations may lead to larger and faster compressive stresses transmitted to the tibia, which purportedly increases bone microdamage accumulation and likelihood of tibial stress fracture.[11-14,17,18,63,78] However, considering internal muscle force may be the largest contributor to tibial compression, further study is warranted to determine if faster vGRFs coincide with a substantial increase in older adults' ankle joint moments, particularly late in stance, that may increase tibial compression impulse and bone damage.

The compromised ankle neuromuscular function commonly exhibited by older adults did not lead to significant alterations in ankle joint stiffness during the current locomotor tasks.[78,102,103] Contrary to our hypothesis, older adults did not exhibit consistent increases in ankle joint stiffness compared to the young adults. The older adults, for instance, exhibited an approximate 10% increase in ankle joint stiffness during the overground walk, while the young adults had a 9% stiffer ankle during the stair descent task. The current discrepancy in ankle joint stiffness may be attributed to specific alterations in ankle joint biomechanics exhibited by the older adults during each task. Considering joint stiffness is calculated as change in joint angle for an applied joint moment, the substantial, but statistically insignificant, 38% reduction in peak ankle flexion angle without a concurrent decrease in ankle flexion moment may lead to the 10% stiffer ankle exhibited by the older adults during the overground walk.[45] Conversely, the older adults had a significant 10% reduction in peak ankle joint moment during the stair descent, which may contribute to the 9% stiffer ankle observed for younger adults during the task. Interestingly, the older adults also exhibited an insignificant ~3% and ~7% reduction in peak ankle joint moment during walk and stair ascent tasks, which contradicts the previous significant 9% to 45% decrease in older adult ankle joint moments observed during similar locomotor tasks.[15,22,37] Considering walk speed impacts peak lower limb joint moments, the fact the older adults' self-selected walking speed was not significantly slower than their younger counterparts may contribute to the lack of significant cohort differences observed for peak ankle joint moments.[14,15,22,37,104,111] Older adults, in fact, typically walk slower than their younger counterparts to allow for spatiotemporal and lower limb biomechanics changes necessary to compensate for age-related changes in balance and gait variability.[104,111-115] Consistent with previous work, the older adults exhibited an, albeit statistically insignificant, 17% to 43% reduction in peak ankle joint flexion angle compared to the young adults during the current locomotor tasks.[22,36] This limited ankle flexion may be a specific gait strategy adopted by older adults to help maintain their balance and safely complete the task. However, future research is warranted to determine if older adults do, indeed, alter ankle biomechanics to limit their risk of musculoskeletal or fall-related injury, or whether it is merely a consequence of the compromised ankle neuromuscular function.

Contrary to our hypotheses, a consistent strong linear relation between ankle biomechanics and tibial compression metrics was not observed. In fact, only the relationship between maximum tibial compression and peak ankle flexion moment exhibited a consistent moderate to strong linear relation (overground walk: $r = -0.69 \pm 0.26$; stair descent: $r = -0.48 \pm 0.32$; stair ascent: $r = -0.72 \pm 0.25$, respectively). In agreement with Matijevich at al. and Walker et al., these experimental outcomes highlight the important contribution the internal ankle muscle forces (i.e., ankle joint flexion moment), rather than external joint forces (i.e., ground reaction forces), make towards tibial compression.[19,51] Yet, despite its relation to maximum tibial compression, peak ankle flexion moment did not exhibit a similar relation to the impulse of tibial compression (overground walk: r = -0.30 ± 0.36; stair descent: r = 0.10 ± 0.38; stair ascent: r = 0.008 ± 0.39, respectively). While the reason for this discrepancy is not immediately evident, it may be that time-dependent increases (i.e., impulse) in tibial compression may result from elevated joint moments across the entirety of stance, rather than an increase in discrete (i.e., peak) joint moments.

Older adults typically exhibited a stronger relation between ankle biomechanics and tibial compression than the young adults. For example, older adults exhibited a moderate linear relation between ankle joint stiffness and peak ankle joint moment with impulse of tibial compression during the walk (r = 0.44 ± 0.48 and r = -0.47 ± 0.47), and peak ankle joint moment with maximum tibial compression (r = -0.48 ± 0.47) during stair descent task. Whereas young adults exhibited a weak relation between the same ankle biomechanical and tibial compression measures (r = 0.23, -0.20, and -0.27, respectively) during the walk and stair descent tasks. It may be age-related changes to the ankle neuromuscular system, such as the approximately 52% and 33% decrease in peak ankle plantar- and dorsi-flexion strength currently exhibited by older adults (Appendix A), that require alterations in ankle biomechanics to safely complete the current locomotor tasks and hinder the already compromised musculature's ability to attenuate elevated joint loads in general, and bone stress specifically.[10,28,36,37,78,102,116-118] For instance, the weaker older adults may limit ankle flexion to safely perform the task, but consequently reduce musculature's capacity to absorb energy and attenuate impact ground reaction forces.[22,36,37,78] Yet considering internal, and not external joint forces may increase tibial compression, future work should determine if older adults limited ankle flexion leads to greater transmission of internal forces to the tibia, and elevated bone stress.

