Naomi Frankston

The effects of dynamic loading on *in vivo* second metatarsal bone stress in runners

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A race is a work of art that people can look at and be affected by in as many ways as they're capable of understanding.

Steve Prefontaine

Abstract

Background: Bone stress injuries (BSIs) occur in populations that engage in repetitive loading, such as distance runners and military recruits. In these populations, the second metatarsal is a common site for BSIs. Risk of BSI may be influenced by modifiable factors including running speed, load carriage, and prolonged running by affecting internal metatarsal loading. Internal metatarsal bone stress can be estimated during running using a two-dimensional beam theory model and experimentally measured inputs. Variability in the model inputs including experimentally measured geometry, kinetics, and kinematics can influence the within-day repeatability and between-day reliability of bone stress estimates.

Aims: The overall aims of this thesis were to determine how modifiable factors including running speed, load carriage, and prolonged running influence internal loading in the second metatarsal during running. The thesis also aimed to quantify the within-day repeatability and between-day reliability of estimates of metatarsal loading.

Methods: Data for all three studies were collected from a single cohort of participants over two visits to the biomechanics laboratory. Participants were 28 healthy (14 female/14 male), rearfoot strike runners aged 18 – 40 years. During the first laboratory visit, participants were screened for foot strike type and, if eligible, completed six running conditions at three speeds with and without load carriage. During the second laboratory visit, participants completed overground running trials at their preferred speed before and after a demanding 10 km treadmill run. Kinematic and kinetic data were collected during barefoot overground running using motion capture cameras, a pressure plate, and a force plate. Bone stress was calculated at the dorsal and plantar surfaces of the second metatarsal midshaft using a two-dimensional beam theory model.

Results: Second metatarsal bone stress estimates demonstrated excellent within-day repeatability and between-day reliability. Running speed and load carriage were found to

independently influence second metatarsal bone stress. Bone stress increased with both increasing running speed and added load carriage. Between-participant variability was observed in peak bone stress, with individual responses to conditions differing from group trends. The magnitude of ground reaction force under the second metatarsal was very strongly correlated with second metatarsal bone stress. Second metatarsal stress when running at preferred speed overground did not change following completion of a demanding 10 km treadmill run, despite localized changes in foot and ankle kinematics and kinetics.

Conclusions: Together, the findings indicate that second metatarsal bone stress is responsive to changes in the external loading environment and that the model demonstrates excellent reliability for estimating second metatarsal internal loading.

Running at faster speeds and with load carriage increased *in vivo* second metatarsal stress, which may increase the risk of metatarsal BSI. These findings help explain the high prevalence of second metatarsal BSIs in runners and military populations who frequently engage in repetitive loading and load carriage.

A similar effect was not observed following a prolonged run where second metatarsal stress remained unchanged. This indicates either that prolonged running may not alter peak metatarsal loading in the same way as speed and load carriage or that the beam theory model lacked sensitivity to the observed reductions in distal forces.

Sammendrag (Abstract in Norwegian)

Bakgrunn: Belastningsskader i bein (BSI) oppstår hos grupper som utsettes for repetitiv belastning, for eksempel langdistanseløpere og militære rekrutter. Hos disse gruppene er den andre metatarsalen et vanlig skadested for BSI. Risikoen for BSI kan påvirkes av modifiserbare faktorer, inkludert løpehastighet, bæring av last og langvarig løping, ved at disse påvirker den interne belastningen i metatarsalen. Den interne belastningen på metatarsalen kan estimeres under løping ved hjelp av en todimensjonal stråleteorimodell basert på eksperimentelt målte inndata. Variasjon i modellinndataene, inkludert eksperimentelt målt geometri, kinetikk og kinematikk, kan påvirke repeterbarheten innenfor en dag og reliabiliteten mellom dager for estimatene av belastningen i bein.

Mål: Hovedmålet med denne avhandlingen var å undersøke hvordan modifiserbare faktorer, inkludert løpehastighet, bæring av last og langvarig løping, påvirker den interne belastningen i den andre metatarsalen under løping. Avhandlingen hadde også som mål å kvantifisere repeterbarheten innenfor samme dag og reliabiliteten mellom dager for estimatene av metatarsalbelastningen.

Metoder: Dataene for alle tre studiene ble samlet inn fra én deltakergruppe i løpet av to besøk ved biomekanikklaboratoriet. Deltakerne var 28 friske (14 kvinner/14 menn) løpere med bakfotslanding i alderen 18 – 40 år. Under det første laboratoriebesøket ble deltakerne screenet for landingsmønster, og de som kvalifiserte gjennomførte seks løpebetingelser ved tre ulike hastigheter, både med og uten bæring av last. Under det andre laboratoriebesøket gjennomførte deltakerne løpsforsøk over bakken i sin foretrukne hastighet før og etter et 10 km-løp på tredemølle med høy anstrengelse. Kinematiske og kinetiske data ble samlet inn under barfotløping på bakken ved hjelp av et optisk bevegelsessystem, en trykkplate og en kraftplattform. Beinbelastningen ble beregnet på de dorsale og plantare overflatene av midtskaftet i den andre metatarsalen ved hjelp av en todimensjonal stråleteorimodell.

Resultater: Estimeringene av belastningen i den andre metatarsalen viste høy repeterbarhet innenfor samme dag og høy reliabilitet mellom dager. Løpehastighet og bæring av last påvirket beinbelastningen i den andre metatarsalen uavhengig av hverandre. Beinbelastningen økte både med økende løpehastighet og ved bæring av last. Det ble observert variasjon mellom deltakerne i maksimal belastning, der individuelle responser på betingelsene avvek fra gruppetrendene. Den maksimale bakkereaksjonskraften under den andre metatarsalen var svært sterkt korrelert med beinbelastningen i den andre metatarsalen. Beinbelastningen i den andre metatarsalen under løping over bakken i foretrukket hastighet endret seg ikke etter et 10 km-løp på tredemølle med høy anstrengelse, til tross for lokale endringer i fot- og ankelkinematikk og -kinetikk.

Konklusjoner: Samlet sett viser funnene at beinbelastningen i den andre metatarsalen påvirkes av endringer i den ytre belastningen, og at modellen viser høy reliabilitet ved estimering av intern belastning i den andre metatarsalen.

Løping i høyere hastigheter og med bæring av last økte beinbelastningen i den andre metatarsalen in vivo, noe som kan øke risikoen for belastningsskader i den andre metatarsalen (BSI). Disse funnene bidrar til å forklare den høye forekomsten av BSI i den andre metatarsalen hos løpere og militærpersonell som ofte utsettes for repetitiv belastning og bæring av last.

En tilsvarende effekt ble ikke observert etter en langvarig løpetur, der beinbelastningen i den andre metatarsalen forble uendret. Dette indikerer enten at langvarig løping ikke påvirker maksimal metatarsalbelastning på samme måte som hastighet og bæring av last, eller at stråleteorimodellen hadde begrenset sensitivitet for de observerte reduksjonene i distale krefter.

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List of Papers

The dissertation is based on the following original research papers:

I. **Frankston N**, Ellison M, Riise JN, Seynnes O, Farris D, & Rice H. Second metatarsal bone stress during running is independently influenced by running speed and load carriage. *Scandinavian Journal of Medicine & Science in Sports*. 2025: e70152. https://doi.org/10.1111/sms.70152.

II. **Frankston N**, Ellison M, Riise JN, Seynnes O, Farris D, Rice H. A demanding 10 km treadmill run alters lower limb kinetics and kinematics but not second metatarsal bone stress. (Under revision in Sports Biomechanics)

III. **Frankston N**, Ellison M, Farris D, Rice H. Between-Day Reliability and Within-Day Repeatability of Estimates of Second Metatarsal Bone Stress During Running. (Manuscript)

Original research papers related to but not contained in the dissertation:

I. **Frankston N**, Ellison M, Bill K, Sventeckis A, Mai P, Warden S, Rice H. Factors affecting loading in the second metatarsal during running with implications for bone stress injury: A systematic review. (Manuscript)

II. Riise JN, Mai P, **Frankston N**, Schuster R, Kelly L, Rice H. The biomechanical function of the foot is robust to fatigue-related alterations following a demanding 10km run. (Submitted to Gait & Posture)

List of Abbreviations

2D	two-dimensional
3D	three-dimensions
ANOVA	analysis of variance
BSI	bone stress injury
CI	confidence interval
CV	coefficient of variation
GRF	ground reaction force
Hz	hertz
hGRF	horizontal ground reaction force
ICC	intraclass correlation coefficient
JRF	joint reaction force
kg	kilograms
km	kilometer
N	Newtons
Nm	Newton meters
m	meters
MRI	magnetic resonance imaging
ms	milliseconds
MT2	second metatarsal
MTP	metatarsophalangeal
RPE	rate of perceived exertion
SD	standard deviation
vGRF	vertical ground reaction force

List of Symbols

α	second metatarsal pitch angle
ε	distance from the inferior metatarsal head to the start of the plantar tendon
F_{ax}	axial force
F_{jz}	vertical joint reaction force
F_{jy}	horizontal joint reaction force
F_{mz}	vertical force under the second metatarsal
F_{my}	horizontal force under the second metatarsal
F_{sh}	shear force
F_{tz}	vertical force under the second toe
F_{ty}	horizontal force under the second toe
l_m	second metatarsal length
l_t	second toe length
M_{be}	bending moment
P	plantar tendon force due to the toe flexor muscles
r_1	lower radius of the metatarsal head
r_2	upper radius of the metatarsal head
θ	plantar tendon angle

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1 Introduction

1.1 Overview

A bone stress injury (BSI) is an overuse injury which occurs when repetitive mechanical loading exceeds the bone's capacity for repair, resulting in structural damage, bone pain, and tenderness.[1,2] BSIs occur along a continuum from stress reactions to stress fractures, with the potential to progress to complete fractures.[1] Due to the repetitive loading experienced during training, BSIs are a common overuse injury in runners and military recruits.[3,4] BSIs can require a minimum of six weeks break from the inciting activity.[5]

For injury prevention and management, it is important to understand risk factors for BSIs. Risk factors are multifactorial and involve mechanical and biological factors, including both modifiable (e.g., training load) and non-modifiable (e.g., sex) factors.[1,2] The mechanical load applied to bone and the ability of bone to resist deformation from loading are important factors in BSI development. Bone loading refers to the application of external or internal mechanical forces to bone that generate internal stresses and strains within the bone. Bone loading during running typically occurs from the application of ground reaction forces (GRF) and from muscle and tendon forces acting directly on the bone.[6] Biomechanics can be used to investigate how modifiable factors such as running speed, load carriage, and prolonged running influence internal bone loading.

Internal loads, however, are difficult to measure and are typically quantified using either surgically implanted strain gauges or musculoskeletal modeling. While the metatarsals, especially the second metatarsal, are common locations for BSIs in runners and military recruits,[7,8] few studies have quantified internal metatarsal stresses or strains during repetitive loading activities such as running.[9–12] The second metatarsal is particularly at risk for BSIs due to higher loading[10,13] and smaller average bone geometry[14] compared to other metatarsals and the tibia. When using modeling approaches based on experimentally

measured inputs to estimate internal loads, it is also important to quantify the within-day repeatability and between-day reliability of metatarsal bone stress estimates to distinguish true differences between conditions from measurement variability.

1.2 Bone Stress Injuries

1.2.1 Etiology and Continuum of Bone Stress Injuries

Bone adapts to mechanical loading, with normal physiological loading being necessary for bone to maintain its current volume and structural integrity.[15] Relatively few loading cycles can induce the formation of new bone by stimulating remodeling, while repetitive submaximal loading can lead to damage accumulation.[16] Without adequate repair, bone can fail from repetitive loading cycles at stress levels well below the bone's ultimate failure strength. Bone stress injuries are associated with repetitive loading activities such as running, where bone experiences submaximal cyclic loads at frequencies as low as 1-3 Hz.[17]

Under normal physiological loading conditions, bone is constantly accumulating microscopic damage (microdamage) in the form of microcracks (Figure 1). The accumulation of microdamage is not pathological but a normal physiological process.[15] Bone stress injuries occur when microdamage accumulation from repetitive submaximal mechanical loads exceeds the remodeling capacity of the bone.[1] There are three general stages of bone failure that occur due to repetitive loading: microcrack initiation which occurs as a response to loading, microcrack growth which occurs when bone remodeling is insufficient, and coalescence or merging of smaller cracks.[2] Bone remodeling has been shown to preferentially occur in areas with moderate microdamage accumulation.[17]

Bone stress injuries occur along a continuum with stress reactions being an early and less severe form of BSI and stress fractures being a more severe form, which can lead to complete bone fractures.[1] BSIs can be classified into four magnetic resonance imaging (MRI) grades, with grades one to three indicating bone edema of the periosteum (the connective tissue on

the outer bone surface) consistent with a stress reaction and grade four indicating involvement of the cortical bone (the compact outer layer of bone) with a visible fracture line on imaging, consistent with a stress fracture.[18]



Figure 1: Microscope image of a cross-section of human compact bone stained with fuchsin. Arrows indicate linear microcracks which occurred under physiological loading conditions. Image reproduced from Seref-Ferlengez et al.[62] with permission of the authors.

1.2.2 Epidemiology and Burden of Bone Stress Injuries in Runners and Military Recruits

Running is one of the most popular forms of physical activity worldwide.[19] This widespread participation is accompanied by a high prevalence of musculoskeletal overuse injuries,[20–23] with the overall injury prevalence in runners reported to be as high as 85%.[24,25] The high injury incidence may be explained by the repetitive loading in running. Running involves external

loads of about two to three times body weight.[6] The cyclic application of loads[26] over the course of a run means that bones and joints are loaded in a repetitive manner.

BSIs are one of the most prevalent injury types in both runners[23,27] and military recruits.[3,28] The repetitive loading experienced during running and military tasks, including load carriage, can lead to microdamage and elevated BSI risk in the absence of adequate recovery.[29] BSIs commonly occur in the tibial shaft and metatarsals in runners and military recruits.[1,30–32]

The incidence of BSIs in competitive runners has been reported to range from 5 to 21% during a 1-year period.[1] In military populations, the incidence of BSIs ranges from 1 to 31%,[33–35] with the highest rates observed during the first 4 – 6 weeks of basic training.[34,35] In addition to a high incidence, BSIs impose a substantial burden due to prolonged time loss from activity. The mean time to return to unrestricted activity can range from 6 to 27 weeks.[5,36]

1.3 Risk Factors for Bone Stress Injuries

Bone stress injury risk depends on the load applied to bone and the ability of bone to resist deformation. Warden et al.[1] divides risk factors for BSI into two categories: 1) factors modifying the load applied to bone (mechanical risk factors), and 2) factors affecting the ability of bone to resist load without accumulating damage (biological risk factors).

1.3.1 Mechanical Risk Factors

Microdamage occurs as a result of increasing load magnitudes and the number of loading cycles experienced by bone.[1] When considering running, bone loading is affected by both external and internal forces which can be described by the following loading characteristics: magnitude, rate, duration, frequency, and direction.[1] Increases in training load are related to increased risk of BSI through modification of these factors.

Running at faster speeds increases the magnitude of loading by increasing both external GRFs[37] and internal muscle forces.[38] Faster running speeds have also been linked to

increased probability of tibial stress fracture.[39] In contrast, there is very limited evidence associating GRF loading rates and peak accelerations with a history of tibial stress fracture,[40] with more recent prospective and mechanical fatigue studies showing that loading rates have little or no influence on bone fatigue and BSI risk.[41,42] Duration describes the length of a run or the number of loading cycles experienced by bone. The risk of bone stress injury increases with increasing loading cycles at a given load magnitude due to the accumulation of microdamage in bone (Figure 2).[26,43]

Loading frequency describes how often bone is loaded relative to recovery time. When bone is loaded frequently with insufficient recovery time, bone remodeling can become imbalanced, with osteoclastic resorption exceeding osteoblastic bone formation. Since bone formation takes longer than resorption, repeated loading during this period can lead to microdamage accumulation and increased risk of bone stress injury.[44,45]

The direction of loading also influences the mechanical response of bone. Bone is anisotropic, meaning that its properties differ when loaded in different directions.[46] The direction of bone loading affects risk of BSI since bone is stronger under compression than tension.[46,47] Bone is moderately strong under bending which consists of compression on the concave side and tension on the convex side.[46] Cortical bone is weakest under shear loading compared to compression and tension.[48] Metatarsal BSIs typically occur in the diaphysis (shaft) of cortical bone[49] which experiences both shear and bending forces during running.[9,50]

Running kinematics such as lower-extremity joint angles have the potential to affect BSI risk by altering the magnitude and direction of bone loading.[2,51] However, evidence of kinematic risk factors for BSI is limited and inconsistent. There is very limited evidence, for example, of an association between increased peak rearfoot eversion angle and increased peak hip adduction angle and a history of tibial BSI. There is also conflicting evidence of the relationship between high and low foot arches and BSI risk.[52]

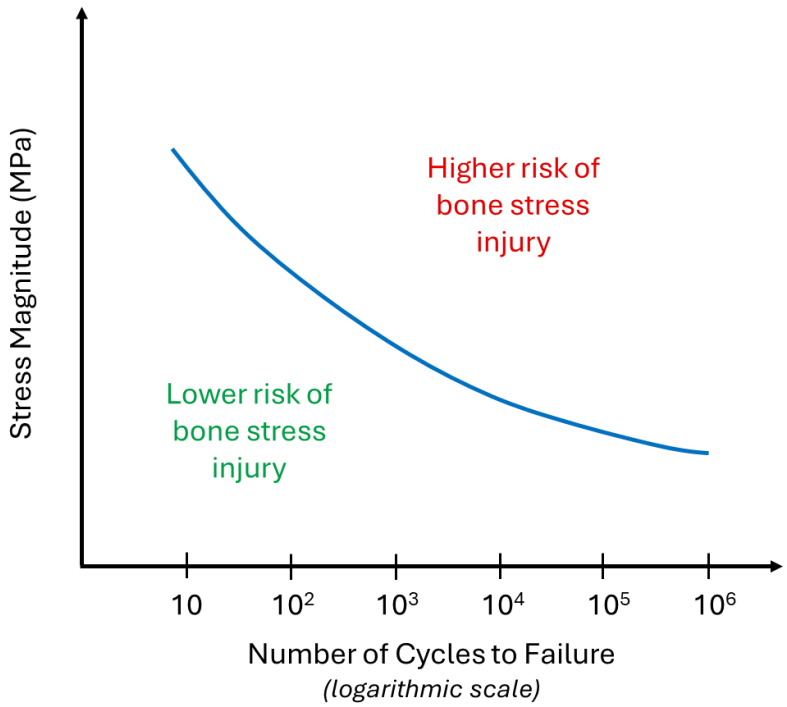


Figure 2: Theoretical S-N curve illustrating the relationship between stress magnitude and the number of loading cycles to failure in bone.[27,63] Above the curve, cyclic loading results in damage accumulation that exceeds the repair capacity of bone, increasing the risk of bone stress injury. Below the curve, damage accumulation remains within the repair capacity of bone, reducing the risk of bone stress injury.

Training factors influence BSI risk by modifying loading magnitude, duration, and frequency. Runners are at particularly high risk due to the combination of high numbers of loading cycles and high loading magnitudes.[2] Rapid increases in training load are associated with elevated BSI risk, particularly in military recruits and athletes during early training periods.[51] Load carriage tasks in military populations further amplify risk by increasing loading magnitude.[53] Increases in training volume or session frequency raise BSI risk by increasing loading cycles or loading frequency, while higher running speeds increase risk through greater magnitudes

of external GRFs[37] and internal muscle forces.[38] When physical activity levels are increased too rapidly or frequently compared to the level an individual is adapted to, the balance between bone microdamage formation and bone repair can be disrupted, increasing the risk of BSI.[1,51]

1.3.2 Biological and Behavioral Risk Factors

Biological factors related to BSI risk affect the ability of bone to resist load. More compliant bones experience greater deformation than more rigid bones under the same applied load and are thus more susceptible to microdamage and BSIs.[1] Biological characteristics which influence bone rigidity include bone mass, structure (distribution of bone mass), and quality of bone material.[2,51] Prospective studies have shown that smaller bone mass, smaller cross-sectional size, and lower bone mineral content were associated with increased risk of BSI.[1,54] Muscles apply load to bone but may also play a protective role for BSI risk. Prospective studies have shown that smaller muscle size and strength are associated with increased BSI risk.[2,51] Muscle fatigue may reduce the protective effect of muscles as bone strain has been shown to increase during exertional running.[2,51] Additional biological factors which have been associated with BSI risk include female sex, older age, both low and high body mass index.[51,53] A history of BSI is one of the strongest predictors of future BSI risk,[1,2,55] likely reflecting an underlying susceptibility to bone injury.

Modifiable behavioral factors relate to BSI risk in both military and athletic populations by influencing the ability of bone to resist loading. A limited history of physical activity has been linked to increased risk of BSI in both populations,[1,2,51,53,56] likely because the bone is not adapted to tolerate higher levels of loading. Additionally, low energy availability and nutritional deficiencies (low calcium and vitamin D levels) have been associated with increased BSI risk.[1,2,53] Sleep quality and duration may also influence BSI risk but this relationship is poorly understood.[2,57,58] Psychological factors including psychological stress may also be related to BSI risk but this relationship is also not well understood.[2,51,59]

1.4 Second Metatarsal Bone Stress Injuries

The second metatarsal bone in the forefoot is one of the most common injury sites for BSI in runners and military recruits.[7,8] In a population of Royal Marine recruits, 65% of stress fractures reported were in the metatarsals, of which 27% were in the second metatarsal,[7] while in a cohort of competitive collegiate track and field athletes and cross-country runners, 21% of BSIs occurred in the metatarsals, with 23% in the second metatarsal.[60] The relatively high incidence of second metatarsal BSIs may be influenced by bone geometry, with the second metatarsal being on average the longest metatarsal with the narrowest diaphyseal width (Figure 3).[14] The second metatarsal has also been shown to experience higher vertical plantar force, bending strain, shear force,[10] compression strain,[13,61–63] and tension strain [13,62] during dynamic loading compared to the other metatarsals and the tibia.

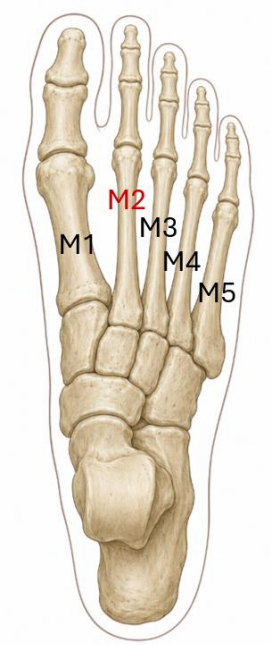


Figure 3: Dorsal view of the bony anatomy of the human foot, with the first through fifth metatarsals labeled (M1-M5) and the second metatarsal highlighted in red. Image generated using OpenAI ChatGPT.

Second metatarsal BSIs most commonly occur in the shaft of cortical bone.[49,64,65] BSIs of the second metatarsal shaft are considered low risk for complications, with an average return to activity of 8 weeks and return to full activity of 10 – 11 weeks.[66,67] It remains unclear whether diaphyseal metatarsal BSIs occur more commonly at the dorsal or plantar aspect of the bone. Bone stresses at both the dorsal and plantar aspects of the mid-shaft of the second metatarsal were therefore analyzed in **Papers I, II, and III**.

1.5 Quantifying Second Metatarsal Loading

Quantifying internal bone loading allows us to better understand factors contributing to the risk of second metatarsal BSI. Internal bone loading is, however, difficult to measure and is typically quantified using surgically implanted strain gauges or computational modeling. Arndt et al.[68] used implanted strain gauges to measure *in vivo* second metatarsal strain during barefoot treadmill walking and found increased compression strain with walking-induced fatigue and load carriage. However, only walking was examined and not running, despite runners having a high prevalence of metatarsal BSI. Donahue and Sharkey[69] used strain gauges in cadaveric feet with a dynamic gait simulator to measure second metatarsal strain and found that simulated muscle fatigue of the plantar flexors and cutting the plantar fascia both increased dorsal strain. However, the study only included cadaveric feet and may not reflect dynamic loading *in vivo*.

Milgrom et al.[13] used strain gauges to measure *in vivo* second metatarsal and tibial strains and found that metatarsal compression strains were greater than tibial strains during treadmill walking and running, both barefoot and in running shoes. The study, however, included only two participants and did not examine the effects of modifiable gait factors on metatarsal loading. Together, these strain gauge studies provide evidence that second metatarsal loading is influenced by external loading conditions and fatigue. However, they are limited by small sample sizes, were mostly non-running or cadaveric studies, and did not investigate how modifiable gait factors influence *in vivo* metatarsal loading during running.

While strain gauges offer a direct method for quantifying metatarsal loading, they are limited in that they involve invasive surgical implantation and only measure highly localized strain at the bone surface. More recent studies have employed computational modeling approaches to estimate second metatarsal loading non-invasively. Ellison et al. ran two separate analyses of the same dataset using both a two-dimensional (2D) beam theory model[9] and a three-dimensional (3D) finite element model[11] to estimate *in vivo* second metatarsal stress during barefoot running. Both modeling approaches found no difference in metatarsal stress between rearfoot and non-rearfoot strike runners. While finite element modeling considers the internal stress distribution within the bone and allows for more participant-specific model customization, it is computationally intensive and has been shown to produce similar relative differences and conclusions to a simpler 2D beam theory model using the same dataset.[70] However, only foot strike was investigated and it remains unclear how other modifiable gait factors which may be relevant for injury risk influence metatarsal stress.

A study by Firminger et al.[12] used musculoskeletal and finite element modeling to estimate *in vivo* second metatarsal strain during running and found that reducing stride length did not significantly affect strain, while wearing minimalist compared to traditional shoes increased strain and the probability of bone failure. However, the study did not consider how running speed, load carriage, or prolonged running influence metatarsal loading. Together, these modeling studies indicate that second metatarsal loading during running is not substantially influenced by foot strike pattern or stride length but may increase with minimalist with footwear. It remains unclear how other modifiable factors including running speed, load carriage, and prolonged running influence *in vivo* second metatarsal loading during running and thus may contribute to injury risk.

Few studies have investigated second metatarsal loading *in vivo*, and it is unknown how well external loading approximates internal loading in the metatarsal during running. In the tibia, external loading metrics, including GRFs, have been shown to be poor surrogates for internal tibial loading during running.[71] It therefore remains unclear how well external loading metrics,

including vertical force under the second metatarsal and vertical GRF, and variables related to external loading, including body mass and running speed, relate to second metatarsal stress during running. Understanding how well these external loading metrics relate to internal metatarsal loading may be useful for researchers, clinicians, and coaches interested in estimating metatarsal loading using simpler, more accessible metrics.

1.6 Within-Day Repeatability and Between-Day Reliability of Stress Estimates

Second metatarsal bone stress during running was estimated in this thesis using a 2D beam theory model of the second metatarsal.[9] The model relies on experimentally measured geometric, kinetic, and kinematic inputs as outlined in Section 2.4, including forces under the second metatarsal and the metatarsal pitch angle. The model inputs can vary due to both normal physiological fluctuations in running mechanics and methodological factors such as marker placement. Differences in marker placement between laboratory visits may influence stress estimates since marker positions are used to determine second metatarsal and second toe lengths. A prior sensitivity analysis of the model showed that a 10% increase in second metatarsal length modestly increased peak dorsal and plantar stresses.[70] Due to potential variability in model inputs, it is unknown how repeatable metatarsal bone stress estimates are between trials within a single laboratory visit and how reliable these estimates are across separate laboratory visits. Quantifying both within-day repeatability and between-day reliability is important for distinguishing normal physiological variability from meaningful experimental effects and for contextualizing changes in estimated metatarsal bone stress.

1.7 Modifiable Factors Influencing Second Metatarsal Loading

1.7.1 Running Speed

Running speed is a modifiable factor which may affect second metatarsal bone stress by altering the magnitude of external and internal loading. In the tibia, probabilistic modeling

using experimental data showed that reducing running speed by 1 m/s reduced the estimated risk of tibial stress fracture by 7 – 10%.[39] A beam theory model with experimentally measured inputs showed that increasing running speed by 20% increased tibial loading.[72] However, it is unknown whether these findings translate to the second metatarsal. In the second metatarsal, peak plantar pressure measured with pressure-sensing insoles or a pressure plate has been shown to increase with increasing running speed.[73,74] However, external loading measures such as plantar pressure may not reflect internal bone loading. It remains unknown how altering running speed influences *in vivo* second metatarsal bone stress.

1.7.2 Load Carriage

Load carriage is another modifiable factor which may affect second metatarsal bone stress by increasing the magnitude of external and internal loading. Load carriage has previously been shown to alter gait kinematics and kinetics.[75] Military recruits commonly engage in load carriage activities,[76] potentially contributing to the high incidence of second metatarsal bone stress injuries in military populations. Despite this, there has been limited investigation of the effects of load carriage on internal metatarsal loading. The only known study to investigate this used strain gauges and reported increased *in vivo* second metatarsal dorsal strain during treadmill walking with a 20 kg backpack compared to walking with no additional load.[77] However, strain gauges only measure localized strain at the implantation site and strain gauge implantation may alter mechanics. Additionally, the study did not investigate running which may increase injury risk due to higher external and internal loading compared to walking. It remains unknown how load carriage influences *in vivo* second metatarsal bone stress during running.

Since both running speed and load carriage have the potential to increase loading in the second metatarsal, the combined effect of speed and load carriage may be greater than the sum of the individual factors. This is particularly relevant for military personnel and individuals who run while carrying loads. If there is an interaction effect between speed and

load carriage, then running at faster speeds with load carriage could disproportionately increase injury risk. The potential interaction between running speed and load carriage on second metatarsal bone stress has not been previously investigated.

1.7.3 Prolonged Running

Prolonged running may induce muscle fatigue which may influence second metatarsal bone stress by reducing the capacity of muscles to attenuate mechanical loading. Muscle fatigue can be defined as the reduced ability of muscles to generate force,[78] which may influence internal bone loading and thus risk of BSI. Treadmill walking and marching fatigue protocols have been shown to increase second metatarsal stress and strain and tibial loading.[77,79–81] Running fatigue protocols have been shown to increase[80] or decrease[82,83] tibial loading, suggesting an influence of fatigue on internal bone loading. Cadaveric studies have demonstrated that simulated muscle fatigue during walking increased second metatarsal strain,[69] while simulated muscle contraction reduced second metatarsal dorsal strain,[84] suggesting a direct influence of muscle forces on bone strain. Muscle fatigue may thus increase the risk of BSI by altering internal bone loading.

Ten kilometer running protocols performed at high levels of perceived exertion have previously been used to investigate the effects of prolonged running on lower-extremity biomechanics. These protocols have been shown to reduce muscle activity in the gastrocnemius and vastus lateralis muscles,[86] alter running kinematics,[87] and redistribute joint work from distal to proximal joints.[88] These findings indicate that prolonged running can alter lower limb gait neuromechanics, which may influence injury risk. However, despite the high prevalence of second metatarsal BSIs in runners,[31,85] it is unknown how prolonged, exertional running influences *in vivo* second metatarsal loading.

Foot and ankle kinematics and kinetics may also change with running-induced fatigue. Ankle dorsiflexion angle at initial contact[89,90] and ankle plantarflexor moment[88,90–92] have been observed to decrease with running-induced fatigue, indicating a reduced ability of the plantar

flexors to generate force when fatigued. It remains unclear how a prolonged running protocol performed at high levels of perceived exertion influences metatarsophalangeal (MTP) joint kinematics and kinetics. Additionally, it is unclear how altered MTP mechanics affect metatarsal loading and BSI risk.

1.8 Aims

The overall aims of this doctoral thesis were to determine how modifiable factors including running speed, load carriage, and prolonged running influence *in vivo* second metatarsal bone stress during running, and to quantify the within-day repeatability and between-day reliability of bone stress estimates using a beam theory model with experimentally measured inputs. The specific aims of the three papers were:

Paper I

The aim of this study was to determine how running speed and load carriage influence bone stress in the second metatarsal during running, and to quantify whether these factors interact. Specifically, the effects of running at preferred speed, 90% of preferred speed, and 110% of preferred speed, with and without an additional 10% body mass, were quantified.

It was hypothesized that second metatarsal bone stress would increase with both increasing running speed and load carriage, and that there would be an interaction effect so that the combined influence would increase stress to a greater extent than the sum of the individual factors.

Paper II

The aim of this study was to determine how an exertional, prolonged run influences second metatarsal bone stress. Specifically, changes in bone stress and MTP joint mechanics were quantified following a 10 km treadmill run at 105% of estimated 10 km race time.

It was hypothesized that second metatarsal bone stress would increase following an exertional 10 km run. A directional hypothesis was not made regarding MTP joint mechanics due to limited previous research in this area.

Paper III

The aim of this study was to determine how repeatable second metatarsal bone stress estimates were within a day and how reliable estimates were between days. Specifically, within-day repeatability and between-day reliability of bone stress estimates using a beam theory model with experimentally measured inputs were quantified.

A hypothesis was not specified due to the lack of prior studies assessing the reliability of bone stress estimates.

2 Methods

This thesis is based on data collected from the same cohort of participants during two separate visits to the biomechanics laboratory. **Paper I** is based on data collected during the first laboratory visit which involved running at three different speeds with and without load carriage. **Paper II** is based on data from the second visit which involved an exertional 10 kilometer treadmill run. Data from preferred speed running during both visits were analyzed in **Paper III** to assess within-day repeatability and between-day reliability of second metatarsal bone stress estimates. This methods section will describe the methods for all three studies.

2.1 Participants

2.1.1 Sample Size

Participant recruitment was conducted jointly for all three studies (Papers I-III), with the same cohort being included in all studies. The minimum sample sizes required to adequately power each study were calculated separately due to each paper containing different statistical analyses. Sample size analyses for **Papers I** and **II** were conducted in G*Power software (Düsseldorf, Germany).[93]

For **Paper I**, an *a priori* sample size analysis was conducted for a within-subjects repeated measures analysis of variance (ANOVA) with running speed at three levels and load carriage at two levels. Statistical significance was set at $\alpha = 0.05$ and the study was powered at 80%. To estimate the corresponding effect size for a repeated-measures ANOVA, Cohen's f was calculated as half the Cohen's d reported by Ellison et al.[9] for dorsal second metatarsal stress. The resulting minimum required sample size was 19 participants.

For **Paper II**, the minimum required sample size was determined for a paired t-test with two dependent means (pre- and post-treadmill run). For the sample size calculation, a

significance level of $\alpha = 0.05$ was used with a power of 90%. Cohen's d effect size was calculated to be 0.62 based on data from a study which quantified differences in peak tibial bending moments pre- and post- a 10 km treadmill run.[82] The results of the sample size estimate yielded a minimum 24 participants to adequately power the study.

Since the data for **Paper III** had already been collected, a post hoc precision-based approach was performed to assess if the sample size was adequate to power the analysis.[94] The observed intraclass correlation coefficient (ICC) for second metatarsal dorsal stress was used to determine the minimum sample size required to achieve a 95% confidence interval width of 0.10 for this ICC. Calculations were performed in R (v4.5.1, R Foundation for Statistical Computing, Vienna, Austria) and resulted in a minimum sample size of 17 participants to adequately power the study.

Since **Paper II** had the largest sample size requirement with a minimum of 24 participants, it was used as the basis for participant recruitment. A conservative sample of 28 participants were included in the study cohort to account for any potential study drop out or in case of possible poor data quality for some participants.

2.1.2 Inclusion and Exclusion Criteria

Participants were included who were any gender, ages 18-40 years, ran at least 15 km per week, ran at least three times per week, had been running on a regular basis for at least three years, and were rearfoot strikers. Study participation was limited to rearfoot strikers since foot strike pattern influences running biomechanics[95] and rearfoot strikers typically maintain a rearfoot strike when fatigued.[96] Participants had also completed a maximal-effort run of any distance from 5 km to a marathon in the last 2 years, which allowed them to self-estimate their maximal-effort 10 km time. Participants were comfortable running 10 km on a treadmill and could run barefoot without pain.

