TIBIAL BONE LOAD DURING RUNNING:

ESTIMATION METHODS AND RELATIONSHIPS TO CLINICAL MEASURES

by

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DEDICATION

To Jesse, Aysa, and TT,
without whom this adventure would not be possible.
I am so grateful for your unconditional support,
constant encouragement,
and your patience.
I love you!

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ABSTRACT

Introduction: Tibial bone stress injuries (BSI) are an overuse injury in endurance runners, influenced by a complex set of risk factors including excessive mechanical loads. However, current methods for estimating these loads are not clinically accessible. It is important to establish surrogate measures of tibial bone load that are clinically accessible to identify runners who may be at risk of tibial BSI. Additionally, tibial loads are sensitive to methodology; however, models for quantifying tibial force specifically have yet to be compared. Therefore, the aims of this dissertation are as follows:

Aim one: To evaluate the influence of different modeling methodologies on tibial force estimation during running.

Aim two: To determine whether clinical measures can predict tibial loads during running in healthy and injured recreational running populations.

Methods: For Aim one, a systematic review and meta-analysis was conducted to determine whether differences in peak tibial axial force during running exist between three modeling approaches. Additionally, methods for estimating Achilles tendon moment arm were compared to investigate how length differences influence plantarflexor and tibial force estimates during running. For Aim two, two studies were conducted to investigate relationships between strength, balance, and endurance assessments and tibial bone load during running in healthy and injured populations. Results: Model methodology accounted for 50% of the variability in tibial force estimates, with a static optimization model producing 3.45 BW greater forces than a distal model. Achilles tendon moment arm length was also method-dependent and influenced estimates of Achilles and tibial axial forces during running. Clinical measures of ankle strength, dynamic balance, and plantarflexor endurance cannot be used as surrogates for tibial bone force or other tibial load metrics during running. Preliminary results suggest that runners who have recently returned to running after tibial BSI exhibit decreased plantarflexor strength and endurance, along with increased impulse and cumulative load compared to healthy runners. Conclusions: Estimates of tibial bone load during running is method-dependent; model design and assumptions should be carefully considered when selecting methods. Study conditions such as speed, strike pattern, load carriage, surface grade, and step frequency contribute to variability in load estimates and should be accounted for in study design and analysis. Common clinical assessments of strength, dynamic balance, and endurance are not reliable surrogates for tibial bone load in healthy runners; however, injured runners present with deficits in these measures during the early return to run period following tibial BSI.

CHAPTER ONE

INTRODUCTION

Bone stress injuries (BSI) are a sport-related injury common in an endurance running population. These injuries exist on a continuum from mild marrow or periosteal edema to a fracture and result in pain, tenderness, and inability to participate in sport.[1,2] BSIs often result in significant consequences in terms of time away from sport and risk of recurrent injury. While risk factors for BSI are multifaceted, repetitive mechanical loading - such as the load experienced during running - is an integral component to the development of BSI. Therefore, understanding the relationship between mechanical load and BSI is essential to advancing our understanding of risk factors and informing interventions.

Incidence of Bone Stress Injury

Up to 20% of all athletic injuries are BSI, although BSIs occur most commonly in high school and collegiate runners as well as military personnel.[2–4] High school girls' cross-country athletes have an incidence of 10.5 per 100,000 athlete-exposures while women's collegiate cross-country athletes have the highest reported incidence at 28.6/100,000 athlete exposures, and competitive track and field athletes were found to have a 21% yearly incidence of BSI.[4–6] BSIs affect up to 6.4% of female collegiate cross-country athletes and up to 6.8% of female military recruits.[7,8] The majority of BSIs occur in-season, and while most BSIs require participation restriction, up to 22% are season-ending injuries.[5,9] Unfortunately, after initial diagnosis, an

athlete has a one in five risk of recurrent injury and a military recruit is 3.5 times as likely to sustain a subsequent BSI compared to a recruit without a history of BSI.[5,8]

The tibia is among the most common locations of BSI in both running and military populations. The tibia accounts for up to 40% of all BSIs in high school cross country athletes, up to 68% of all BSIs in collegiate athletes, and up to 46% in competitive track and field athletes.[3–7] Likewise, in a miliary population, up to ~58% of all BSIs are located in the tibia.[8,10] Given the high prevalence of tibial injury among BSIs, it is valuable to understand how mechanical load during running is applied to the tibia as these loads likely have a role in the development and recurrence of tibial BSI.

The Mechanical Load of Running

Running is an endurance sport that generates cumulative mechanical load on the skeletal system due to repetitive activity.[11] The cumulative load can lead to tissue fatigue and consequently localized microdamage within the bone, including microcracks or microfractures.[1,12] This microdamage is a necessary stimulus for a targeted physiologic remodeling process that results in replacement of damaged bone.[11] While the end result of remodeling is adaptation to load, the remodeling process creates a temporary increase in localized porosity. When coupled with continued repetitive activity and inadequate rest, a disproportionate accrual in microdamage *and* the temporary porosity can contribute to the development of a BSI.[13–15] Additionally, when the microdamage and increased porosity accumulate, the material properties of the bone become negatively affected.[12,15–17] A reduction in bone stiffness and strength due to the localized microdamage along with continued repetitive activity can perpetuate the development of a BSI.[13,17] The accumulation of microdamage and

progressive decline in material properties defines mechanical fatigue of bone.[13,18] Symptoms of pain and reduced tolerance for weightbearing activity often accompany this fatigue, which frequently leads to a medical consultation and formal diagnosis of BSI. The necessity of repetitive activity as a precursor to BSI suggests that cumulative load is a potentially important variable to investigate when establishing relationships between tibial bone load and BSI.[19]

In addition to cumulative load, the magnitude of load has been referenced as an important contributing factor to microdamage and ensuing potential for BSI.[13,20–22] While mechanical fatigue is dependent on the number of loading cycles, or the cumulative load, it is also influenced by the magnitude of the load.[13,15] As load magnitude increases, the number of cycles to fatigue is disproportionately reduced.[13,22] The application of this phenomenon to running is described as a 10% *reduction* in load magnitude results in a 100% increase in the number of cycles to failure.[13,23] Likewise, a 10% *increase* in load will result in a 50% reduction in cycles to failure.[24] These examples illustrate how excessive load magnitude can result in disproportionate accumulation of microdamage; therefore, it is reasonable to speculate on the relationship between load magnitude and the development of BSI.