This study may be limited by the current tibial compression calculation. Although this calculation has previously quantified tibial compression during walking and running, the Achilles tendon moment arm (used to quantify the internal force contributions) is treated as a constant 5 cm, and may over or underestimate individual tendon length, altering the quantified compression values during stair negotiation tasks.[19,46,47,51] Additionally, previous researchers projected the vertical ground reaction force, rather than the ankle joint reaction force, up the long axis of the tibia.[19,51] Although projecting the vertical ground reaction force up the tibia may be a suitable method to calculate tibial compression, we chose to project the ankle joint reaction force, as this negated the potential for the absorption of ground reaction forces in the foot and underestimation of tibial compression. Further, the chosen locomotor tasks may be a limitation, as only ascending and descending two stairs may limit the real-world applicability.

In conclusion, older adults may place larger compressive forces on the tibia, and increase likelihood of suffering a stress fracture compared to young adults. The 3% to 10% larger impulse of tibial compression exhibited by older adults may increase tibial microdamage accumulation and lead to faster fatigue failure of the bone. Yet, despite exhibiting a stronger relation between ankle biomechanics and tibial compression than their younger counterparts, there was not a specific alteration in older adults' ankle biomechanics that may predict the substantial change in their tibial compression.

CHAPTER FOUR: CONCLUSION

Introduction

This study sought to determine whether older adults exhibit differences in: (1) tibial compression and (2) ankle joint stiffness and associated biomechanics during an overground walk and stair negotiation tasks compared to young adults, and (3) examine the linear relation between tibial compression and ankle biomechanics, and whether these relations differ for young and older adults. Key findings partially support the hypotheses that older adults exhibit significant differences in tibial compression, and ankle joint stiffness and biomechanics compared with young adults, but contradict the hypothesis that there would be a strong linear relation between ankle biomechanics and tibial compression. The older adults, however, typically exhibited a stronger relation between ankle biomechanics and tibial compression.

Key Findings

Older adults may place larger compressive forces on the tibia than young adults. Although older adults exhibited smaller maximum tibial compression, they exhibited large 3% to 10% increases in impulse of tibial compression compared to the young adults. These large increases may be problematic for older adults and increase their likelihood of suffering tibial stress fracture. Additionally, the compromised ankle neuromuscular function typically exhibited by older adults may lead to specific alterations in ankle joint stiffness and biomechanics, including reductions in ankle joint flexion moment and angle, adopted to help maintain their balance and safely complete the locomotor tasks.

During the locomotor tasks, only peak ankle flexion moment exhibited a moderate to strong linear relation with maximum tibial compression (r between -0.48 and -0.72). This highlights the important contribution of internal ankle muscle forces (i.e., ankle joint flexion moment) to tibial compression. Yet, older adults typically exhibited a stronger relation between ankle biomechanics, particularly peak ankle joint moment, and tibial compression compared to their younger counterparts. Age-related decrements in ankle neuromuscular function in general, but particularly decreased plantarflexion and dorsiflexion strength may reduce older adults' ability to attenuate impact forces during locomotor tasks, and thereby require greater joint moments and subsequent tibia compressive forces to complete the task.

Significance

These findings support the tenet that the compromised ankle neuromuscular function, commonly exhibited by older adults, leads to changes to ankle biomechanics and tibial compression that increase tibial stress fracture risk. Specifically, this study documented that older adults apply greater time-dependent loading (i.e., impulse), rather than maximum tibial compression, compared to young adults during the walk and stair negotiation tasks. Considering this elevated tibial loading may increase cortical bone microdamage, leading to faster fatigue failure for older adults, it may place them at a greater risk for tibial stress fracture. We also documented that older adults alter joint stiffness and ankle biomechanics to complete the locomotor tasks compared with their younger counterparts. These findings can be used by clinicians to reduce the risk of stress fracture development in older adults during routine locomotor tasks. Specifically, these outcomes can be used to identify specific ankle biomechanics, such as peak ankle flexion moment and angle (i.e., joint strength and range of motion), to reduce the problematic increase in tibial loading exhibited by older adults. Successful implementation of the knowledge provided herein may result in a reduction of lower limb musculoskeletal overuse injury in older adult populations and may improve the quality of life through pain reduction and decrease the high out of pocket medical expenses for a rapidly ageing population.[50]

Limitations

This study may be limited by the current tibial compression calculation. Although this calculation has been used previously to quantify tibial compression during walking and running tasks, it relies upon the estimation of the Achilles moment arm from cadaveric data. The calculation estimates the Achilles tendon moment arm to be a constant 5 cm, which may over or underestimate individual tendon length and alter the internal force contributions for each individual.[19,46,47,51] Further, the previous calculation of tibial compression relied on the vertical ground reaction force projected up the long axis of the tibia to provide an estimation of the external force contribution, whereas the current calculation used the ankle joint reaction force projected up the long axis of the tibia.[19,51] While this may obfuscate the comparison with previous tibial compression data, it will negate potential absorption of ground reaction forces in the foot and provide a more accurate measure of stance phase tibial compression.