Exclusion criteria included any neurological, musculoskeletal, or movement disorder, or current pregnancy. Participants were excluded if they had any lower extremity injury within the previous six months, any previous foot or ankle surgery, any symptomatic foot pathology, or any illness or injury that prevented running. Body mass greater than 100 kg was also an exclusion criterion since the maximum carrying capacity of the weight vest used in the load carriage conditions was 10 kg and the weight vest was set to 10% of a participant's body mass.

2.1.3 Recruitment and Ethics

Participants were recruited through flyers placed around the local area in Oslo, posts on social media, and a project page on the school's website. The study was approved by the ethics committee at the Norwegian School of Sport Sciences (Approval number 281 – 150623) and Sikt – The Norwegian Agency for Shared Services in Education and Research (Appendix I). All participants provided written informed consent prior to participation.

2.1.4 Questionnaires

Participants completed an online screening survey prior to visiting the biomechanics laboratory. The screening survey assessed basic study eligibility including age, running habits, injury within the previous six months, comfort with running 10 km on a treadmill, etc. (Appendix II). Answers from the screening questionnaire were only used to determine potential study eligibility and were not analyzed. Participants also completed the Modified Physical Activity Readiness Questionnaire (PAR-Q)[97] to screen for underlying health conditions prior to performing running trials in the lab (Appendix III). Any "yes" answers on the PAR-Q disqualified participants from inclusion in the study. Participants completed a running history questionnaire during their laboratory visit which asked about participants' age, sex, leg dominance, weekly running volume, running frequency, years of running experience, previous injuries, etc. (Appendix IV). A previous injury was defined as a running-related injury which caused the participant to miss three or more planned running sessions.[98]

Leg dominance was determined by asking participants which leg they would use to kick a ball.

2.2 Procedures

2.2.1 Participant Screening

Participants were screened for foot strike type at the beginning of the first laboratory visit. The screening trials were used to assess study eligibility and also served as a warm-up and familiarization for eligible participants. For the screening trials, participants ran at a self-selected comfortable running speed that approximated their preferred running speed for a typical length run. Participants first ran in their own running shoes and then barefoot since footwear can influence preferred foot strike pattern.[95]

Participants ran over a pressure plate (HR Mat, TekScan, Norwood, MA, USA, 0.49 m 0.45 m 0.01 m, with 8,448 sensors, 185 Hz, 3.9 sensors/cm^2). The pressure plate was set in a 14.4-m ethylene-vinyl acetate (EVA) runway (0.02 m thick) with a gap where the pressure plate was located. An EVA runway was used due to its similarity to the midsole of a shoe and to encourage runners to retain their natural foot strike pattern. To monitor running speed, wireless infrared timing gates (TCi System, Brower Timing Systems, Draper, UT, USA) were placed 5-m apart and centered around the pressure plate. Participants did not receive feedback on running speed during the screening procedure.

At least three shod screening trials were conducted with the participant's entire left or right foot landing within the bounds of the pressure plate, no obvious targeting of the plate, and no obvious accelerating or decelerating. If the foot strike pattern differed between trials, more trials were collected until a majority pattern could be determined. Participants repeated the screening process while barefoot until at least three acceptable barefoot trials were collected. Participants qualified for the study if they ran with a rearfoot strike while both shod

and barefoot. A rearfoot strike was defined as the center of pressure during the first frame of stance being located in the rear third of the foot in the pressure data.

No instructions or feedback were given to participants regarding foot strike unless a participant demonstrated a rearfoot strike during the shod trials but switched foot strike types during the barefoot trials. In the case of switching to and maintaining a non-rearfoot strike, the participant was instructed to try landing with a rearfoot strike while running barefoot. The participant was excluded from the study if they reported that landing with a rearfoot strike did not feel natural during barefoot running. The average running speed during the barefoot screening trials was used to determine the participant's preferred running speed since the overground testing trials were conducted barefoot.

2.2.2 Marker Placement

Barefoot running was chosen for the overground running trials to allow for placement of markers directly on the foot. Fifteen retro-reflective spherical markers (six mm in diameter) were placed on participants' right foot in a modification of the Rizzoli multi-segment foot marker set.[99] The Rizzoli marker set allowed for multi-segment kinematic and kinetic analysis and was modified to facilitate measurement of the second ray. The marker set was modified by adding markers on the head of the middle phalanx of the second toe and on the base and head of the third metatarsal, while removing markers on the tibial tuberosity and fibular head (Appendices V and VI). Key markers include those on the second metatarsal head at the dorso-medial aspect of the second metatarso-phalangeal joint, second metatarsal base at the dorso-medial aspect of the second metatarso-cuneiform joint, and the head of the middle phalanx of the second toe at the most distal and dorsal point and were used to define the second metatarsal and second toe segments. Six-millimeter markers were used to minimize tracking challenges such as marker merging which can occur from the close placement of markers on the foot, particularly at the metatarsal heads.

Additional markers (10-12 mm) were positioned on the right medial and lateral malleoli, right knee medial and lateral joint lines (medial and lateral to the estimated average axis of rotation of the knee joint[100]), bilateral anterior superior iliac spines, iliac crests, posterior superior iliac spines, suprasternal notch, C7, and bilateral acromioclavicular joints. Rigid cluster shells with four noncolinear markers were placed on the right thigh and shank segments and wrapped with elastic bandages to minimize movement relative to the segment. Markers were placed on segments proximal to the foot to allow for investigation of proximal segment kinematics including knee and hip kinematics.

All markers were adhered directly to skin and not to clothing. Marker locations were marked on the skin with surgical pen to assist with marker replacement following the 10 km run and in the case of markers falling off. Markers were secured using double sided tape. Marker locations were all determined by the same researcher (NF) for consistency in marker placement. The right leg was arbitrarily selected for all participants. A static standing trial was taken before the start of the overground running trials to define reference marker positions and segment lengths.

2.2.3 Overground Running

All overground running trials were conducted barefoot to allow for the placement of markers directly on the foot and to eliminate the influence of footwear on running biomechanics.[95] Participants ran overground while kinematics data were collected in 3D at 185 Hz using 23 motion capture cameras (Oqus 500 and 700, Qualisys, Göteborg, Sweden) and ground reaction forces were recorded in 3D at 1850 Hz from a floor-embedded force plate (AMTI, Watertown, MA, USA). The pressure plate was located directly over the floor-embedded force plate so that ground reaction forces and plantar pressures could be recorded from the same running step. For all trials, the sampling rate of the motion capture cameras was the same as the pressure plate, with the force plate capturing at 10 times that frequency. For the first five participants, the pressure plate recorded at less than 185 Hz (92.5 to 174 Hz) due to technical

issues. Data from these first five participants were checked before being included in the analysis as detailed in the data analysis section.

A successful trial was defined as a trial in which the whole foot landed within the bounds of the pressure plate, the participant did not appear to target the plate, the running speed was within ± 5% of the target speed, and the participant landed with a rearfoot strike (the center of pressure was within the rear third of the foot in the first frame of stance). Five to eight successful overground trials were collected for each running condition to ensure at least five usable trials.

2.2.4 Running Conditions

During the first laboratory visit, after the screening trials to determine if participants were rearfoot strike runners, eligible participants performed six different running conditions. The running conditions were performed barefoot overground with and without a weight vest carrying 10% of the participant's body mass. The mass in the weight vest was evenly distributed between the front and back of the vest. Participants ran at three different running speeds: preferred speed (determined during the screening trials), 90% of preferred speed (slower speed), and 110% of preferred speed (faster speed). Trials at each running speed were performed both with and without the weight vest.

Five to seven successful trials were collected for each running condition during the first visit. The order of the running conditions was randomized for each participant using an online random sequence generator. Participants had a ~10 minute break halfway through testing on the first visit to complete the digital running history survey (Appendix IV). If a participant consistently switched to a non-rearfoot strike type at any point in the study, they were given verbal instructions to land with a rearfoot strike.

For the load carriage conditions, ten percent of body mass was chosen to investigate if even a relatively small increase in load carriage would increase metatarsal stress. The relatively

low mass was also chosen to ensure that it was not overly physically demanding for participants who may not be accustomed to running with added mass to complete all six running conditions.

On the second laboratory visit, five to eight successful barefoot overground trials were performed pre-10 km treadmill run at the same preferred running speed as the first visit. Another five to eight successful barefoot overground trials were conducted following the 10 km treadmill run, again at preferred speed.

2.2.5 10 km Treadmill Run

During the second visit, after the pre-treadmill overground trials, participants were asked to run for 10 kilometers on a non-instrumented treadmill. Participants ran in their own running shoes on the treadmill. The treadmill was set to 105% of each participant's self-reported maximal-effort time for a 10 kilometer run. The 10 km treadmill protocol was based on a previously published protocol which resulted in changes to running kinematics and kinetics.[87,88]

The treadmill speed was set to each participant's pre-determined speed for the duration of the 10 km run unless the participant requested a reduction in speed if they felt that they could not complete the 10 km at the current speed. Participants could request a reduction in speed as many times as they felt necessary. For each request, the speed was reduced by a set amount of 0.05 m/s. Participants rated their perceived exertion (RPE) on the Borg scale[101] every kilometer of the treadmill run. Participants were given a choice between an English or Norwegian version of the Borg scale (Appendix VII).

After completion of the treadmill run, shoes were removed, paper towels were used to remove sweat from the foot, and the foot and ankle markers were reattached using the pen marks for guidance. The transition time from the treadmill to the post-treadmill overground trials was recorded. A new static standing trial was captured, and the overground running

protocol was repeated for five to eight successful post-treadmill trials at the same preferred running speed (± 5%) as the pre-treadmill trials.

2.3 Data Processing

2.3.1 Signal Processing and Filtering

For the forces involved in the calculation of second metatarsal stress, the stance phase of running was defined using a force threshold of 5 N for both the pressure plate and force plate. The threshold was set to retain as many frames of data as possible from the pressure plate due to the lower capture frequency compared to the force plate. For lower extremity kinematics and kinetics, a vertical force threshold of 20 N was used to define stance phase and ground contact time as these data were not dependent on the pressure plate.

For data from the first visit, marker trajectories and ground reaction force data were filtered with a fourth-order lowpass Butterworth filter with cut-off frequencies of 12 and 50 Hz, respectively in Visual3D (HAS Motion, Kingston, Ontario, Canada). For data from the second visit, due to the analysis of joint moments, marker trajectories and GRF were filtered with a fourth-order Butterworth filter with cut-off frequencies of 20 Hz. Using the same cut-off frequencies for kinematic and force data is recommended to minimize signal noise while retaining physiological signals when analyzing joint moments using inverse dynamics.[102] For Paper III, data from the first visit was refiltered with a cut-off frequency of 20 Hz for both marker trajectories and force data to facilitate comparison the data from the second visit.

Data were interpolated to 101 frames of stance phase. Data from the five participants where the pressure plate capture frequency was less than 185 Hz were included in the analysis based on visual comparison of the summed total foot pressure data time-series with the vertical force plate data time-series. The within-participant study design also ensured that participants were compared to themselves, minimizing the potential influence of differences in capture frequency.

2.3.2 Calculation of Model Inputs

Strideway software (Strideway 7.80, TekScan, Norwood, MA, USA) was used to automatically apply regional anatomical masks to the pressure data to determine the second metatarsal and second toe regions. Regional masks were manually adjusted when necessary to ensure that each region of interest was fully contained within its respective mask. To try to improve consistency between trials, the minimum number of adjustments were made to define each region (Figure 4).

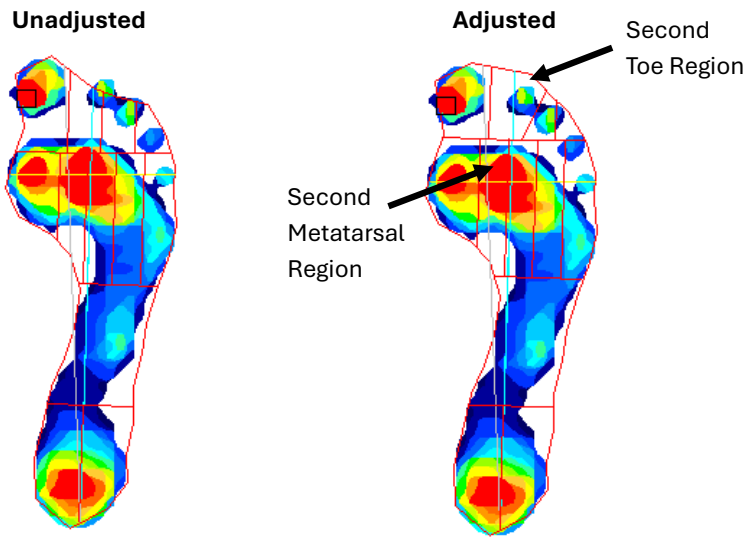


Figure 4: Example from a single trial demonstrating the manual adjustments made to the automatic regional masks in Strideway v7.80 (TekScan, Norwood, MA, USA). The left image shows the unadjusted regional masks and the right image shows the masks after manual adjustment for the second metatarsal and second toe regions.

To determine the vertical and horizontal ground reaction forces acting under the second metatarsal and second toe, the whole body vertical and horizontal GRFs were scaled

according to the percentage of force under those regions in the pressure data. The second metatarsal pitch angle was calculated by taking the inverse sine of the ratio of the vertical distance between the second metatarsal head and base markers divided by the 3D vector distance between the two markers. The length of the second metatarsal was defined as the 3D vector distance between the markers on the second metatarsal head and base, while the length of the second toe was defined as the 3D vector distance between the marker on the second metatarsal head and the head of the middle phalanx of the second toe.

2.3.3 Gait Event Detection

To calculate stride length and stride frequency, foot strikes were defined when the vertical velocity of the inferior posterior calcaneus marker crossed zero from negative to positive. Additionally, an anteroposterior heel marker velocity threshold of near zero (6.5×10^{-5}) was set to prevent false foot strike detections during swing phase. One stride per trial was analyzed with a minimum of five strides required per participant for inclusion in the analysis.

2.4 Beam Theory Model

2.4.1 Model Overview

A previously published 2D beam theory model of the second metatarsal was used to estimate dorsal and plantar stresses at the bone midshaft (Figure 5).[9] The model treated the metatarsal as a cantilevered beam fixed at the proximal end and was based on previous models from Stokes et al.[50] and Gross and Bunch.[10] The Stokes model estimated second metatarsal forces during walking using cadaver bone geometries scaled to each participant, forefoot orientation relative to the ground from 2D video, and vertical forces measured under the metatarsal and toe. Gross and Bunch modified the model to estimate second metatarsal stress and strain during running by representing the metatarsal as a hollow ellipse and using generic bone geometries from literature but did not incorporate metatarsal pitch angle. Ellison et al.[9] further developed the model to incorporate participant-specific bone

geometries from MRI, experimentally measured vertical and horizontal forces under the second metatarsal and toe, and metatarsal pitch angle from 3D motion capture data.

2.4.2 Model Development

Ellison et al.[9,70] modified the previous models by assuming that the short and long plantar tendons share force equally between them and can be represented by a single force vector acting in the direction of the midpoint between the tendons. Ellison's approach enabled a deterministic solution to the plantar tendon contributions, in contrast with Stokes et al. who gave upper and lower force bounds due to indeterminate tendon force distribution.

In contrast with previous models, Ellison et al. also incorporated horizontal forces under the second toe and metatarsal as inputs into the model in addition to vertical forces under these segments. However, the model is most sensitive to vertical force under the second metatarsal and the sensitivity to horizontal forces appears limited,[70] although these horizontal forces have been retained as inputs in the present thesis.

Additionally, Ellison et al. used participant-specific bone geometries based on MRI data. The second metatarsal was segmented from the MRI and a midshaft cross-section (at 50% of the metatarsal length) was analyzed. The cortical bone from the cross-section was divided into a series of triangles from which cross-sectional properties such as the area moment of inertia could be calculated using a custom automated MATLAB script.[9,70]

In the present thesis, mean bone geometries from MRI data in a cohort of 20 healthy recreational runners were used in the model.[9] Participant-specific second metatarsal and toe lengths determined from static trial marker positions were also used in the model.

2.4.3 Model Assumptions

Model assumptions are outlined in Stokes et al.[50] and Ellison et al.[9,70] The assumptions described in Stokes et al. and retained by Ellison et al. were:

- The metatarsophalangeal (MTP) joints were frictionless
- Forces due to acceleration of the components were negligible
- Forces from the toe extensor muscles and MTP joint ligaments were negligible
- The five foot rays acted independently of each other

Assumptions added by Ellison et al. include:

- External forces were applied at points under the distal metatarsal head and toe
- Mediolateral forces and kinematics were negligible
- Flexor muscle forces were represented by two plantar tendons (long and short) with equal load distribution
- The toe segment was parallel to the ground during ground contact
- Segment masses were negligible

2.4.4 Model Equations

As shown in the overall free-body diagram of the second ray (Figure 5), the model equations are as follows and were calculated in MATLAB (R2024b, MathWorks, Natick, MA, USA).

The forces from the short and long plantar tendons were represented by a single force acting in the direction of the midpoint between the two tendons. The angle between tendons was estimated to be 10° based on cadaver measurements,[103] giving the angle of plantar tendon action (θ):

$$\theta = \alpha - \frac{10^\circ}{2} \quad (1)$$

where α is the second metatarsal pitch angle relative to the horizontal.

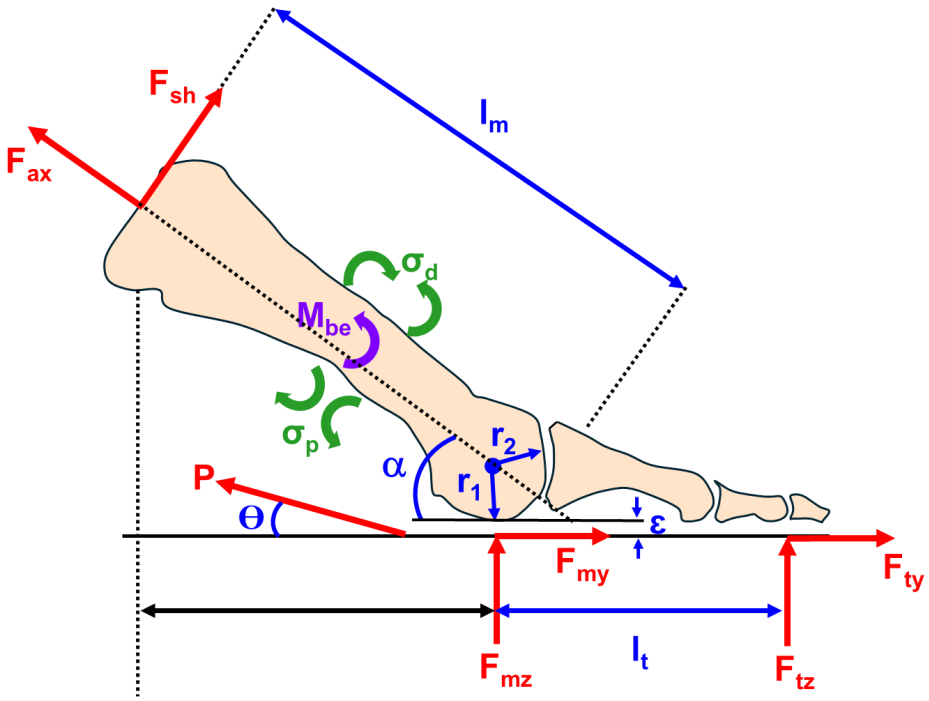


Figure 5: Overview of the 2D beam theory model used to estimate dorsal and plantar stresses at the second metatarsal midshaft. Model inputs included pitch angle (α) which represents the angle between the metatarsal and the horizontal, vertical (F_{mz}) and horizontal (F_{my}) forces under the metatarsal, vertical (F_{tz}) and horizontal (F_{ty}) forces under the second toe, metatarsal length (l_m), and toe length (l_t). Average MRI-derived bone geometries were used from a similar cohort from Ellison et al.[9] (see Appendix VIII) and included the lower (r_1) and upper (r_2) radii of the metatarsal head and the distance between the lower surface of the metatarsal head and the plantar tendon (ε). Calculated variables include the force due to the toe flexor muscles (P), the angle of plantar tendon action (θ), axial force (F_{ax}), shear force (F_{sh}), midshaft bending moment (M_{be}), and dorsal (σ_d) and plantar (σ_p) midshaft stresses. Figure adapted with permission from Ellison et al.[9]

The second toe was first considered in isolation to calculate the forces from the plantar tendons of the toe flexor muscles, flexor digitorum longus and brevis (Figure 6). The joint moments acting about the toe sum to zero in equilibrium, allowing for the calculation of both joint contact forces and plantar tendon forces. The second MTP joint moment in the sagittal plane was calculated from the vertical force under the second toe (F_{tz}) and the length of the second toe (l_t):

$$F_{tz} l_t = P(\varepsilon + r_1) \tag{2}$$

where P is the plantar tendon force acting at a distance of $r_1 + \varepsilon$ from the joint center, r_1 is the lower radius of the metatarsal head, and ε is the distance from the inferior surface of the metatarsal head to the start of the plantar tendon. The average geometries used for r_1 and ε are given in Appendix VIII.

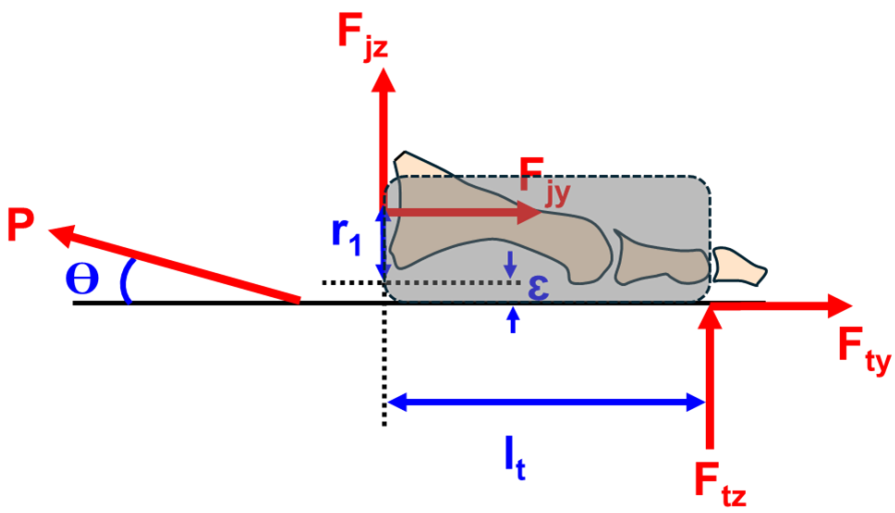


Figure 6: Free body diagram of the second toe (grey rectangle) showing forces (red) and geometric parameters (blue) used in the model. The vertical (F_{jz}) and horizontal (F_{jy}) joint reaction forces act at the MTP joint. Figure adapted with permission from Ellison et al.[8]

The joint reaction forces were calculated using the plantar tendon force, assuming static equilibrium in which the forces sum to zero:

$$F_{jz} + F_{tz} + P\sin\theta = 0 \quad (3)$$

$$F_{jy} + F_{ty} - P\cos\theta = 0 \quad (4)$$

where the F_{jz} and F_{jy} are the vertical and horizontal joint reaction forces, respectively, and F_{tz} and F_{ty} are the vertical and horizontal second toe forces, respectively.

After the calculation of joint reaction forces, the metatarsal can be considered (Figure 7). Vertical and horizontal forces acting on the second metatarsal were resolved into axial and shear components relative to the long axis of the bone, yielding:

$$F_{ax} = F_{mz}\sin\alpha - F_{my}\cos\alpha - F_{jz}\sin\alpha + F_{jy}\cos\alpha \quad (5)$$

$$F_{sh} = F_{mz}cos\alpha + F_{my}\sin\alpha - F_{jz}\cos\alpha - F_{jy}\sin\alpha \quad (6)$$

where F_{ax} and F_{sh} are the axial and shear forces acting on the metatarsal, respectively, and F_{mz} and F_{my} are the vertical and horizontal forces acting under the second metatarsal, respectively.

The midshaft bending moment for the second metatarsal was determined from the shear force and the distance from the center of the metatarsal head to the metatarsal midshaft:

$$M_{be} = \left(\frac{l_m - r_2}{2}\right) F_{sh} \quad (7)$$

where M_{be} is the midshaft bending moment, l_m is the length of the metatarsal, and r_2 is the upper radius of the metatarsal head.

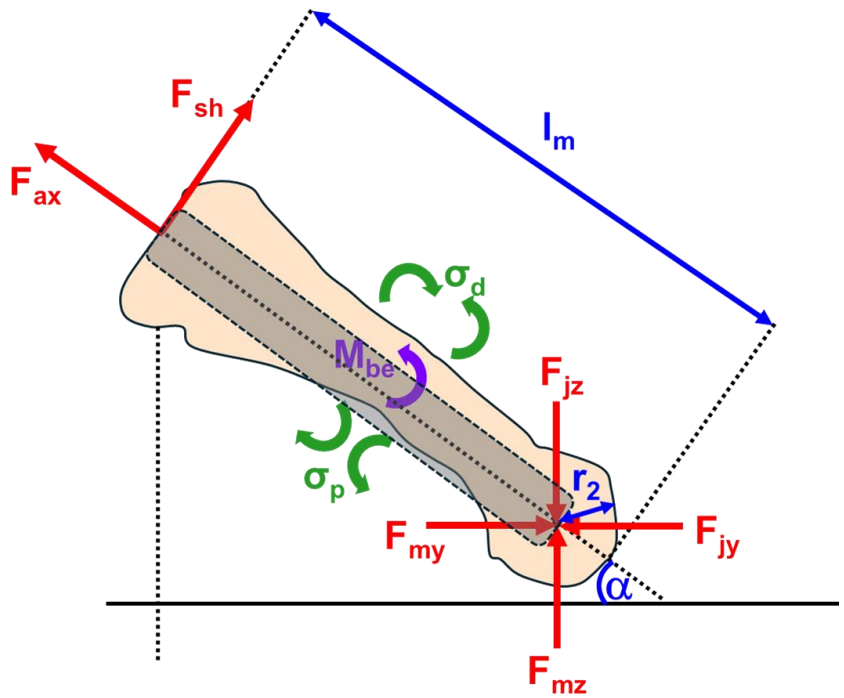


Figure 7: Free body diagram of the second metatarsal (grey rectangle) showing forces (red), geometric parameters (blue), the midshaft bending moment (purple), and dorsal and plantar stresses (green). The vertical and horizontal metatarsal and joint reaction forces are resolved into an axial force acting along the long axis of the bone and a shear force acting perpendicular to the long axis. Figure adapted with permission from Ellison et al.[100]

Axial stress (σ_{ax}) was calculated from the axial force and the average second metatarsal bone cross-sectional area (*CSA*) obtained from a cohort of 20 recreational runners (Appendix VIII).

$$\sigma_{ax} = \frac{F_{ax}}{CSA} \tag{8}$$

Bending stress (σ_{be}) was calculated from the midshaft bending moment, the radial distance from the center of the bone to the surface (r), and the area moment of inertia about the horizontal axis of the bone cross-section (I_{xx}).

$$\sigma_{be} = \frac{M_{be} r}{I_{xx}} \tag{9}$$

The second metatarsal midshaft stresses at the dorsal (σ_d) and plantar (σ_p) surfaces of the bone were calculated by summing and subtracting the axial and bending stresses, respectively.

$$\sigma_d = \sigma_{ax} + \sigma_{be} \tag{10}$$

$$\sigma_p = \sigma_{ax} - \sigma_{be} \tag{11}$$

2.5 Data Analysis

2.5.1 Overview of Outcome Variables

Peak second metatarsal dorsal and plantar stresses were the primary outcome variables in **Papers I** and **II**, while the repeatability and reliability of stress estimates within and between visits were assessed in **Paper III**. Stress-related variables were those involved in the calculation of bone stress, such as axial force and pitch angle, and were also analyzed in each article to provide context to the bone stress results. Additional non-stress-related metrics such as stride characteristics and lower-extremity kinematics and kinetics were calculated in the articles to further contextualize the bone stress results. Data were processed in Visual3D (HAS Motion, Kingston, Ontario, Canada) and MATLAB, and statistical analyses were performed in R (v4.5.1, R Foundation for Statistical Computing, Vienna, Austria).

Peak dorsal and plantar stresses were selected as the primary outcome variables because stress magnitude is a primary determinant of the probability of bone failure, with higher stress values associated with increased failure probability.[104,105] Dorsal and plantar stresses were expressed in megapascals (MPa) and were multiplied by negative one, so that a positive value indicated tensile stress and a negative value indicated compressive stress, consistent with engineering convention.

2.5.2 Running Speed and Load Carriage Analysis (Paper I)

Linear mixed-effects models were used to assess the influence of running speed (90% of preferred, preferred, 110% of preferred) and load carriage (body mass, 110% body mass) on peak dorsal and plantar stresses, with participant included as a random effect. Two-way repeated measures ANOVAs were used to assess the significance of fixed effects. In the case of significant interaction effects, simple effects were evaluated by comparing levels of one factor within each level of the other. In the absence of significant interaction effects, comparisons were performed on the main effects average over the other factor. When significant main effects were found, post hoc comparisons were conducted using estimated marginal means without adjustment for multiple comparisons.

To quantify between-participant variability in bone stress, coefficients of variation (CV) were calculated as the standard deviation (SD) divided by the mean for peak dorsal and plantar stresses. Pearson correlations were conducted to examine associations between peak dorsal stress and selected loading variables for the baseline condition (preferred speed without load carriage). The strength of the correlations were interpreted as follows: 0.10-0.39 weak, 0.40-0.69 moderate, 0.70-0.80 strong, and 0.90-1.00 very strong.[106]

2.5.3 Pre- and Post-10 km Treadmill Run Analysis (Paper II)

To compare differences pre- and post-10 km treadmill run, paired t-tests were conducted when data were normally distributed. Cohen's d effect sizes were calculated to quantify the

magnitude of differences. For non-normally distributed variables, paired Wilcoxon signed-rank tests were used to compare pre- and post-10 km run differences. Wilcoxon effect sizes (r) assessed the magnitude of effects. Statistical significance was set at $\alpha = 0.05$ for all tests unless otherwise noted.

To allow for comparison with previous research on running-induced fatigue, lower-extremity joint angles and moments were analyzed pre- and post-10 km treadmill run. Joint angles were defined as the distal segment relative to the proximal segment using an XYZ Cardan sequence, with neutral joint positions defined in the static standing trials. The first MTP sagittal plane angle was defined as the planar angle between the vectors defined by the first metatarsal segment and the hallux segment projected into the sagittal plane with positive angles indicating MTP dorsiflexion.[107] The second MTP sagittal plane angle was the planar angle between the second metatarsal segment and the second toe segment projected into the sagittal plane with positive angles indicating dorsiflexion. External joint moments including ankle plantarflexion and first MTP dorsiflexion moments were calculated using Newton-Euler inverse dynamics resolved in the coordinate system of the proximal segment.[108] For the first MTP joint moment, reference values were used to estimate foot segment masses as a proportion of each participant's body mass.[109] The first MTP joint moment was analyzed during only the last 50% of the stance phase since participants were rearfoot strikers and the center of pressure did not approach the MTP joint until late stance.

2.5.4 Repeatability and Reliability Analysis (Paper III)

Intraclass correlation coefficients (ICC) were calculated to assess within-day repeatability and between-day reliability for peak stresses, stress-related variables, and stride characteristics. ICCs were calculated using the irr package[110] in R for a two-way mixed-effects model with absolute agreement and average measures. ICC values were interpreted as poor (< 0.5), moderate (0.5 – 0.75), good (0.75 – 0.9), or excellent (> 0.9).[111]

Bone stress for each visit was calculated using the model, with second metatarsal and second toe segment lengths from the static trial of the given visit used as inputs. To assess how variability in marker-based segment length estimates influenced dorsal and plantar stresses, bone stress was also computed using constant bone lengths by applying the first-visit lengths to both visits.

The change in relative marker positions between days indicates variability in marker placement between days, which can influence segment length estimates. The relative distances between the second metatarsal head and base markers and reference markers (first and fifth metatarsal head and posterior inferior calcaneus markers) were determined by calculating the 3D vector distances between these markers. Mean and percent differences of relative marker distances were calculated.

Stride characteristics including ground contact time, stride length, and stride frequency were calculated to determine the repeatability and reliability of running gait and to understand how variability in running technique might contribute to the repeatability and reliability of bone stress estimates.

3 Results

3.1 Participants and Testing Conditions

All 28 participants completed testing on both days, with a mean (± SD) of 21 ± 26 days (range: 1 – 139 days) between laboratory visits. Participant characteristics are summarized in Table 1. A small number of trials were excluded from the analysis due to protocol deviations (e.g., non-rearfoot strike pattern).

Running speeds were a mean (± SD) of 3.25 ± 0.36 m/s for preferred speed, 2.93 ± 0.32 m/s for the slower speed, and 3.58 ± 0.41 m/s for the faster speed. The treadmill running speed (3.74 ± 0.48 m/s) was faster than preferred speed for all participants.

All participants completed the 10 km treadmill run. The total running time on the treadmill was a mean (± SD) of 45:16 ± 6:02 minutes (range: 35:30 – 63:00 min), with five participants requesting at least one speed reduction. During the treadmill run, participants' RPE started at a mean (± SD) of 12.5 ± 1.1 out of 20 on the Borg scale in the first kilometer and increased to 16.6 ± 1.9 at 10 km which corresponds to a perceived exertion of "very hard". The transition time from the treadmill to the overground trials, during which shoes were removed and foot and ankle markers were reapplied, lasted a mean of 4:55 ± 1:11 minutes (range: 3:22 – 8:30 min).

Table 1: Participant information presented as mean ± standard deviation with range in parenthesis for all participants.

	Participants (n=28)
Age (years)	29.6 ± 4.7 (23 - 40)
Height (m)	1.73 ± 0.09 (1.54 - 1.90)
Mass (kg)	68.8 ± 9.7 (50.0 - 85.8)
Weekly Running Volume (km/week)	48.0 ± 24.9 (15 - 115)
Running Experience (years)	7.8 ± 5.6 (2 - 22)

3.2 Within-Day Repeatability and Between-Day Reliability of Second Metatarsal Stress Estimates (Paper III)

Before considering the results of Papers I and II, it is important to establish the reliability of the beam theory model used to estimate bone stress. Since metatarsal stress estimates are calculated from geometric and experimentally measured kinetic and kinematic inputs, the reliability of the model depends on the reliability of the model inputs. Quantifying the within-day repeatability and between-day reliability of bone stress estimates helps to contextualize whether changes in metatarsal stress observed in Papers I and II reflect true experimental effects or variability in the model inputs.

3.2.1 Repeatability and Reliability of Metatarsal Stress Estimates and Related Variables

Within-day repeatability was excellent for peak second metatarsal dorsal and plantar stresses (ICC > 0.9) (Table 4). Between-day reliability was also excellent for peak dorsal and plantar stresses (Table 4).

Within-day repeatability was excellent for peak midshaft bending moment and normalized peak whole body vertical GRF. Within-day repeatability was good for peak shear force, peak axial force, peak vertical and horizontal force under the second metatarsal, second metatarsal pitch angle at the time of peak stress, normalized peak whole body horizontal GRF, percent of GRF under the second metatarsal at the time of peak stress, and maximum percent of GRF under the second metatarsal.

Between-day reliability was excellent for all stress-related variables except for second metatarsal and second toe lengths (calculated form static trial marker positions) which showed moderate reliability (Table 4). Second metatarsal length was longer by a mean of 1.9% (mean difference ± standard error (SE) of the difference: 1.2 ± 1.2 mm) when measured on the second visit compared to the first visit, while second toe length was longer by a mean

of 2.6% (mean difference ± SE: 1.2 ± 0.9 mm). When the same second metatarsal length was used for both visits, between-day reliability increased by 6.0% for peak dorsal stress and 7.0% for plantar stress (Table 5). When the same second toe length was used for both visits, between-day reliability did not change for peak dorsal stress and increased by 0.1% for peak plantar stress (Table 5).