A few studies have reported on the association between the magnitude of tibial load while running and a history of BSI. For example, Meardon, et al. (2015) reported greater distal tibial stress in runners with a history of tibial BSI as compared to healthy runners.[25] Similarly, increased free moment and tibial shock have been suggested as surrogates of tibial load as they have also been found to be elevated in runners with a history of tibial BSI.[26–28] Other metrics of tibial load, such as loading rates, have been reported to differ between injured and healthy runners. For example, Popp, et al. (2023) found women with a history of multiple BSIs had

greater increases in vertical average and instantaneous loading rates with exertion compared to healthy runners.[28] While these examples are few in number, they support the concept that metrics of tibial bone load are elevated when running in runners with a history of tibial BSI.

Modeling Tibial Bone Load

Magnitude and cumulative load are clearly important metrics to consider when investigating biomechanical variables related to BSI risk. Therefore, quantification of tibial bone load is an important methodological consideration. A few studies report on the direct quantification of tibial loads, while the majority of approaches rely on biomechanical modeling. Surgically implanted strain gauges have been utilized to investigate *in vivo* strain, or the relative deformation under load, on the tibia during running.[19,29–34] While this technique allows for direct tibial load measurement, it cannot accommodate large sample sizes, and quantification of load is limited to the superficial periosteum of the area at which the gauge is implanted. Study participants are also exposed to risk beyond that of traditional motion capture data collection because the implantation of the strain gauge is highly invasive, may involve pain and/or numbness, and raises the risk of infection.[31,35]

Strain gauges can quantify tibial load directly; however, less invasive techniques that use biomechanical models to estimate tibial bone load are much more common within running-related research. These techniques include finite element analysis (FEA) or beam theory to estimate strain or stress, and mathematical models based on inverse dynamics for estimating axial force. Traditionally used to estimated bone strain, FEA is a technique which distributes load onto small elements that comprise a larger 3D structure.[36–39] This requires advanced imaging such as computed tomography or magnetic resonance to acquire bone geometry from which to

create the 3D model. While FEA is useful for determining regions of high intensity strain on the tibia, it is computationally expensive and often not practical for large cohorts.[38,40] An alternative to estimating strain is to calculate stress which is the force per unit area.[34] Beam theory models the tibia as a hollow-ellipse and uses the cross-sectional area to calculate stress at specific locations of the diaphysis.[14,25,41,42] The cross-sectional area can be acquired through previously published literature or, when participant specific geometry is preferred, via musculoskeletal imaging such as X-ray, computed tomography, or magnetic resonance imaging.

An alternative approach to strain or stress for quantification of bone load is estimating tibial axial force. Although tibial force estimation does not incorporate participant specific imaging, it is less computationally expensive while still accounting for external and internal contributions to tibial load.[43–45] Therefore, it is a suitable method for large cohorts. Estimating tibial force utilizes joint moments and reaction forces from inverse dynamics along with muscle forces estimated from a variety of methodologies. Muscle forces may be represented solely as plantarflexor force acting on the distal tibial,[46–49] as contributions from the gastrocnemius, quadriceps, and hamstrings acting on the proximal tibia,[50–53] or as multiple muscle contributions estimated from static optimization procedures.[42,43,45,54–57]

Importantly, 80% of internal load on the tibia during locomotion is attributable to plantarflexor muscle forces.[39] Given these substantial contributions to internal tibial load, the plantarflexor muscle forces significantly influence the net tibial force. During running, the external forces from ground reaction force generate a negative bending moment, producing posterior tibial tension and anterior tibial compression (Figure 1.1 A).[14] In contrast, the gastroc-soleus contraction produces axial compression, posterior compression, and anterior tension

resulting in a net positive bending moment (Figure 1.1 B,C).[14] Consequently, the plantarflexor forces act to offset the tensile loading on the posterior tibia by shifting the posterior force toward compression. Because tensile loads are more strongly associated with development of BSI, the reduction in posterior tensile force may act as a key protective mechanism for posterior tibial BSI.

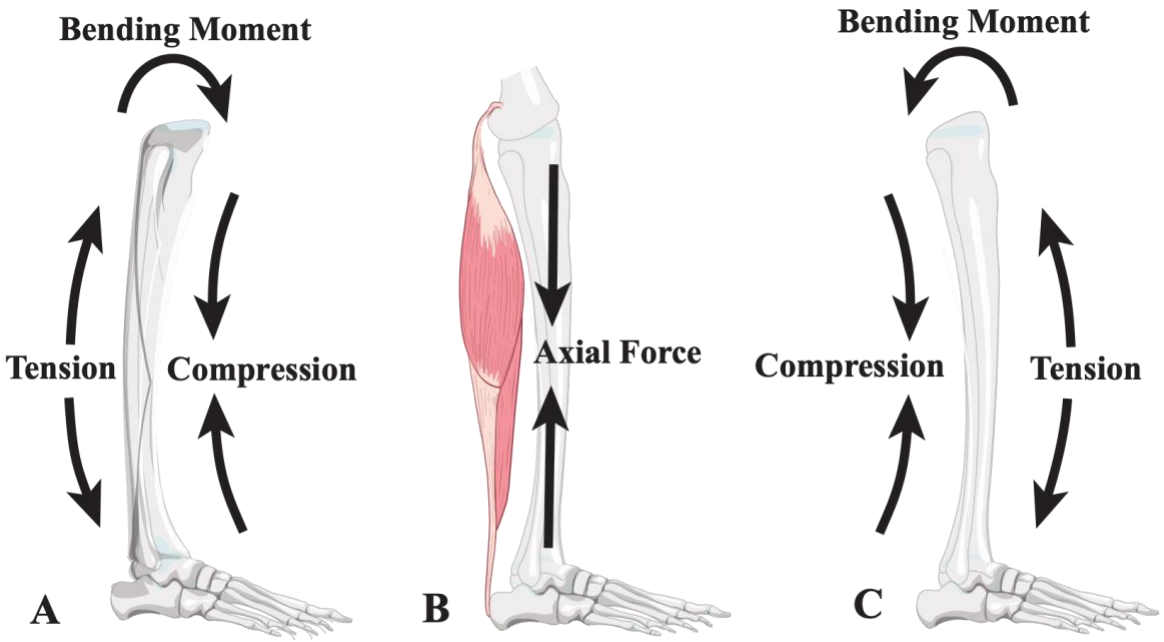


Figure 1.1 Forces acting on the tibia during running.

Perhaps the most challenging aspect of modeling tibial bone load through axial force is the inability to validate force outcomes. A few studies have implanted force transducers into a tibiofemoral joint replacement to investigate *in vivo* forces. Yet these studies tend to be very small in number ($n < 5$), in older cohorts (>60 years), and measure forces in a previously

pathologic joint.[58–60] While at least two studies report *in vivo* tibial forces during running activity in older adults, the average running speed of 2.23 m/s was substantially less than that of the slowest speeds from reports on tibial force in younger adult participants.[46,49,50,57–59] Therefore, comparison of tibial bone force to previously reported data is often reliant on force estimations from a variety of model methodologies.