The chosen locomotor tasks may be a limitation. Although each activity is routinely performed in the real world, producing an exact replication of each in the lab environment is challenging. The overground walk task was performed over a short distance (approximately 10 m), which may not reflect extended walking in real settings, particularly as only the flat, normal surface trials were examined in this study. Further, the stair descent and ascent tasks were performed using only a small set consisting of two stairs, which may not accurately reflect the stair negotiation activity in realistic settings. It is possible that the constraints of performing these tasks in a laboratory may potentially alter lower limb, particularly ankle, biomechanics and lead to an over- or under-estimation of the current dependent variables.

The chosen participant population may be a limitation. The current older adult participants had to have experienced an accidental fall within the previous 12 months to be included. However, older adults who have not experienced a similar fall within the past 12 months could possibly exhibit different ankle biomechanics and tibial compression metrics than the chosen population.

Finally, the self-selected walking speed may be a limitation. Typically, older adults walk slower than their younger counterparts. Considering walking speed has been shown to influence several lower limb biomechanics, including those used in the tibial compression calculation, the fact our older adult sample did not walk slower than the young adults may not capture differences in ankle biomechanics and tibial compression that exist in the larger older adult population.

Future Work

Ageing resulted in greater, yet statistically insignificant, increases in impulse of tibial compression values during all locomotor tasks by older adults. As such, future work is warranted to determine whether older adults exhibit waveform, rather than discrete, differences of the tibial compression. Further, assessing individuals with a history of tibial stress fracture may provide additional insight into differences of tibial compression and biomechanical changes that may lead to this type of injury.

Older adults decreased their peak ankle flexion moment during all locomotor tasks compared to the young adults. However, this study limited its assessment of age-related changes in lower limb biomechanics to the ankle joint, and it is unclear what influence, if any, other lower limb biomechanics may have on tibial compression. As such, future work is warranted to determine whether changes in lower limb biomechanics at other joints influence tibial loading and stress fracture risk. Also, additional work is needed to determine whether the alterations in ankle biomechanics adopted by older adults are necessary to limit their risk of musculoskeletal or fall-related injury.

Lastly, considering tibial compression reportedly stems from internal muscle forces, further study is needed to improve the accuracy of the current tibial compression calculation. For instance, better estimates of the Achilles tendon moment arm may provide more accurate measures of tibial compression.[19,51] In addition, future work assessing the ankle flexion moment (i.e., internal force components) waveform may provide additional insight into how tibial compression is altered across stance or impacted by age.

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APPENDIX A

Strength Testing

Ankle Strength

Each participant had maximal isometric ankle strength recorded using an isokinetic dynamometer (HUMAC NORM, CSMI, Stoughton, MA, USA). Specifically, participants lay prone with a neutral ankle (i.e., 0 degrees of plantarflexion), and either maximally plantar- or dorsi-flexed their foot three times for 5 seconds.[36] Participants were provided 15 seconds of rest between each maximal contraction.[37] Maximum torque produced during each contraction was recorded for analysis, and this value was normalized to body mass and recorded as N*m. Normalized values were submitted to an independent t-test in SPSS (v26, IBM, Armonk, NY) to assess the strength differences between young and older adults, with alpha level set to a *priori* at $p < 0.05$.

<u>Results</u>

Older adults' maximal ankle dorsi- (0.34 ± 0.11 Nm/kg) and plantarflexion (0.56 ± 0.23 Nm/kg) was significantly weaker ($p = 0.002$; $p < 0.001$) than the maximal ankle dorsi- (0.50 ± 0.11 Nm/kg) and plantarflexion (1.15 ± 0.40) exhibited by young adults.