In addition to stress-related variables, gait characteristics including stride length, stride frequency, and ground contact time showed excellent between-day reliability and within-day repeatability.

Table 2: Within- and between-day intraclass correlation coefficients for second metatarsal stress and key variables involved in the estimation of stress.

Variable	Visit 1 ICC	Visit 2 ICC	Between-Day ICC
Peak Dorsal Stress (MPa)	0.916	0.930	0.932
Peak Plantar Stress (MPa)	0.919	0.933	0.924
Second Metatarsal Pitch Angle at TPS (°)	0.851	0.933	0.936
Peak vGRF under Second Metatarsal (N)	0.882	0.890	0.979
Second Metatarsal Length (mm)	---	---	0.656
Second Toe Length (mm)	---	---	0.680

MPa = megapascals. TPS = time of peak stress. vGRF = vertical ground reaction force. ICC = intraclass correlation coefficient.

Table 3: Between-day reliability for second metatarsal dorsal and plantar stresses using fixed- compared to visit-specific bone lengths. Fixed-length conditions used first-visit second metatarsal and second toe lengths for both visits.

Variable	Between-Day ICC	% Δ ICC
Peak Dorsal Stress, Visit-Specific Lengths (MPa)	0.932	---
Peak Plantar Stress, Visit-Specific Lengths (MPa)	0.924	---
Peak Dorsal Stress, Fixed Second Metatarsal Length (MPa)	0.988	6.0
Peak Plantar Stress, Fixed Second Metatarsal Length (MPa)	0.989	7.1
Peak Dorsal Stress, Fixed Second Toe Length (MPa)	0.932	0.0
Peak Plantar Stress, Fixed Second Toe Length (MPa)	0.925	0.1

MPa = megapascals. ICC = intraclass correlation coefficient. % Δ = percent difference from visit-specific lengths.

3.2.2 Between-Day Reliability of Relative Marker Distances

To assess potential sources of variability in second metatarsal and second toe length measurements, between-day reliability of relative marker distances was examined for the second toe and second metatarsal head and base markers. When comparing relative marker distances between days, reliability was moderate to excellent (ICC range: 0.579 – 0.932). The second toe marker showed greater reliability than the second metatarsal head or base markers (Table 6). The mean difference in relative marker positions between days was less than 2.5 millimeters or 2.1% (Table 6).

Table 4: Between day differences and reliability of relative 3D vector distances between motion capture markers. Mean and percent differences were calculated as Visit 2 relative to Visit 1, with positive differences indicating greater distances for Visit 2.

Marker	Relative To	Mean Difference ± SE (mm)	% Difference	ICC
SMH	FMH	0.4 ± 0.7	1.0	0.621
	VMH	0.6 ± 0.8	0.8	0.698
	CAL	-0.1 ± 0.7	-0.0	0.932
SMB	FMH	1.4 ± 1.1	2.1	0.579
	VMH	1.1 ± 1.0	1.3	0.734
	CAL	-1.5 ± 0.9	-1.0	0.868
ST	FMH	0.7 ± 0.6	1.7	0.812
	VMH	0.9 ± 0.7	0.8	0.862
	CAL	2.4 ± 0.4	0.3	0.987

SMH = second metatarsal head. SMB = second metatarsal base. ST = second toe. FMH = first metatarsal head. VMH = fifth metatarsal head. CAL = posterior inferior calcaneus.

SE = standard error of the difference. ICC = intraclass correlation coefficient.

3.3 Effects of Running Speed and Load Carriage on Bone Stress (Paper I)

3.3.1 Effects of Speed and Load Carriage on Metatarsal Stress

Peak second metatarsal dorsal and plantar stresses increased with both increasing running speed and added load carriage, with no interaction between these factors (Table 2). Changing running speed by 10% had a significant effect on both peak dorsal stress ($F(2, 157) = 27.35$, $p < 0.001$, $\eta_p^2 = 0.26$) and peak plantar stress ($F(2, 157) = 27.66$, $p < 0.001$, $\eta_p^2 = 0.26$). Adding load carriage of 10% body mass also had a significant effect on both peak dorsal stress ($F(1, 157) = 22.56$, $p < 0.001$, $\eta_p^2 = 0.13$) and peak plantar stress ($F(1, 157) = 18.68$, $p < 0.001$, $\eta_p^2 = 0.11$).

When averaged across loading conditions, reducing running speed by 10% from preferred to slower speed decreased peak dorsal stress by 2.10 ± 5.21% (2.54 ± 6.5 MPa, 95% confidence interval (CI): [0.76, 4.32], p = 0.030) (Figure 8a) and plantar stress by 2.32 ± 5.08% (2.59 ± 5.88 MPa, 95% CI: [0.98, 4.19], p = 0.017). When averaged across loading conditions, increasing running speed by 10% from preferred to faster speed increased peak dorsal stress by 4.22 ± 5.70% (5.74 ± 7.73 MPa, 95% CI: [3.61, 7.87], p < 0.001) and peak plantar stress by 4.20 ± 5.69% (5.17 ± 6.88 MPa, 95% CI: [3.28, 7.07], p < 0.001).

When averaged across speeds, increasing load from body mass to 110% of body mass increased peak dorsal stress by 3.13 ± 6.32% (4.37 ± 8.61 MPa, 95% CI: [2.47, 6.28], p < 0.001) (Figure 8b) and peak plantar stress by 2.87 ± 6.33% (3.68 ± 7.95 MPa, 95% CI: [1.92, 5.44], p < 0.001).

3.3.2 Variability Between Participants in Peak Stress

Between-participant variability in peak dorsal stress, quantified using the coefficient of variation, was 24.6 ± 0.9% (range: 23.6 – 25.7%) when averaged across the six running conditions. Between-participant variability for peak plantar stress, quantified using the CV, was 25.3 ± 0.9% (range: 24.3 – 26.5%) when averaged across the six conditions. Visualizing peak dorsal stress across running speeds (Figure 9a-b), both non-normalized and normalized to body mass, demonstrates the between-participant variability in stress magnitudes and within-participant changes across running speeds.

Table 5: Peak second metatarsal dorsal and plantar stresses across six running speed and load carriage conditions. Data are presented as mean ± standard deviation with confidence intervals displayed in brackets. Negative stress values indicate compressive stress and positive values indicate tensile stress.

Stress (MPa)	Preferred Speed	Preferred Speed + 10% Mass	Faster Speed	Faster Speed + 10% Mass	Slower Speed	Slower Speed + 10 % Mass	P value		
							Speed x Load	Speed	Load
Peak Dorsal Stress	-138 ± 32	-143 ± 35	-141 ± 32	-147 ± 38	-135 ± 35	-139 ± 36	0.882	<0.001*	<0.001*
Peak Plantar Stress	125 ± 30	130 ± 33	128 ± 30	133 ± 35	122 ± 33	126 ± 33	0.722	<0.001*	<0.001*

MPa= megapascals. *= significant effect ($p < 0.05$).

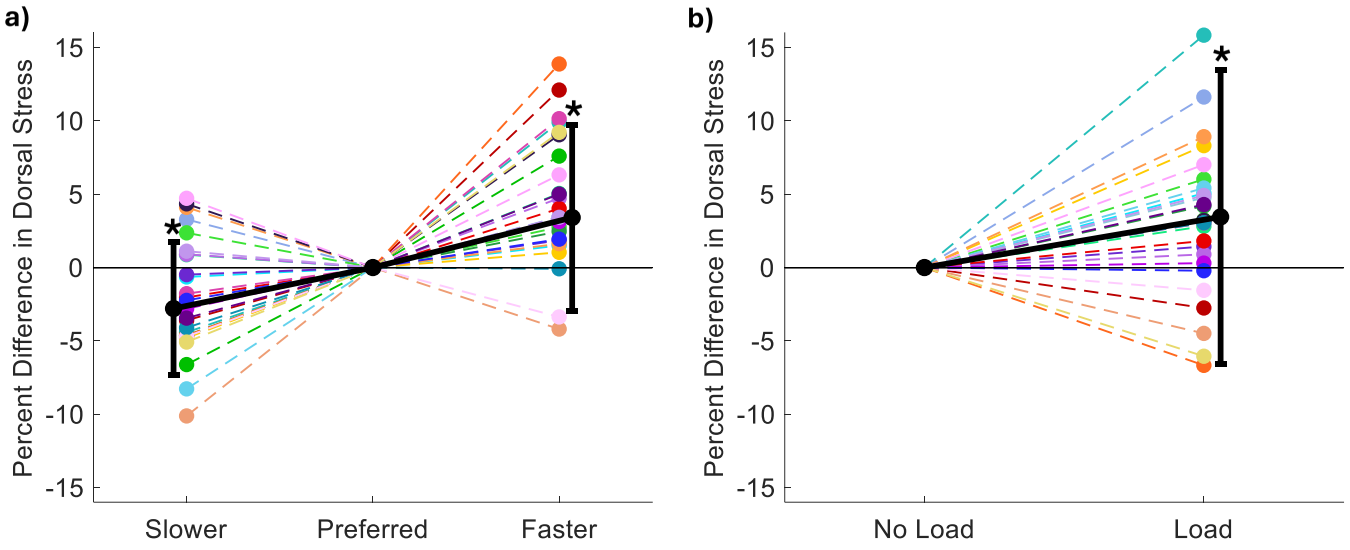


Figure 8: Percent differences in peak dorsal stress for (a) running speed, (b) load carriage, and (c) pre- and post-10 km treadmill run conditions. Colored circles represent participant means (averaged across trials) and black circles represent group means. Black error bars represent ± standard deviation. An asterisk indicates a significant difference from the reference condition. **a)** Slower and faster speeds relative to preferred speed (averaged across loading conditions). **b)** Loaded relative to unloaded condition (averaged across running speeds). **c)** Post- relative to pre-10 km run.

3.3.3 Correlations Between Peak Dorsal Stress and Potential Explanatory Variables

Peak dorsal stress was very strongly correlated with peak second metatarsal vertical force when running at preferred speed without load carriage, ($r(26) = 0.91$, $p_{adj} < 0.001$, 95% CI: [0.80, 0.96] (Figure 10a)). Peak dorsal stress and peak horizontal second metatarsal force were moderately correlated, ($r(26) = 0.64$, $p_{adj} < 0.001$, 95% CI: [0.35, 0.82]). The correlation between peak dorsal stress and vertical force under the second toe was weak, ($r(26) = 0.36$, $p_{adj} = 0.058$, 95% CI: [-0.01, 0.65]), and the correlation between peak dorsal stress and horizontal force under the second toe was also weak, ($r(26) = 0.37$, $p_{adj} = 0.058$, 95% CI: [-2.6×10^{-5}, 0.65]). Peak vertical whole-body ground reaction force was moderately correlated with peak dorsal stress when running at the preferred speed without load carriage, ($r(26) = 0.66$, $p_{adj} < 0.001$, 95% CI: [0.38, 0.83] (Figure 10b)). Running speed was moderately correlated with peak dorsal stress, ($r(26) = 0.46$, $p_{adj} = 0.018$, 95% CI: [0.11, 0.71] (Figure 10c)), while body mass was also moderately correlated with peak dorsal stress, ($r(26) = 0.48$, $p_{adj} = 0.015$, 95% CI: [0.14, 0.73] (Figure 10d)).

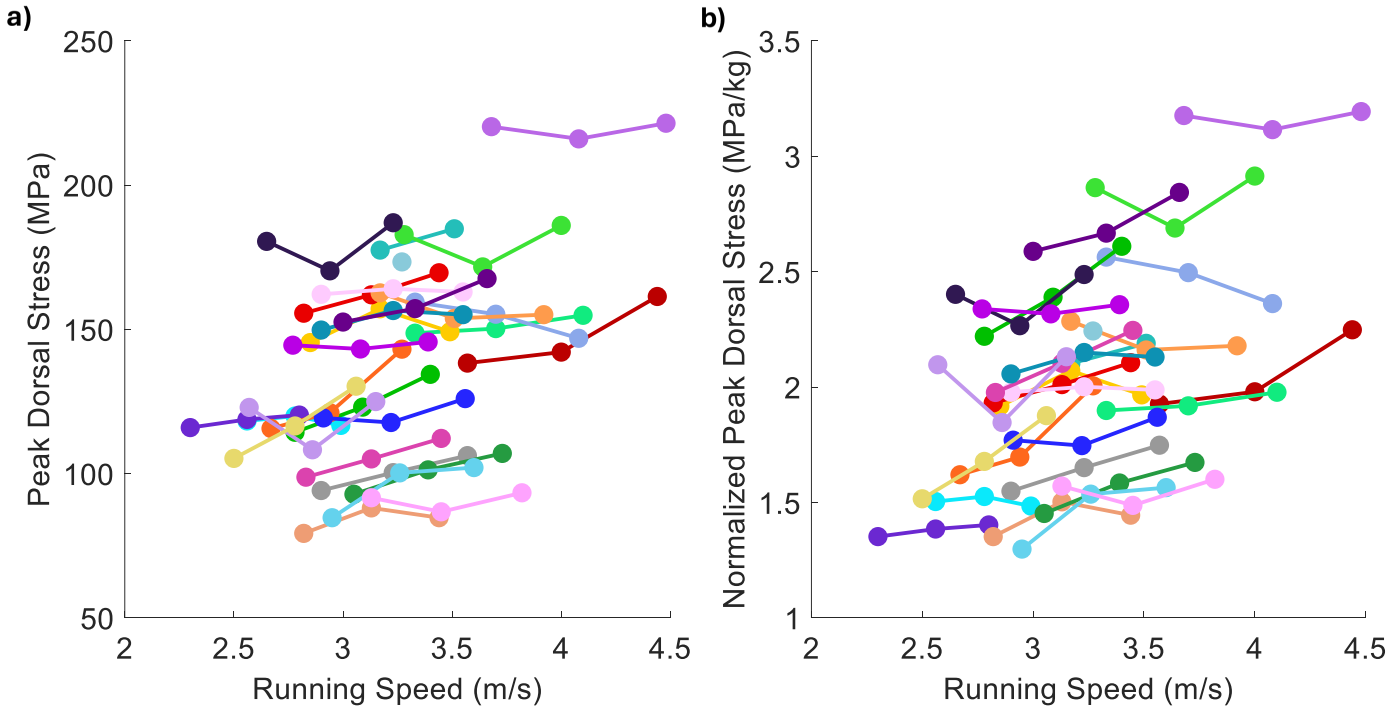


Figure 9: Average peak dorsal stress across running speeds (slower, preferred, and faster) for the no-load conditions. Each participant is shown in a different color, with circles representing participant means for each speed condition. **a)** Peak dorsal stress across running speeds, **b)** Peak dorsal stress normalized to body mass.

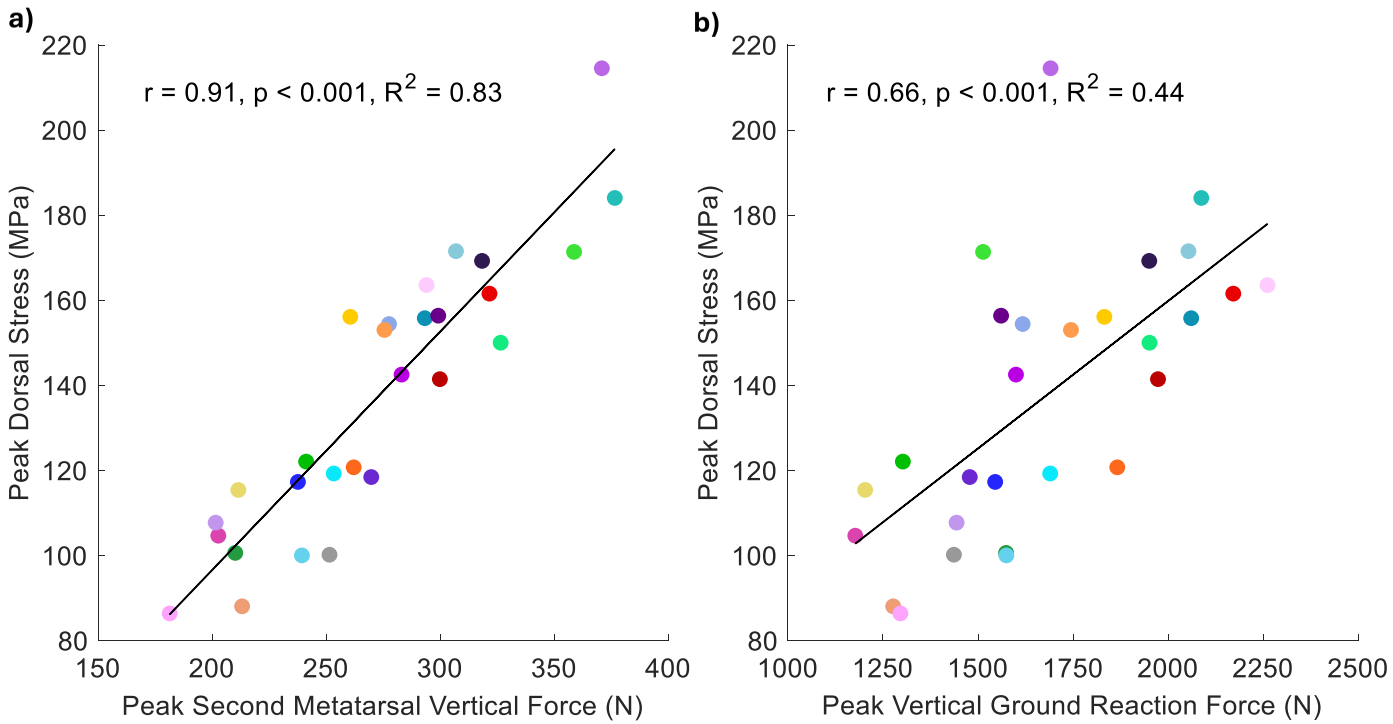
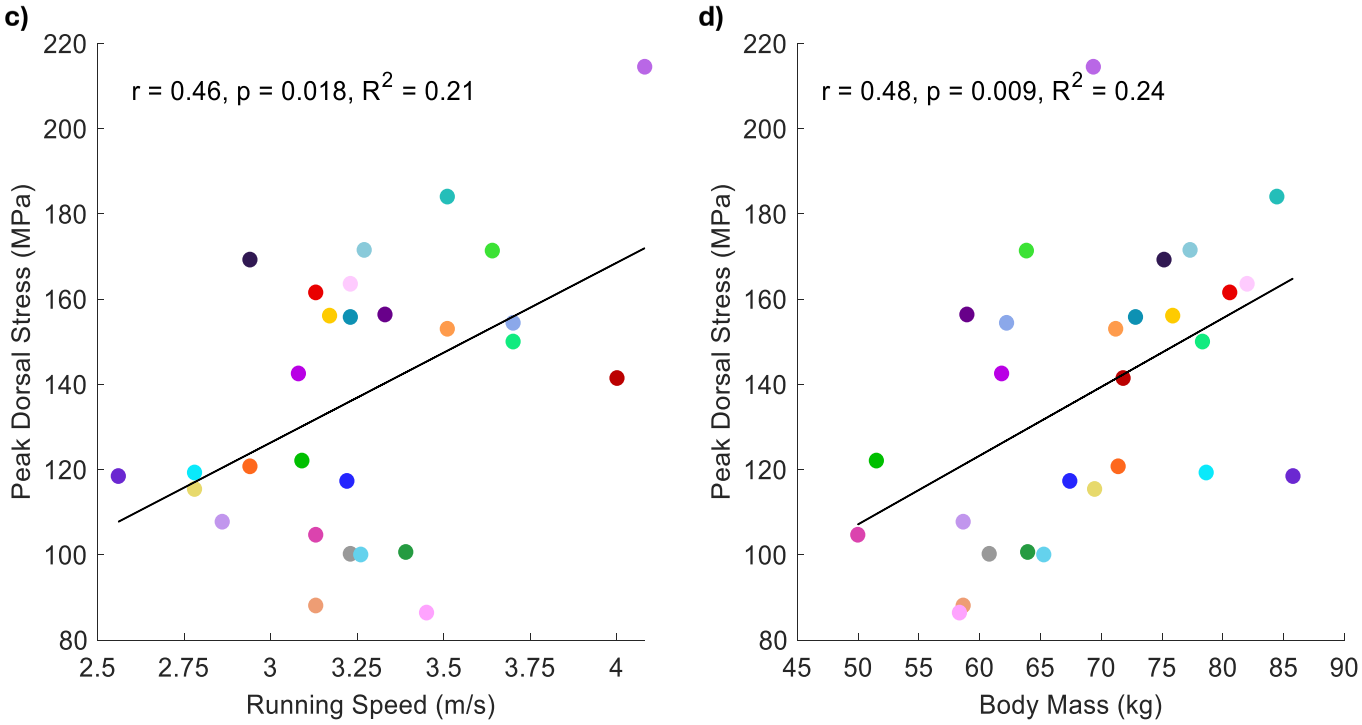


Figure 10: Correlation plots for the preferred speed no load condition. Each colored circle represents a different participant. Peak dorsal stress values are absolute values. **a)** Very strong positive correlation between peak dorsal stress and peak second metatarsal vertical force (r = 0.91). **b)** Moderate positive correlation between peak dorsal stress and peak vertical ground reaction force (r = 0.66). **c)** Moderate positive correlation between peak dorsal stress and preferred running speed (r = 0.46). **d)** Moderate positive correlation between peak dorsal stress and body mass (r = 0.48).


c)
r = 0.46, p = 0.018, R2 = 0.21
Peak Dorsal Stress (MPa)
Running Speed (m/s)
d)
r = 0.48, p = 0.009, R2 = 0.24
Peak Dorsal Stress (MPa)
Body Mass (kg)

3.4 Effect of a 10 km Treadmill Run on Second Metatarsal Stress (Paper II)

3.4.1 Effect of a 10 km Run on Metatarsal Stress

No significant difference was observed in second metatarsal peak dorsal stress post- compared to pre-10 km treadmill run ($t(27) = 1.88$, $p = 0.071$, $d = 0.05$) (Table 3). Peak plantar stress also did not change significantly post- compared to pre- treadmill run ($t(27) = 0.84$, $p = 0.409$, $d = 0.02$) (Table 3).

Peak bending stress, midshaft bending moment, and shear force were unchanged post- compared to pre-treadmill run (Table 3), while peak axial stress and force decreased by 9% (stress: 0.60 ± 0.81 MPa; force: 22.8 ± 30.4 N) ($t(27) = 3.96$, $p < 0.001$, $d = 0.33$). Second metatarsal pitch angle at the time of peak stress decreased by a mean of 0.8 ± 2.0° post- compared to pre-10 km run ($t(27) = 2.12$, $p = 0.043$, $d = 0.21$) (Table 3), indicating a less inclined metatarsal post-run. Peak vertical and horizontal force under the second metatarsal were unchanged post- compared to pre-run (Table 3).

Peak MTP vertical joint reaction force (JRF) decreased by 21% (23.2 ± 30.3 N) ($t(27) = 4.06$, $p < 0.001$, $d = 0.51$), while peak MTP horizontal JRF decreased by 13% (13.0 ± 22.4 N) ($t(27) = 3.07$, $p = 0.005$, $d = 0.33$). Peak vertical force under the second toe decreased by 15% (5.07 ± 7.91 N) ($z = 3.03$, $p = 0.002$, $r = 0.57$), while peak horizontal force under the second toe decreased by 26% (2.62 ± 3.45 N) ($z = 3.53$, $p < 0.001$, $r = 0.67$). Peak plantar tendon force decreased by 14% (17.0 ± 27.3 N) ($t = 3.31$, $p = 0.003$, $d = 0.37$).

Table 6: Peak second metatarsal stresses, forces, and pitch angle presented as mean ± standard deviation. Effect sizes are Cohen's d unless indicated. Negative stress values indicate compressive stress.

Variable	Pre-Run	Post-Run	P value	Effect Size
Peak Dorsal Stress (MPa)	-144 ± 37	-142 ± 35	0.071	0.05
Peak Plantar Stress (MPa)	131 ± 34	130 ± 33	0.409	0.02
Axial Stress (MPa)	6.8 ± 2.0	6.2 ± 1.7	<0.001*	0.33
Bending Stress (MPa)	138 ± 35	136 ± 34	0.168	0.04
Peak Shear Force (N)	266 ± 54	264 ± 50	0.192	0.05
Peak Axial Force (N)	256 ± 74	233 ± 63	<0.001*	0.33
Second Metatarsal Pitch Angle (°)†	29.5 ± 3.7	28.8 ± 3.7	0.043*	0.21
Peak Second Metatarsal vGRF (N)	279 ± 56	276 ± 53	0.114	0.05
Peak Second Metatarsal hGRF (N)	51.4 ± 15.4	53.1 ± 15.2	0.196	-0.11
Peak MTP Vertical JRF (N)	111 ± 50	88 ± 40	<0.001*	0.51
Peak MTP Horizontal JRF (N)	103 ± 44	90 ± 35	0.005*	0.33
Peak Second Toe vGRF (N)	34.8 ± 14.7	29.7 ± 11.5	0.002* [a]	0.57 [b]
Peak Second Toe hGRF (N)	10.2 ± 5.4	7.6 ± 4.4	<0.001* [a]	0.67 [b]
Plantar Tendon Force (N)	124 ± 51	107 ± 42	0.003*	0.37

MPa = megapascals. N = newtons. v = vertical. h = horizontal. GRF = ground reaction force. JRF = joint reaction force.

* = significant difference post- compared to pre-10 km run ($p < 0.05$). † = at the time of peak stress. a = non-parametric Wilcoxon sign-rank test performed; b = non-parametric Wilcoxon effect size.

3.4.2 Effects of a 10 km Run on Lower-Extremity Kinematics and Kinetics

Peak ankle plantarflexion moment decreased by a mean of 6% (-11.7 ± 16.9 Nm) post- compared to pre-10 km treadmill run ($t(27) = -3.68$, $p = 0.001$, $d = -0.25$) (Figure 11a). Peak second MTP dorsiflexion moment decreased by 14% (-0.24 ± 0.35 Nm) post- compared to pre-10 km run ($t(27) = 3.52$, $p = 0.002$, $d = 0.38$) (Figure 11b). Peak first MTP dorsiflexion angle increased by a mean of 3.3 ± 4.9° post- compared to pre-10 km run ($t(27) = -3.62$, $p = 0.001$, $d = -0.56$) (Figure 11c).

There was no change in peak rearfoot eversion angle, peak knee abduction angle, peak hip adduction angle, peak knee internal rotation angle, sagittal plane angle between the foot and ground at initial contact, peak second MTP dorsiflexion angle, or peak first MTP dorsiflexion moment post- compared to pre- 10 km treadmill run.

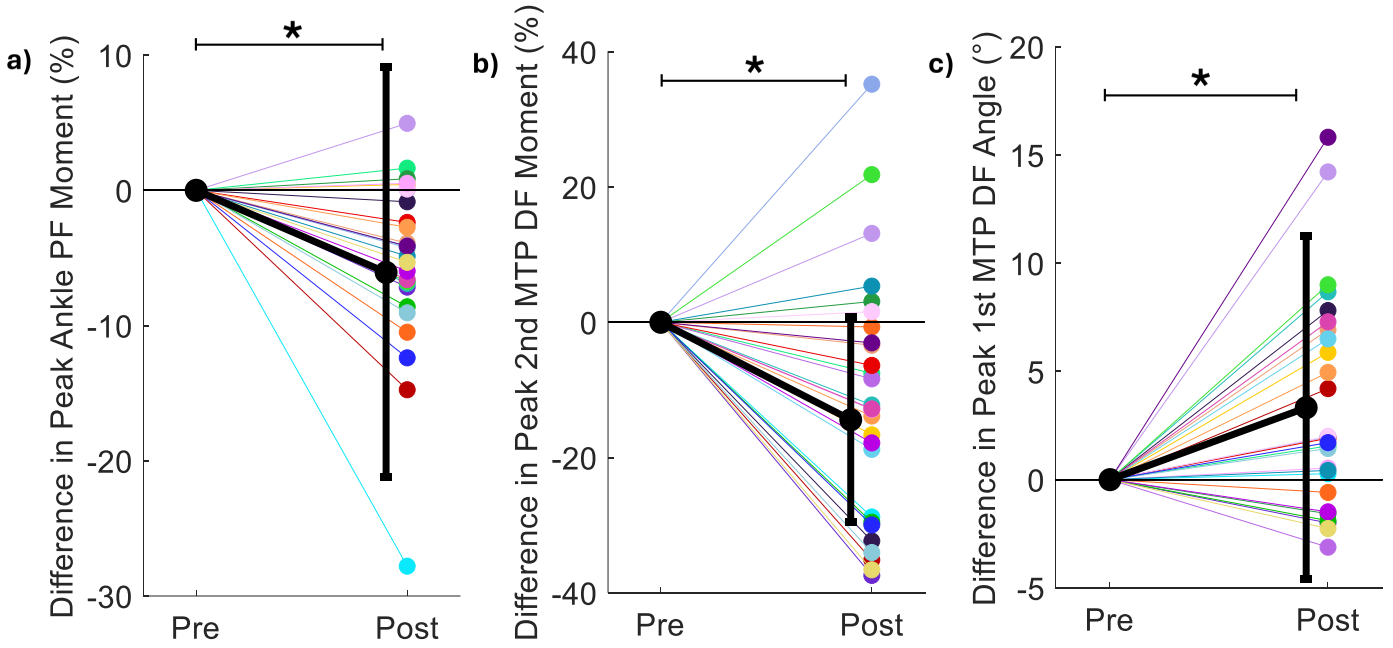


Figure 11: Percent and mean differences in kinematic and kinetic variables post- compared to pre-10 km treadmill run. Individual participants are shown in different colors. Larger black circles show group means while vertical bars show standard errors of the differences. * = significant difference compared to pre trials. Percent differences are shown for **a)** peak ankle plantarflexion (PF) moment and **b)** peak second metatarsophalangeal (MTP) dorsiflexion (DF) moment. Mean difference is shown for **c)** peak first MTP dorsiflexion angle.

4 Discussion

4.1 Overall Findings

The overall findings from this thesis indicate that *in vivo* second metatarsal bone stress during running is influenced by relatively small changes in external loading conditions including running speed and load carriage, while remaining stable when external loading remained unchanged. This is further supported by the very strong correlation between vertical GRF under the second metatarsal and second metatarsal bone stress, indicating that localized external forces may be closely related to internal loading in the second metatarsal. This close association may also indicate an overreliance of the model on vertical force, which may be accentuated by the use of average bone geometries. Additionally, the beam theory model assumes that forces are applied to the bone at single points and does not consider more complex patterns of loading distribution. The model also does not consider the effects of bone deformation or the surrounding soft tissues on bone stress.

Running speed and load carriage independently increased peak dorsal and plantar stresses, as indicated by the lack of an interaction effect between these factors. This suggests that gait speed and load contribute additively to metatarsal stress, with each factor independently influencing external loading. A demanding 10 km run did not alter peak dorsal and plantar stresses, likely due to a lack of change in the vertical force under the second metatarsal. Participants, however, did show some localized foot and ankle kinematic and kinetic alterations post-treadmill run, indicating that the 10 km run did elicit gait changes. Taken together, the lack of change in bone stress could be explained by protocol limitations which may have allowed for partial recovery of mechanics following the treadmill run.

The excellent within-day repeatability and between-day reliability of second metatarsal stress estimates support that the observed changes in stress likely reflect true effects rather than measurement variability. Participants ran in a highly consistent manner within and

between days, which is also reflected in the excellent repeatability and reliability of the model inputs. Bone stress was reliable despite small changes in marker positioning between days. In addition, an acclimation effect to study procedures was observed with a slight increase in repeatability during the second laboratory visit.

4.2 Repeatability and Reliability of Second Metatarsal Bone Stress Estimates (Paper III)

4.2.1 Reliability of Second Metatarsal Stress Estimates

Bone stress estimates demonstrated excellent between-day reliability despite only moderate between-day reliability of second metatarsal and second toe lengths. A previous sensitivity analysis demonstrated that the model is sensitive to larger changes (10%) in metatarsal length,[70] suggesting that the between-day differences observed in bone lengths were too small (~2% or less) to meaningfully influence bone stress reliability. The relatively small between-day differences in bone lengths were likely due to consistent marker placement by the same researcher, as marker placement reliability has been shown to vary between researchers.[112] Overall, these findings suggest that stress estimates are robust to typical between-day variability in estimated bone lengths from marker-based measurements. However, this apparent robustness may also reflect limited sensitivity of the model to small changes in input variables, potentially limiting the model's ability to detect subtle differences in loading between conditions.

While bone stress estimates demonstrated excellent between-day reliability, using the same bone segment lengths for both visits further improved the reliability of bone stress. Using the same second metatarsal length for both visits modestly improved between-day reliability for peak dorsal and plantar stresses (6 – 7%), while using the same second toe length for both visits had minimal to no effect on bone stress reliability. When first-visit metatarsal and toe lengths were applied to the second visit, dorsal and plantar stresses were slightly reduced

compared to using visit-specific lengths, indicating that small differences in estimated bone lengths can systematically influence stress estimates. Future research comparing bone stress estimates between days should consider using fixed bone lengths to minimize the influence of between-day variability in marker placement and further improve reliability.

Since second metatarsal length was calculated from marker positions, between-day differences in marker positioning will influence the reliability of bone stress. The reliability of between-day relative marker distances was moderate to excellent. Despite the relative consistency in marker positions between days, even small (less than ~2%) differences in relative marker positions between visits influenced the reliability of peak stress by introducing variation in estimated bone lengths. These findings indicate that while the variation in segment lengths was not large enough to meaningfully reduce bone stress reliability, stress estimates were sensitive to variations in marker placement between days.

4.2.2 Reliability of Kinematic, Kinetic, and Stride Variables

Variables related to stress and gait which also demonstrated excellent within-day repeatability and between-day reliability included second metatarsal pitch angle at the time of peak stress, peak vertical GRF under the second metatarsal, and stride characteristics (stride length, stride frequency, and ground contact time), with stride frequency showing good repeatability on the first visit. These findings indicate that participants ran in a highly consistent manner across trials and between days. This suggests that the high consistency of metatarsal kinematics, metatarsal loading, and stride characteristics likely contributed to the excellent reliability of bone stress estimates within and between days.

The reliability or repeatability of model-derived estimates of bone stress during gait has not previously been reported. The excellent between-day reliability and within-day repeatability of second metatarsal stress estimates is consistent with previous studies which demonstrated good to excellent reliability of kinematics, kinetics, and GRFs during walking and running.[100,113–115] The excellent reliability of bone stress estimates likely reflects the

model's reliance on kinematic and GRF inputs, which have been shown to demonstrate high within- and between-day reliability. Taken together, these findings support the use of model-derived bone stress estimates as a reliable outcome measure for investigating bone stress during gait.

4.3 The Influence of Running Speed and Load Carriage on Second Metatarsal Stress (Paper I)

4.3.1 Effects of Speed and Load Carriage on Metatarsal Stress

Second metatarsal bone stress was sensitive to relatively small changes in running speed and load carriage, with a 10% increase in running speed increasing metatarsal stress by ~4%, a 10% reduction in speed decreasing stress by ~2%, and a 10% increase in carried mass increasing stress by ~3%. These findings suggest that the second metatarsal is sensitive to changes in the magnitude of external loading which occur with changes in running speed and load carriage. While running speed and load carriage each influenced metatarsal loading, they acted as independent factors, contributing additively to metatarsal stress rather than interacting to produce greater effects than the sum of their individual contributions.

Greater second metatarsal stress with increasing running speeds is consistent with previous studies which observed increased plantar pressure under the second metatarsal when running at faster speeds.[73,74] The increases in estimated stresses are likely driven by increases in second metatarsal vertical forces, which are a main contributor to peak bone stress in the model. Similarly, in the tibia, increasing running speed has also been shown to result in increased peak tibial bending moments, tibial strain, and probability of tibial bone failure.[39,72]

Greater second metatarsal stress with load carriage is consistent with previous findings that peak second metatarsal dorsal strain increased during treadmill walking with a 20 kg backpack compared to no load.[77] The bone stress findings from this thesis are comparable

to strain findings since stress and strain are proportional below the yield threshold of bone. That second metatarsal stress increased even with a relatively light load of 10% body mass is consistent with previous findings in the tibia that load carriage of ~20% body mass increased tibial bending moments, cumulative impulse, von Mises strain, and probability of stress fracture.[72,116]

The changes observed in bone stress with changes in running speed and load carriage likely reflect true physiological effects rather than measurement variability, as supported by the excellent within-day repeatability and between-day reliability of bone stress estimates. The lack of change in metatarsal stress observed following the 10 km treadmill run (Paper II) does not necessarily indicate a lack of model sensitivity, but may instead reflect limitations in the protocol which could have limited changes in the metatarsal loading environment, as discussed in Section 4.4.1.