Each method for quantifying tibial bone load during running has benefits and challenges. While subject-specific methods are desirable, these methods are not always possible given the imaging and computational requirements. Therefore, a simplified musculoskeletal model to estimate tibial force may be favorable with large cohorts or in the absence of imaging equipment.

Dissertation Aims

Tibial BSI is a multifaceted overuse injury that primarily affects endurance runners and has a complex set of risk factors including excessive mechanical loads. These loads may be sensitive to methodology, however, methodological considerations among the numerous available models for quantifying tibial force have yet to be compared. Additionally, the relationship between tibial bone load and BSI has been established, yet the clinical utility of this finding is impractical if bone load is not a clinically accessible variable. Therefore, it is important to establish surrogate measures of tibial bone load that are clinically accessible to identify runners who may be at risk of tibial BSI. Therefore, the aims of this dissertation are as follows:

Aim one: To evaluate the influence of different modeling methodologies on tibial load variables during running.

Aim two: To determine whether clinical measures can predict tibial loads during running in healthy and injured recreational running populations.

Aim one will be addressed with a systematic review and meta-analysis comparing methodologies for estimating tibial force during running in healthy participants. This aim will also include a comparison study examining methods for estimating the Achilles tendon moment arm, a key component tibial force estimation models. Aim two will be addressed with a cross-sectional study on healthy participants as well as a preliminary study on runners recently diagnosed with tibial BSI, with both studies investigating strength, endurance, and dynamic balance as predictors of tibial load variables during running.

Together, these reports will contribute to the existing literature by identifying how different modeling approaches influence tibial force estimates. Additionally, they will evaluate whether common clinical measures can serve as surrogate measures of tibial load, potentially enabling identification of runners with elevated loads in a clinical setting. Establishing such surrogate measures may support the use of tibial load a clinically meaningful variable in an effort to reduce the risk of recurrent BSI.

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CHAPTER TWO

ESTIMATION OF TIBIAL AXIAL FORCE DURING RUNNING USING THREE METHODOLOGIES: A SYSTEMATIC REVIEW AND META-ANALYSIS

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Contributions: Study conceptualization, methodology development, consulted on statistical analysis, read and revised final manuscript

Abstract

Introduction**:** The peak magnitude of tibial bone load is a variable related to the development of bone stress injuries, yet current methods for estimating tibial bone force result in a wide range of outcomes making the exact relationship difficult to characterize. Purpose**:** The purpose of this systematic review and meta-analysis was to quantify differences in tibial axial force estimations during running between three methodologies. Methods: A systematic search of electronic databases was conducted to identify published peer-reviewed literature reporting quantitative values of tibial axial forces during running in healthy participants. A random effects meta-analysis was used to analyze differences in peak force calculated from three methodologies used to estimate the internal forces that contribute to resultant tibial force. Results**:** Fifteen studies of moderate to low risk of bias were included within the review and meta-analysis. Weighted pooled means of 7.83 (95% CI: 6.31-9.35), 9.79 (95% CI 8.70-10.88), and 11.29 (95% CI: 9.95-12.29) body weights were found for a distal method, a proximal method, and a static optimization method, respectively. Estimation method significantly influenced peak tibial axial force ($QM = 15.23$, $p < 0.001$, $R^2 = 49.78$) which was also impacted by running speed, foot strike pattern, surface incline, stride length, and load carriage. Conclusion**:** Estimation method accounted for ~49.78% of the variability in peak tibial axial force between studies. A substantial amount of between study variability was unaccounted for yet is likely influenced by confounding variables such as running experience, speed, step frequency, surface, grade, and load carriage.

Introduction

The mechanical load experienced by the tibia during running is considered a key mechanical risk factor for bone stress injury (BSI) and is commonly quantified in terms of stress,[1,2] strain,[3,4] or force.[5–7] While stress and strain account for participant specific geometry, tibial force is often used as a surrogate measure of internal tibial load because it can be quantified non-invasively in large cohorts and reflects the combined contribution of internal and external forces acting on the bone. Tibial axial forces, therefore, have become a primary outcome measure in running biomechanics studies examining how gait modifications influence internal tibial bone load, including stride length and step width modifications,[8–10] use of external load carriage,[11] or changes in running speed,[2,12–14] grade,[15] and foot strike pattern.[7,16]

Despite widespread use of tibial axial force as an outcome measure, reported peak values during running range from 6.4 to 13.4 body weights.[12,17] This wide variability may be partially attributed to the method used to estimate tibial force. Three methods are commonly used in running literature: distal, proximal, and static optimization (SO). The distal method estimates axial force at the distal tibia by summing the ankle joint reaction force with estimated Achilles tendon forces, which are derived from ankle moments and Achilles tendon moment arms.[11,12,14,18] This model assumes the internal tibial load arises entirely from the plantarflexor muscle force and ignores other muscle contributions. The proximal approach also relies on inverse-dynamics, however, estimates force at the tibiofemoral joint as opposed to the distal approach which estimates force at the ankle joint. The proximal method sums knee joint reaction forces along with gastrocnemius, hamstring and quadricep muscle force estimates derived from ankle, knee and hip moments and assumed moment arms.[12,19] Finally, the SO method estimates lower

extremity muscle forces by minimizing a cost function, typically the sum of squared muscle activations subject to experimentally measured joint moment constraints, and then sums these forces with ankle joint reaction forces to determine tibial force.[2,5,7,13,15–17]

While the three approaches described have a common objective of estimating tibial axial force, the methodological differences may influence the magnitude of tibial forces, thus complicating comparisons across studies. A comparison of methods has not yet been performed, and it is currently unknown whether different estimation methods for tibial axial force produce meaningfully different values. Therefore, the purpose of this systematic review and meta-analysis is to determine if differences in peak tibial axial force during running exist between 1) a distal approach, 2) a proximal approach, and 3) a SO approach. By quantifying method dependent differences, this review aims to inform interpretations of tibial force outcomes and promote greater methodological consistency in future research.