APPENDIX B

Additional Figures

Tibial Compression Figures

Overground Walk

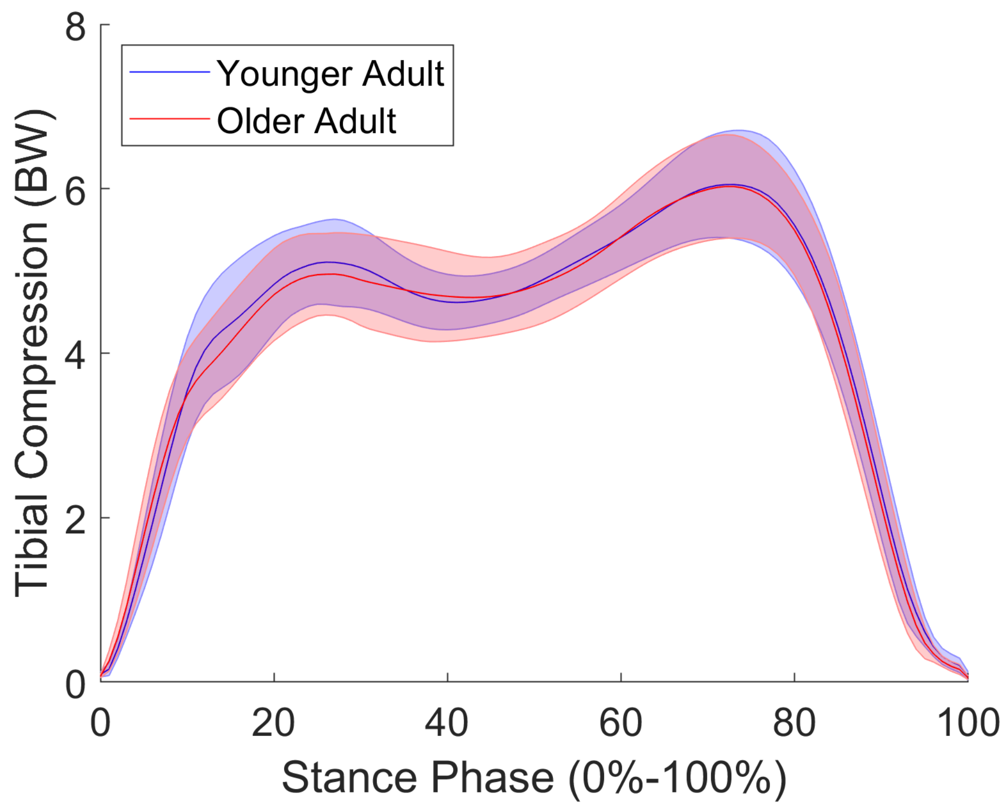

Figure B.1 Mean ± SD stance phase (0% - 100%) tibial compression for younger and older adults during the overground walking task

Stair Ascent

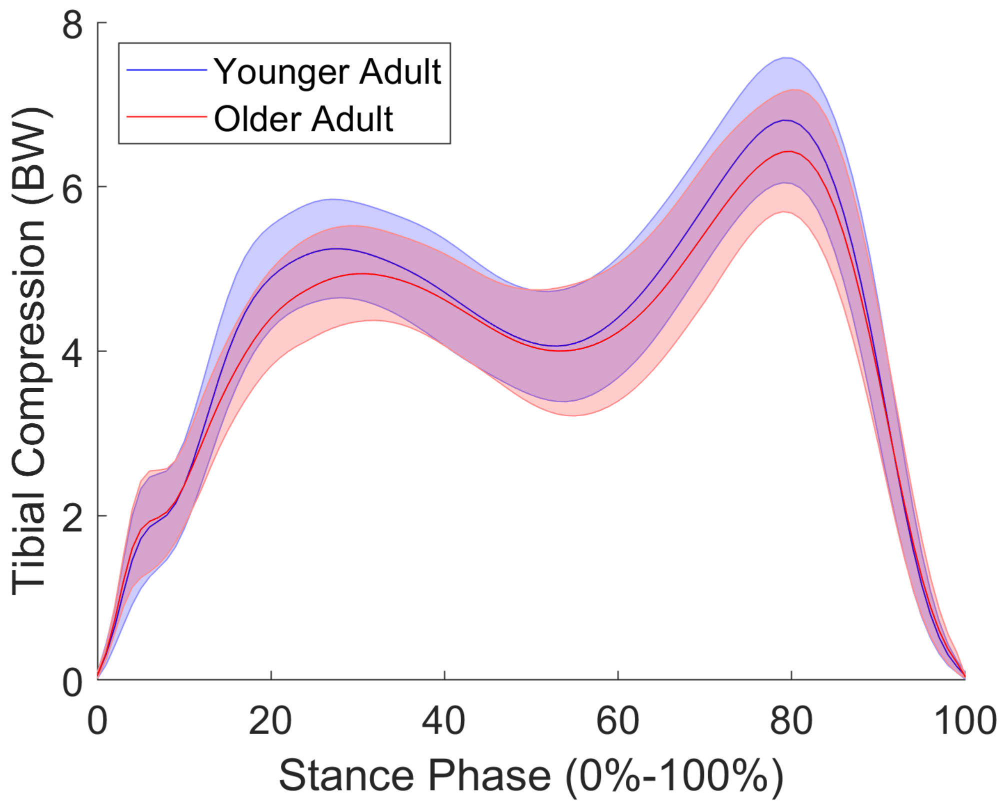

Figure B.2 Mean ± SD stance phase (0% - 100%) tibial compression for younger and older adults during the stair ascent task