4.3.2 Between-Participant Variability in Peak Second Metatarsal Stress

The between-participant variability observed in peak second metatarsal dorsal stress was not explained by running speed and body mass (Figure 9). The CV values were smaller than those calculated from a previous study of second metatarsal stress in 12 rearfoot runners.[9] The smaller CV values may be explained by the larger sample size (n = 28) in the present thesis and the use of generic rather than participant-specific bone geometries, which may have reduced between-participant variability. The CVs were similar in magnitude to those from previous studies on peak second metatarsal plantar forces.[73,74] Participant-specific differences in peak bone stress could exceed 100 MPa even when running at similar speeds under the same load carriage condition, highlighting the importance of considering the individual when assessing bone stress injury risk.

4.3.3 Associations Between Potential Explanatory Variables and Peak Second Metatarsal Stress

Peak dorsal stress was very strongly correlated with peak vertical force under the second metatarsal, indicating that peak vertical second metatarsal force may be a reasonable surrogate for peak metatarsal stress. This contrasts with tibial loading studies where ground reaction forces were poor estimates of tibial internal loading,[71] and may indicate differences between the metatarsal and tibia. This difference may be due to plantar pressure data allowing for the measurement of localized forces acting directly under the individual metatarsal, whereas whole-body GRFs used to estimate tibial loading reflect acceleration of the whole-body center of mass.

Peak whole-body vGRF was only moderately correlated with metatarsal stress, in agreement with previous findings in the tibia.[71] Whole-body external loads do not fully reflect internal tissue loads since internal loads result from a combination of internal muscle forces and external GRFs.[6,117] Further, increases in internal loading observed with increasing running speed are likely due to the greater muscular activity required to maintain faster speeds.[6] This highlights the importance of considering localized compared to whole-body external loading when estimating internal metatarsal loads.

Body mass was only moderately correlated with peak dorsal stress and normalizing to body mass did not explain the between-participant variability (Figure 9b). This suggests that body mass alone is not a primary determinant of metatarsal stress. These findings, however, may be influenced by the use of standard bone geometries in the model with the exception of individualized bone lengths. Using standard bone geometries for participants with different body masses may under or overestimate the bone's ability to resist deformation by misrepresenting the cross-sectional area, moment of inertia, and resistance to bending of the bone.[118] This is because individuals with greater body mass tend to possess more robust bone geometries as an adaptive response to withstand higher loads while maintaining

comparable stresses.[118] Since standard bone geometries were used in the model, the model does not account for these size-related structural adaptations, potentially obscuring the relationship between body mass and metatarsal stress. A limitation of this thesis is that bone geometries were not scaled to each participant, which may have clarified the relationship between body size and metatarsal stress.

Running speed was also only moderately correlated with peak dorsal stress and while speed may influence metatarsal stress, it is not a primary determinant of stress on its own. This suggests that individual responses to changes in running speed may be moderated by other biomechanical factors that influence how loads are applied in the foot. Although stride characteristics were highly repeatable and foot strike was controlled, individual differences in lower-extremity joint kinematics and muscle activity may affect how forces are absorbed and distributed in the metatarsal. Differences in metatarsal geometry may further influence how loads are transmitted in the bone. Thus, even similar running speeds may result in different metatarsal stress magnitudes between individuals.

4.4 The Influence of a 10 km Run on Second Metatarsal Stress (Paper II)

4.4.1 Effect of a 10 km Run on Metatarsal Stress

Second metatarsal bone stress remained unchanged following the 10 km treadmill run, despite a reduction in ankle plantarflexion moment and an increase in peak first MTP dorsiflexion angle. The change in these kinetic and kinematic variables indicates that the 10 km protocol was sufficient to induce localized gait changes. The lack of change in metatarsal stress is likely explained by the lack of change in peak second metatarsal vertical force, which was shown to be very strongly correlated with bone stress in Paper I. The absence of change may also reflect methodological limitations, including differences in footwear, running speed, and running surface between the 10 km treadmill run and the overground testing trials. The approximately five minute transition time between the treadmill and overground running may have also allowed for partial recovery of the effects of the 10 km run.

Peak second metatarsal stress remained unchanged after the 10 km run because peak bending stress and peak shear force did not change, despite a reduction in peak axial stress and force. The magnitude of bending stress is far greater than the magnitude of axial stress in the second metatarsal (Table 6), so that axial stress has a negligible contribution to dorsal and plantar stresses. Midshaft bending stress is influenced by internal joint reaction forces, external forces acting under the metatarsal, and the metatarsal pitch angle. The reduction in peak second toe forces post-10 km run occurred along with reduced plantar tendon and MTP joint reaction forces. This reduction in distal forces counteracted the concurrent reduction in pitch angle, resulting in no change in shear force post-10 km run. However, the results overall suggest a change in the loading environment of the second ray following the 10 km run. The absence of a corresponding change in metatarsal stress may either be explained by these counteracting mechanisms or may reflect a lack of sensitivity of the metatarsal model to detect small changes, such as those which could occur from altered forces acting under the second toe.

The lack of change in second metatarsal stress following the 10 km treadmill run is contrary to previous studies which reported increased strain in the second metatarsal and tibia with exercise-induced fatigue.[69,77,79,80] Previous studies also found increased vertical force under the second metatarsal following a fatiguing run,[119–121] in contrast to the unchanged second metatarsal vertical force in this thesis. The contrast in findings may be explained by methodological differences between the 10 km protocol which was run shod on a treadmill at a faster speed and the testing trials which were run barefoot overground at participants' preferred speed. Differences in footwear, running surface, and running speed may have altered loading patterns, and the post-treadmill transition period may have allowed for partial recovery from the 10 km run. There is limited evidence that different running surfaces may affect bone loading, as a small strain gauge study in the tibia showed substantially lower bone strain during treadmill compared to overground running.[122] While it is unclear whether these findings in the tibia translate to the metatarsal, it is possible that the treadmill elicited

lower bone loading than overground running. This may have contributed to the unchanged metatarsal stress in this thesis.

4.4.2 Effects of a 10 km Run on Lower-Extremity Kinematics and Kinetics

The reduction in peak ankle plantarflexion moment after the 10 km run is similar to that reported by Sanno et al.[88] during the last kilometer of the same 10 km treadmill protocol. Participants' RPE in the final kilometer of the 10 km run was also comparable to that reported by Sanno et al.[88] The plantarflexion moment and RPE findings indicate a similar effect of the 10 km run as in the prior study, despite data being collected overground after the 10 km run in this thesis compared to continuously during the treadmill run in the previous study. This suggests that runners had not fully recovered their distal lower-extremity mechanics prior to the post-10 km trials and that the unchanged metatarsal stress may indicate a true effect.

No differences, however, were found in proximal lower-extremity kinematics in contrast with Willwacher et al.[87] who found reductions in proximal kinematics during the last kilometer of the same 10 km protocol. The difference in findings might be explained by differences in running speed during treadmill and overground running in this thesis, which can affect running kinematics.[123] In addition, the approximately five minute transition time between the treadmill run and overground trials may have allowed for recovery of proximal lower-extremity kinematics. The reduced plantarflexion moment along with unchanged proximal kinematics suggest that either kinetic effects may be slower to recover following a demanding 10 km run or that kinematics are more sensitive to changes in running modalities. Further research should determine how differences in running conditions including changes in footwear, speed, and surface influence the recovery of lower-extremity kinetics and kinematics following prolonged running and how long these changes persist.

Changes in MTP mechanics were also observed following the 10 km treadmill run. The increase in first MTP dorsiflexion angle indicates potential fatigue of the hallux flexor muscles which eccentrically control MTP joint dorsiflexion prior to push-off. This increase in the first

MTP peak dorsiflexion angle in combination with the lack of change in the first MTP peak dorsiflexion moment suggests reduced MTP joint stiffness following the 10 km treadmill run. The second MTP joint also showed reduced stiffness through a combination of an unchanged peak second MTP dorsiflexion angle and a decreased second MTP dorsiflexion moment after the 10 km run. The reduction in second MTP moment resulted from reduced forces under the second toe, which is consistent with previous studies which also reported reductions in forces under the lesser toes after demanding running protocols.[119,121,124] The overall MTP findings indicate that although the first and second MTP joints showed different responses to the 10 km run, both exhibited changes consistent with reduced joint stiffness which may be indicative of muscular fatigue. Despite these changes at the MTP joint, including reduced forces under the second toe and reduced MTP joint reaction forces, no change was observed in second metatarsal stress. The lack of change in metatarsal stress may reflect a compensatory mechanism whereby a concurrent reduction in metatarsal pitch angle resulted in unchanged shear force and bending stress, or a limited sensitivity of the model to changes in distal forces.

4.5 Implications of Second Metatarsal Stress for Bone Stress Injury Risk

Peak dorsal stress magnitudes were consistently greater than peak plantar stress magnitudes across all study conditions, in agreement with previous observations in both rearfoot and non-rearfoot strike runners.[9] The implications of these loading patterns for determining site-specific risk for BSI within the metatarsal remain unclear, as injury risk likely depends on the interaction between loading magnitude and distribution, bone geometry, and bone strength. Because common locations of BSIs within the metatarsal shaft are not well documented, further research is needed to determine how regional loading patterns influence injury susceptibility.

The relationship between metatarsal stress magnitude and injury risk remains unclear. The Weibull probabilistic fatigue model predicts that the probability of second metatarsal bone

failure will increase with increasing bone stress.[104] Thus, increases in bone stress as a result of increasing running speed or load carriage may increase the risk of second metatarsal BSI. However, validating models against *in vivo* estimates remains challenging and prospective injury data are required to better understand how stress estimates relate to injury risk. Notably, not all participants demonstrated an increase in peak stress with increasing running speed or load carriage (Figure 8a-b), suggesting that additional factors such as lower-extremity joint mechanics and muscle activity may mediate the relationship between speed, load carriage, and bone stress.

While military recruits often carry loads heavier than 10% body mass, these results suggest that even relatively modest loads can increase internal bone stress. These findings are particularly relevant in military contexts as the second metatarsal is one of the most common sites for bone stress injuries in military populations.[7,54,125] Further research could provide insight into whether heavier load carriage results in proportional or non-linear increases in metatarsal bone stress, which would have implications for BSI risk in military populations.

4.6 Strengths and Limitations

A strength of this thesis was the repeated measures design where participants served as their own controls, minimizing the effect of between-participant variability. The effect of condition order was minimized by randomizing the order of the six running conditions on the first visit. Inclusion criteria were selected to limit the influence of factors such as age and running volume which have been shown to affect running biomechanics.[126] Additionally, testing was conducted barefoot to allow for direct placement of markers on the foot, eliminate the effects of footwear, and enable a mechanistic investigation of metatarsal stress.

The simplicity of the model used to estimate bone stress may have influenced the results by over- or underestimating metatarsal stress. The model assumes that forces are applied at single points and does not account for the distribution of load across the metatarsal, which

may misrepresent the distribution of internal stresses. Bone stress was also estimated at the mid-shaft of the metatarsal, which is assumed to be the area with the smallest cross-sectional area and thus highest risk of bone stress injury. It is unknown, however, whether this is the most common location for metatarsal bone stress injuries in runners. The two-dimensional beam theory model did not account for second metatarsal eversion angle, mediolateral forces, or bone torsion as these were assumed to have a negligible effect on bone stress. In addition, the only participant-specific bone geometries used in the model were second metatarsal and second toe lengths, with modest errors introduced by estimating these lengths from the positions of skin-mounted markers.

A limitation of conducting the testing trials barefoot was that it limits ecological validity since participants were habitually shod runners. Another limitation was that the pressure plate was a harder surface than the softer EVA runway which may have caused participants to alter their gait pattern when landing on the pressure plate. The study cohort of healthy young adult runners may also limit the generalizability of results to other populations including less experienced runners, different age groups, and clinical populations.

One of the main limitations of **Paper II** was that the 10 km running protocol was performed shod on a treadmill at a faster speed, while testing was conducted during barefoot overground running at participants' preferred speed. Treadmill running may induce lower bone loading compared to overground running, since limited evidence in the tibia showed substantially lower bone strain during treadmill compared to overground running.[122] Treadmill running is also associated with differences in stride mechanics compared to overground running,[127] which may influence loading patterns. Testing was conducted overground to allow for simultaneous collection of pressure and force plate data, enabling estimation of localized vertical and horizontal forces under the second metatarsal. In addition, there was a transition time of approximately five minutes from finishing the treadmill run to starting the post-treadmill overground trials, which may have allowed for partial recovery of the effects of the 10 km run.

A limitation of **Paper III** was the potential effect of acclimation to study procedures, where participants may have been more consistent in their running technique on the second visit. Additionally, the testing procedures were not identical on both days, with the preferred speed condition being one of six conditions performed in a randomized order on the first visit whereas it was always the first condition on the second visit. The time between testing days was also not controlled for and may have affected the results due to the large variability in the time interval between visits.

5 Conclusion

This thesis contributes to the current understanding of factors affecting internal loading in the second metatarsal during running and the repeatability and reliability of these internal loading estimates. The main conclusions are:

- Excellent within-day repeatability and between-day reliability of second metatarsal bone stress estimates support the ability of the model to detect meaningful changes in loading.
- Running speed and load carriage each independently influenced peak second metatarsal bone stress, with increases in both speed and load carriage increasing bone stress.
- Peak second metatarsal bone stress was sensitive to relatively small changes (10%) in running speed and load carriage.
- Between-participant variability in peak bone stress demonstrates the importance of considering the individual when assessing metatarsal loading.
- Peak second metatarsal vertical force was very strongly correlated with internal second metatarsal stress, suggesting that second metatarsal vertical force may be a useful indicator of internal metatarsal loading.
- A 10 km treadmill run did not alter peak second metatarsal bone stress during overground running at preferred speed, although localized kinetic and kinematic changes were observed in the foot and ankle post-10 km run.

6 References

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Paper I

Second metatarsal bone stress during running is independently influenced by running speed and load carriage

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Abstract

Bone stress injuries occur in populations who engage in repetitive loading such as distance runners and military recruits. The second metatarsal is a common bone stress injury site. Running speed and load carriage are modifiable factors which may influence metatarsal loading. The aim of this study was to determine how running at preferred, 90% of preferred, and 110% of preferred speed with and without a weight vest carrying 10% body mass influences second metatarsal bone stress. Data were collected from 28 runners (14 female, 14 male) during barefoot running using motion capture cameras, a pressure plate, and a force plate. Bone stress was calculated at the dorsal and plantar surfaces of the midshaft of the second metatarsal using a two-dimensional beam theory model. No interaction effect was found between running speed and load carriage. Running at 90% of preferred speed decreased peak dorsal and plantar stress by 2.1 and 2.3%, respectively, compared to preferred speed, while running at 110% of preferred speed increased peak dorsal and plantar stress by 4.2%. Running with 110% of body mass increased peak dorsal and plantar metatarsal stress by 3.1 and 2.9%, respectively, compared to running with no additional mass. The mean coefficient of variation across all conditions was 25%, indicating variation in peak stress between participants. Running at faster speeds and with additional loads may increase the risk of second metatarsal bone stress injury. The high variation in peak bone stress between participants demonstrates the importance of considering the individual when assessing injury risk.

Introduction

Running is a popular form of physical activity but is associated with high rates of running-related overuse injuries, which have been reported to range from 15% to 85%.[1] Bone stress injuries are one of the most common running injuries[2] with stress fractures being the most severe type.[3] Bone stress injuries are burdensome and can result in a time-loss of 4-8 weeks from the causative activity.[4] Bone stress injuries are caused by repetitive loading such as occurs in long distance running and military training.[5] Risk factors for bone stress injuries are multifactorial and can include factors such as an individual's training load, anthropometrics, and gait characteristics.[6]

The second metatarsal bone in the forefoot is one of the most common injury sites for bone stress injuries in track and field athletes, distance runners, and military recruits.[7,8] In a population of Royal Marines recruits, 65% of stress fractures reported were in the metatarsals, of which 27% were in the second metatarsal.[9] The relatively high incidence of second metatarsal injuries may be influenced by the bone geometry, since the second metatarsal is reportedly the longest of the metatarsals on average with the narrowest diaphyseal width.[10] The second metatarsal has been shown to have the highest vertical plantar force, bending strain, and shear force compared to the other metatarsals and the tibia.[11–14]

Running speed and load carriage are two modifiable factors that may affect bone stress injury risk. Running speed may be related to bone stress injury risk since external and internal loads increase with increasing running speed.[15,16] Recently, Rice et al.[16] found that increasing running speed by 20% increased peak tibial bending moment and tibial impulse per kilometer in male distance runners. Similarly, reducing running speed has been shown to reduce the probability of tibial stress fracture when using a probabilistic stress fracture model in male runners.[17] Since previous research focused on the tibia, it remains unknown how changing running speed influences bone stress in the metatarsal.

Load carriage has been shown to alter gait kinematics and kinetics[18] and may therefore also influence second metatarsal loading. Military recruits commonly engage in load carriage activities,[19] potentially contributing to the high incidence of second

metatarsal bone stress injuries in military recruits. A study by Arndt et al.[20] found that second metatarsal dorsal strain increased during treadmill walking with a 20 kg backpack compared to walking with no additional load when using implanted strain staples to measure *in vivo* bone strain. However, it is unknown how load carriage influences second metatarsal bone stress during running. Since both running speed and load carriage have the potential to increase loading of the second metatarsal, the combined effect of speed and load carriage may be greater than the sum of the individual factors. The potential interaction between running speed and load carriage on second metatarsal bone stress has not been previously investigated.

Plantar loading under the second metatarsal during running is subject to high between-participant variability, which may be influenced by running style - including foot strike technique, as well as structural anatomy and tissue geometry. Nunns et al.[21] reported a between-participant coefficient of variation (CV) for peak pressure under the second metatarsal of 32.3% for example, even when restricting the analysis to those who ran with a rearfoot strike. However, externally-measured plantar pressures and forces are not always indicative of internal bone loading.[22,23] Ellison et al.[22] observed that despite differences in scaled vertical ground reaction forces under the metatarsal heads in two distinct groups of runners, there were no significant differences in second metatarsal stress between groups, likely due to the influence of geometry on estimates of stress. Variables such as peak horizontal and vertical second metatarsal and second toe forces, peak ground reaction forces, running speed, and body mass may help to explain between-participant variability in bone stress. Quantifying the between-participant variability and exploring the factors that explain this variability will aid in the understanding of the extent to which broad training recommendations can be made for runners regarding modifiable training factors and risk of bone stress injury.

The aim of this study was to determine how speed and load carriage influence bone stress in the second metatarsal during running, and to quantify whether these factors interact. Specifically, this study quantified the influence of running at preferred speed, 90% of preferred speed (slower), and 110% of preferred speed (faster) with and without an additional 10% body mass on bone stress in the second metatarsal during running. Furthermore, the between-participant variability was quantified and

associations with potential contributors to metatarsal stress were explored to aid understanding. It was hypothesized that second metatarsal bone stress would increase with both increasing running speed and load carriage, and that there would be an interaction between running speed and load carriage such that the combined influence of increased running speed and load carriage would increase metatarsal stress to a greater extent than the sum of the individual factors.

Methods

Participants

Twenty-eight participants were recruited for the current study. An a priori sample size analysis was conducted using G*Power software[24] for a within factors repeated measures ANOVA with effect size f = 0.25, α = 0.05, and power = 0.80, which yielded a minimal sample size of 19. Cohen's f effect size was determined as half the Cohen's d reported for dorsal stress by Ellison et al.[22] Recruitment for the current study was conducted jointly with another study which required a minimum of 24 participants, so a conservative sample of 28 participants was recruited for both studies.

Participants were included if they were any gender, aged 18-40 years who had completed a run of any distance from 5 km to a marathon at a maximal effort in the last 2 years, ran at least 15 km per week and ran at least 3 times per week, had been running on a regular basis for at least 3 years, ran with a rearfoot strike, were comfortable running a 10 km on a treadmill, and could run barefoot without pain. Exclusion criteria were any neurologic, musculoskeletal, or movement disorder, current pregnancy, any lower extremity injury within the previous 6 months, any previous foot or ankle surgery, any symptomatic foot pathology, and any illness or injury that prevented running. Exclusion criteria also included body mass greater than 100 kg since the weight vest carrying capacity was 10 kg and the weight vest was set to 10% of a participant's body mass.

Descriptive information including age, sex, height, and body mass were also obtained for each participant. Running history including weekly running volume, running frequency, and years of running experience were also obtained. Previous injury was defined as a previous running-related injury which caused the participant to miss three

or more planned running sessions.[25] Leg dominance was defined by asking participants which leg they would use to kick a ball.

Participants were recruited through flyers placed around the local area in Oslo, posts on social media, and a project page on the Norwegian School of Sports Sciences' (NIH) website. The study was approved by Sikt – The Norwegian Agency for Shared Services in Education and Research and the ethics committee at the Norwegian School of Sport Sciences (Approval number 281 – 150623). All participants provided written informed consent prior to participation.

Protocol and Equipment

Inclusion in the study was limited to participants with a rearfoot strike as foot strike type is known to influence running biomechanics.[26] Rearfoot striking has been reported to be the most common foot strike type among distance runners.[27] Participants were screened for foot strike type to assess eligibility and this screening also served as a warm-up and familiarization to the lab for eligible participants. Barefoot running was chosen to allow for the placement of markers directly on the foot and to eliminate the influence of footwear on running biomechanics.[26]

To assess foot strike, participants ran first in their own running shoes and then barefoot at a self-selected comfortable running speed over a pressure plate (HR Mat, TekScan, Norwood, MA, USA, 0.49 m 0.45 m 0.01 m, with 8,448 sensors, 185 Hz, 3.9 sensors/cm^2). The pressure plate was set in a 14.4-m ethylene-vinyl acetate (EVA) runway (0.02 m thick) with a gap where the pressure plate was located. An EVA runway was used due to its similarity with a shoe midsole and to encourage runners to retain their natural foot strike pattern, which may change when running barefoot on a hard surface.[28] Timing gates (TCi System, Brower Timing Systems, Draper, UT, USA) were placed 5-m apart centered around the pressure plate to monitor running speed. No feedback on speed was given to participants during the screening trials.

At least three shod screening trials were collected in which the participant's entire left or right foot landed within the bounds of the pressure plate, the participant did not obviously target the plate, and the participant was not obviously accelerating or decelerating. If the foot strike pattern differed between trials, more trials were collected

until a majority pattern could be determined. Participants repeated the screening process while barefoot until at least three acceptable barefoot trials were collected.

Participants qualified for the study if they ran with a rearfoot strike both with and without shoes. A rearfoot strike was defined as the center of pressure during the first frame of stance in the pressure data being located in the rear 1/3 of the foot. The average running speed during the barefoot screening trials was used to determine the participant's preferred running speed. No instruction or feedback was given to participants about foot strike unless a participant demonstrated a rearfoot strike during the shod trials but switched foot strike type during the barefoot trials, in which case the participant was instructed to attempt landing with a rearfoot strike while running barefoot. The participant was excluded from the study if they reported that landing with a rearfoot strike did not feel natural while running barefoot.

The three-dimensional position of the second metatarsal was tracked using retro-reflective spherical markers (6 mm in diameter) attached to the skin of participants' proximal and distal second metatarsals. The markers were part of a set of seventeen markers positioned on participants' right foot and ankle which were required for use in another study. The foot marker set was modified from the Rizzoli marker set[29] by adding markers on the head of the middle phalanx of the second toe and on the base and head of the third metatarsal, while removing markers on the tibial tuberosity and fibular head. An additional 20 markers were positioned from the trunk to the right leg, also required for use in another study. Markers were secured using double sided tape. All marker locations on each participant were determined by the same researcher (NF) for consistency in placement. The right leg was selected arbitrarily for all participants.

The experimental protocol required participants to run barefoot overground with and without a weight vest carrying 10% of body mass at three different running speeds: preferred speed, 90% of preferred (slower speed), and 110% of preferred (faster speed). The mass was evenly distributed between the front and back of the vest. The order of the six conditions were randomized for each participant using an online random sequence generator. Ten percent of body mass was chosen to investigate if even a relatively small increase in load would increase metatarsal stress. The relatively low mass was also

chosen to ensure that it was not overly physically demanding for participants who may not be used to running with added mass to complete all six running conditions.

Twenty-three motion capture cameras (Oqus 500 and 700, Qualisys, Göteborg, Sweden) collected kinematic data in three-dimensions at 185 Hz while ground reaction forces were recorded in three-dimensions at 1850 Hz with a floor-embedded force plate (AMTI, Watertown, MA, USA). The pressure plate was located directly over the floor-embedded force plate so that both ground reaction forces and plantar pressures were recorded from the same running step. Due to technical issues with the pressure plate, data from the first five participants were collected at capture frequencies lower than 185 Hz (range: 92.5-174 Hz).

For each running condition, 5-7 successful trials were collected. A successful trial was defined as one where the whole foot landed within the bounds of the pressure plate, the participant did not appear to target the plate, the running speed was within ±5% of the target speed, and the participant landed with a rearfoot strike (the center of pressure was within the rear 1/3 of the foot in the first frame of stance). If a participant consistently switched to a non-rearfoot strike type at any point in the study, they were given verbal instructions to land with a rearfoot strike.

Data Analysis

Stance phase was defined using a force threshold of 5 N for both the pressure plate and force plate. The threshold was determined to retain as many frames of data from the pressure plate as possible. Kinematic and ground reaction force data were filtered with a fourth-order lowpass Butterworth filter with cut-off frequencies of 12 and 50 Hz, respectively. Data during each running stance phase were interpolated to 101 frames. Pressure data from the five participants where the capture frequency was less than 185 Hz were included in the analysis based on visual comparison of the time-series for the summed total foot pressure data with the time-series for the vertical force plate data and the within-participant study design. Regional anatomical masks were automatically applied to the foot in the pressure plate software to determine the regions under the second metatarsal and second toe (Strideway 7.80, TekScan, Norwood, MA, USA). The regional masks were manually adjusted if deemed necessary to define the desired regions with the least number of adjustments necessary. Second metatarsal and

second toe forces were calculated by scaling the vertical and horizontal ground reaction forces obtained from the force plate to the percentage of total foot force under those areas according to the pressure plate data. Second metatarsal pitch angle was calculated by taking the inverse sine of the ratio of the vertical distance between the proximal and distal second metatarsal markers divided by the three-dimensional (3D) vector distance between the two markers. Second metatarsal and second toe lengths were defined as the 3D vector distance between markers placed on the proximal and distal second metatarsal and distal second metatarsal and head of the middle phalanx of the second toe, respectively.

Dorsal and plantar bone stress were estimated using a two-dimensional beam theory model of the second metatarsal using participant-specific metatarsal and toe lengths and with the remaining bone geometries obtained from the mean of a cohort of 20 healthy recreational runners.[22] Model assumptions have previously been reported by Ellison et al.[22] Briefly, key model assumptions were that 1) the flexor muscle forces can be represented by a long and short plantar tendon with an equal distribution of force between them; 2) the external forces are applied at points under the distal metatarsal head and second toe; 3) the second toe segment is parallel to the ground during ground contact; and 4) mediolateral forces and kinematics are negligible.

Model calculations are fully described by Ellison et al.[22] The model estimates the plantar tendon force by calculating the joint moment at the metatarsophalangeal (MTP) joint using the vertical force acting under the second toe and length of the second toe as inputs. The angle of plantar tendon action was calculated by subtracting the mean angle between the short and long plantar tendons (5°) from the second metatarsal pitch angle.[22] Joint reaction forces at the MTP joint were calculated by assuming that the linear joint forces from the plantar tendons, second toe forces, and joint reaction forces sum to zero in equilibrium. The axial and shear forces acting at the second metatarsal were then calculated by summing the axial and shear components of both the joint reaction forces and forces under the second metatarsal.

The bending moment at the midshaft of the second metatarsal was calculated from the shear force and the distance from the midpoint of the metatarsal to the center of the metatarsal head. Axial stress was the axial force divided by the mean bone cross-

sectional area of $3.77x10^{-5}$ m^2 obtained from Ellison et al.'s[22] dataset (Supplement 1, Table I). Bending stress was calculated from the bending moment, the mean radial distance of $3.74x10^{-2}$ m from the center of the bone to the surface, and the mean area moment of inertia about the horizontal axis of the bone cross-section of $2.05x10^{-10}$ m^4 (Supplement 1, Table I).[22] The midshaft stresses at the dorsal and plantar bone surfaces of the second metatarsal were then calculated by summing and subtracting, respectively, the axial and bending moments. Dorsal and plantar stress were multiplied by negative one so that a positive value indicated dorsal stress and a negative value indicated compressive stress, consistent with engineering convention. Peak dorsal and plantar stresses were used for data analysis. To quantify differences in peak stress between individual participants, coefficient of variation (CV) was calculated. All calculations were performed using custom scripts in MATLAB (R2024b, MathWorks, Natick, MA, USA).

Statistics

All statistical analyses were conducted in R (v4.2.2, R Foundation for Statistical Computing, Vienna, Austria). To assess the influence of running speed (90% of preferred, preferred, 110% of preferred) and load carriage (body mass, 110% body mass) on peak dorsal and plantar stress, linear mixed-effects models were run to account for the random effect of participant and to allow for the inclusion of participants who were missing data for one or more conditions. The significance of fixed effects were assessed using ANOVAs. The Shapiro-Wilk test was used to test normality. In the case of significant interaction effects, simple effects were evaluated by comparing levels of one factor within each level of the other. If there was no significant interaction effect, comparisons were performed on the main effects averaged over the other factor. When significant main effects were found, post hoc pairwise comparisons were conducted using estimated marginal means without adjustment for multiple comparisons. The results of the ANOVAs for dorsal and plantar stress and the post hoc pairwise comparisons for speed and load carriage were not adjusted for multiple comparisons since these were a priori hypothesis tests. The Benjamini-Hochberg procedure was used to control the false discovery rate for additional ANOVAs run on secondary variables of interest which included peak bending moment, peak shear and axial forces, peak vertical and horizontal ground reaction forces under the second metatarsal, second metatarsal pitch angle at

the time of peak stress, and the timing of peak dorsal and plantar stresses (% stance). Significance was set at $\alpha = 0.05$ for all tests.

Pearson correlations were conducted between the following variables and the absolute value of peak dorsal stress to aid understanding of between-participant differences in peak dorsal stress: peak vertical and horizontal second metatarsal and second toe forces, peak ground reaction forces, running speed, and body mass. The correlation analyses were only conducted during running at the preferred speed without load carriage as this was considered the baseline condition. The Benjamini-Hochberg procedure was used to control the false discovery rate due to running multiple correlations and statistical significance was evaluated based on the adjusted p values (p_{adj}, $\alpha = 0.05$). The strength of the correlations were interpreted as follows: 0.10-0.39 weak, 0.40-0.69 moderate, 0.70-0.80 strong, and 0.90-1.00 very strong.[30]

Results

The faster speed condition for one participant was excluded due to a non-rearfoot strike pattern. The faster and slower conditions with and without the weight vest were excluded for another participant due to an error in calculating faster and slower running speeds.

Participant descriptive information is presented in Table 1. Participants' preferred running speed was (mean ± standard deviation) 3.25 ± 0.36 m/s (range: 2.56 – 4.08 m/s), slower running speed was 2.93 ± 0.32 m/s (range: 2.30 – 3.68 m/s), and faster running speed was 3.58 ± 0.41 m/s (range: 2.80 – 4.48 m/s). The dorsal and plantar aspects of the metatarsal were under compression and tension respectively during stance phase (Figure 1, Table 2).

Table 1

Participant information presented as mean ± standard deviation with range in parenthesis for all participants and by sex.

	All Participants (n=28)	Females (n=14)	Males (n=14)
Age (years)	29.6 ± 4.7	28.9 ± 4.5 (23 - 38)	30.3 ± 5.1 (24 - 40)
Height (m)	1.73 ± 0.09	1.67 ± 0.06 (1.54 - 1.75)	1.81 ± 0.06 (1.70 - 1.90)
Mass (kg)	68.8 ± 9.7	62.7 ± 8.8 (50.0 - 85.8)	74.8 ± 6.1 (62.2 - 84.4)
BMI (kg/m^2)	22.8 ± 3.3	22.8 ± 4.4 (19.2 - 36.2)	22.9 ± 2.0 (19.6 - 26.0)
Average Weekly Running Volume (km/week)	48.0 ± 24.9	45.0 ± 24.6 (25 - 115)	51.0 ± 25.7 (15 - 115)
Average Number of Runs per Week	4.6 ± 1.8	4.4 ± 2.1 (3 - 11)	4.7 ± 1.4 (3 - 8)
Running Experience (years)	7.8 ± 5.6	7.6 ± 4.4 (3.5 - 20)	7.9 ± 6.8 (2 - 22)
Previous Injury, n (%)	19 (68%)	9 (64%)	10 (71%)
Right Leg Dominant, n (%)	22 (79%)	12 (86%)	10 (71%)

Peak Dorsal Stress

Peak dorsal stress was normally distributed for all conditions ($p > 0.05$). There was no interaction effect between speed and load carriage (Table 2). There was a significant main effect of running speed on peak dorsal stress ($F(2, 157) = 27.35$, $p < 0.001$, $\eta_p^2 = 0.26$). When averaged across loading conditions, increasing running speed by 10% from preferred to faster speed increased peak dorsal stress by 4.22 ± 5.70% (5.74 ± 7.73 MPa, 95% confidence interval (CI): [3.61, 7.87], $p < 0.001$) and reducing running speed by 10% from preferred to slower speed decreased peak dorsal stress by 2.10 ± 5.21% (2.54 ± 6.5 MPa, 95% CI: [0.76, 4.32], $p = 0.030$) (Figure 2a). There was a significant main effect of load carriage on peak dorsal stress ($F(1, 157) = 22.56$, $p < 0.001$, $\eta_p^2 = 0.13$). When averaged across speeds, increasing load from body mass to 110% of body mass

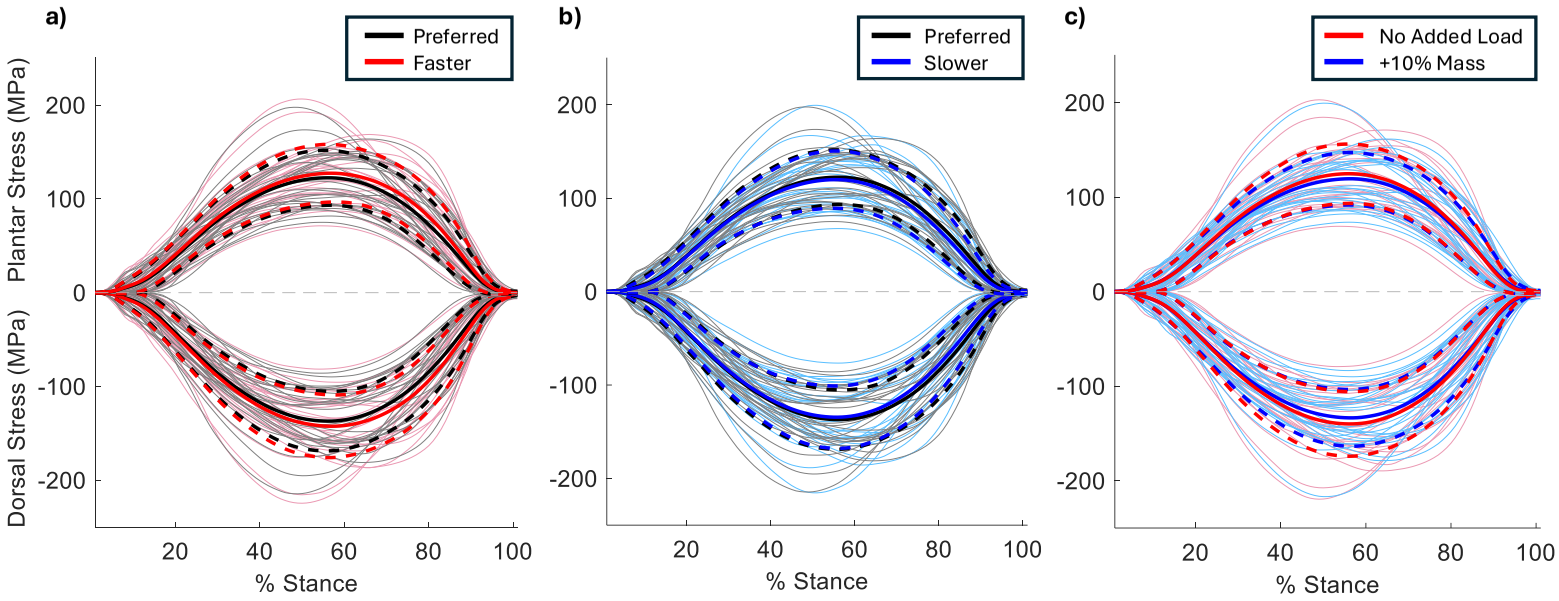


Figure 1: Dorsal stress (negative) and plantar stress (positive) for 100% of stance phase during running. Thin lines represent the average of all trials for a single participant. Thick lines represent the mean for all participants and thick dashed lines show standard deviations. **a)** Preferred speed averaged across loading conditions (black) compared to fast speed averaged across loading conditions (red). **b)** Preferred speed averaged across loading conditions (black) compared to slow speed averaged across loading conditions (red). **c)** Trials with no load carriage averaged across three speeds (blue) compared to load carriage trials averaged across three speeds (red).