Methods

Ethics Statement

This systematic review and protocol were registered with PROSPERO database (CRD42023399532) and completed according to the Preferred Reporting Items for Systematic Reviews and Meta-Analysis (PRISMA) guidelines.20–22

Search Strategy

A systematic search of published peer-reviewed literature on tibial axial force during running was conducted by searching the PubMed and SPORTDiscus electronic databases in December 2025. Databases were queried using the following search terms and strategy described

in Figure 1: (tibia* OR tibio*) AND (load* OR force*) AND (runn*). The search terms were determined from commonly used terms in previously published studies on tibial bone load in runners. Any articles acquired from database searches from all years up to December 2025 were considered for further evaluation. Two reviewers screened articles for inclusion criteria based initially on title, second by abstract review, and finally by a review of the full text. Potential articles to include within the review were compared and any disagreements in article selection were resolved by joint review and discussion.

Selection Criteria

To be eligible for inclusion in the systematic review and meta-analysis, articles were publicly available original data in full text, peer-reviewed, and published in the English language. The primary outcome of interest for this review and meta-analysis was peak tibial axial force per step in healthy human participants during running and calculated via proximal or distal inverse dynamics or static optimization at the distal third of the tibia. Quantitative values of peak tibial axial force or graphs in which peak tibial axial force may be determined using a Web Plot Digitizer[23] were required for inclusion. Studies that did not meet inclusion criteria were excluded from this review and meta-analysis.

Data Extraction

Data extraction was completed by a single author and cross-checked by a second author for accuracy. Publication details, number of participants, a brief description of study protocol, primary estimation methodology for tibial force and extracted data including study conditions, speed, and peak values of tibial bone force are described in Table 1.1. A web-based plot digitizer was used to extract peak values for bone load when numerical values were not explicitly reported

within the article.[23] The web-based plot digitizer has been reported as a valid and reliable tool for digitally extracting numerical data from graphs.[22,24]

Table 1.1 Systematic Review Study Specific Methods and Data

Study	n	Protocol	Model	Condition	Speed (m/s)	Mean Force ± SD (BW)
Baggaley et al. 2022	9 M 8 F	Instrumented treadmill run at five grades	SO	10 deg decline 5 deg decline 0 deg 5 deg incline 10 deg incline	3.33	10.40 ± 2.33* 10.83 ± 2.07* 12.30 ± 2.09* 11.96 ± 2.06* 11.11 ± 1.93*
Boyer et al. 2018	30 M 8 F	Seven overground run trials on a 30 m runway with habitual strike pattern, seven trials with alternate strike pattern	SO	hRFS-RFS hFFS-RFS hRFS-FFS hFFS-FFS	3.35	9.6 ± 1.0 9.5 ± 1.2 10.8 ± 1.1 11.1 ± 1.1
Edwards et al. 2009	10 M	28.5 m overground run x10 trials at PSL and -10% PSL	SO	PSL -10% PSL	4.4	13.4 ± 1.2 12.7 ± 1.0
Edwards et al. 2010	10 M	Overground run x10 trials at three speed conditions	SO	Speed	2.5 3.5 4.5	10.73 ± 1.03 12.63 ± 1.19 13.80 ± 1.63
Hunter et al. 2019	14 M 29 F	Overground run on 50-m indoor track x3 laps each speed (9 laps total)	D	Speed	2.70 3.27 4.08	6.30 ± 0.27** 6.74 ± 0.89** 7.28 ± 0.33**
McDonnell et al. 2019	10 M 10 F	Overground running trials	P		2.68	11.31 ± 1.02
Meardon et al. 2021	20 M 20 F	Instrumented treadmill run at 3 running speeds	SO	M slow M preferred M fast	90% 2.97 110%	9.92 ± 1.53* ** 9.98 ± 1.41* ** 1.11 ± 1.15* **

Table 1 Continued

Rooney et al. 2013	15 sex nr	Ten overground run trials with habitual strike pattern, ten trials with alternate strike pattern	SO	hRFS - RFS hRFS - FFS hFFS - RFS hFFS – FFS	4.25 4.27	11.1 ± 0.98 12.0 ± 0.91 10.7 ± 1.06 12.4 ± 0.78
Sasimontonkul et al. 2007	10 M	Overground run on 10m runway x6 trials	SO		3.84	9.0 ± 1.13
Thakkar et al. 2019	50 F	Instrumented treadmill	P	≤ 166 SPM 168-176 SPM ≥ 178 SPM	2.98	9.65 ± 1.13 9.22 ± 1.16 8.15 ± 1.12
van Middelaar et al. 2025	8 M 6 F	3x 60 second runs on instrumented treadmill at three speeds	D		2.5 3.1 3.6	6.34 ± 0.37 6.86 ± 0.38 7.22 ± 0.54
Willy & Meardon et al. 2016	16 M 14 F	Instrumented treadmill run at varying step rates	P	RT: hSR RT: +7.5% hSR CON: hSR	3.21 3.16	10.32 ± 1.57 9.37 ± 1.35 9.85 ± 1.51
Willy & Willson et al. 2016	16 M	Unloaded run trial for 8 minutes on an instrumented treadmill, followed by 15kg load trials and +7.5% cadence	P	Unloaded 15 kg Unloaded +7.5% 15kg + 7.5%	2.7	9.61 ± 1.34 10.09 ± 1.23 8.76 ± 1.19 9.32 ± 1.20
Willy et al. 2025	17 M 17 F	Unloaded run trial for 2-minutes on an instrumented treadmill, 2-minute run with 15 kg	D	0 kg LC 15 kg LC	2.7	9.8 ± 1.5 10.7 ± 1.3

Table 1 Continued

Zandbergen et al. 2023	8 M 4 F	90-second running trial on an instrumented treadmill at self-selected step frequency	D	3.33 m/s	7.6 ± 0.6

Abbreviations: male (M), female (F), habitual (h), rearfoot strike (RFS), forefoot strike (FFS), preferred stride length (PSL), load carriage (LC), static optimization (SO), proximal model (P), distal model (D) Retraining (RT), control (CON), step rate (SR), steps per minute (SPM), not reported (nr)

*Denotes BW units calculated from reported peak N and average participant mass **Denotes values obtained via web-based plot digitizer

Risk of Bias Assessment

Risk of bias at the study level in the selected articles was assessed by a single reviewer and cross-checked by a second reviewer using a modified Downs and Black Quality Index.[25] This modified scale was adapted from the original tool and has been used previously in reviews of similar nature on running related injuries.[26–28] Bias was evaluated on the following criteria: study objective, outcome description and accuracy, subject characteristics, cofounding variable description and analysis, statistical test selection and reporting, description of subjects lost to follow-up, and population representation. All items were scored as "yes" = 1, "no" or "unable to determine" = 0, with the exception of Item 5 which was scored as "yes" = 2, "partially" = 1, and "no" = 0. Risk of bias was determined based on collective score for each article with 5 or less points indicating a low-quality study due to high risk of bias, 6-10 points indicating moderate risk of bias, and 11 points or more implying low risk of bias therefore a high-quality study. If risk of bias was unclear or discrepancies existed in scoring among the reviewers, resolution was determined by discussion and review of the article in question.