Ankle Torque Figures

Overground Walk

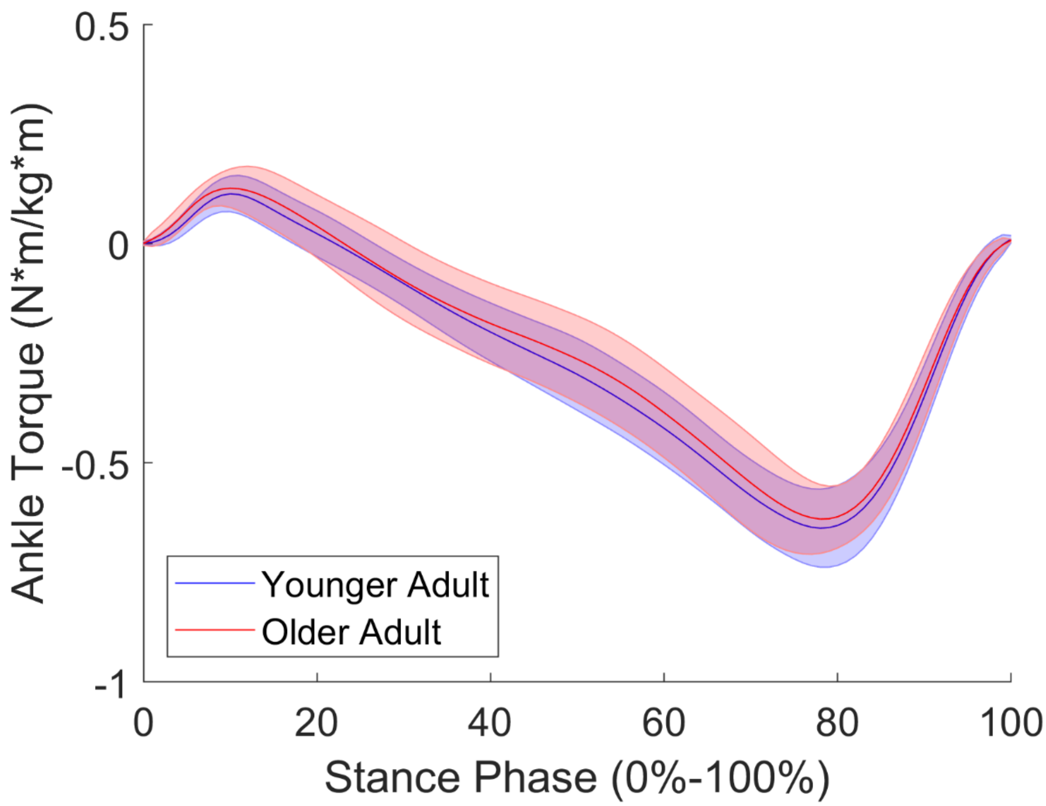

Figure B.3 Mean ± SD stance phase (0% - 100%) ankle torque (moment) for younger and older adults during the overground walk task