Table 2

Mean ± standard deviation for peak stresses, forces, angles, and the timing of peak stresses for the second metatarsal. Confidence intervals displayed in brackets.

Variable	Preferred Speed	Preferred Speed +10% Mass	Faster Speed	Faster Speed +10% Mass	Slower Speed	Slower Speed +10% Mass	P value		
							Speed x Load	Speed	Load
Peak Dorsal Stress (MPa)	-138 ± 32 [-125, -151]	-143 ± 35 [-129, -157]	-141 ± 32 [-129, -153]	-147 ± 38 [-133, -162]	-135 ± 35 [-121, -148]	-139 ± 36 [-125, -153]	0.882	<0.001*	<0.001*
Peak Plantar Stress (MPa)	125 ± 30 [137, 113]	130 ± 33 [143, 117]	128 ± 30 [140, 117]	133 ± 35 [147, 119]	122 ± 33 [135, 109]	126 ± 33 [138, 113]	0.722	<0.001*	<0.001*
Peak Bending Moment (Nm)	7.2 ± 1.7 [6.6, 7.9]	7.5 ± 1.9 [6.8, 8.2]	7.4 ± 1.7 [6.7, 8.0]	7.6 ± 2.0 [6.8, 8.4]	7.0 ± 1.9 [6.3, 7.7]	7.2 ± 1.9 [6.5, 8.0]	0.823	<0.001*	<0.001*
Peak Shear Force (N)	261 ± 50 [242, 280]	271 ± 56 [249, 292]	269 ± 52 [249, 289]	279 ± 60 [255, 302]	255 ± 56 [233, 277]	263 ± 57 [241, 285]	0.823	<0.001*	<0.001*
Peak Axial Force (N)	251 ± 72 [223, 279]	260 ± 76 [230, 289]	248 ± 71 [221, 276]	274 ± 82 [243, 306]	248 ± 79 [218, 279]	260 ± 84 [228, 292]	0.349	0.025*	<0.001*
Peak Vertical GRF under Second Metatarsal (N)	273 ± 52 [253, 293]	281 ± 59 [258, 304]	279 ± 52 [259, 299]	288 ± 62 [263, 312]	267 ± 58 [244, 289]	274 ± 60 [250, 297]	0.806	<0.001*	0.002*
Peak Horizontal GRF under Second Metatarsal (N)	50 ± 15 [44, 56]	52 ± 17 [46, 59]	57 ± 17 [50, 63]	57 ± 19 [50, 65]	45 ± 15 [39, 51]	47 ± 16 [41, 53]	0.486	<0.001*	0.025*
Second Metatarsal Pitch Angle at Time of Peak Stress (°)	29.4 ± 3.4 [28.1, 30.8]	29.2 ± 3.3 [28.0, 30.5]	29.7 ± 3.5 [28.4, 31.1]	29.9 ± 3.9 [28.4, 31.4]	29.4 ± 3.7 [28.0, 30.9]	29.4 ± 3.6 [28.0, 30.8]	0.823	0.008*	0.740
Timing of Peak Dorsal Stress (% stance)	56 ± 5 [54, 58]	57 ± 5 [55, 59]	59 ± 5 [56, 61]	58 ± 6 [55, 60]	56 ± 5 [54, 58]	56 ± 6 [54, 58]	0.315	<0.001*	0.823
Timing of Peak Plantar Stress (% stance)	56 ± 5 [54, 58]	57 ± 5 [55, 59]	58 ± 5 [56, 61]	58 ± 6 [55, 60]	56 ± 5 [54, 58]	56 ± 5 [54, 58]	0.213	<0.001*	0.823

GRF = ground reaction force. MPa = megapascals. N = Newtons. Nm = Newton meters. * = significant effect ($p < 0.05$).

increased peak dorsal stress by 3.13 ± 6.32% (4.37 ± 8.61 MPa, 95% CI: [2.47, 6.28], $p < 0.001$) (Figure 2b).

Peak Plantar Stress

Peak plantar stress was normally distributed for all conditions ($p > 0.05$). There was no interaction effect between speed and load (Table 2). There was a significant main effect of speed on peak plantar stress ($F(2, 157) = 27.66$, $p < 0.001$, $\eta_p^2 = 0.26$). When averaged over loading conditions, increasing speed by 10% from preferred to faster speed increased peak plantar stress by 4.20 ± 5.69% (5.17 ± 6.88 MPa, 95% confidence interval (CI): [3.28, 7.07], $p < 0.001$) and reducing running speed by 10% from preferred to slower speed decreased peak plantar stress by 2.32 ± 5.08% (2.59 ± 5.88 MPa, 95% CI: [0.98, 4.19], $p = 0.017$). There was a significant main effect of load carriage on peak plantar stress ($F(1, 157) = 18.68$, $p < 0.001$, $\eta_p^2 = 0.11$). When averaged across speeds, increasing load from body mass to 110% of body mass increased peak plantar stress by 2.87 ± 6.33% (3.68 ± 7.95 MPa, 95% CI: [1.92, 5.44], $p < 0.001$).

Variability Between Participants in Peak Stress

For peak dorsal stress, the mean CV between participants across all conditions was 24.6 ± 0.9% (range: 23.6 – 25.7%). For peak plantar stress, the mean CV between participants across conditions was 25.3 ± 0.9% (range: 24.3 – 26.5%). Visualizing peak dorsal stress for each participant against running speed, both non-normalized (Figure 3a) and normalized to body mass (Figure 3b) demonstrates the between-participant variability in both stress magnitudes and the change in stress between running speeds within-participant.

Correlation between Peak Dorsal Stress and Potential Explanatory Variables

When running at the preferred running speed without load carriage, peak dorsal stress was very strongly correlated with peak second metatarsal vertical force, ($r(26) =$

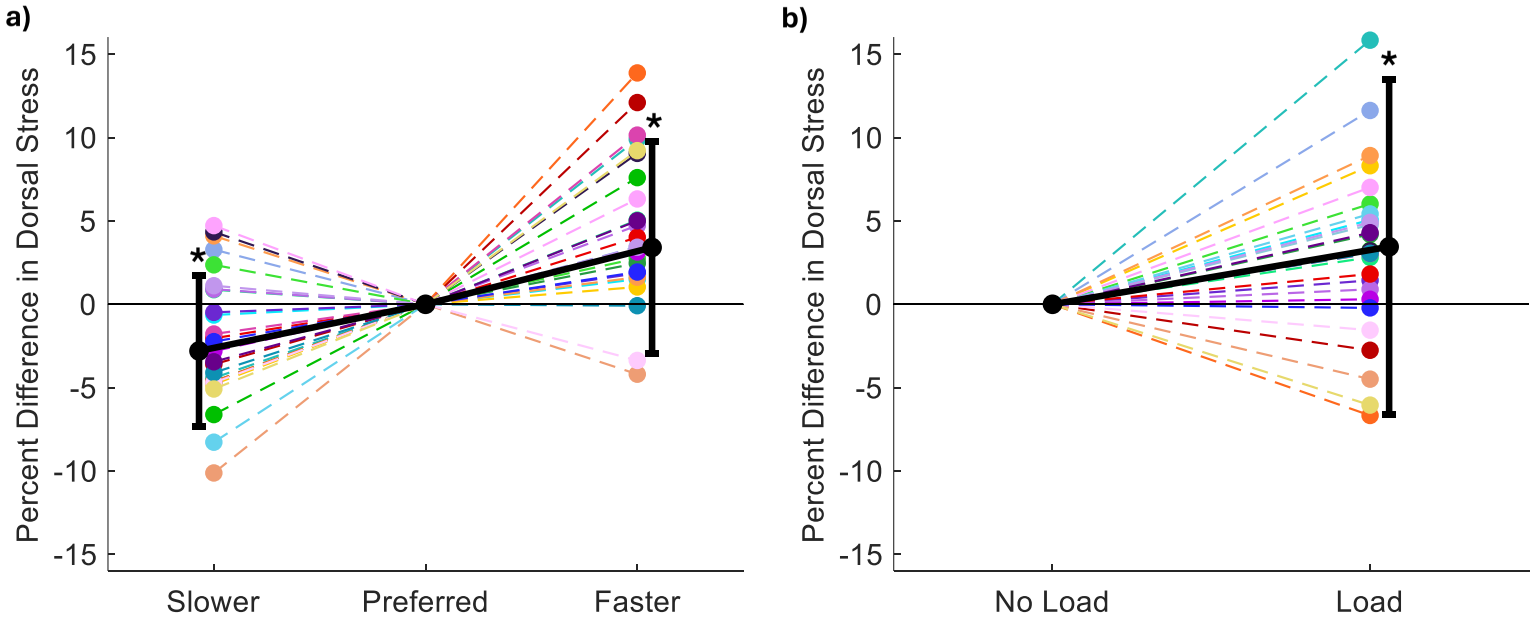


Figure 2: Percent difference in peak dorsal stress between speed (a) and load carriage (b) conditions. Colored circles represent average percent difference in dorsal stress for each participant (averaged across trials). Black circles represent means for each condition averaged across participants. Black error bars represent ± standard deviation. **a)** Percent difference in dorsal stress between slow speed compared to preferred speed and between fast speed compared to preferred speed when averaged over loading conditions. An asterisk indicates a significant difference from the preferred speed condition. **b)** Percent difference in dorsal stress between the load condition and no load condition when averaged across running speeds. An asterisk indicates a significant difference from the no load condition.

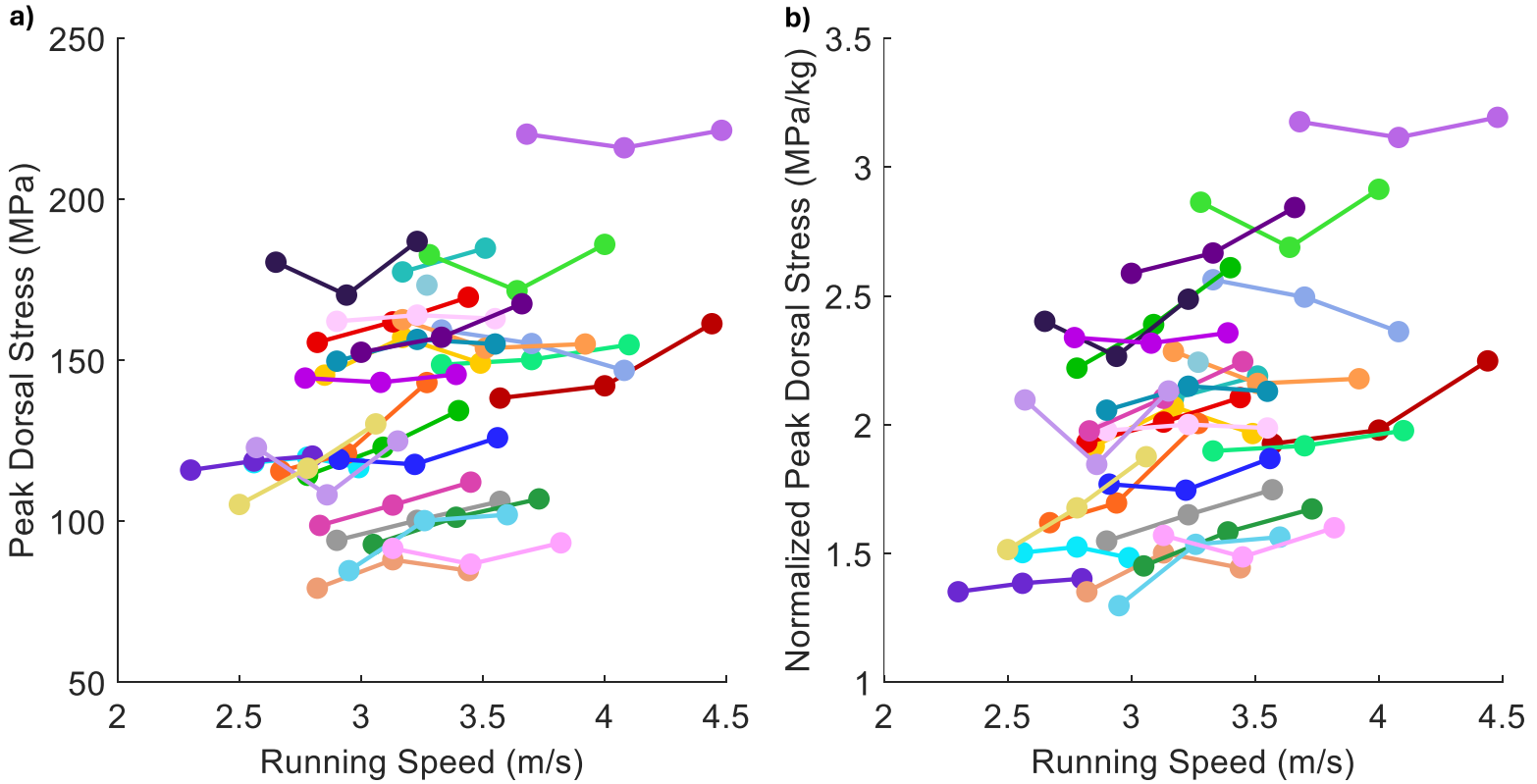


Figure 3: Average peak dorsal stress is shown according to running speed for slower, preferred, and faster speeds with no load carriage. Each participant is shown in a different color. Dots represent average peak dorsal stress for each speed condition for each participant. **a)** Peak dorsal stress for 3 running speeds, **b)** Peak dorsal stress normalized to body mass for each participant.

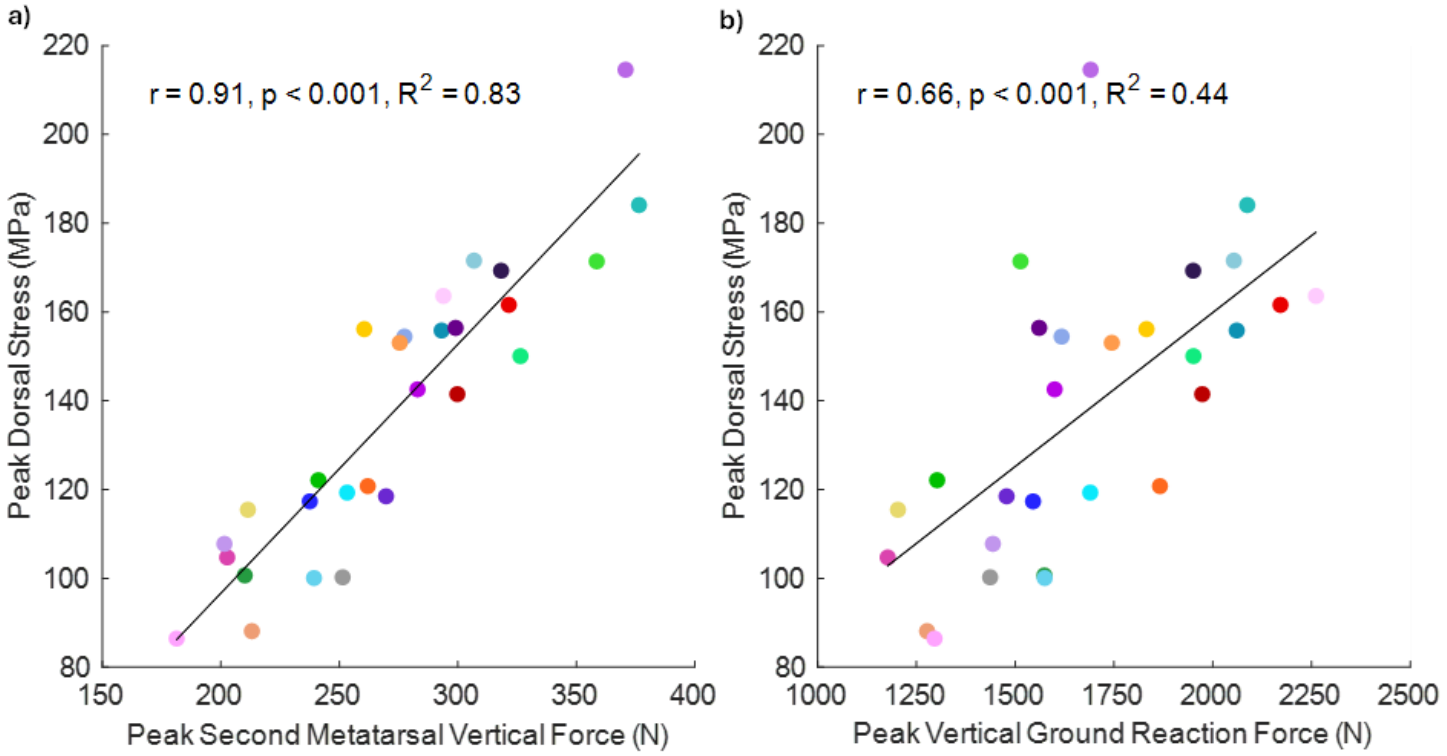
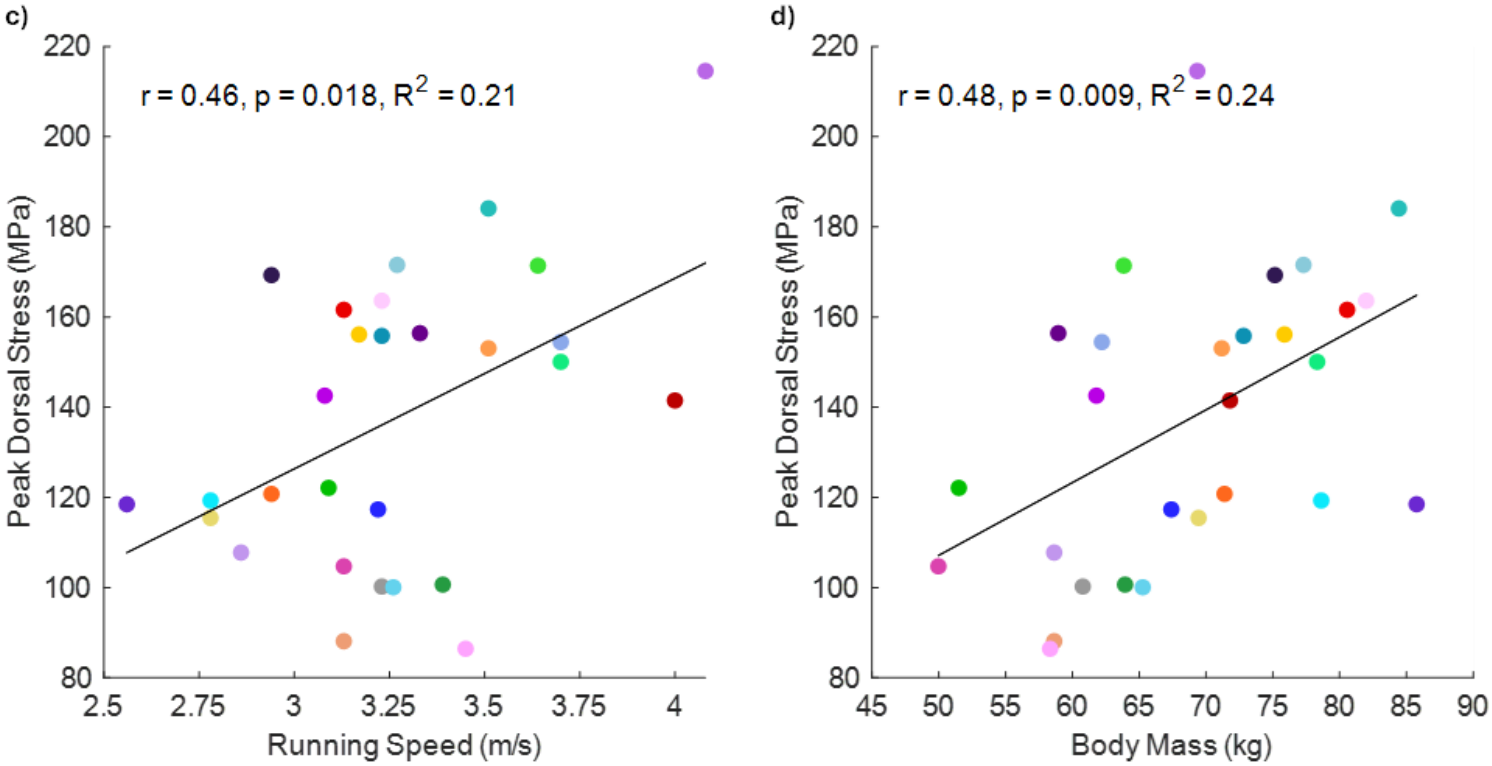


Figure 4: Correlation plots for the preferred speed no load condition. Each color represents a different participant. For plots a and b, each dot represents a single trial. For plots c and d, each dot represents a single participant. **a)** A very strong correlation of r = 0.954 was found between peak dorsal stress and peak second metatarsal vertical force. **b)** No significant correlation (r = -0.018) was found between peak dorsal stress and vertical ground reaction vertical force. **c)** A moderate correlation of r = 0.465 was found between peak dorsal stress and preferred running speed. **d)** A moderate correlation of r = 0.485 was found between peak dorsal stress and body mass.


c)
r = 0.46, p = 0.018, R2 = 0.21
Peak Dorsal Stress (MPa)
Running Speed (m/s)
d)
r = 0.48, p = 0.009, R2 = 0.24
Peak Dorsal Stress (MPa)
Body Mass (kg)

0.91, $p_{adj} < 0.001$, 95% CI: [0.80, 0.96] (Figure 4a)). Peak dorsal stress and peak horizontal second metatarsal force were moderately correlated, ($r(26) = 0.64$, $p_{adj} < 0.001$, 95% CI: [0.35, 0.82]). The correlation between peak dorsal stress and vertical force under the second toe was weak, ($r(26) = 0.36$, $p_{adj} = 0.058$, 95% CI: [-0.01, 0.65]), and the correlation between peak dorsal stress and horizontal force under the second toe was also weak, ($r(26) = 0.37$, $p_{adj} = 0.058$, 95% CI: [-2.6×10^{-5}, 0.65]). Peak vertical ground reaction force was moderately correlated with peak dorsal stress when running at the preferred speed without load carriage, ($r(26) = 0.66$, $p_{adj} < 0.001$, 95% CI: [0.38, 0.83] (Figure 4b)). Running speed was moderately correlated with peak dorsal stress, ($r(26) = 0.46$, $p_{adj} = 0.018$, 95% CI: [0.11, 0.71] (Figure 4c)), while body mass was also moderately correlated with peak dorsal stress, ($r(26) = 0.48$, $p_{adj} = 0.015$, 95% CI: [0.14, 0.73] (Figure 4d)).

Discussion

This study examined how running speed and load carriage influence second metatarsal bone stress during running. The hypotheses that second metatarsal bone stress would increase with increasing running speed and with load carriage were confirmed. The hypothesis that there would be an interaction effect between running speed and load carriage was not supported. The effects of speed and load carriage on second metatarsal bone stress were therefore independent of each other.

The magnitude of peak dorsal stress was greater than the magnitude of peak plantar stress across all conditions, which is supported by Ellison et al.[22] who found greater magnitudes of peak dorsal stress compared to plantar stress in both rearfoot and non-rearfoot strike runners. While peak dorsal compressive stress values were greater than peak plantar tensile stress values, bone is stronger under compression than tension.[31] It is unknown which aspect of the metatarsal bone most frequently incurs bone stress injuries. Further research documenting the distribution of bone stress injury sites within the second metatarsal would help improve the understanding of metatarsal bone stress injury risk.

The change in metatarsal stress of approximately 2-4% that was observed when changing running speed by 10% demonstrates that this approach of estimating metatarsal stress is sensitive to running speed, even when the change in running speed

is relatively small. It has previously been reported that peak plantar pressure under the second metatarsal increased with increasing running speed when comparing speeds of 2.8, 3.3, and 3.8 m/s.[32] The current study supports these previous findings since increased second metatarsal pressure when running at faster speeds was the main contributor to bone stress in the bone stress model used. Increasing running speed has also been demonstrated to result in increased peak tibial bending moments [16] as well as peak principal compressive tibial strain and probability of tibial bone failure.[17] While failure probability was not considered in the present study, Taylor et al.'s[33] Weibull probabilistic model predicts that the probability of second metatarsal bone failure will increase with increasing bone stress and stressed bone volume. Thus, increases in bone stress may increase the risk of second metatarsal bone stress injury. Furthermore, the increased magnitude of metatarsal loading may increase the risk of stress injury due to the relative importance of the magnitude of loading to risk of failure .[34] It is also important to note that there were individual participants who did not demonstrate an increase in peak stress with increasing running speed (Figure 2a), suggesting that other factors may mediate the relationship between speed and bone stress.

In terms of the influence of load carriage, our findings support those of Arndt et al.[20] who found that dorsal second metatarsal strain increased during treadmill walking with a 20 kg backpack compared to no load. Our findings are comparable to Arndt et al.[20] since strain is proportional to stress while load is below the yield strength of bone. The finding that peak metatarsal bone stress increased even with a relatively light load of 10% body mass supports previous findings in the tibia that load carriage (~20% body mass) increased bending moments and cumulative impulse;[16] and von Mises strain and probability of stress fracture.[35] Load carriage may thus increase the risk of lower extremity bone stress injury by increasing the magnitude of loading. While military recruits often carry loads heavier than 10% body mass, our results suggest that even relatively modest loads can increase internal bone stress. These findings are relevant since the second metatarsal is one of the most common sites for bone stress injuries in military recruits.[8,9,36] Investigation of the influence of heavier load carriage would provide further understanding.

Internal tissue loads should not be conflated with external measures of loading. [23] Internal bone loads likely result from a combination of muscular forces and the contribution from the ground reaction forces. Increases in internal loading with increasing running speed may reflect the greater muscular activity required to maintain faster speeds.[37] In the present study, peak dorsal stress was highly correlated with peak vertical force under the second metatarsal, indicating that peak vertical second metatarsal force may be a reasonable surrogate for peak metatarsal stress. This finding contrasts with tibial loading studies where ground reaction forces were poor estimates of tibial internal loading,[23] and may indicate differences between the metatarsal and tibia. This difference may be due to the metatarsals being closer to the location of external load application, resulting in less opportunity for attenuation of external forces. Additionally, plantar pressure data allows for quantification of forces acting directly under individual metatarsals, whereas using external ground reaction force data as a surrogate for tibial loading relies on whole-body center of mass acceleration. Due to challenges of evaluating and validating modelling approaches against *in vivo* estimates — which in themselves have limitations — it remains unclear the extent to which quantification of metatarsal stress can be used to infer understanding about injury risk. Prospective injury data is required to better understand how estimates of bone stress relate to injury risk.

Body mass was only moderately correlated with peak dorsal stress (Figure 4d), and normalizing peak stress to body mass did not explain the between-participant variability (Figure 3b). These findings suggest that body mass alone is not a primary determinant of metatarsal stress, although these findings may be influenced by the use of standard bone geometries with the exception of individualized bone lengths in the model. Risk factors for bone stress injuries are multifactorial and loading history, changes in training load, nutrition status, sleep, and other factors should be taken into consideration when assessing injury risk.[38]

Large between-participant variation in peak bone stress was observed, even when accounting for running speed and body mass (Figure 3). Coefficients of variation were on average 24.5% for peak dorsal stress and 25.3% for peak plantar stress when averaged across conditions, which are smaller than the corresponding CV values of 43% for peak dorsal and plantar stress reported by Ellison et al.[22] from 12 rearfoot runners. The lower

CVs in the present study may be explained by the larger sample size of 28 participants in the present study and our use of generic rather than participant-specific bone geometries. The CVs in our study were similar to CVs found for peak external plantar forces under the second metatarsal of ~21-24%.[32,39] Differences in peak bone stress between individual participants thus exist even when running at similar speeds and under the same load carriage condition, highlighting the importance of considering the individual when assessing bone stress injury risk.

The beam theory model is primarily driven by the input parameter of vertical second metatarsal force, which was demonstrated by the very strong correlation between peak dorsal stress and second metatarsal vertical force (Figure 4a). The model's reliance on vertical metatarsal force may be influenced by the use of mostly standardized bone geometries. The model thus likely oversimplifies bone stress by overlooking the role of individual bone morphology.[22] When external forces are held constant, longer metatarsal length and smaller upper metatarsal head radius increase bending moments and thus bending stress. Additionally, a larger mid-shaft radius and smaller area moment of inertia also increase bending stress at the bone surface. A sensitivity analysis by Ellison[40] showed that a 10% change in one of these bone geometry variables can alter peak stress by 1-12%, with cumulative errors potentially having a compounding effect. The model is also sensitive to metatarsal pitch angle.[40] The pitch angle at the time of peak stress in the present study increased with faster running speeds, which likely contributed to increased bending moments and bone stress. Pitch angle, however, was not affected by load carriage, where changes in bone stress were driven primarily by changes in vertical second metatarsal force.

The only participant-specific bone geometries used in the current study were second metatarsal and second toe lengths, with errors potentially introduced by estimating these lengths from the positions of skin-mounted markers. Since the current study used a repeated measures experimental design where participants served as their own controls across the six speed and load carriage conditions, using standardized bone geometries was not expected to influence the evaluation of the main study aims. Development of statistical shape models that allow reconstruction of metatarsal cortical and trabecular geometry from limited individualized imaging data could allow for

identification of between-participant differences as well as further understanding of loading patterns along the entirety of the bone through finite element analyses for example.

A strength of this study was the repeated measures design where participants served as their own controls. The repeated measures design minimized the effect of individual participant differences. The effect of condition order was minimized by randomizing the order of the 6 conditions. Inclusion criteria were intended to minimize the effect of factors such as age, running volume, and running experience which have been shown to influence running biomechanics.[41,42] The study was performed barefoot so that markers could be placed directly on the foot and to eliminate the effects of footwear. Barefoot running in a controlled laboratory setting, however, limits the ecological validity of the study since participants were habitually shod runners.

Another limitation of the study was the variability in preferred running speeds across participants. Running at participants' relative running speeds allowed for greater ecological validity but resulted in variability in absolute running speeds. Limiting preferred running speed in the inclusion criteria would have restricted study recruitment and limited generalizability. An additional limitation was that participants who switched to a non-rearfoot foot strike pattern were instructed to land with a rearfoot strike, which might not be representative of their natural running style. This was the case for 10 of the participants, eight of whom displayed inconsistent foot strike patterns across the different speed and load carriage conditions. Another limitation was that the pressure plate was a hard surface within a soft EVA runway which may have caused participants to alter their gait pattern when landing on the pressure plate.

The calculation of metatarsal stress in the present study relied on a combination of generic bone geometries and participant-specific metatarsal and second toe lengths. It should be noted that the use of skin-mounted markers to estimate second metatarsal and second toe lengths were likely subject to error which would influence the between-subject variability values. The simplicity of the model used to estimate bone stress may have also influenced the results. A simple two-dimensional cantilever model was used which did not account for second metatarsal eversion angle, mediolateral forces, or bone torsion. A sensitivity analysis of a finite element model for calculating second metatarsal

bone stress showed that bone stress was not sensitive to second metatarsal eversion angle.[43] An assumption of the model was that mediolateral forces were negligible. It was also assumed that any bone torsion would be negligible. Bone stress was only estimated at the mid-shaft of the metatarsal, which is assumed to be the location of minimal cross-sectional area and thus the highest bone stress and injury risk. However, it is possible that bone stress injuries occur at other locations within the metatarsal.

In conclusion, running speed and load carriage each independently influenced peak bone stress in the second metatarsal. Reducing running speed to 90% of preferred speed decreased peak dorsal and plantar second metatarsal stress by 2.1 and 2.3%, respectively, while increasing running speed to 110% of preferred speed increased peak dorsal and plantar stress by 4.2%. The addition of 10% of body mass during running increased peak dorsal and plantar second metatarsal stress by 3.1 and 2.9%, respectively. A coefficient of variation of ~25% demonstrates the between-participant variability in peak stress values, illustrating the importance of considering individual differences when evaluating metatarsal bone stress and injury risk.

Perspective

On average, running at faster speeds and with added load carriage increased second metatarsal bone stress, although this pattern was not observed for every runner. Our findings show that modifiable factors can therefore contribute to bone stress injury risk through altered loading magnitude. Our findings may help to explain why military recruits have a high incidence of metatarsal bone stress injuries, since even relatively small increases in running speed and load carriage each independently increased metatarsal bone stress, likely due to increases in vertical force under the metatarsal. While second metatarsal vertical force was strongly related to peak dorsal stress, vertical ground reaction force was only moderately related to bone stress, demonstrating the importance of force application location when determining internal bone stresses. Prospective research is needed to determine whether higher magnitudes of metatarsal loading during running are associated with greater occurrence of bone stress injury.

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Supplement 1

Table I

Means and standard deviations (SD) for bone geometry values provided by Ellison et al.[23] from magnetic resonance images of 20 healthy runners. Mean values were used in the bone stress calculations.

Variable	Mean	SD
ε (mm)	4.07	6.18
r1 (mm)	9.66	1.61
r2 (mm)	7.65	0.93
Cross-Sectional Area (mm^2)	$3.77x10^{-2}$	$6.00x10^{-3}$
Ixx (mm^4)	$2.05x10^{-7}$	$8.29x10^{-8}$
r (mm)	3.74	0.48

ε = distance between the inferior metatarsal head and the plantar tendon

r1= lower radius of the metatarsal head

r2 = upper radius of the metatarsal head

Ixx = area moment of inertia about the horizontal axis of the bone cross-section

r = radial distance from the center of the bone to the surface

Paper II

A demanding 10 km treadmill run alters lower limb kinetics and kinematics but not second metatarsal bone stress

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Abstract

Prolonged running may alter internal loading of the second metatarsal , which could increase risk of bone stress injury. The purpose of this study was to determine how second metatarsal bone stress and lower-extremity mechanics change following a demanding 10 km treadmill run. Participants were 28 healthy, rearfoot strike runners (14 female/14 male) aged 18 - 40 years. Participants ran barefoot overground at their preferred running speed before and after a shod 10 km treadmill run at 105% of their estimated maximal 10 km race time. Lower-extremity kinematics and kinetics were recorded during overground running using motion capture cameras, a pressure plate, and a force plate. A two-dimensional beam theory model of the second metatarsal was used to calculate bone stress pre- and post-10 km run. Peak dorsal and plantar second metatarsal bone stress did not change post-10 km run. Peak ankle plantarflexion moment decreased, while peak first metatarsophalangeal flexion angle increased post-10 km run. Thus, changes in lower-extremity running mechanics were observed when running at preferred speed following a demanding 10 km run, while loading in the second metatarsal was not altered.