Data Synthesis and Meta-Analysis

Statistical analysis was conducted in RStudio (v2025.9.1.401, Posit Team, Bosten, MA). Studies were categorized as "distal", "proximal", or "SO" according to methodology used to estimate tibial axial force. Meta-analysis was conducted RStudio package "metafor"[29] to examine whether estimation method influenced peak tibial axial forces during running. To reduce confounding from heterogeneous experimental conditions and to maintain statistical independence of effect sizes, only one condition per study was included in the meta-analysis. When multiple tibial axial force outcomes were reported, force values from the following

conditions were selected: speed closest to the overall mean running speed across studies (3.30 m/s), preferred stride length, 0-degree incline, 0 kg load carriage, and when available, a rearfoot strike pattern were selected. A random effects model with estimation method as a moderator was used to determine if method was a significant predictor of peak tibial axial force outcome. This model was chosen due to expected heterogeneity in study populations and methodologies; heterogeneity was assessed using Higgins I^2 statistic and were interpreted as low (0-40%), moderate (30-60%), substantial (50-90%), and considerable (75-100%).[30] The mean difference and confidence interval (CI) for tibial axial force between methods was calculated, as well as Hedge's *g* for effect size. Hedge's g was interpreted as small effect size .2, medium effect size 0.5, or large effect size 0.8.[31] Values of $p < .05$ were considered statistically significant.

Results

Study Selection

The initial literature search resulted in 1500 total records (Figure 2.1). After removal of duplicates and addition of one pertinent study not indexed in search databases, 1100 records remained. After a survey of titles, 70 abstracts and full texts were reviewed. Of these, 51 studies did not meet inclusion criteria and were removed, two were omitted as they were abstracts only, one was omitted due to inability to extract data, and one was removed due to duplicate data. The final literature search resulted in fifteen full text articles to be included in the review and meta-analysis.

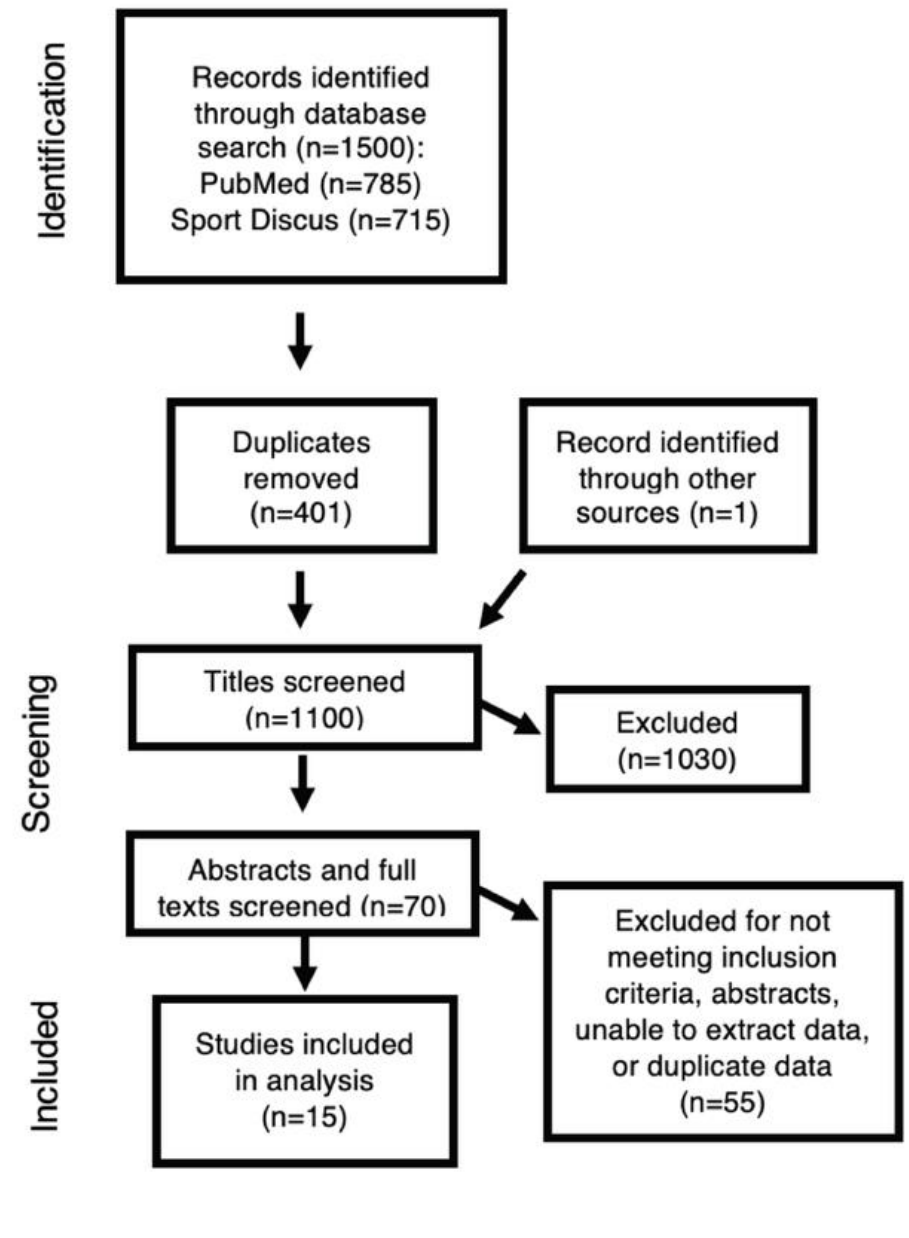


Figure 2.1 PRISMA flow diagram for search strategy

Risk of Bias

Modified Downs & Black Quality Index scores ranged between 9-14 points (Table 1.2). Eight studies[5,7,12,13,16–18,32] scored in the moderate risk of bias range indicating moderate quality studies, and seven studies[2,6,11,14,15,33,34] scored in the low risk of bias range indicating high quality studies. Internal study bias primarily occurred within categories concerning confounders as seven studies provided tables of potential confounding variables,[2,6,7,11,32–34] and eight studies reported adjustments for potential confounders within the statistical analysis.[2,5,6,14,16,32–34] Out of the studies that considered confounders, twelve controlled or adjusted for running speed, seven for strike pattern, and four for sex. The external validity category also had high risk of bias as only five studies reported concise methods for participant recruitment.[2,6,11,12,34]