Stair Ascent

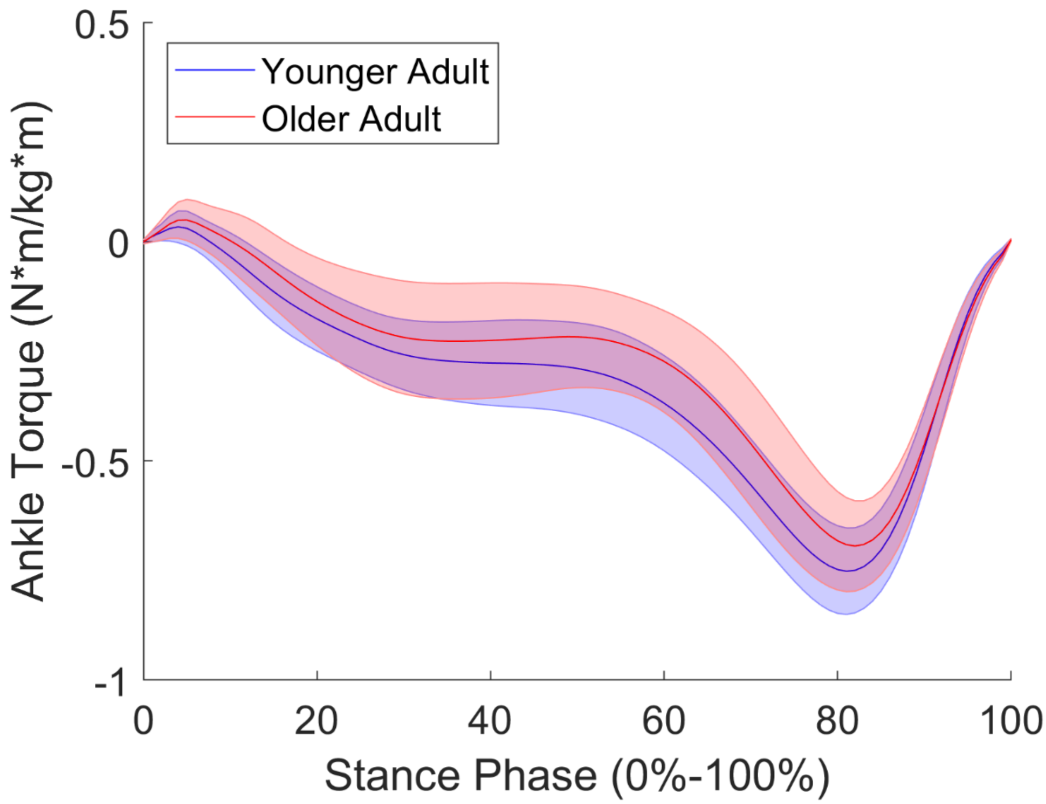

Figure B.4 Mean ± SD stance phase (0% - 100%) ankle torque (moment) for younger and older adults during the stair ascent task

Correlation Scatterplots

<u>Overground Walk</u>

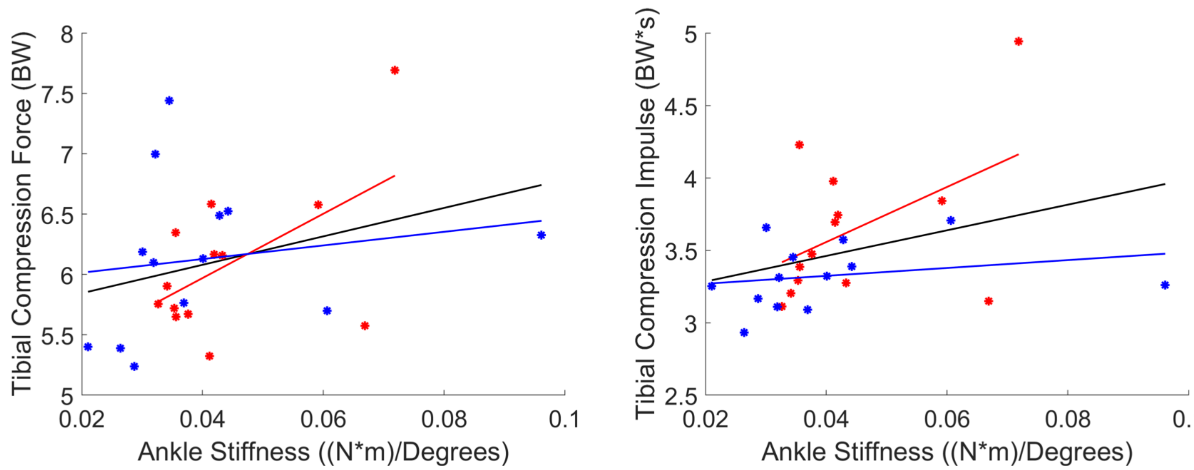

Figure B.5 Scatterplot of ankle joint stiffness (x-axis) and tibial compression metrics (y-axis), maximum tibial compression (right) and impulse of tibial compression (left), during the overground walk task for all participants (black), older adults (red) and younger adults (blue)