Keywords

Biomechanics; running-induced fatigue; foot biomechanics; bone stress injury

Introduction

Bone stress injuries are a common overuse injury in distance runners. The most severe type of bone stress injury are stress fractures (Warden et al., 2023), which are one of the ten most common types of running injuries (Francis et al., 2019). The foot is one of the most frequent sites of bone stress injury in competitive track and field athletes (Hoenig et al., 2024), with a particularly high incidence in the metatarsals (Bennell et al., 1996; Brukner et al., 1996; Hoenig et al., 2024; Nattiv et al., 2013). When compared to the other metatarsals and the tibia, the second metatarsal in particular has been shown to experience the highest vertical plantar force, bending strain, shear force (Gross & Bunch, 1989), compression strain (Donahue et al., 2000; Milgrom et al., 2002, 2011; Niu et al., 2015), and tension strain (Milgrom et al., 2002, 2011) during dynamic loading. Bone stress injuries occur when microdamage accumulates in the bone, for example from repetitive loading which occurs during running if adequate rest is not taken to allow for bone to repair. Bone stress injuries have a high burden due to time loss from the causative activity, typically requiring a reduction in running activity of four to eight weeks or more to allow sufficient time for the bone to repair (Hatch et al., 2007).

Fatigue during exercise has been described as a physical or cognitive change that limits perceived and physical function (Enoka & Duchateau, 2016; Phillips, 2024). More specifically, muscle fatigue can be defined as the reduced ability of muscles to generate force (Bigland-Ritchie & Woods, 1984), which may influence internal bone loading and thus risk of bone stress injury. In cadaveric studies, simulated muscle fatigue during walking increased second metatarsal strain (Donahue & Sharkey, 1999), while simulated muscle contraction reduced second metatarsal dorsal strain (Sharkey et al., 1995). The findings indicate a direct influence of muscle forces on bone strain. Muscle fatigue may thus increase the risk of bone stress injury by altering internal bone loading.

Exercise-induced fatigue has been shown to influence second metatarsal loading, with stress and strain increasing with walking- and marching-induced fatigue (Arndt et al., 2002; Gefen, 2002). Additionally, second metatarsal plantar pressure has been shown to increase following a fatiguing run (Anbarian & Esmaeili, 2016; Weist et al., 2004; Willems et al., 2012). In the tibia, running fatigue protocols have been shown to increase (Milgrom et al., 2007) or decrease (Rice et al., 2024, 2025) tibial loading, suggesting an influence of running-induced fatigue on internal bone loading. Despite a high incidence of second metatarsal bone stress injuries in runners (Bennell et al., 1996; Brukner et al., 1996), it is unknown how prolonged running influences internal second metatarsal loading.

A 10 kilometre (km) running time trial has been shown to result in reduced muscle activity in the gastrocnemius and vastus lateralis muscles of well-trained cross-country runners (Paavolainen et al., 1999). A more recent study found that a 10 km treadmill run at 105% of participants' seasons best 10 km time led to changes in peak hip adduction angle, peak knee abduction angle, peak ankle eversion angle, and peak knee internal rotation angle (Willwacher et al., 2020) and a redistribution of positive joint work (Sanno et al., 2018), indicating that running-induced fatigue can alter lower limb gait mechanics.

Foot and ankle kinematics and kinetics may also change with fatigue. Ankle dorsiflexion angle at initial contact has been shown to decrease with running-induced fatigue (Quan et al., 2021; Willer et al., 2021), perhaps as a mechanism to reduce the positive work requirements at the ankle joint. Ankle plantarflexor moments have also been observed to decrease following a fatiguing run (Melaro et al., 2021; Sanno et al., 2018; Tam et al., 2017; Willer et al., 2021), indicating a reduced ability of the plantar flexors to generate force about the ankle when fatigued. It remains unclear how metatarsophalangeal (MTP) joint kinematics and kinetics are influenced by a demanding running protocol. A change in MTP joint loading

mechanics may influence stress in the metatarsals which could have implications for bone stress injury risk.

The purpose of this study was to determine how second metatarsal bone stress and MTP joint mechanics change following a 10 km treadmill run at 105% of estimated 10 km race time. We hypothesized that second metatarsal bone stress would increase following a 10 km run. We did not make a directional hypothesis regarding MTP joint mechanics due to limited previous literature in this area.

Materials and Methods

Participants

Sample size was determined using an a priori analysis with G*Power software (Düsseldorf, Germany) for a paired t-test between two dependent means with $\alpha = 0.05$, power of 0.90, and Cohen's d effect size of 0.62. The effect size was calculated based on differences in peak tibial mediolateral bending moment pre- vs post- the same 10 km treadmill running protocol as in the present study (Rice et al., 2024). Based on the sample size estimation, a minimum of 24 participants were needed to power the study. We recruited a conservative sample size of 28 participants.

Participants were females and males aged 18-40 years, had completed any distance from five kilometres to a marathon at a maximal effort in the last two years, ran at least 15 km per week, ran at least three times per week, had been running on a regular basis for at least three years, ran with a rearfoot strike, were comfortable running 10 km on a treadmill, and could run barefoot without pain. Exclusion criteria were any neurological, musculoskeletal, or movement disorder, current pregnancy, any lower extremity injury within the previous six months, any previous foot or ankle surgery, any symptomatic foot pathology, and any illness or injury that precludes running. Exclusion criteria also included body mass greater than 100

kg because study recruitment was conducted jointly with another study with this criterion (Frankston et al., 2025). Descriptive information and running history for each participant were obtained through a digital questionnaire.

Study participation was limited to rearfoot strikers due to the influence of foot strike pattern on running biomechanics (Lieberman et al., 2010). When fatigued, some runners may switch to a rearfoot strike but is it not typical for runners to switch to a non-rearfoot strike when fatigued (Bovalino & Kingsley, 2021).

Participants were recruited through flyers placed around the local area in Oslo, posts on social media, and a project page on the school website. The study was approved by the local ethics committee at the Norwegian School of Sport Sciences (approval number 281 – 150623) and Sikt – The Norwegian Agency for Shared Services in Education and Research. All participants provided written informed consent prior to participation.

Participant Screening

Screening trials to determine foot strike type were conducted during a separate initial laboratory visit as described previously (Frankston et al., 2025). Briefly, during the screening visit participants ran over a pressure plate (HR Mat, TekScan, Norwood, MA, USA, 0.49 m 0.45 m 0.01 m, with 8,448 sensors, 185 Hertz (Hz), 3.9 sensors/cm^2) set in a 14.4-m ethylene-vinyl acetate (EVA) runway (0.02 m thick) with a gap where the pressure plate was located. The EVA running surface was used due to its similarity to a running shoe midsole. To monitor running speed, timing gates (TCi System, Brower Timing Systems, Draper, UT, USA) were placed 5-m apart and centred around the pressure plate. Participants did not receive feedback on running speed during the screening procedure.

Three acceptable screening trials were performed in participant's own running shoes and an additional three trials were run barefoot to determine foot strike type. Participants qualified for the study if they ran with a rearfoot strike while both shod and barefoot. A rearfoot

strike was defined as a foot-ground contact where the centre of pressure in the first frame of stance was located under the rear third of the foot. The average running speed during the barefoot screening trials was used to determine the participant's preferred running speed.

Marker Placement

Seventeen markers were placed on participants' right foot and ankle in a modification of the Rizzoli marker set (Portinaro et al., 2014). The marker set was modified by adding markers on the head of the middle phalanx of the second toe and on the base and head of the third metatarsal, while removing markers on the tibial tuberosity and fibular head. Additional markers were positioned on the right knee medial and lateral joint lines (medial and lateral to the estimated average axis of rotation of the knee joint (Kadaba et al., 1990)), bilateral anterior superior iliac spines, iliac crests, posterior superior iliac spines, suprasternal notch, C7, and bilateral acromioclavicular joints. Rigid cluster shells with four noncolinear markers were placed on the right thigh and shank segments and wrapped with elastic bandages to minimize movement relative to the segment.

All markers were adhered directly to skin and not to clothing. Markers placed on the foot were six millimetres in diameter which allowed for close placement of markers, while markers at other locations were eight to ten millimetres in diameter. Marker locations were marked on the skin with surgical pen to assist with marker replacement following the 10 km run. The markers were secured using double sided tape. Marker locations were all determined by the same researcher (NF) for consistency in marker placement. The right leg was arbitrarily selected for all participants. A static standing trial was taken before the start of the overground running trials.

Overground Running

All data were collected during barefoot running which eliminated the influence of footwear on running biomechanics (Lieberman et al., 2010) and allowed for placement of

markers directly on the foot for accurate measurement of foot kinematics. Participants ran barefoot overground at their preferred running speed while kinematic data were collected in three-dimensions at 185 Hz using 23 motion capture cameras (Oqus 500 and 700, Qualisys, Göteborg, Sweden) and ground reaction forces were recorded in three-dimensions at 1850 Hz with a floor-embedded force plate (AMTI, Watertown, MA, USA). The pressure plate was located directly over the floor-embedded force plate so that ground reaction forces and plantar pressures could be recorded from the same running step. For the first five participants, the pressure plate recorded at less than 185 Hz (170 or 174 Hz) due to technical issues. For all trials, the sampling rate of the motion capture cameras was the same as the pressure plate, with the force plate capturing at 10 times that frequency.

A successful trial was defined as a trial in which the whole foot landed within the confines of the pressure plate, the participant did not appear to target the plate, the running speed was within ± 5% of the preferred speed, and the participant landed with a rearfoot strike. Five to eight successful overground running trials were collected pre-treadmill run for each participant. Up to eight trials were collected to ensure a minimum of five usable trials.

10 km Treadmill Run

Following the initial overground trials, participants were asked to run 10 km on a treadmill in their own running shoes. The treadmill run was performed shod to minimize the risk of discomfort due to prolonged running in an unfamiliar barefoot condition in a cohort of habitually shod runners. Since the study protocol was designed to mechanistically assess changes to the metatarsal loading environment, its ecological validity is limited.

Participants ran on the treadmill at 105% of their self-reported estimated maximal-effort 10 km time (Willwacher et al., 2020). The treadmill was set to each participant's pre-determined speed for the duration of the 10 km, unless participants requested a reduction in speed if they felt that they could not complete the 10 km at the given speed. Participants could

request a reduction in speed as many times as they felt necessary. For each request, the speed was reduced by a set amount of 0.05 m/s. Participants were asked to rate their perceived exertion (RPE) on the Borg scale (Borg, 1970) every kilometre during the treadmill run.

Following the treadmill run, shoes were removed, paper towels were used to remove sweat from the foot, and foot and ankle markers were reattached while the participant was seated. The transition time from the treadmill to the overground trials was recorded. A new static standing trial was taken and the overground running protocol was repeated for five to eight successful trials post-treadmill run at the same running speed (±5%) as each participant's pre-10 km running speed.

Data Analysis

For metatarsal stress-related variables, a vertical force threshold of 5 Newtons (N) was used to define stance phase for both the pressure and force data in order to retain as many frames of pressure data as possible. For lower extremity kinematics and kinetics, a vertical force threshold of 20 N was used to define stance phase and ground contact time. Kinematic and force plate data were filtered with a fourth-order lowpass Butterworth filter with a cut-off frequency of 20 Hz in Visual3D (HAS Motion, Kingston, Ontario, Canada). Data from each stance phase were interpolated to 101 frames. Pressure data from five participants where the capture frequency was less than 185 Hz (either 170 or 174 Hz) were included in the analysis after visual comparison of the time-series for the summed total foot pressure data with the time-series for the vertical force plate data and due to the within-participant study design.

Strideway software (Strideway 7.80, TekScan, Norwood, MA, USA) was used to automatically apply regional anatomical masks to the pressure data to determine the second metatarsal and second toe regions. The regional masks were manually adjusted when necessary to define the desired regions with the minimum number of adjustments. External forces under the second metatarsal and second toe were required to compute second metatarsal stress and

were calculated by scaling the vertical and horizontal whole body ground reaction forces based on the percentage of force underneath those regions in the pressure data. The pitch angle of the second metatarsal was calculated by taking the inverse sine of the ratio of the vertical distance between the second metatarsal head and base markers divided by the three-dimensional (3D) vector distance between the two markers. The length of the second metatarsal was defined as the 3D vector distance between the markers placed on the second metatarsal head and base, while the length of the second toe was defined as the 3D vector distance between the marker on the second metatarsal head and the head of the middle phalanx of the second toe.

Kinematic and Kinetic Analysis

Custom scripts in Visual3D were used to compute joint angles and moments. Joint angles were defined in three dimensions as the distal segment relative to the proximal segment using an XYZ Cardan sequence. The coordinate system was defined as +X medial, +Y posterior, and +Z superior, with rotations following the right-hand convention. Static standing trials were used to define neutral joint positions. The first MTP sagittal plane angle was defined as the planar angle between the vectors defined by the first metatarsal segment and the hallux segment projected into the sagittal plane with positive angles indicating MTP dorsiflexion (Leardini et al., 2007). The second MTP sagittal plane angle was the planar angle between the second metatarsal segment and the second toe segment projected into the sagittal plane with positive angles indicating dorsiflexion.

External joint moments including the ankle plantarflexion and first MTP dorsiflexion moments were calculated using Newton-Euler inverse dynamics resolved in the coordinate system of the proximal segment (Robertson, 2014). For the first MTP joint moment, reference values were used to estimate foot segment masses as a proportion of each participant's body mass (Matsumoto et al., 2022). The first MTP joint moment was analysed during only the last 50% of the stance phase.

Gait Characteristics

To determine stride length and frequency, foot strikes were detected by determining when the vertical velocity of the inferior posterior calcaneus marker crossed zero from negative to positive. To avoid false foot strike detections during swing phase, an anteroposterior heel marker velocity threshold of near zero (6.5×10^{-5}) was set for event detection. One stride per trial was included in the analysis with a minimum of five strides required per participant for inclusion in the analysis.

Beam Theory Model

A previously published two-dimensional beam theory model of the second metatarsal (M. A. Ellison et al., 2020) was used to estimate dorsal and plantar stress at the bone midshaft using average bone geometries obtained from a cohort of 20 healthy recreational runners (M. A. Ellison et al., 2020) as previously described (Frankston et al., 2025). Participant-specific metatarsal and toe lengths were used in the present study, obtained from marker data during standing trials. Key model assumptions were that 1) the flexor muscle forces can be represented by a long and short plantar tendon with an equal distribution of force between them, 2) the external forces are applied at points under the distal metatarsal head and second toe, 3) the second toe segment is parallel to the ground during ground contact, and 4) mediolateral forces and kinematics are negligible.

To calculate second metatarsal dorsal and plantar stresses, axial and bending stresses were first calculated from the midshaft bending moment and the axial and shear forces acting on the second metatarsal. To determine the forces acting on the second metatarsal, the second toe forces were first resolved to obtain the MTP joint reaction forces. At the second toe, the MTP sagittal plane joint moment counteracts the moment generated by the plantar tendon force. The second MTP sagittal plane joint moment was the product of the vertical force under the second toe and the length of the second toe (M. A. Ellison et al., 2020). Plantar tendon force

was calculated by dividing the second MTP joint moment by previously reported perpendicular distances between the centre of the metatarsal head and the plantar tendon (M. A. Ellison et al., 2020; Frankston et al., 2025). Vertical and horizontal MTP joint reaction forces (JRF) were calculated from the vertical and horizontal forces under the second toe and the components of the plantar tendon force. The metatarsal axial and shear forces were calculated by summing the axial and shear components of both the joint reaction forces and forces under the second metatarsal. Axial stress was the ratio of the axial force to the bone cross sectional area. The midshaft bending moment was the product of the shear force and the distance from the midpoint of the metatarsal to the centre of the metatarsal head. Bending stress was the product of the midshaft bending moment and the distance from the bone centre to the surface, divided by the area moment of inertia.

The main outcome variables were the metatarsal midshaft stresses at the dorsal and plantar bone surfaces and were calculated by summing and subtracting, respectively, the axial and bending stresses. Peak dorsal and plantar stresses were analysed. To better understand factors contributing to bone stress, we also analysed variables involved in the calculation of second metatarsal stress. All calculations were performed using custom scripts in MATLAB (R2024b, MathWorks, Natick, MA, USA).

Statistics

Statistical analyses were conducted in R (v4.2.2, R Foundation for Statistical Computing, Vienna, Austria). Normality of all variables was assessed using the Shapiro-Wilk test. For normally distributed variables, paired t-tests were conducted to compare differences pre- and post-10 km treadmill run and Cohen's d effect sizes quantified the magnitude of differences. For non-normally distributed variables, paired Wilcoxon signed-rank tests were used to compare differences pre- and post-10 km run and Wilcoxon effect sizes (r) assessed the magnitude of effects. Statistical significance was set at $p < 0.05$.

Results

All participants (Table 1) successfully completed the 10 km treadmill run. Five participants requested a reduction in treadmill speed at least once during the run, with speed reductions occurring 12 times across all participants. Total running time on the treadmill ranged from 35:20 to 63:00 minutes. The mean (± standard deviation) treadmill running speed was 3.74 ± 0.48 m/s while the mean overground preferred running speed was 3.26 ± 0.35 m/s. During the treadmill run, participants' RPE started at 12.5 ± 1.1 (range: 10 – 15) out of 20 on the Borg scale at 1 km and increased to 16.6 ± 1.9 (range: 13 – 20) at 10 km which corresponds to a perceived exertion of "very hard". Participants had an average transition time of 4:55 ± 1:11 minutes (range: 3:22 – 8:30) from the treadmill to the overground trials.

Table 1

Participant information for all participants (14 female/14 male) presented as mean ± standard deviation with range in parenthesis.

	All Participants (n=28)
Age (years)	29.6 ± 4.7 (23 - 40)
Height (m)	1.73 ± 0.09 (1.54 - 1.90)
Mass (kg)	68.8 ± 9.7 (50.0 - 85.8)
BMI (kg/m^2)	22.8 ± 3.3 (19.2 - 36.2)
Average Weekly Running Volume (km/week)	48.0 ± 24.9 (15 - 115)
Average Number of Runs per Week	4.6 ± 1.8 (3 - 11)
Running Experience (years)	7.8 ± 5.6 (2 - 22)
Estimated 10 km Race Time (Self-Reported) (min:sec)	43:08 ± 5:44 (33:30-60:00)
Previous Injury, n (%)	19 (68%)
Right Leg Dominant, n (%)	22 (79%)

Second Metatarsal Stress-Related Variables

All metatarsal stress-related variables were normally distributed ($p > 0.05$) except for vertical and horizontal force under the second toe. No significant difference was observed in

second metatarsal peak dorsal stress post- compared to pre-10 km treadmill run ($t(27) = 1.88$, $p = 0.071$, $d = 0.05$) (Table 2, Figure 1a). Peak second metatarsal plantar stress also did not significantly differ post- compared to pre- treadmill run ($t(27) = 0.84$, $p = 0.409$, $d = 0.02$) (Table 2, Figure 1b).

Peak axial stress and force decreased by 9% (axial stress: 0.60 ± 0.81 MPa; axial force: 22.8 ± 30.4 N) post- compared to pre-treadmill run ($t(27) = 3.96$, $p < 0.001$, $d = 0.33$) (Table 2, Figure 1c), while peak bending stress, bending moment, and shear force were unchanged (Table 2, Figure 1d).

Peak vertical and horizontal force under the second metatarsal were not different post- compared to pre-run. Peak MTP vertical JRF decreased by 21% (23.2 ± 30.2 N) ($t(27) = 4.06$, $p < 0.001$, $d = 0.51$), while peak MTP horizontal JRF decreased by 13% (13.0 ± 22.4 N) ($t(27) = 3.07$, $p = 0.005$, $d = 0.33$) (Table 2, Figure 1e-f). Peak vertical force under the second toe decreased by 15% (5.07 ± 7.91 N) ($z = 3.03$, $p = 0.002$, $r = 0.57$), while peak horizontal force under the second toe decreased by 26% (2.62 ± 3.45 N) ($z = 3.53$, $p < 0.001$, $r = 0.67$) (Table 2). Peak plantar tendon force decreased by 14% (17.0 ± 27.3 N) ($t = 3.31$, $p = 0.003$, $d = 0.37$) (Table 2).

Second metatarsal pitch angle at the time of peak stress decreased by a mean of 0.8 ± 2.0° post- compared to pre-10 km run ($t(27) = 2.12$, $p = 0.043$, $d = 0.21$) (Table 2, Figure 1g), indicating a less inclined metatarsal post-run.

Peak whole body vertical ground reaction force (GRF) decreased by 2% (39.8 ± 41.1 N) ($t(27) = 5.12$, $p < 0.001$, $d = 0.14$) post- compared to pre-10 km run, but peak whole body horizontal GRF was not different (Table 2). There were no differences in the timing of peak dorsal or plantar stress post- compared to pre-10 km run ($p > 0.05$).

Table 2

Peak stresses, bending moment, forces, and pitch angle for the second metatarsal presented as mean ± standard deviation with 95% confidence intervals displayed in brackets. Effect sizes are Cohen's d unless indicated with 95% confidence intervals shown in brackets.

Variable	Pre-Run	Post-Run	P value	Effect Size
Peak Dorsal Stress (MPa)	-144 ± 37 [-130, -158]	-142 ± 35 [-129, -156]	0.071	0.05 [-0.48, 0.59]
Peak Plantar Stress (MPa)	131 ± 34 [144, 118]	130 ± 33 [143, 118]	0.409	0.02 [-0.51, 0.56]
Axial Stress (MPa)	6.79 ± 1.96 [6.03, 7.55]	6.18 ± 1.68 [5.53, 6.83]	<0.001*	0.33 [-0.21, 0.87]
Bending Stress (MPa)	138 ± 35 [124, 151]	136 ± 34 [123, 149]	0.168	0.04 [-0.50, 0.57]
Peak Bending Moment (Nm)	7.55 ± 1.94 [6.79, 8.30]	7.48 ± 1.84 [6.76, 8.19]	0.168	0.04 [-0.50, 0.57]
Peak Shear Force (N)	266 ± 54 [246, 287]	264 ± 50 [244, 283]	0.192	0.05 [-0.49, 0.58]
Peak Axial Force (N)	256 ± 74 [227, 285]	233 ± 63 [209, 258]	<0.001*	0.33 [-0.21, 0.87]
Second Metatarsal Pitch Angle at Time of Peak Stress (°)	29.5 ± 3.7 [28.1, 31.0]	28.8 ± 3.7 [27.3, 30.2]	0.043*	0.21 [-0.32, 0.75]
Peak Vertical GRF under Second Metatarsal (N)	279 ± 56 [257, 300]	276 ± 53 [255, 297]	0.114	0.05 [-0.48, 0.59]
Peak Horizontal GRF under Second Metatarsal (N)	51.4 ± 15.4 [45.4, 57.3]	53.1 ± 15.2 [47.2, 59.0]	0.196	-0.11 [-0.65, 0.42]
Peak MTP Vertical JRF (N)	111 ± 50 [92, 131]	88 ± 40 [73, 104]	<0.001*	0.51 [-0.03, 1.06]
Peak MTP Horizontal JRF (N)	103 ± 44 [86, 120]	90 ± 35 [77, 104]	0.005*	0.33 [-0.21, 0.87]
Peak Vertical GRF under Second Toe (N)	34.8 ± 14.7 [29.1, 40.5]	29.7 ± 11.5 [25.2, 34.1]	0.002* [a]	0.57 [b] [0.25, 0.80]
Peak Horizontal GRF under Second Toe (N)	10.2 ± 5.4 [8.1, 12.3]	7.6 ± 4.4 [5.9, 9.3]	<0.001* [a]	0.67 [b] [0.40, 0.88]
Plantar Tendon Force (N)	124 ± 51 [104, 143]	107 ± 42 [90, 123]	0.003*	0.37 [-0.17, 0.91]
Peak Whole Body Vertical GRF (N)	1678 ± 290 [1566, 1791]	1639 ± 285 [1528, 1749]	<0.001*	0.14 [-0.40, 0.67]
Peak Whole Body Horizontal GRF (N)	248 ± 42 [231, 264]	244 ± 44 [227, 261]	0.143	0.09 [-0.44, 0.63]

MPa = megapascals. Nm = Newton metres. N = Newtons. GRF = ground reaction force. JRF = joint reaction force. * = significant difference post- compared to pre-10 km run ($p < 0.05$), a = non-parametric Wilcoxon sign-rank test performed, b = non-parametric Wilcoxon effect size.

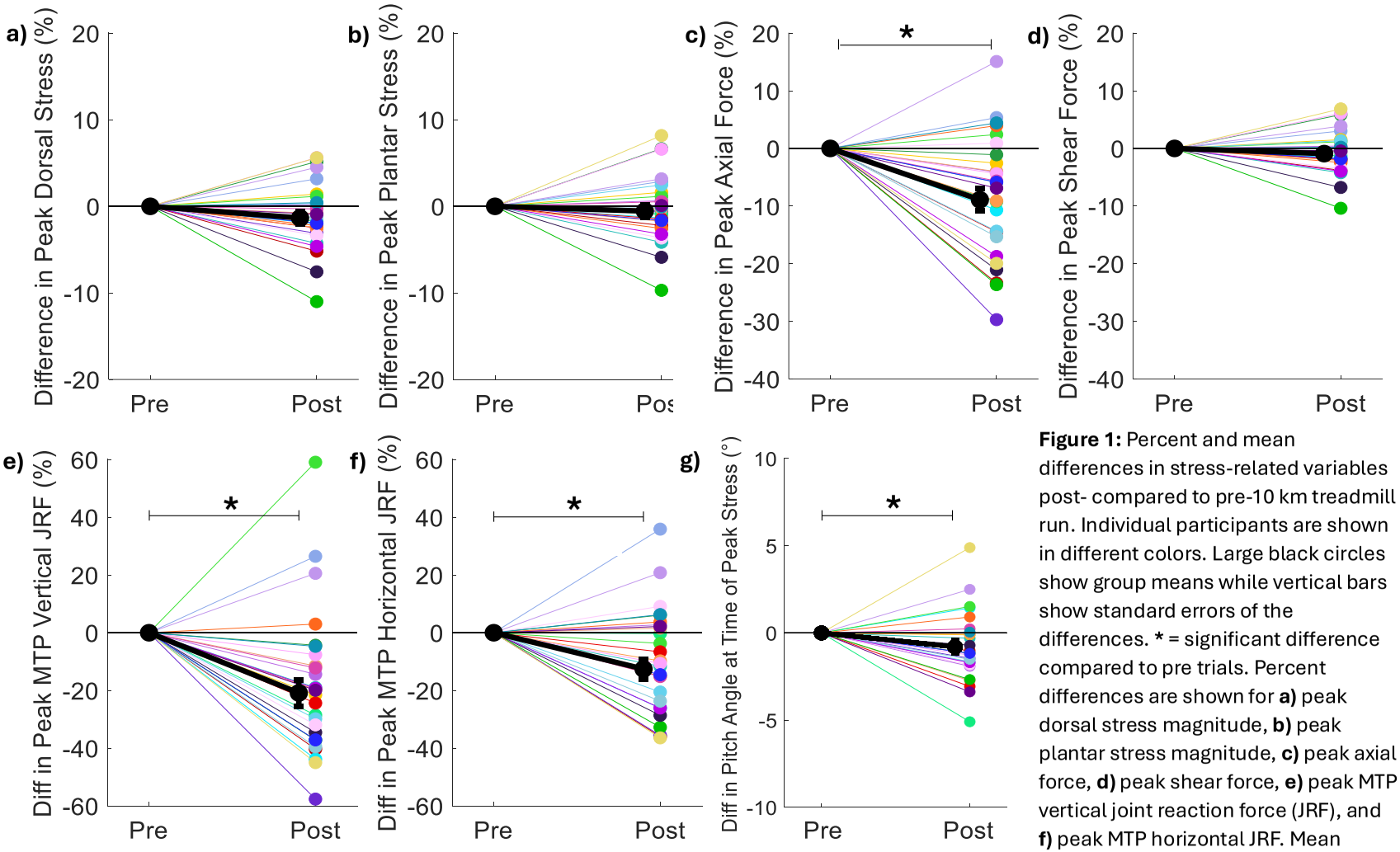


Figure 1: Percent and mean differences in stress-related variables post- compared to pre-10 km treadmill run. Individual participants are shown in different colors. Large black circles show group means while vertical bars show standard errors of the differences. * = significant difference compared to pre trials. Percent differences are shown for **a)** peak dorsal stress magnitude, **b)** peak plantar stress magnitude, **c)** peak axial force, **d)** peak shear force, **e)** peak MTP vertical joint reaction force (JRF), and **f)** peak MTP horizontal JRF. Mean difference is shown for **g)** pitch angle at time of peak stress.

Joint Angles and Moments

All peak joint angles and moments were normally distributed ($p > 0.05$). Peak ankle plantarflexion moment decreased by a mean of 6% (11.7 ± 16.9 Nm) post- compared to pre-10 km treadmill run ($t(27) = -3.68$, $p = 0.001$, $d = -0.25$) (Table 3, Figure 2b, Figure 3a). Peak first MTP dorsiflexion angle increased by a mean of 3.3 ± 4.9° post- compared to pre-10 km run ($t(27) = -3.63$, $p = 0.001$, $d = -0.56$) (Table 3, Figure 2c, Figure 3c). Peak second MTP dorsiflexion moment decreased by 14% (0.24 ± 0.35 Nm) post- compared to pre-10 km run ($t(27) = 3.52$, $p = 0.002$, $d = 0.38$) (Table 3, Figure 2d, Figure 3b).

There was no change in peak rearfoot eversion angle, peak knee abduction angle, peak hip adduction angle, peak knee internal rotation angle, sagittal plane angle between the foot and ground at initial contact, peak second MTP dorsiflexion angle, or peak first MTP dorsiflexion moment post- compared to pre- 10 km treadmill run ($p > 0.05$, Table 3).

Gait Characteristics

Stride length was normally distributed ($p > 0.05$), while stride frequency and ground contact time were not ($p < 0.05$). Stride length (pre: 2.41 ± 0.23 m; post: 2.39 ± 0.24 m; $p = 0.301$), stride frequency (pre: 1.38 ± 0.09 Hz; post: 1.36 ± 0.11 Hz; $p = 0.864$), and ground contact time (pre: 225 ± 24 ms; post: 227 ± 21 ms; $p = 0.202$) did not change significantly post- compared to pre-10 km run.

Table 3

Peak joint angles and moments presented as mean ± standard deviation with 95% confidence intervals displayed in brackets. Effect sizes are Cohen's d unless indicated with 95% confidence intervals shown in brackets.

Variable	Pre-Run	Post-Run	P value	Effect Size
Peak Rearfoot Eversion Angle (°)	-7.5 ± 2.9 [-8.7, -6.4]	-7.1 ± 2.9 [-8.3, -6.0]	0.129	-0.13 [-0.67, 0.40]
Peak Knee Abduction Angle (°)	-3.7 ± 3.2 [-5.0 -2.5]	-3.9 ± 3.8 [-5.4, -2.4]	0.644	0.05 [-0.48, 0.59]
Peak Hip Adduction Angle (°)	11.0 ± 3.9 [9.5, 12.6]	11.3 ± 3.7 [9.9, 12.7]	0.374	-0.08 [-0.62, 0.45]
Peak Knee Internal Rotation Angle (°)	7.7 ± 5.7 [5.5, 9.9]	7.0 ± 5.9 [4.7, 9.3]	0.363	0.11 [-0.42, 0.65]
Foot to Ground Sagittal Angle at Initial Contact (°)	-14.6 ± 4.4 [-16.3, -12.9]	-14.9 ± 4.0 [-16.4, -13.3]	0.412	0.07 [-0.46, 0.61]
Peak First MTP Dorsiflexion Angle (°)	40.1 ± 5.7 [37.9, 42.3]	43.4 ± 6.1 [41.0, 45.8]	<0.001*	-0.58 [-0.96, 0.12]
Peak Second MTP Dorsiflexion Angle (°)	48.2 ± 5.8 [45.9, 50.4]	48.3 ± 5.6 [46.1, 50.4]	0.866	-0.02 [-0.55, 0.52]
Peak Ankle Plantarflexion Moment (Nm)	-193 ± 50 [-213, -174]	-182 ± 42 [-198, -165]	0.001*	-0.25 [-0.79, 0.28]
Peak First MTP Dorsiflexion Moment (Nm)†	1.8 ± 0.6 [2.0, 1.6]	1.8 ± 0.5 [2.0, 1.6]	0.761	0.03 [-0.51, 0.56]
Peak Second MTP Dorsiflexion Moment (Nm)	1.6 ± 0.7 [1.4, 1.9]	1.4 ± 0.5 [1.2, 1.6]	0.002*	0.38 [-0.16, 0.92]

MTP = metatarsophalangeal, Nm = Newton meters, * = significant effect ($p < 0.05$), † = peak during the last 50% of stance.

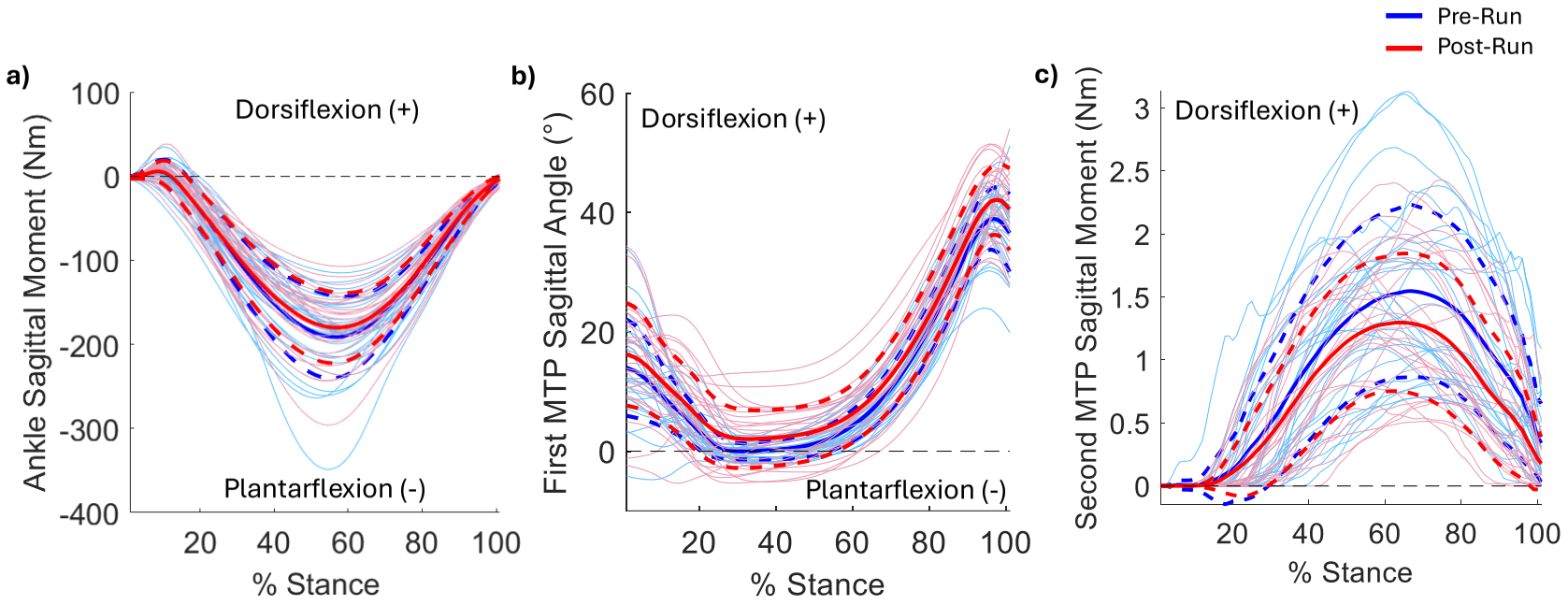


Figure 2: Time series during the stance phase of running comparing overground kinematics and kinetics pre- (blue) and post- (red) 10 km treadmill run. Thick lines represent the mean for all participants (n = 28), dashed lines show standard deviations, and thin lines represent individual participants averaged over trials. Time series are shown for **a)** ankle sagittal moment, **b)** first metatarsophalangeal (MTP) sagittal plane angle, and **c)** second MTP sagittal moment.