Table 1.2 Risk of Bias Assessment from Downs & Black Quality Index

Criteria	1	2	3	5	6	7	9	10	11	16	18	20	25	**Total**
Baggaley et al. 2022	1	1	1	1	1	1	0	1	0	1	1	1	0	10
Boyer et al. 2018	1	1	1	0	1	1	1	1	0	1	1	1	1	11
Edwards et al. 2009	1	1	1	1	1	1	0	1	0	1	1	1	0	10
Edwards et al. 2010	1	1	1	0	1	1	0	1	0	1	1	1	0	9
Hunter et al. 2019	1	1	1	0	1	1	0	1	1	1	1	0	0	9
McDonnell et al. 2019	1	1	1	2	1	1	0	0	0	1	1	1	1	11
Meardon et al. 2021	1	1	1	2	1	1	0	1	1	1	1	1	1	13
Rooney et al. 2013	1	1	0	2	1	1	0	0	0	1	1	1	0	9
Sasimontonkul et al. 2007	1	1	1	1	1	1	0	1	0	1	1	1	1	11
Thakkar et al. 2019	1	1	1	2	1	1	1	1	0	1	1	1	1	13
Van Middelaar et al. 2025	1	1	1	1	1	1	0	1	0	1	1	1	1	11
Willy & Meardon et al. 2016	1	1	1	2	1	1	1	1	1	1	1	1	1	14
Willy & Willson et al. 2016	1	1	1	2	1	1	0	1	1	1	1	1	1	13
Willy et al. 2025	1	1	1	2	1	1	1	1	1	1	1	1	0	13
Zandbergen et al. 2023	1	1	1	1	1	1	1	0	0	1	1	1	0	10

Scoring: Items 1-3, 6-26: "yes" = 1 "no" = 0 "unable to determine" = 0 Item 5: "yes" = 2, "partially" = 1 "no" = 0. 1: Is the hypothesis/objective of the study clearly described? 2: Are the main outcomes to be measured clearly described in the Intro or Methods section? 3: Are the characteristics of the patients included in the study clearly described? 5: Are the distributions of principle cofounders in each group of subjects to be compared clearly described? 6: Are the main findings of the study clearly described? 7: Does the study provide estimates of the random variability in the data for the main outcomes? 9: Have the characteristics of patients lost to follow-up been described? 10: Have actual probability values been reported for the main outcomes except where the probability value is less than .001? 11. Were the subjects asked to participate in the study representative of the entire population from which they were recruited? 16: If any of the results of the study were based on "data dredging", was this made clear? 18: Were the statistical tests used to assess the main outcomes appropriate? 20: Were the main outcome measures used accurate (valid & reliable)? 25: Was there adequate adjustment for confounding in the analysis from which the main findings were draw?

Study Characteristics

The fifteen articles that met inclusion criteria were published between July 2007 and October 2025. Sample sizes ranged from 10-50 participants with *a priori* power analysis for sample size reported in three studies.[11,12,34] All studies were cross-sectional by design and were conducted on healthy adults who ran regularly to competitive runners with two exceptions: one study reported on participants who were healthy active adults without requirement of regular running activity,[11] and one study reported on military cadets.[34] Four studies included only male participants,[5,13,17,34] one study included only females,[33] one study did not specify sex[7] and all other studies included both male and female participants. Average ± SD for participant demographics were as follows: age 25.0 ± 5.0 years, height 1.76 ± 0.06 m, mass 69.9 ± 5.71 kg. When reported, the average weekly running volume of participants ranged from 22.65 to 96.4 km per week; however, six studies did not report activity volume,[13–15,17,32,33] and two studies reported average Tenger scores ranging from 5.9 to 6.9 out of 10.[11,34]

Study Protocols

Data collection was conducted using an instrumented treadmill in nine studies[2,6,11,14,15,18,33,34] while an overground track or runway ranging from ten to fifty meters was used in the remaining studies.[5,7,12,13,16,17,32] Of the treadmill studies, participants ran 60 seconds to eight-minutes prior to data collection, one study did not quantify run time,[33] and a single study reported data collection began after each participant had subjectively accommodated to the treadmill without quantification of duration.[15] All overground running studies reported the number of successful trials which ranged from three to ten per condition. Running speed averaged 3.30 m/s across studies and ranged from 2.20 to 4.50 m/s.

Data Processing

Inverse dynamics was used in all studies to estimate joint reaction forces and moments, while tibial axial forces were estimated by summing the joint reaction forces with muscle forces from the chosen musculoskeletal model. Four studies utilized the distal musculoskeletal model to determine plantarflexor muscle force as the quotient of the internal plantarflexor moment and an estimated ATMA.[11,12,14,18] Within these studies, ATMA was estimated as 20% of foot length,[12] as a fixed 5.0 cm,[14,18] and as a function of sagittal ankle kinematics.[11] Four studies used the proximal approach to estimate tibial force by considering the gastrocnemius, hamstring, and quadricep forces.[6,32–34] Last, seven studies used SO to estimate lower limb muscle forces as input into the tibial force model.[2,5,7,13,15–17] Of these, two studies used a modified Arnold model[15,16,35] while five studies used a SIMM model developed by Delp, et al.[2,5,7,13,17,36] All studies using the proximal or SO methods acquired tendon moment arms from previously published values based on a variety of techniques including cadaver studies, *in vivo* imaging, and a tendon excursion method.

Data Reporting

Outcomes reported in this review were reserved for peak tibial axial forces while acknowledging that additional variables not included within this review may be reported within the fifteen reviewed studies.

Study conditions with a main effect on peak tibial axial force included running speed, strike pattern, stride length, surface grade, and load carriage. Peak tibial force increased with running speed in all studies that included varying speed conditions.[2,12–14] Greater peak forces were also reported with a forefoot strike pattern during running regardless of habitual foot strike

pattern.[7,16] Peak force was reduced in the declined as well as the +10-degree incline condition as compared to the 0-degree condition.[15] A shortened stride length or increased step frequency as compared to a habitual stride length or step frequency also reduced peak tibial forces.[6,16,17,33,34] Finally, running with a 15 kg load increased proximal peak tibial force by ~5.6% and distal peak tibial force by ~.9 BW, as compared to running without external load.[11,34]

Data Synthesis and Meta-Analysis Results

Individual study estimates and pooled means are illustrated in Figure 2.2. Weighted pooled mean peak tibial axial force was 7.83 BW (95% CI: 6.31-9.35) in the distal models, 9.79 BW (95% CI: 8.70-10.88) for proximal models, and 11.29 BW (95% CI: 9.95-12.29) for the SO models.