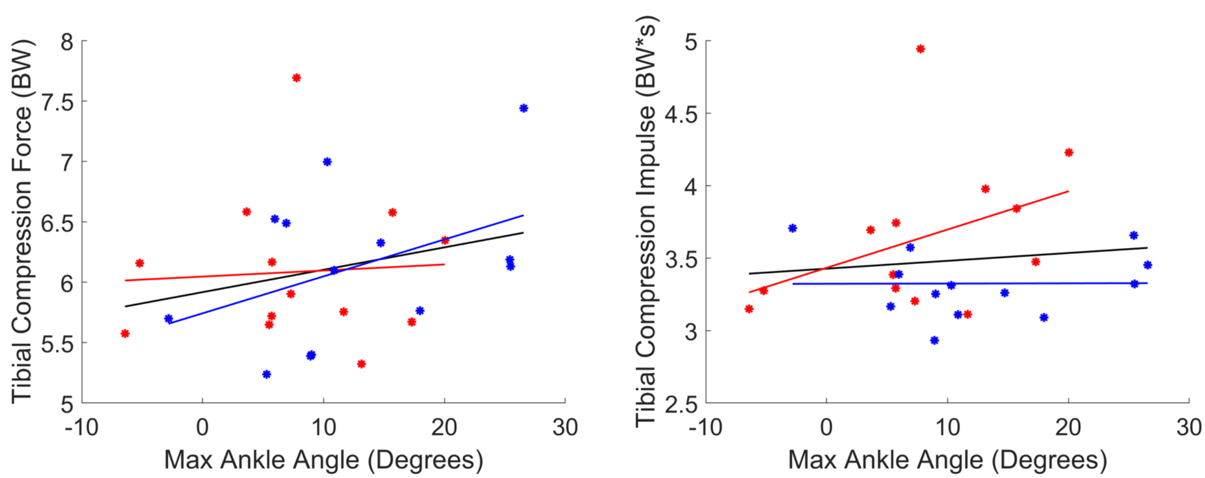

Figure B.6 Scatterplot of maximum ankle angle (x-axis) and tibial compression metrics (y-axis), maximum tibial compression (right) and impulse of tibial compression (left), during the overground walk task for all participants (black), older adults (red) and younger adults (blue)

Stair Ascent

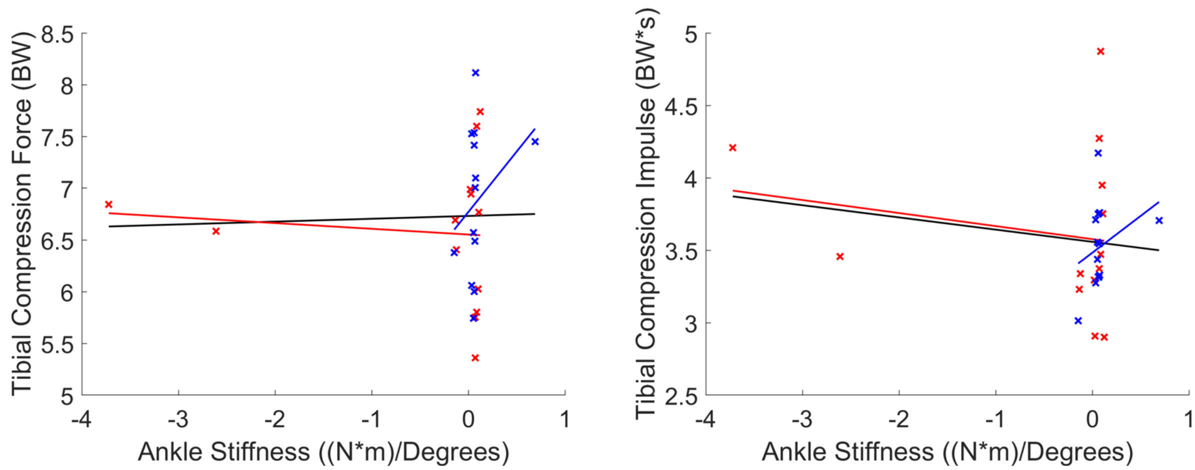

Figure B.7 Scatterplot of ankle joint stiffness (x-axis) and tibial compression metrics (y-axis), maximum tibial compression (right) and impulse of tibial compression (left), during the stair ascent task for all participants (black), older adults (red) and younger adults (blue)

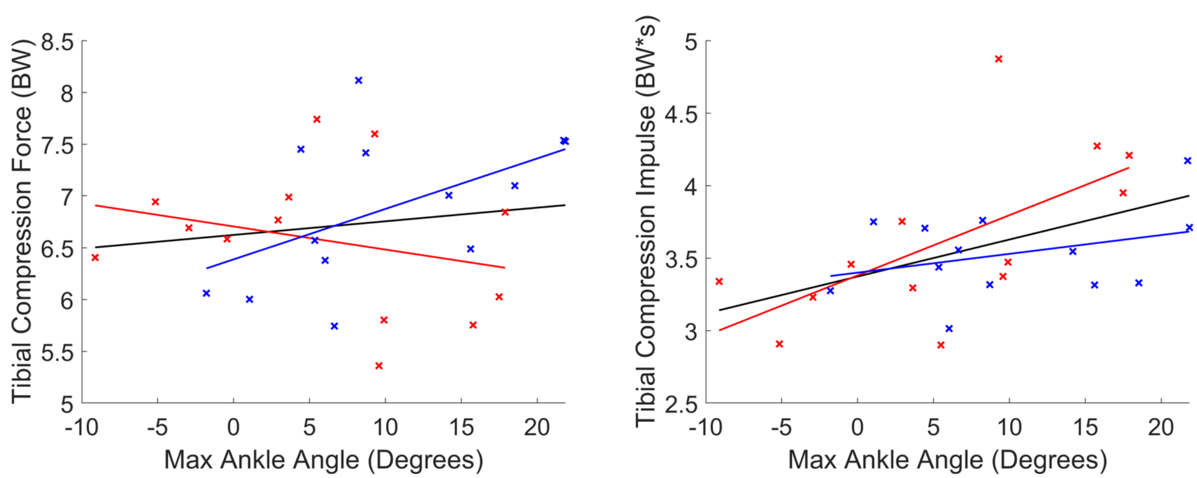

Figure B.8 Scatterplot of maximum ankle angle (x-axis) and tibial compression metrics (y-axis), maximum tibial compression (right) and impulse of tibial compression (left), during the stair ascent task for all participants (black), older adults (red) and younger adults (blue)