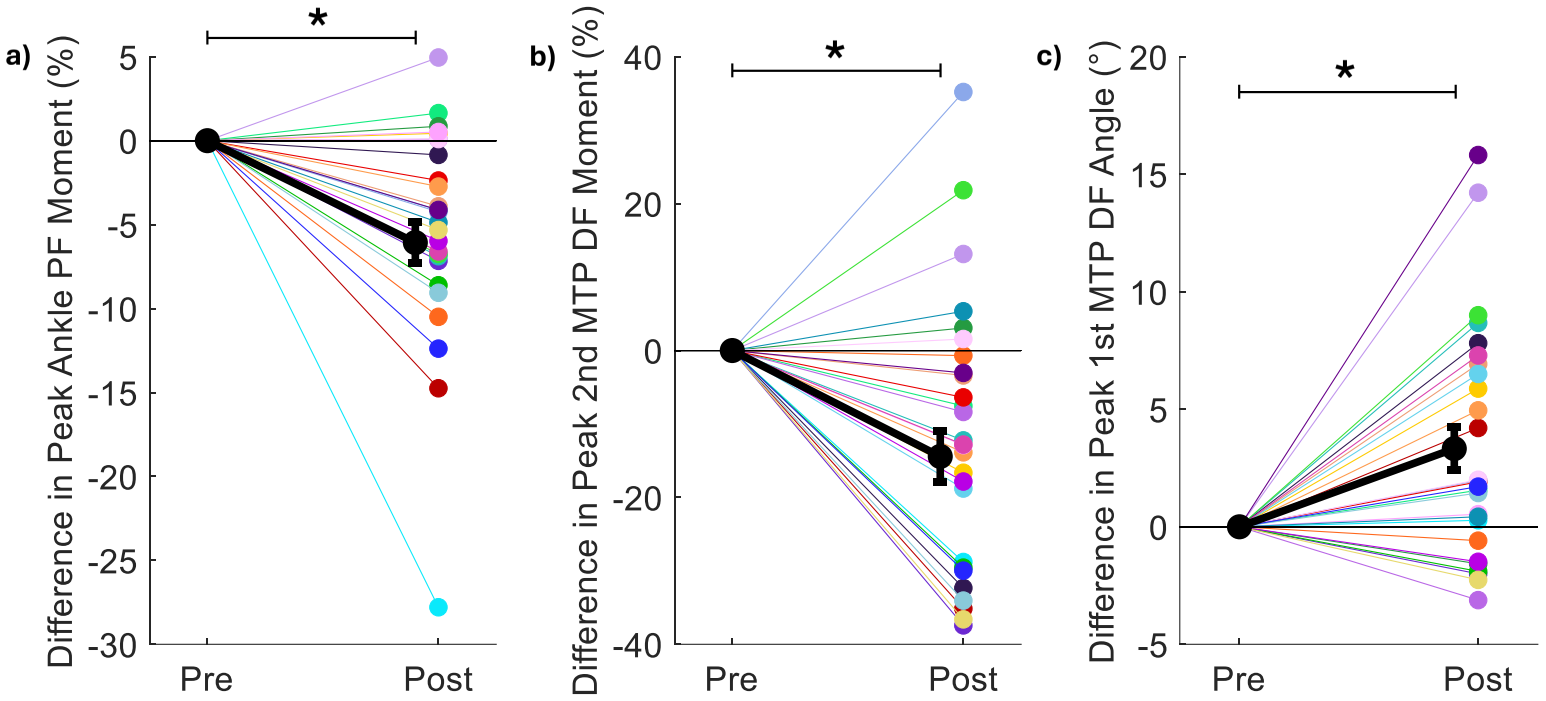


Figure 3: Percent and mean differences in kinematic and kinetic variables post- compared to pre-10 km treadmill run. Individual participants are shown in different colours. Larger black circles show group means while vertical bars show standard errors of the differences. * = significant difference compared to pre trials. Percent differences are shown for **a)** peak ankle plantarflexion (PF) moment and **b)** peak second metatarsophalangeal (MTP) dorsiflexion (DF) moment. Mean difference is shown for **c)** peak first MTP dorsiflexion angle.

Discussion and Implication

This study quantified changes in second metatarsal bone stress and lower-extremity joint mechanics during overground running at preferred speed following a demanding 10 km treadmill run. Our hypothesis that second metatarsal dorsal and plantar stress would increase following a demanding 10 km treadmill run was not supported as no differences were found. Despite the lack of change in peak metatarsal stress, the reduction in peak ankle plantarflexion moment and the increase in peak first MTP dorsiflexion angle indicate that the protocol was effective at inducing localized gait changes even given the study's methodological limitations. Methodological limitations include that the 10 km run was performed shod on a treadmill at a faster speed while the measurements were conducted overground during barefoot running at preferred speed. Additionally, the approximately five minute transition time between the treadmill and overground running may have allowed for partial recovery.

Our findings differ from previous studies which have found increased bone loading with exercise-induced fatigue. Peak second metatarsal strain has previously been shown to increase *in vivo* during walking following walking-induced fatigue (Arndt et al., 2002; Donahue & Sharkey, 1999) and marching-induced fatigue (Gefen, 2002). In the tibia, compression and tension strain have also been shown to increase following a fatiguing march and a fatiguing run (Milgrom et al., 2007). The lack of change in peak stresses in the present study can likely be explained by the lack of change in peak vertical force under the second metatarsal. Our previous work supports this since we showed peak dorsal stress to be very strongly correlated with peak second metatarsal vertical force (Frankston et al., 2025). We have also shown the model to be sensitive to relatively small changes in second metatarsal vertical force (M. Ellison, 2020, p. 99). Our results differ from previous studies which found increased vertical force under the second metatarsal following a fatiguing run (Anbarian & Esmaeili, 2016; Weist et al., 2004; Willems et al., 2012). The difference in findings may be due to

methodological differences including footwear conditions (barefoot vs shod), gait type (walking, marching, or running), and running surface (treadmill vs overground). In combination with the approximately five-minute transition period, these methodological differences may have allowed for partial attenuation of the effects of the 10 km run.

The second metatarsal pitch angle at the time of peak stress was more parallel to the ground post-10 km treadmill run. While in isolation this change would have resulted in greater bending stress, bending stress remained unchanged post-10 km run. The lack of change in bending stress can be explained by the concurrent reduction in peak forces under the second toe post-run, which resulted in reduced plantar tendon and joint reaction forces and counteracted the influence of reduced metatarsal pitch angle. While peak bending stress and shear force did not change, peak second metatarsal axial stress and force decreased post-10 km treadmill run, although with a small effect size. Since the contribution of axial stress to dorsal and plantar stress is negligible compared to the contribution of bending stress (Table 2), the lack of change in dorsal and plantar stress following the 10 km run can be explained by the lack of change in bending stress. Overall, the results suggest a change in the loading environment of the second ray following the 10 km run. The absence of a corresponding change in metatarsal stress may either be explained by these counteracting mechanisms or may reflect a lack of sensitivity of the metatarsal model to detect small changes, such as those which could occur from altered forces acting under the second toe.

Our finding of reduced peak ankle plantarflexion moment after the 10 km run supports the findings of Sanno et al. (2018), who found a similar reduction in the final kilometre of a 10 km treadmill run, although we observed only a small effect size in the present study whereas the previous authors reported a large effect size. Participants' RPE in the present study was comparable to that reported by Sanno et al. (2018) when using the same 10 km protocol. The RPE results and the decrease in ankle plantarflexion moment following the run in both the

present study and Sanno et al.'s (2018) study indicate a similar effect of the running protocol on lower-extremity mechanics. Similar effects were found despite data being collected during barefoot overground running in the present study compared to shod treadmill running in Sanno et al.'s study, although the difference in effect size may be indicative of a recovery effect in the protocol used in the present study.

We found no change in peak rearfoot eversion angle, peak hip and knee abduction angles, or peak knee internal rotation angle after the 10 km run in contrast to Willwacher et al. (2020) who found kinematic changes in these variables during the final kilometre of a 10 km treadmill run. The difference in findings may be explained by methodological differences including the approximately five minutes of recovery time participants had between treadmill and overground running, which was not the case in Willwacher et al.'s (2020) study. Additionally, participants ran at different speeds during the treadmill and overground running, with the treadmill speed being a mean of ~0.48 m/s faster than overground running. It seems therefore that running-induced changes in kinematics may be more sensitive to the testing protocol than the kinetics are.

The increase in first MTP dorsiflexion angle following the 10 km run was associated with a medium effect size. The average increase of 3.3° may indicate altered movement strategies or fatigue of the hallux flexor muscles which eccentrically control MTP joint dorsiflexion prior to push-off. The unchanged first MTP peak dorsiflexion moment along with the increase in first MTP peak dorsiflexion angle suggests reduced MTP joint stiffness following the 10 km treadmill run. The second MTP joint also showed reduced stiffness through a combination of an unchanged peak second MTP dorsiflexion angle and a decreased second MTP dorsiflexion moment after the 10 km run. The reduction in second MTP moment resulted from reduced forces under the second toe, which is consistent with previous studies which also reported reductions in forces under the lesser toes after demanding running

protocols (Anbarian & Esmaeili, 2016; Liu et al., 2023; Willems et al., 2012). The overall MTP findings in the present study indicate that although the first and second MTP joints showed different responses to the 10 km run, both exhibited changes consistent with reduced joint stiffness which may be indicative of muscular fatigue or adaptations in movement strategies.

One of the main limitations of the present study was that the 10 km protocol was conducted on a treadmill while kinematics and kinetics were measured during overground running. Data were captured overground due to the use of the pressure plate and the force plate together to allow for collection of localized vertical and horizontal forces under the second metatarsal. Participants also ran in their own running shoes during the treadmill run for ecological validity but ran barefoot during data collection to allow for marker placement directly on the skin and to eliminate the influence of footwear on gait mechanics. Additionally, the pressure plate was a harder surface than the softer EVA runway, which may have caused participants to alter their gait when landing on the pressure plate. Another limitation was that the overground trials were run at a slower speed than the treadmill run. Reducing running speed may alter muscle recruitment patterns and potentially mitigate the effects of muscle fatigue on metatarsal bone stress. Future research should determine how running at the same speed as during a fatiguing running protocol influences second metatarsal bone stress.

The model used was a simple two-dimensional cantilever model which did not account for second metatarsal eversion angle, mediolateral forces, or bone torsion as these were assumed to have a negligible effect on bone stress. Bone stress was estimated at the mid-shaft of the metatarsal which is assumed to be the area with the smallest cross-sectional area and thus highest risk of bone stress injury. It is unknown, however, whether this is the most common location for metatarsal bone stress injuries in runners. The model also does not account for altered load distribution across the metatarsal, which could have resulted from changes such as the reduced metatarsal pitch angle post-10 km run.

Conclusion

Completion of a demanding 10 km treadmill run did not change peak second metatarsal dorsal or plantar stress, likely due to the peak vertical force under the second metatarsal remaining unchanged. The reduction in peak ankle plantarflexion moment and the increase in peak first metatarsophalangeal dorsiflexion angle following the 10 km run indicate that the protocol induced gait changes. Therefore, while a demanding 10 km treadmill run may cause changes in lower-extremity running mechanics, it may not influence metatarsal loading during overground running at preferred speed.

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Paper III

Within-Day Repeatability and Between-Day Reliability of Estimates of Second Metatarsal Bone Stress During Running

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Abstract

Background

Internal metatarsal bone stress can be estimated during running using a two-dimensional beam theory model. Variability in the model inputs including experimentally measured geometry, kinetics, and kinematics can influence the reliability of bone stress estimates within and between days.

Purpose

This study aimed to quantify the within-day repeatability and between-day reliability of estimates of second metatarsal bone stress when using a beam theory model to estimate *in vivo* stress.

Methods

Twenty-eight (14 F/14 M) healthy runners visited the laboratory on two separate occasions. During each visit, participants performed barefoot running trials at their preferred speed while motion capture, ground reaction forces, and plantar pressure data were recorded. Intraclass correlation coefficients (ICC) were calculated to determine how repeatable metatarsal bones stress estimates were between five to eight trials within a single laboratory visit and between visits.

Results

Second metatarsal dorsal and plantar stress demonstrated excellent (ICC > 0.9) within-day repeatability and between-day reliability. Between-day reliability of the relative positions of the motion capture markers used to define the second metatarsal and second toe lengths was moderate to excellent. Replacing visit-specific bone lengths with the same bone lengths for each visit improved between-day reliability of bone stress by ~6-7%.

Significance

Quantifying within-day repeatability and between-day reliability of metatarsal bone stress during running is important for distinguishing normal physiological variability from methodological effects and for contextualizing changes in estimated metatarsal bone stress.

Keywords

Metatarsal Stress; Measurement Reliability; Modeling; Foot Biomechanics

Introduction

Runners have a high incidence of bone stress injuries in the foot (Hoenig et al., 2024), particularly in the second metatarsal (Bennell et al., 1996; Brukner et al., 1996; Nattiv et al., 2013). The second metatarsal has been shown to experience high vertical plantar force, bending strain, shear force (Gross & Bunch, 1989), compression strain (Donahue et al., 2000; Milgrom et al., 2002, 2011; Niu et al., 2015), and tension strain (Milgrom et al., 2002, 2011) during dynamic loading when compared to the other metatarsals and the tibia. By estimating internal bone stress, we can better understand factors contributing to the risk of metatarsal bone stress injury. Internal bones stress is, however, difficult to measure and requires either surgical implantation of strain gauges (Arndt et al., 2002; Donahue et al., 2000; Milgrom et al., 2002) or computational modeling (M. A. Ellison, Akrami, et al., 2020; M. A. Ellison, Kenny, et al., 2020; Firminger et al., 2017).

A two-dimensional beam theory model of the second metatarsal has previously been used to estimate internal metatarsal bone stress during running (M. A. Ellison, Kenny, et al., 2020; Frankston et al., 2025, 2026). The model relies on experimentally measured geometric, kinetic, and kinematic inputs including the forces under the second metatarsal and the pitch angle of the metatarsal to estimate bone stress (M. A. Ellison, Kenny, et al., 2020). These model inputs can vary due to both normal physiological fluctuations in running mechanics and methodological factors such as marker placement. Differences in motion capture marker placement between laboratory visits may influence bone stress estimates due to the use of marker positions to determine second metatarsal and second toe lengths. A previous sensitivity analysis of the same model demonstrated that a 10% increase in second metatarsal length increased peak

dorsal stress by 10.4% and peak plantar stress by 11.7%, while a 10% increase in second toe length decreased peak dorsal stress by -0.1% and peak plantar stress by -0.6% (M. Ellison, 2020, p. 99). Due to potential variability in model inputs, it is unknown how repeatable metatarsal bone stress estimates are between trials within a single laboratory visit and how reliable these estimates are across separate laboratory visits. Quantifying both within-day repeatability and between-day reliability is important for distinguishing normal physiological variability from meaningful experimental effects and for contextualizing changes in estimated metatarsal bone stress.

Within-day repeatability of the model reflects how consistent outcome measurements are across different running steps that occur during the same testing session when physiological factors such as fatigue and training load are relatively stable. Between-day reliability represents how consistent outcome measurements are between testing sessions when physiological factors may fluctuate and running technique may differ more between days than within a single session. The purpose of the present study was to quantify the within-day repeatability and between-day reliability of second metatarsal bone stress when estimated using experimental data and a beam theory model of the metatarsal. We did not make a directional hypothesis due to the lack of prior studies assessing the reliability of bone stress estimates.

Methods

Participants

Because study data had already been collected, a post hoc precision-based approach (Bonett, 2002) was used to assess sample size adequacy. The observed ICC of 0.93 for dorsal stress was used to calculate the minimum sample size required to achieve

a 95% confidence interval width of 0.10. A minimum required sample size of 17 participants was calculated in R (v4.5.1, R Foundation for Statistical Computing, Vienna, Austria).

The present study analyzed data previously collected in related studies (Frankston et al., 2025, 2026). Participants were 14 females and 14 males aged 18-40 years, ran at least 15 km per week, ran at least three times per week, had been running on a regular basis for at least three years, ran with a rearfoot strike, had completed any distance from five kilometers to a marathon at a maximal effort in the last two years, and could run barefoot without pain. Exclusion criteria were any neurological, musculoskeletal, or movement disorder, current pregnancy, any lower extremity injury within the previous six months, any previous foot or ankle surgery, any symptomatic foot pathology, and any illness or injury that precludes running.

Running history and descriptive information for each participant were obtained through a digital questionnaire. The study was approved by the local ethics committee at the Norwegian School of Sport Sciences (approval number 281 – 150623) and all participants provided written informed consent prior to participation.

Procedures

Participants visited the biomechanics laboratory on two separate occasions a mean (± standard deviation) of 21 ± 26 days apart (range: 1 – 139 days). During the first laboratory visit, participants were screened for foot strike type as previously described (Frankston et al., 2025) so that only rearfoot strike runners were included in the study. The foot strike screening trials also served as a warmup and acclimatization to the testing

procedures. Preferred running speed for each participant was determined as the mean running speed from the barefoot screening trials (Frankston et al., 2025).

The right foot was selected arbitrarily for all participants. On both visits, retro-reflective spherical markers (6 mm in diameter) were attached to the skin of participants' right foot at the second metatarsal head at the dorso-medial aspect of the second metatarso-phalangeal joint, second metatarsal base at the dorso-medial aspect of the second metatarso-cuneiform joint, and the head of the middle phalanx of the second toe at the most distal and dorsal point to define the second metatarsal and second toe in a modification of the Istituto Ortopedico Rizzoli marker set (Leardini et al., 2007). Additional markers defined participants' right foot, shank, thigh, pelvis, and trunk as previously described (Frankston et al., 2026). Marker locations for each participant were determined by the same researcher (NF) through palpation of bony landmarks.

On both laboratory visits, each participant ran barefoot overground at the same preferred running speed (± 5%) as determined for that participant during the screening trials. During the running trials, kinematics were recorded in three-dimensions at 185 Hz using 23 motion capture cameras (Oqus 500 and 700, Qualisys, Gothenberg, Sweden). A static standing trial was recorded before the running trials to define reference marker locations and segment lengths. A pressure plate (HR Mat, TekScan, Norwood, MA, USA, 0.49 m 0.45 m 0.01 m, with 8,448 sensors, 185 Hz, 3.9 sensors/cm^2) recorded plantar forces during running and was set in a 14.4-m ethylene-vinyl acetate (EVA) runway (0.02 m thick) with a gap where the pressure plate was located. The pressure plate was located directly over a floor-embedded force plate which recorded three-dimensional forces from the same step at 1850 Hz (AMTI, Watertown, MA, USA).

A successful trial was defined as a trial where the whole foot landed within the bounds of the pressure plate, the participant did not appear to target the plate, the running speed was within ± 5% of the preferred speed, and the participant landed with a rearfoot strike. Five to eight successful running trials were collected for each participant on each visit. Up to eight trials were collected to ensure a minimum of five usable trials. For the first laboratory visit, the preferred speed running trials were one of six running conditions conducted in a randomized order (Frankston et al., 2025), while for the second visit the preferred speed trials were the first running condition conducted (Frankston et al., 2026). Participants' training load and physiological state (such as fatigue levels) were not controlled for.

Data Analysis

To determine stance phase during running, a vertical force threshold of 5 N was used for both the pressure and force data to retain as many frames of pressure data as possible. Marker positions and ground reaction force data were filtered with a fourth-order lowpass Butterworth filter with a cut-off frequency of 20 Hz in Visual3D (HAS Motion, Kingston, Ontario, Canada). Data were interpolated to 101 frames of stance phase. For the pressure data, regional masks were automatically applied using Strideway software (Strideway 7.80, TekScan, Norwood, MA, USA) to define second metatarsal and second toe regions. Regional masks were manually adjusted by the same researcher (NF) to ensure that each region of interest was fully contained within its respective mask. To maintain consistency between trials, the minimum number of adjustments were made to define each region (Figure 1).

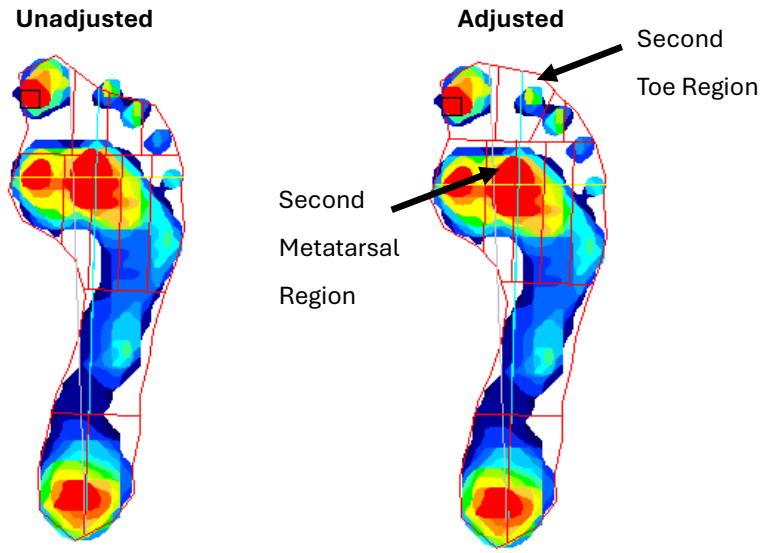


Figure 1: Example from a single trial demonstrating the manual adjustments made to the automatic regional masks in Strideway v7.80 (TekScan, Norwood, MA, USA). The left image shows the unadjusted regional masks and the right image shows the masks after manual adjustment for the second metatarsal and second toe regions using the minimum number of adjustments to define each region.

For the calculation of stride length and frequency, foot strikes were detected when the vertical velocity of the inferior posterior calcaneus marker crossed zero from negative to positive. Additionally, an anteroposterior heel marker velocity threshold of near zero (6.5×10^{-5}) was set to prevent false foot strike detections during swing phase. One stride per trial was analyzed with a minimum of five strides required per participant for inclusion in the analysis.

Beam Theory Model

A two-dimensional beam theory model of the second metatarsal was used to estimate dorsal and plantar stress at the bone midshaft (M. A. Ellison, Kenny, et al., 2020). Participant-specific metatarsal and toe lengths calculated from the static trial were used

in the model along with mean bone geometries from a cohort of 20 healthy recreational runners (M. A. Ellison, Kenny, et al., 2020) as previously described (Frankston et al., 2025). Model assumptions have been reported previously (M. A. Ellison, Kenny, et al., 2020). Briefly, the model assumes that external forces are applied at points under the second metatarsal head and second toe and that mediolateral forces and kinematics are negligible.

Model calculations have previously been described in detail (M. A. Ellison, Kenny, et al., 2020). Briefly, second metatarsal axial and shear forces were calculated by summing the axial and shear components of the joint reaction forces and external forces acting under the second metatarsal. Bending moment was the product of shear force and distance from the metatarsal midpoint to the center of the metatarsal head. Dorsal stress was the sum of axial and bending stress and plantar stress was the difference between axial and bending stress. Peak values were analyzed.

Model Inputs

External forces acting under the second metatarsal and second toe were input into the model and were calculated by scaling the vertical and horizontal whole body ground reaction forces by the percentage of force under those regions in the pressure data. Second metatarsal pitch angle was also required by the model and was calculated from the inverse sine of the ratio of the vertical distance between the second metatarsal head and base markers divided by the three-dimensional (3D) vector distance between those markers. Second metatarsal and second toe lengths were input into the model and were quantified from the static trial as the 3D vector distance between the second metatarsal head and base markers and as the 3D vector distance between markers on

the second metatarsal head and the head of the middle phalanx of the second toe, respectively.

Repeatability and Reliability Analyses

Intraclass correlation coefficients (ICC) were used to assess within-day repeatability and between-day reliability for stress-related variables and stride characteristics. ICCs were calculated using the irr package (Gamer et al., 2019) in R (v4.5.1, R Foundation for Statistical Computing, Vienna, Austria) for a two-way mixed-effects model with absolute agreement and average measures. ICC values were interpreted as poor (< 0.5), moderate (0.5 – 0.75), good (0.75 – 0.9), or excellent (> 0.9) (Koo & Li, 2016). Standard error of measurement (SEM) was calculated to estimate the magnitude of random measurement error in the same units as the variable of interest. SEM was calculated between days as SD * $\sqrt{}$(1 – ICC) where SD was pooled standard deviation between days. SEM was also reported as a percentage of the mean value for each variable. The minimal detectable change (MDC) was used to represent the smallest change which exceeds measurement error and indicates a true change. MDC was calculated as SEM*1.96*$\sqrt{}$2.

Bone stress for each visit was calculated using the second metatarsal and second toe lengths from the static trial of the given visit. To assess the influence of metatarsal and toe lengths on dorsal and plantar stress, bone stress was also computed using constant bone lengths by applying the first-visit lengths to both visits.

To determine the change in relative marker positions between days, the relative distances between the second metatarsal head and base markers and reference markers (first and fifth metatarsal head and posterior inferior calcaneus markers) were

determined by calculating the 3D vector distances between these markers. Mean and percent differences of relative marker distances were calculated. All calculations were performed in MATLAB (R2024b, MathWorks, Natick, MA, USA).

Bland-Altman plots were created in R to provide an overview of agreement between visits for representative stress, force, and geometric variables related to second metatarsal stress.

Results

All 28 participants completed the testing on both days. The number of days between laboratory visits varied between participants, with 14% of participants following up in 1 day, 32% within 7 days, 36% within 14 days, 86% within 30 days, and 96% within 45 days. Participants had a mean (± standard deviation) age of 29.6 ± 4.7 years, body mass of 68.8 ± 9.7 kg, height of 1.73 ± 0.09 m, body mass index of 22.8 ± 3.3 $kg \cdot m^{-2}$, running experience of 7.8 ± 5.6 years, and weekly running volume of 48 ± 25 km.

Within-day repeatability was excellent for peak dorsal and plantar stress, peak bending moment, and normalized peak vertical force (Table 1). Within-day repeatability was good for peak shear force, peak axial force, peak vertical and horizontal force under the second metatarsal, second metatarsal pitch angle at the time of peak stress, normalized peak whole body horizontal ground reaction force (GRF), percent of GRF under the second metatarsal at the time of peak stress, and maximum percent of GRF under the second metatarsal (Table 1).

Between-day reliability was excellent for all stress-related variables except for second metatarsal and second toe lengths which showed moderate reliability (Table 2). Second metatarsal length was longer by a mean of 1.9% (mean difference ± standard

Table 1

Within-day repeatability metrics and percent difference in ICC between visits for stress-related variables.

	Visit 1	Visit 2	
Variable	ICC [95% CI]	ICC [95% CI]	% Δ ICC
Peak Dorsal Stress (MPa)	0.916 [0.807, 0.957]	0.930 [0.845, 0.961]	1.5
Peak Plantar Stress (MPa)	0.919 [0.800, 0.956]	0.933 [0.875, 0.967]	1.4
Peak Bending Moment (Nm)	0.918 [0.795, 0.958]	0.932 [0.883, 0.975]	1.5
Peak Shear Force (N)	0.877 [0.654, 0.935]	0.890 [0.796, 0.943]	1.4
Peak Axial Force (N)	0.840 [0.647, 0.919]	0.817 [0.636, 0.892]	-2.7
Peak Vertical GRF under Second Metatarsal (N)	0.882 [0.791, 0.936]	0.890 [0.793, 0.943]	0.9
Peak Horizontal GRF under Second Metatarsal (N)	0.862 [0.750, 0.918]	0.873 [0.752, 0.941]	1.3
Second Metatarsal Pitch Angle at Time of Peak Stress (°)	0.851 [0.705, 0.940]	0.933 [0.875, 0.959]	9.6
Normalized Peak Whole Body Vertical GRF (N/kg)	0.956 [0.903, 0.977]	0.944 [0.867, 0.971]	-1.3
Normalized Peak Whole Body Horizontal GRF (N/kg)	0.857 [0.735, 0.933]	0.805 [0.575, 0.878]	-6.1
Percent of GRF under Second Metatarsal at Time of Peak Stress (%)	0.740 [0.539, 0.860]	0.839 [0.730, 0.907]	13.4
Maximum Percent of Force under Second Metatarsal (%)	0.801 [0.597, 0.889]	0.816 [0.704, 0.891]	1.9

MPa = Megapascals. Nm = Newton meters. N = Newtons. GRF = Ground reaction force. ICC = intraclass correlation coefficient. 95% CI = 95% confidence interval. % Δ ICC = Percent difference in ICC between visits.

error (SE) of the difference: 1.2 ±1.2 mm) when measured on the second visit compared to the first visit, while second toe length was longer by a mean of 2.6% (mean difference ± SE: 1.2 ± 0.9 mm). When the same second metatarsal length was used for both visits, between-day reliability increased by 6.0% for peak dorsal stress and 7.0% for plantar stress (Table 3). When the same second toe length was used for both visits, between-

day reliability did not change for peak dorsal stress and increased by 0.1% for peak plantar stress (Table 3).

Table 2

Between-day reliability metrics for stress-related variables.

Variable	Mean ± SD		ICC	95% CI of ICC		SEM	% SEM	MDC
	Visit 1	Visit 2		Lower	Upper			
Peak Dorsal Stress (MPa)	138 ± 32	144 ± 37	0.932	0.847	0.969	9.0	6.4	25.0
Peak Plantar Stress (MPa)	125 ± 30	131 ± 34	0.924	0.830	0.965	8.9	7.0	24.8
Peak Bending Moment (Nm)	7.2 ± 1.7	7.5 ± 1.9	0.928	0.839	0.967	0.5	6.7	1.8
Peak Shear Force (N)	261 ± 50	266 ± 54	0.983	0.960	0.992	6.8	2.6	18.8
Peak Axial Force (N)	251 ± 72	256 ± 50	0.953	0.900	0.978	15.8	6.2	43.7
Peak Second Metatarsal vGRF (N)	273 ± 52	279 ± 56	0.979	0.950	0.991	7.8	2.8	21.7
Peak Second Metatarsal hGRF (N)	50 ± 15	51 ± 15	0.976	0.946	0.989	2.3	4.6	6.5
Second Metatarsal Pitch Angle at TPS (°)	29.4 ± 3.4	29.5 ± 3.7	0.936	0.861	0.970	0.9	3.1	2.5
Normalized Peak Whole Body vGRF (N/kg)	24.4 ± 2.6	24.4 ± 2.6	0.985	0.968	0.993	0.3	1.3	0.9
Normalized Peak Whole Body hGRF (N/kg)	3.6 ± 0.5	3.6 ± 0.5	0.924	0.836	0.964	0.1	3.8	0.4
Percent of GRF under Second Metatarsal at TPS (%)	18.4 ± 3.4	18.7 ± 3.5	0.965	0.926	0.984	0.0	3.5	0.0
Maximum Percent of Force under Second Metatarsal (%)	29.1 ± 13.0	28.4 ± 10.3	0.929	0.846	0.967	0.0	10.9	0.1
Second Metatarsal Length (mm)	63 ± 6	64 ± 7	0.656	0.262	0.840	3.7	5.9	10.4
Second Toe Length (mm)	48 ± 5	49 ± 5	0.680	0.321	0.850	2.7	5.6	7.5

GRF = ground reaction force. MPa = megapascals. N = newtons. Nm = newton meters. h = horizontal. v = vertical. TPS = time of peak stress. SD = standard deviation. ICC = intraclass correlation coefficient. 95% CI = 95% confidence interval. SEM = standard error of the measurement. MDC = minimal detectable change.

Table 3

Between-day reliability metrics for second metatarsal dorsal and plantar stress when second metatarsal length, second toe length, or both were fixed. Fixed-length conditions used first-visit length measurements and were compared with bone stresses calculated using visit-specific lengths.

Variable	ICC [95% CI]	% Δ ICC	SEM	% Δ SEM	MDC	% Δ MDC
Peak Dorsal Stress, Visit-Specific Lengths (MPa)	0.932 [0.847, 0.969]	---	9.0	---	25.0	---
Peak Plantar Stress, Visit-Specific Lengths (MPa)	0.924 [0.830, 0.965]	---	8.9	---	24.8	---
Peak Dorsal Stress, Fixed Metatarsal and Toe Lengths (MPa)	0.988 [0.972, 0.995]	6.0	9.0	-60.3	25.0	-60.4
Peak Plantar Stress, Fixed Metatarsal and Toe Lengths (MPa)	0.989 [0.973, 0.995]	7.0	8.0	-64.0	24.8	-64.2
Peak Dorsal Stress, Fixed Second Metatarsal Length (MPa)	0.988 [0.972, 0.995]	6.0	3.6	-60.3	9.9	-60.4
Peak Plantar Stress, Fixed Second Metatarsal Length (MPa)	0.989 [0.974, 0.995]	7.1	3.2	-64.4	8.8	-64.5
Peak Dorsal Stress, Fixed Second Toe Length (MPa)	0.932 [0.848, 0.969]	0.0	9.0	-0.3	24.9	-0.6
Peak Plantar Stress, Fixed Second Toe Length (MPa)	0.925 [0.832, 0.966]	0.1	8.8	-1.0	24.4	-1.6

MPa = Megapascals. ICC = Intraclass correlation coefficient. 95% CI = 95% confidence interval. % Δ = Percent difference from non-fixed value. SEM = Standard error of the measurement. MDC = Minimal detectable change.

When comparing relative marker distances between days, reliability ranged from moderate (ICC = 0.579) to excellent (ICC = 0.932), with the second metatarsal head marker showing greater between-day reliability in relative marker distances than the metatarsal base marker (Table 4). The mean difference in relative marker positions between days was less than two millimeters (2.1%) (Table 4).

Table 4

Between-day differences and reliability of relative three-dimensional vector distances between optical motion capture markers. Mean and percent differences were calculated as Visit 2 relative to Visit 1, with positive differences indicating greater distances for Visit 2.

Marker	Relative To	Mean Difference ± SE (mm)	Percent Difference (%)	ICC [95% CI]	SEM	MDC
SMH	FMH	0.4 ± 0.7	1.0	0.621 [0.328, 0.805]	2.5	7.0
	VMH	0.6 ± 0.8	0.8	0.698 [0.446, 0.848]	2.9	8.0
	CAL	-0.1 ± 0.7	-0.0	0.932 [0.858, 0.968]	2.8	7.6
SMB	FMH	1.4 ± 1.1	2.1	0.579 [0.278, 0.779]	4.0	11.0
	VMH	1.1 ± 1.0	1.3	0.734 [0.507, 0.867]	3.7	10.1
	CAL	-1.5 ± 0.9	-1.0	0.868 [0.734, 0.937]	3.5	9.8

SMH = second metatarsal head. SMB = second metatarsal base. FMH = first metatarsal head. VMH = fifth metatarsal head. CAL = posterior inferior calcaneus. SE = Standard error of the difference. ICC = Intraclass correlation coefficient. 95% CI = 95% confidence interval. SEM = Standard error of the measurement. MDC = Minimal detectable change.

Stride length and frequency showed excellent between-day reliability and within-day repeatability (Table 5).

Bland-Altman plots showed good agreement with small mean biases relative to the limits of agreement, indicating minimal systematic bias (Figure 1).

Table 5

Within-day repeatability and between-day reliability of stride characteristics during preferred speed running.

Variable	Visit 1		Visit 2		Between Day		
	Mean ± SD	ICC [95% CI]	Mean ± SD	ICC [95% CI]	ICC [95% CI]	SEM	MDC
Ground Contact Time (ms)	225 ± 22	0.930 [0.868, 0.959]	225 ± 24	0.932 [0.828, 0.961]	0.956 [0.905, 0.980]	4.9	13.6
Stride Length (m)	2.40 ± 0.24	0.953 [0.852, 0.981]	2.40 ± 0.24	0.937 [0.811, 0.971]	0.983 [0.963, 0.992]	0.0	0.1
Stride Frequency (strides/min)	83 ± 5	0.845 [0.642, 0.936]	83 ± 5	0.938 [0.864, 0.968]	0.939 [0.8685, 0.971]	1.2	3.3

ms = Milliseconds. m = Meters. min = Minute. SD = Standard deviation. ICC = Intraclass correlation coefficient. 95% CI = 95% confidence interval. SEM = Standard error of the measurement. MDC = Minimal detectable change.