Meta-analysis of tibial force outcomes from all fifteen studies revealed that estimation method significantly influenced peak tibial axial force during running (QM = 15.23, $p < 0.001$, $R^2 = 49.78$), with a large amount of heterogeneity between studies (QE (12) = 316.80, $p < 0.001$, $I^2 = 97.33$, $tau^2 = 1.94$). No studies were found to be significantly influential therefore all studies were retained in the analysis. The SO method yielded peak tibial force values that were 3.45 BW greater than the distal method ($p < 0.001$, 95% CI: 1.60-5.30, $g = 3.71$). The proximal method also yielded peak tibial force values greater than those of the distal method by approximately 1.96 BW ($p = 0.049$, 95% CI: 0.01 – 3.91, $g = 1.62$). A Wald χ^2 test for *post-hoc* pairwise comparisons indicated no significant difference in peak tibial forces between proximal and SO methods ($\chi^2(1) = 2.82$, $p = 0.093$).

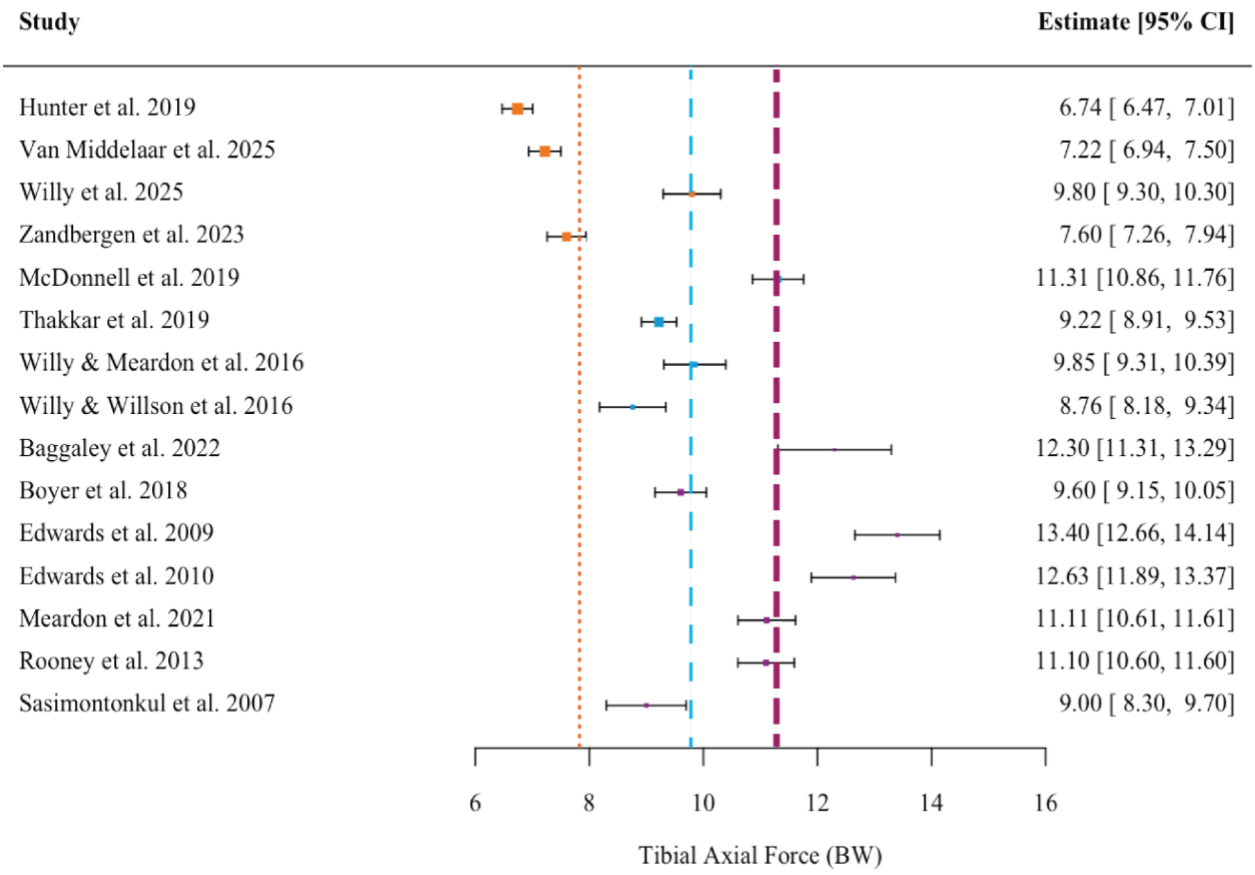


Figure 2.2 Peak tibial axial forces for individual studies. Mean peak tibial axial force (BW) is indicated numerically and by squares: orange (distal), blue (proximal), and purple (SO). The size of the squares represents the influence of the study effect on the pooled estimate. The vertical lines represent weighted pooled means for tibial axial force for distal (dotted), proximal (thin dashed), and SO groups (thick dashed).

Discussion

This systematic review and meta-analysis evaluated differences in peak tibial axial force estimates during running between three musculoskeletal modeling methodologies. Estimation methodology accounted for approximately 50% of between-study variance in peak tibial axial force. Likewise, study design variables that impacted tibial force included running speed, strike pattern, step frequency, surface grade, and load carriage.

Model Influence on Peak Tibial Force

The musculoskeletal models had a significant influence ($p < .05$) on peak tibial axial force. Both the proximal and SO methods resulted in peak forces 1.96 and 3.45 BW greater, respectively, than studies using the distal method. A notable contrast between methods was the use of multiple muscle groups within the proximal and SO methods as compared to a single muscle group in the distal method. Because the distal method constructs internal force contributions solely through the Achilles tendon force, the distal method may underestimate peak tibial forces by omitting other muscles that may contribute to internal tibial load. Similarly, both the proximal and SO methods rely on multiple muscles as contributors to internal force input to tibial load; therefore, total load may be overestimated in these models. Alternatively, the cost minimizing function of SO may underestimate antagonist muscle activity while assigning higher forces to larger muscle groups, resulting in inflated plantarflexor muscle activity and potential to overestimate tibial axial force.[37] Despite the musculoskeletal model accounting for nearly half of the outcome variability, a large portion of unexplained variability remains. Therefore, alternative

factors must contribute to the differences in tibial forces during running in addition to the musculoskeletal model.