Stair Descent

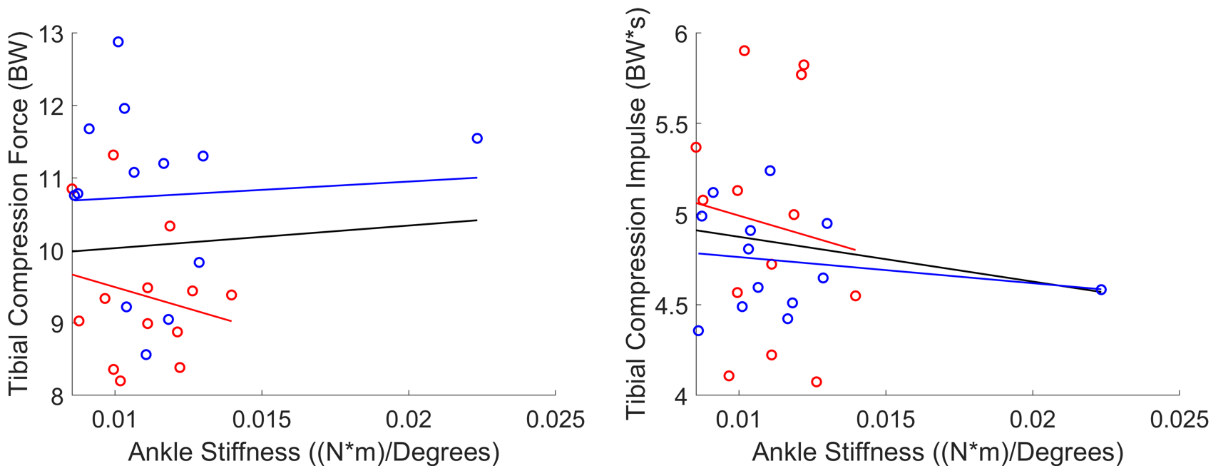

Figure B.9 Scatterplot of ankle joint stiffness (x-axis) and tibial compression metrics (y-axis), maximum tibial compression (right) and impulse of tibial compression (left), during the stair descent task for all participants (black), older adults (red) and younger adults (blue)

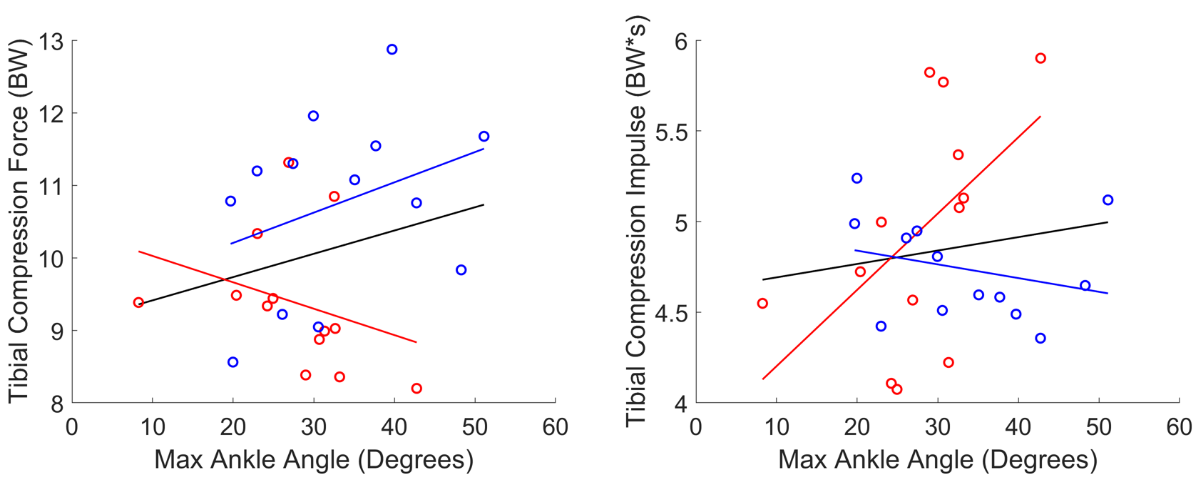

Figure B.10 Scatterplot of maximum ankle angle (x-axis) and tibial compression metrics (y-axis), maximum tibial compression (right) and impulse of tibial compression (left), during the stair descent task for all participants (black), older adults (red) and younger adults (blue)