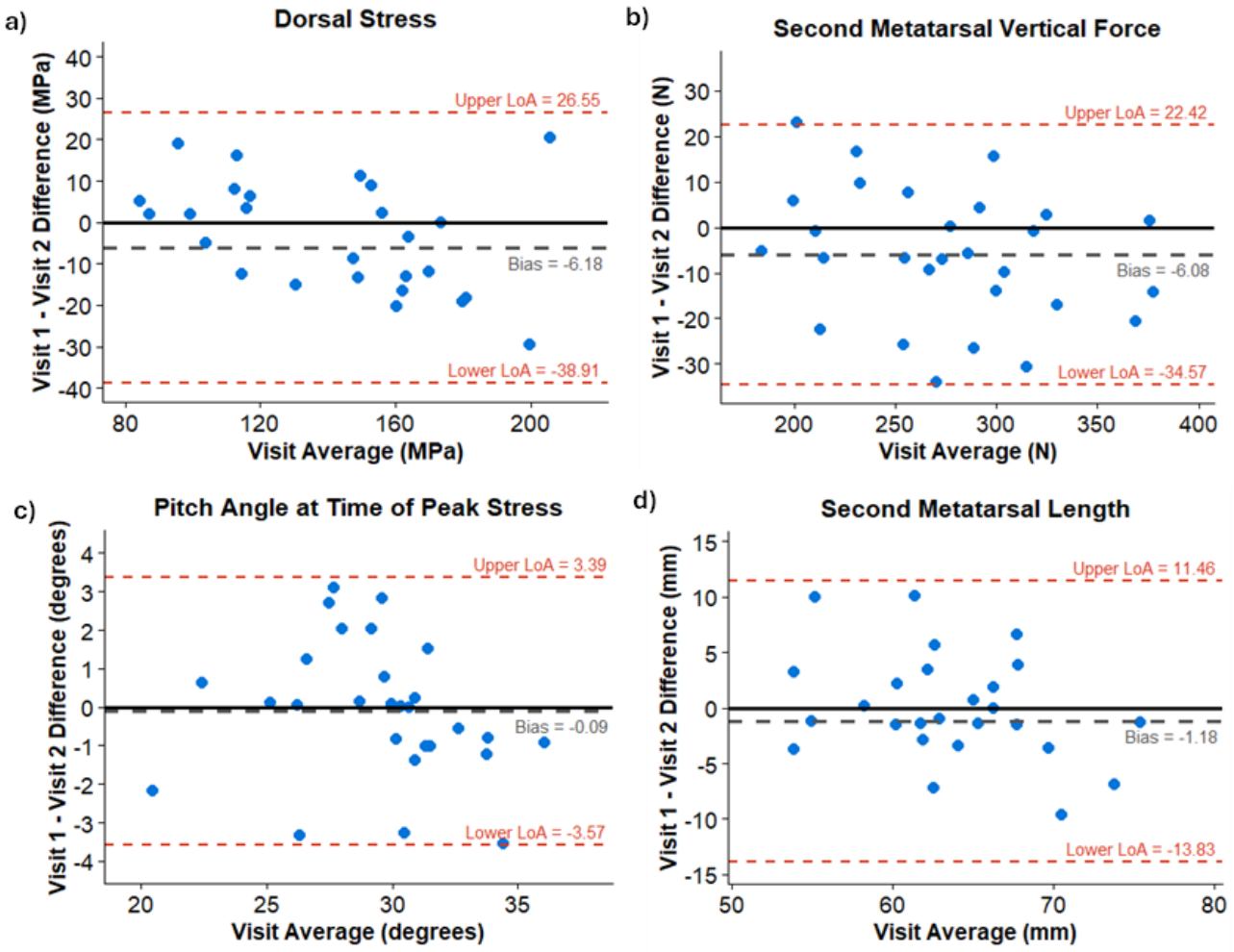
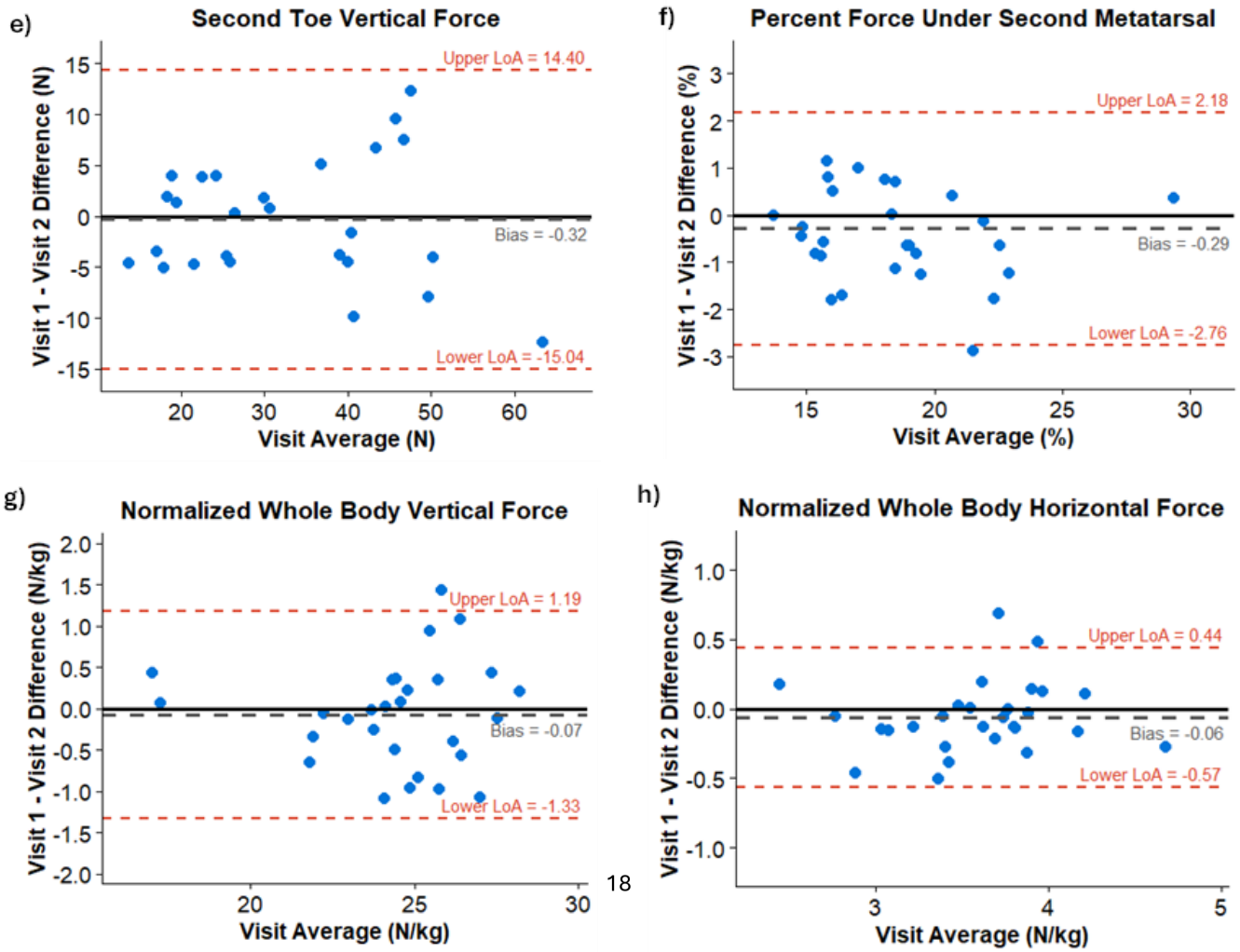


Figure 2: Bland-Altman plots for stress-related variables. Grey dashed horizontal lines show the mean difference (bias) between visits. Red dashed horizontal lines show the lower and upper 95% limits of agreement (LoA) (mean difference ± 1.95 x SD). The black solid line marks zero difference. Each data point is the difference between visits for a single participant plotted against the average of both visits. Plots are shown for **a)** second metatarsal peak dorsal stress, **b)** peak vertical force under the second metatarsal, **c)** pitch angle at the time of peak stress, **d)** length of the second metatarsal, **e)** peak vertical force under the second toe, **f)** percent of force under second metatarsal at the time of peak stress, **g)** whole body vertical ground reaction force normalized to body mass, and **h)** whole body horizontal ground reaction force normalized to body mass.


e)
Second Toe Vertical Force
Upper LoA = 14.40
Bias = -0.32
Lower LoA = -15.04
Visit 1 - Visit 2 Difference (N)
Visit Average (N)
f)
Percent Force Under Second Metatarsal
Upper LoA = 2.18
Bias = -0.29
Lower LoA = -2.76
Visit 1 - Visit 2 Difference (%)
Visit Average (%)
g)
Normalized Whole Body Vertical Force
Upper LoA = 1.19
Bias = -0.07
Lower LoA = -1.33
Visit 1 - Visit 2 Difference (N/kg)
Visit Average (N/kg)
h)
Normalized Whole Body Horizontal Force
Upper LoA = 0.44
Bias = -0.06
Lower LoA = -0.57
Visit 1 - Visit 2 Difference (N/kg)
Visit Average (N/kg)

Discussion

This study quantified the within-day repeatability and between-day reliability of second metatarsal bone stress when estimated using a beam theory model with experimentally collected data on repeat laboratory visits. Bone stress demonstrated excellent within-day repeatability and between-day reliability. The bone stress model thus appears to be robust to repeat measurements of the same running condition, demonstrating excellent consistency over time despite variability in the length of time between visits, ranging from one to 139 days. However, the excellent reliability of bone stress estimates may also indicate a lack of sensitivity of the model to detect true changes in stress between conditions.

Bone stress estimates were reliable despite only moderate between-day reliability of second metatarsal and second toe lengths determined based on the placement of skin-mounted markers. Our findings indicate that the variation in metatarsal and toe lengths between visits was not large enough to significantly influence the reliability of bone stress estimates. The previous sensitivity analysis demonstrated that the model is sensitive to changes in metatarsal length but was conducted with larger length differences of 10% compared to length differences of less than 2% in the present study (M. Ellison, 2020, p. 99).

While bone stress estimates demonstrated excellent between-day reliability, using the same bone lengths for both visits further improved the reliability of bone stress. Using the same second metatarsal length for both visits increased between-day reliability (ICC) by 6.0% for peak dorsal stress and 7.1% for peak plantar stress, while using the same second toe length for both visits had minimal to no effect on between-

day reliability for peak dorsal and plantar stress (Table 3). When using first-visit metatarsal and toe lengths for the second visit, dorsal stress was reduced by 2.4% and plantar stress by 2.5% compared to using visit-specific lengths. Future research comparing bone stress estimates between days should consider using fixed bone lengths to improve between-day reliability.

Since second metatarsal length was calculated from marker positions, between-day differences in marker positioning will influence the reliability of bone stress. The reliability of between-day relative marker distances was moderate to excellent (ICC range: 0.579 – 0.932), with mean differences of less than 2 mm in relative marker positions between visits (Table 4). Thus, even relatively small differences of less than 2 mm in relative marker distances can influence bone stress reliability by 6 – 7%.

We are not aware of any previous studies which have investigated the reliability or repeatability of estimates of bone stress during gait. Our finding of excellent reliability and repeatability for bone stress aligns with previous studies which found good to excellent reliability of kinematics, kinetics, and GRFs during walking and running (Bramah et al., 2021; Ferber et al., 2002; Kadaba et al., 1990; Schache et al., 2002). Our finding of excellent reliability may thus reflect the model's reliance on kinematic and GRF inputs, which have been shown to demonstrate high within- and between-day reliability.

We found greater between- than within-day ICC for peak dorsal stress, shear force, axial force, vertical and horizontal force under the second metatarsal, pitch angle at time of peak stress, normalized peak whole body vertical and horizontal GRF, and percent of GRF under the second metatarsal, which contrasts with previous studies which found kinematics, kinetics, and GRFs to be more repeatable within than between

days (Ferber et al., 2002; Kadaba et al., 1990; Schache et al., 2002). The greater within- compared to between-day repeatability in previous studies may be explained by differences in marker positioning between days. The reason for the greater between- compared to within-day ICCs in the present study is unclear but could be related to differences in procedures between the visits. The preferred speed trials during the first visit were performed in a randomized order as one of six running conditions but were performed first on the second visit. Additionally, the present study involved a variable follow-up period between visits, while previous studies had standard follow-up periods of 1-2 weeks between visits (Ferber et al., 2002; Kadaba et al., 1990; Schache et al., 2002). A familiarization effect to the testing procedures could have also played a role since the ICC values were typically higher for Visit 2 than Visit 1.

Stride characteristics including ground contact time, stride length, and stride frequency showed good or excellent between-day reliability and within-day repeatability (Table 5). Our findings indicate that participants ran in a highly consistent manner across trials and days, suggesting that variability in stress-related variables is unlikely to be explained by changes in running stride.

A strength of the present study is that differences in training load and physiological state such as fatigue levels between visits were not controlled for since these factors are not typically accounted for in biomechanics studies. Limitations of the present study include the potential effect of acclimation to study procedures where participants may have been more consistent in their running technique on the second visit. Another limitation is that testing procedures were not identical on both days, with the preferred speed condition being one of six conditions performed in a randomized order on the first

visit whereas it was always the first condition on the second visit. The time between testing days was also not controlled for and may have affected the results due to the large variability in the time between visits. Additionally, the study cohort of healthy runners may limit the generalizability of results to other populations such as those with less running experience, older age, or unhealthy populations.

Conclusion

Second metatarsal dorsal and plantar stress showed excellent within-day repeatability and between-day reliability during barefoot running when using a beam theory model to estimate stress. Bone stress estimates were highly reliable despite moderate between-day reliability of second metatarsal and second toe lengths measured from skin-mounted markers. Within-day repeatability was greater on the second visit than the first, potentially indicating an acclimation effect to the testing procedures.

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Appendix I. Ethical Approval

Hannah Rice
Institutt for fysisk prestasjonsevne

OSLO 27. juni 2023

Søknad 281 – 150623 – Påvirkningen av løpshastighet, belastning og en intensiv 10 km på tredemølle på andre metatarsal stress hos langdistanseløpere

Vi viser til søknad, prosjektbeskrivelse, informasjonsskriv og innsendtmelding til SIKT/NSD.

I henhold til retningslinjer for behandling av søknad til etisk komite for idrettsvitenskapelig forskning på mennesker har komiteen i møte 15. juni 2023 konkludert med følgende:

Vurdering

Formålet med prosjektet er å forstå hvordan løpshastigheten, belastning og en intensive 10 km test på tredemølle påvirker stress i den andre mellomfotnokkelen i foten. Utvalget er kvinner og menn mellom 18-35 år som løper minimum 3 ganger i uken og mer enn 15 km tilsammen pr uke. Deltakerne skal screenes for sykdom, skader og andre tilstander som ikke er forenelig med gjennomføring av testene. Komiteen ber prosjektleder påse at screeningen også sikrer att deltakerne er friske og skadefrie for andre forhold enn eksklusjonskriteriene når øktene skal gjennomføres. Begrepet «bakfotslag» inngår som inklusjonskriterium i informasjonsskrivet. Dette er ikke et vanlig begrep og bør derfor forklares nærmere (f.eks. «løpe på hælen».

Opplysningen om at NIH er selvassurandør bør formulerers slik at det kommer tydelig frem at det innebærer at *NIH dekker eventuelle skader som deltakerne måtte bli påført* ***som følge av forskningen****.* Komiteen ber prosjektleder vurdere om det i informasjonsskrivet, under punktet «Hva innebærer det for deg å delta», også gis en beskrives fordeler og ulemper ved deltakelse i prosjektet, inkludert risiko for skade. Videre må det presiseres at deltakerne må være friske og skadefrie på testdagene. Komiteen forutsetter at prosjektleder etablerer en beredskapsplan dersom det skulle oppstå skade under øktene i tråd med instituttets praksis ved fysiske intervensjoner.

Komiteens savner for øvrig et punkt i forskningsprotokollen om etiske vurderinger i prosjektet. Det vises her til søknad og det som er redegjort med hensyn til fordeler, ulemper, tiltak og forsvarlighet. Dette skal etter komiteens vurdering også gjenspeiles i forskningsprotokollen.

Vedtak

På bakgrunn av forelagte dokumentasjon finner komiteen at prosjektet er forsvarlig og at det kan gjennomføres innenfor rammene av anerkjente etiske forskningsetiske normer nedfelt i NIHs retningslinjer. Til vedtaket har leder av komiteen lagt følgende forutsetning til grunn:

- *Vilkår fra SIKT/NSD følges*

- *Informasjonsskrivet justeres i tråd med komiteens merknader, og sendes til komiteen til orientering.*

- *Forskningsprotokollen oppdateres i tråd med komiteens merknader, og sendes til komiteens til orientering*

- *Det etableres en beredskapsplan for ev skader under øktene*

Komiteen gjør oppmerksom på at vedtaket er avgrenset i tråd med fremlagte dokumentasjon. Dersom det gjøres vesentlige endringer i prosjektet som kan ha betydning for deltakernes helse og sikkerhet, skal dette legges fram for komiteen før eventuelle endringer kan iverksettes.

Med vennlig hilsen



Professor Anne Marte Pensgaard

Leder, Etisk komite, Norges idrettshøgskole

Besøksadresse: Sognsveien 220, Oslo

Postadresse: Pb 4014 Ullevål Stadion, 0806 Oslo Telefon: +47 23 26 20 00, postmottak@nih.no **www.nih.no**

NIH NORGES IDRETTSHØGSKOLE

NIH NORGES IDRETTSHØGSKOLE

Hannah Rice
Institutt for fysisk prestasjonsevne

OSLO 17. Oktober 2023

Endringsmelding 310-051023- 281 – 150623 – Påvirkningen av løpshastighet, belastning og en intensiv 10 km på tredemølle på andre metatarsal stress hos langdistanseløpere

Vi viser til endringsmelding og justert prosjektbeskrivelse.

I henhold til retningslinjer for behandling av søknad til etisk komite for idrettsvitenskapelig forskning på mennesker har leder av komiteen på fullmakt fra komiteen konkludert med følgende:

Vurdering
Prosjektleder ønsker å endre alder på deltakerne fra 18-35 år til 18-40 år for det skal være lettere å rekruttere deltakere. Leder av komiteen har ingen vesentlige merknader til endringen.

Vedtak
På bakgrunn av forelagte dokumentasjon finner leder av komiteen at endringen er forsvarlig og kan gjennomføres innenfor rammene av anerkjente etiske forskningsetiske normer nedfelt i NIHs retningslinjer. Til vedtaket har leder av komiteen lagt følgende forutsetning til grunn:

- *Vikår fra NSD/SIKT følges*

Leder av komiteen gjør oppmerksom på at vedtaket er avgrenset i tråd med fremlagte dokumentasjon. Dersom det gjøres vesentlige endringer i prosjektet som kan ha betydning for deltakernes helse og sikkerhet, skal dette legges fram for komiteen før eventuelle endringer kan iverksettes.

Med vennlig hilsen



Professor Anne Marte Pensgaard
Leder, Etisk komite, Norges idrettshøgskole

Sikt

Notification form /
The influence of running speed, load carriage, and an intensive 1... /

Assessment of processing of personal data

Reference number
832016
Assessment type
Standard
Date
30.11.2023
Title
The influence of running speed, load carriage, and an intensive 10 km treadmill run on second metatarsal stress in distance runners
Institution responsible for the project
Norges idrettshøgskole / Institutt for fysisk prestasjonsevne
Project leader
Naomi Frankston
Project period
01.06.2023
-
01.06.2028
Categories of personal data
General
Special
Legal basis
Consent (General Data Protection Regulation art. 6 nr. 1 a)
Explicit consent (General Data Protection Regulation art. 9 nr. 2 a)
The processing of personal data is lawful, so long as it is carried out as stated in the notification form. The legal basis is valid until02.06.2028.
Notification Form
Comment
You have updated your Notification Form. We cannot see that you have made any changes to the Notification Form or attachments that will affect our assessment of how personal data are processed in this project.
Read more about which changes should be notified to Data Protection Services before you send in changes in the future: https://sikt.no/en/notify-changes-notification-form
FOLLOW-UP OF THE PROJECT
We will follow up the progress of the project underway (every other year) and at the planned end date in order to determine whether the processing of personal data has been concluded.
Good luck with the project!

Appendix II. Screening Questionnaire

The following questionnaire was given online using SurveyXact. The formatting appears differently here than the online version.

Screening Questionnaire Consent Statement

This screening questionnaire takes less than 5 minutes to complete and contains mostly yes or no questions.

Purpose

This screening questionnaire will be used to determine if you are eligible to participate in a research study at the Norwegian School of Sport Sciences (NIH). We are looking to study distance runners who meet specific eligibility criteria. The purpose of the research is to understand how running speed, running with a weight vest, and completing an intensive 10 km run on the treadmill affect the forces in your foot.

Voluntary

Participation in this screening questionnaire is entirely voluntary. You will be asked questions about yourself and your running habits to determine if you are eligible to participate in the study. You can withdraw your consent at any time without giving a reason. If you withdraw your consent or do not complete the questionnaire, your responses will be deleted.

Data Handling

Your responses to the questions will be stored electronically in SurveyXact. If you are ineligible or choose not to participate in the research study, your responses will be deleted once study recruitment has concluded. If you are eligible for the study and choose to participate, your responses will be saved for the duration of the study. Your responses will be anonymized five years after the end of the project.

Contact Info

If you have any questions or would like more information please contact the study manager, Naomi Frankston, at naomief@nih.no.

I have read the above statement and consent to participate in the screening questionnaire.

(1) ❍ Yes

(2) ❍ No

How old are you?

_____

What is your sex?

(1) ❍ Male

(2) ❍ Female

(3) ❍ Other, please specify _____

How much do you weigh in kg?

_____

Are you currently pregnant?

(1) ❍ Yes

(2) ❍ No

Do you have any illness or injury that prevents you from running?

(1) ❍ Yes

(2) ❍ No

Are you comfortable running 10 km on a treadmill?

(1) ❍ Yes

(2) ❍ No

Can you run barefoot without pain?

(1) ❍ Yes

(2) ❍ No

Have you had a lower extremity injury in the last 6 months?

(1) ❍ Yes

(2) ❍ No

Have you had previous ankle or foot surgery?

(1) ❍ Yes

(2) ❍ No

Do you have any neurologic, musculoskeletal, or movement disorders?

(1) ❍ Yes

(2) ❍ No

On average during the last 4 weeks, did you run at least 15 km per week?

(1) ❍ Yes

(2) ❍ No

On average during the last 4 weeks, did you run at least 3 times per week?

(1) ❍ Yes

(2) ❍ No

What is your average pace for an easy run in minutes per kilometer? This is the pace which feels most comfortable for you to run at.

_____

In the last 2 years, have you completed a running race (or self-timed maximal effort) that is between 5 km and a marathon? This includes running a 5km, 10 km, half marathon, or marathon (or any distance in between) at maximum effort.

(1) ❍ Yes

(2) ❍ No

What is your current estimated time for a 10 km run at maximum effort? This is the total time you think it would take you to run a 10 km at maximum effort. If you are unsure, enter zero.

_____

Have you been running on a regular basis for at least 3 years? Select yes even if you took some breaks but then returned to running.

(1) ❍ Yes

(2) ❍ No

What part of your foot typically contacts the ground first when you run?

(1) ❍ Heel (rearfoot)

(2) ❍ Middle of the foot (midfoot)

(3) ❍ Front of the foot (forefoot)

(4) ❍ Unsure

Would you consider running to be your primary sport or physical activity?

(1) ❍ Yes

(2) ❍ No

Thank you for completing the screening survey.

We would like you to visit the lab at the Norwegian School of Sport Sciences (NIH) for testing.

If you have any questions about the study, you can contact the study manager, Naomi Frankston, at naomief@nih.no.

Please enter your email address here and we will contact you to schedule your visit to NIH Sognsvann:

_____

Thank you for completing the screening survey.

Unfortunately, you do not qualify for the study. If you have any questions regarding the study, you can contact the study manager, Naomi Frankston, at naomief@nih.no.

Your survey answers are stored anonymously in SurveyXact and will be deleted once study recruitment has concluded expect if you enter your email address below.

If you enter your email address below, you consent to having your email address stored along with your survey answers until 01.06.2028.

If you would like to be contacted about future studies, please enter your email address here (optional):

Appendix III. Modified Physical Activity Readiness Questionnaire (PAR-Q)

Subject Code: _____ Date: _____

Please read the questions carefully and answer each one honestly:

☐ Yes	☐ No	1. Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?
☐ Yes	☐ No	2. Do you feel pain in your chest when you do physical activity?
☐ Yes	☐ No	3. In the past month, have you had chest pain when you were not doing physical activity?
☐ Yes	☐ No	4. Do you lose your balance because of dizziness or do you ever lose consciousness?
☐ Yes	☐ No	5. Do you have a bone or joint problem that could be made worse by a change in your physical activity?
☐ Yes	☐ No	6. Is a physician currently prescribing medications for your blood pressure or heart condition?
☐ Yes	☐ No	7. Do you know of any other reason why you should not do physical activity?

Appendix IV. Running History Questionnaire

The following questionnaire was given online using SurveyXact. The formatting appears differently here than the online version.

Enter your subject number

_____

How old are you?

_____

What is your sex?

(1) ❍ Female

(2) ❍ Male

(3) ❍ Other, please specify _____

Which foot would you use to kick a ball?

(1) ❍ Right

(2) ❍ Left

(3) ❍ Both equally

(4) ❍ Unsure

On average during the last 4 weeks, how many kilometers did you run per week?

_____

On average during the last 4 weeks, how many times did you run per week?

_____

How many years have you been running? Count each year in which you ran on a regular basis for more than 6 months of the year.

_____

Over the last 2 weeks, how would you describe your training load compared to your average training load? (also asked on Visit 2)

(6) ❍ 7 Extremely high training load

(11) ❍ 6 Very high training load

(7) ❍ 5 Higher training load

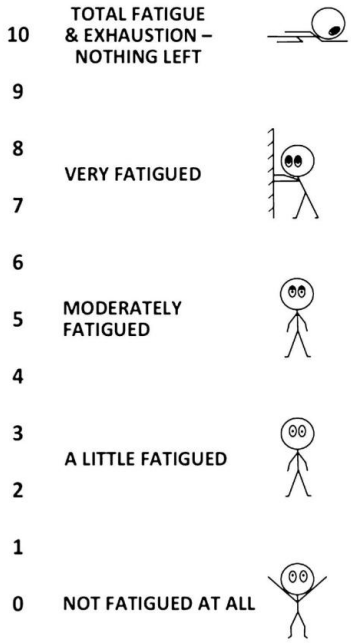
(12) ❍ 4 Average training load

(8) ❍ 3 Lower training load

(9) ❍ 2 Very low training load

(10) ❍ 1 Extremely low training load

How physically fatigued do you feel today? Think about how heavy your legs feel from training and how ready you feel to train right now. (also asked on Visit 2)



(1) ❍ 10

(2) ❍ 9

(3) ❍ 8

(4) ❍ 7

(5) ❍ 6

(6) ❍ 5

(7) ❍ 4

(8) ❍ 3

(9) ❍ 2

(10) ❍ 1

(11) ❍ 0

What is the brand and model of your current running shoes? Example: Brooks Ghost 14

_____

Why did you choose your current running shoes? (optional)

_____
_____
_____
_____
_____
_____

Do you currently or have you ever worn inserts or orthotics in your shoes?

(1) ❍ Yes, I currently use orthotics all of the time or some of the time

(2) ❍ Yes, I used orthotics in the past but I don't use them any longer

(3) ❍ No, I have never used orthotics or inserts

Why did you wear inserts or orthotics?

_____
_____
_____
_____
_____
_____

Were your inserts or orthotics prescribed by a medical professional? Check the box next to comment if you wish to add a comment.

(3) ❑ Yes

(2) ❑ No

(4) ❑ Comment (optional) _____

Did you find that wearing the inserts/orthotics helped? Check the box next to comment if you wish to add a comment.

(3) ❑ Yes

(2) ❑ No

(4) ❑ Comment (optional) _____

Why did you stop wearing inserts or orthotics?

_____
_____
_____
_____
_____
_____

Have you ever had a running-related injury which caused you to miss 3 or more planned running sessions?

(1) ❍ Yes

(2) ❍ No

Which running-related injuries have you had that caused you to miss 3 or more planned running sessions? Please explain which body part was affected and what the injury was. Example: Left foot plantar fasciitis, right femur stress fracture. Please list all relevant running-related injuries.

_____
_____
_____
_____
_____
_____

Which running-related injuries were diagnosed by a medical professional? This can include a physiotherapist, medical doctor, athletic trainer, or similar. You can repeat answers from the previous question if the injury was diagnosed by a medical professional.

_____
_____
_____
_____
_____
_____

Have you ever had surgery related to a running injury?

(1) ❍ Yes

(2) ❍ No

Which body part was affected and which surgery did you have? Example: Right hip labral repair

_____
_____
_____
_____
_____
_____

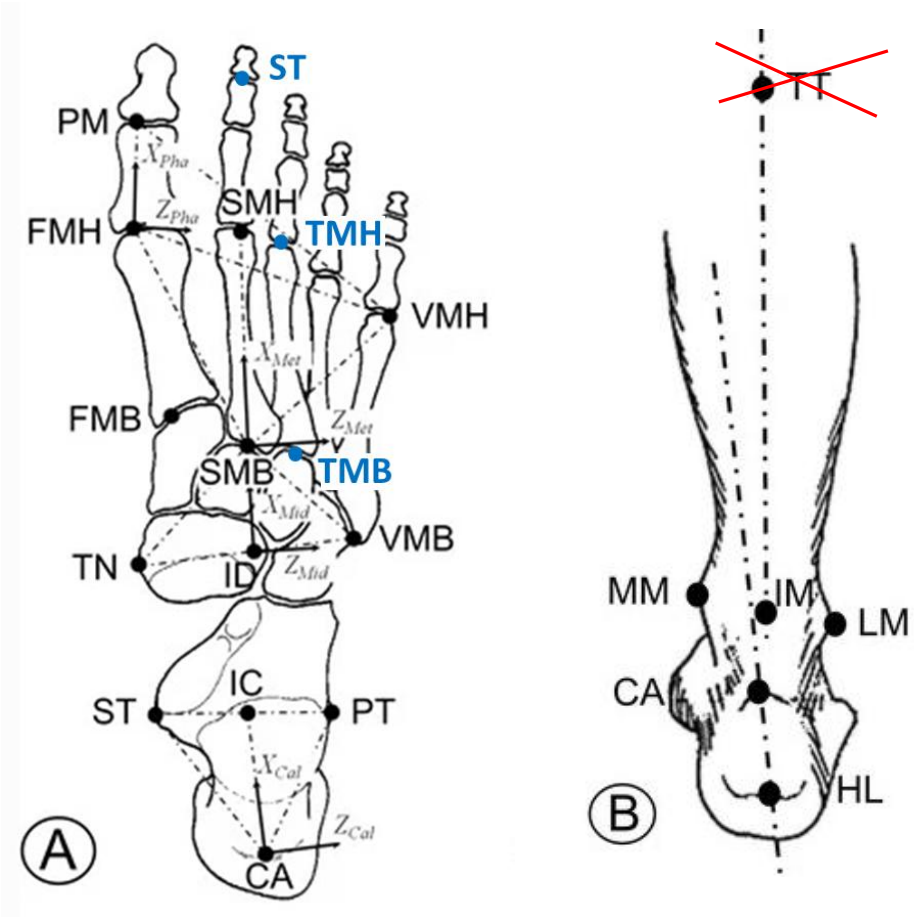
Is there anything else you would like us to know? (optional)

_____
_____
_____
_____
_____
_____

Appendix V. Modified Rizzoli Foot Marker Set

Additions to the Rizzoli foot marker set are shown in blue. Omissions from the marker set are shown with a red X.

Figure Modified From: Portinaro N, Leardini A, et al. *J Foot Ankle Res*. 2014;7(1):754. doi: 10.1186/s13047-014-0057-2. PMID: 25558289.



Appendix VI. Modified Rizzoli Foot Marker Set Guide

Blue text indicates additions to the Rizzoli marker set. Red strikethrough indicates omissions. Virtual markers are indicated with grey text.

Modified from: Leardini A, Benedettu MG, Berti D, et al. *Gait & Posture*. 2007;25(3):453-462. doi: https://doi.org/10.1016/j.gaitpost.2006.05.017

Phalanges (Pha):

PM most distal and dorsal point of the head of the proximal phalanx of the hallux.

ST most distal and dorsal point of the head of the middle phalanx of the second toe.

Metatarsus (Met):

FMB base of the first metatarsal, dorso-medial aspect of the first metatarso-cuneiform joint.

FMH head of the first metatarsal, dorso-medial aspect of the first metatarso-phalangeal joint.

SMB base of the second metatarsal, dorso-medial aspect of the second metatarso-cuneiform joint.

SMH head of the second metatarsal, dorso-medial aspect of the second metatarso-phalangeal joint.

TMB base of the third metatarsal, dorso-medial aspect of the third metatarso-cuneiform joint.

TMH head of the third metatarsal, dorso-medial aspect of the third metatarso-phalangeal joint.

VMB base of the fifth metatarsal, dorso-lateral aspect of the fifth metatarso-cuboid joint.

VMH head of the fifth metatarsal, dorso-lateral aspect of the fifth metatarso-phalangeal joint.

Mid-foot (Mid):

TN most medial apex of the tuberosity of the navicular.

MC the most distal and dorsal aspect of the middle cuneiform, assumed to coincide with SMB.

TC most lateral apex of the tuberosity of the cuboid, assumed to coincide with VMB.

ID intermedius mid-foot, mid point between TN and TC.

Calcaneus (Cal):

CA upper central ridge of the calcaneus posterior surface, i.e. Achilles' tendon attachment.

HL lower central ridge of the calcaneus posterior surface, superior to heel fat pad.

PT lateral apex of the peroneal tubercle.

ST most medial apex of the sustentaculum tali.

IC intermedius calcaneus, mid point between ST and PT.

Shank (Sha):

~~TT most anterior prominence of the tibial tuberosity.~~

~~HF most proximal apex of the head of the fibula.~~

LM distal apex of the lateral malleolus.

MM distal apex of the medial malleolus.

IM intermedius malleoli, mid point between LM and MM.

Appendix VII. Borg Scale (English)

6	
7	Very, very light
8	
9	Very light
10	
11	Fairly light
12	
13	Somewhat hard
14	
15	Hard
16	
17	Very hard
18	
19	Very, very hard
20	

Borg Scale (Norwegian)

6	
7	Veldig, veldig lett
8	
9	Veldig lett
10	
11	Nokså lett
12	
13	Behagelig anstrengende
14	
15	Anstrengende
16	
17	Veldig anstrengende
18	
19	Veldig, veldig anstrengende
20	

Appendix VIII. Bone Geometries from MRI Data

Table I: Means and standard deviations (SD) for bone geometry values provided by Ellison et al.[9] from magnetic resonance images of 20 healthy runners. Mean values were used in the bone stress calculations.

Variable	Mean	SD
ε (mm)	4.07	6.18
r1 (mm)	9.66	1.61
r2 (mm)	7.65	0.93
Cross-Sectional Area (mm^2)	$3.77x10^{-2}$	$6.00x10^{-3}$
Ixx (mm^4)	$2.05x10^{-7}$	$8.29x10^{-8}$
r (mm)	3.74	0.48

ε = distance between the inferior metatarsal head and the plantar tendon
r1= lower radius of the metatarsal head
r2 = upper radius of the metatarsal head
Ixx = area moment of inertia about the horizontal axis of the bone cross-section
r = radial distance from the center of the bone to the surface