Confounding Variables and Study Heterogeneity

Although the musculoskeletal model accounted for ~50% of the reported heterogeneity, a substantial portion (~47%) of the between-study variability remained unexplained by estimation method. Confounding variables are a potential source for the between study variability when estimating tibial force during running. The remaining variability may be explained by differences in participant populations as well as methodological conditions such as speed, strike pattern, load carriage, surface grade, step frequency, running experience, and sex. These conditions have potential to alter musculoskeletal demand, especially on the plantarflexors. Because the plantarflexors contribute up to 80% of the total internal tibial force, any condition that influences plantarflexor activity is also likely to influence tibial force.[7,38–42] Within the reviewed studies, conditions that increased plantarflexor demand included increased running speed, utilizing a forefoot strike pattern, and the addition of load carriage.[2,11–14,34,43] Running speed was on average 3.78 and 3.23 m/s within the SO and distal method studies, respectively. Therefore, a portion of increased tibial force seen in the SO models may be accounted for by running speed due to the direct relationship between speed and tibial force. Another source of potential variability was foot strike pattern. Foot strike was unconstrained in approximately 50% of distal model studies, in all proximal model studies, and 57% of SO studies, creating a plausible source of greater tibial axial force through forefoot striking in both proximal and SO models. Both distal and proximal groups used an instrumented treadmill in 75% of studies while only ~29% of SO studies used an

instrumented treadmill highlighting an additional disparity that may have contributed to outcome variability.[44]

Last, running experience and sex are two variables with potential to influence mechanical loading yet remained largely unaccounted for within the reviewed studies. Participants ranged from recreationally active adults to competitive runners to military cadets, representing a wide range of active populations. Previous studies have reported differences in running mechanics between novice and experienced runners; specifically, novice runners demonstrated increased external moments at the ankle, a variable that has potential to elevate tibial bone force.[45–47] Tibial force has also been reported to be reduced in women compared to men during running. While it is important to acknowledge the potential for running experience and sex to influence tibial loads, it is difficult to quantify differences between studies within this review due to inconsistent reporting.

Limitations and Assumptions

Limitations within this review and meta-analysis include model validity, model assumptions, study heterogeneity, and application of findings. External forces acting on the lower extremity during running have long been investigated; however, model validation for tibial forces is difficult due to challenges associated with acquiring in vivo tendon or bone forces.[48,49] While in vivo methods for collecting tendon force data exist, these methods are limited to a single tendon and small participant numbers due to the invasiveness of the procedure.[50,51] Additionally, in-vivo contact forces have been measured under specific experimental or clinical conditions, however, direct in-vivo measurement of tibial axial force in healthy recreational running populations is currently not reported.[24] Therefore, the validity of the musculoskeletal

models discussed cannot be directly confirmed and researchers must be aware of model assumptions. Tibial force model assumptions introduce additional variability and complicate explanations of the relationship between tibial bone load and running. Each of the three models within this review rely on inverse dynamics to calculate joint moments and reaction forces. Although inverse dynamics is a common musculoskeletal modeling technique, it is unable to account for distribution of forces or moments to individual tissues which likely results in an oversimplification of specific tissue load. The high variability between studies is also a limitation to the meta-analysis as it restricts the meta-analysis to the most homogenous sample possible from each study rather than include all outcomes from all studies. The heterogeneity in sample demographics also limits generalizability of results. Finally, reported outcomes represent acute tibial forces only and findings should not be assumed to be consistent with cumulative tibial loads.

Recommendations and Future Directions

Based on the findings of this systematic review and meta-analysis, it is recommended that future studies estimating tibial bone force during running compare outcomes only to studies with similar musculoskeletal model methodology. Additionally, careful consideration and reporting of confounding variables that may affect tibial bone load is warranted. Running speed, foot strike pattern, surface grade, step frequency (preferred or otherwise), and load carriage may confound outcomes of tibial axial force and should be constrained in methods or adjusted within analysis. Future studies that investigate tibial force outcomes from a direct comparison of methods will aid in our understanding of the direct impact of methodology on tibial force with reduced influence of between study variability.

Conclusion

Tibial axial force during running in healthy adults is reduced when estimated with a distal model as compared to a model using a proximal approach or static optimization. Investigation of fifteen moderate to high quality studies revealed high between-study heterogeneity that likely influenced the outcome of tibial axial force during running and limits the scope of application of the results of meta-analysis. Running speed, foot strike pattern, surface incline, stride length, and load carriage are variables that affect outcomes of estimated tibial bone force during running and are recommended to be reported in future studies. Additional studies are warranted to identify the effect of these variables on tibial axial force in an injured population to expand upon the relationship between tibial axial force and running related injuries.

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Statements and Declarations

The authors have no conflicts of interest to disclose.

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CHAPTER THREE

ACHILLES AND TIBIAL FORCE ESTIMATION DURING RUNNING USING THREE METHODOLOGIES FOR CALCULATING ACHILLES TENDON MOMENT ARM

Contribution of Authors and Co-Authors

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Contributions: Conducted and collected data, processed data, performed statistical analysis, generated graphics, wrote, edited, revised final manuscript

Co-Author: J. Whitten

Contributions: Assisted in methodology for isokinetic dynamometer data collection

Co-Author: J. Becker

Contributions: Study conceptualization; methodology development; consulted on statistical analysis, read and revised final manuscript.

Abstract

Introduction: The magnitude of tibial bone load has been implicated in the development of tibial bone stress injuries (BSI). Several musculoskeletal models have been used to calculate the Achilles tendon moment arm (ATMA), an integral component of modeling Achilles tendon and tibial forces. Whether and how differences in ATMA length propagate to differences in Achilles or tibial bone load remains unclear. Purpose: the purpose of this study was to compare Achilles tendon and tibial axial forces during running using three methods to estimate ATMA. Methods: Sixteen healthy adult participants (M=9, F=7, 23.25 ± 2.32 years, 1.77 ± .08 m, 75.96 ± 12.99 kg) free of lower extremity injury ran across imbedded force plates while whole-body kinematics were recorded. Inverse dynamics was used to calculate ankle joint reaction forces and moments which were used, along with ATMA lengths, as inputs to a musculoskeletal model to calculate Achilles tendon and tibial axial forces. ATMA was estimated using three methods: 1) the tendon excursion method (ATMA-te), 2) as a fixed value of 20% of foot length (ATMA-f), and 3) a regression equation based on ankle angle (ATMA-r). Linear mixed models tested for a fixed effect of ATMA method on ATMA length, Achilles, and tibial axial forces with subject modeled as a random effect. Statistical parameter mapping (SPM) was used to assess temporal differences in forces between methods across the stance phase of running. Results: A significant fixed effect was found for method on ATMA length ($p<.001$); ATMA-r and ATMA-te were significantly shorter than ATMA-f ($p =.002$). Fixed effects of method were also found for peak Achilles and tibial axial forces ($p<.001$). Forces derived from ATMA-r were greater than forces estimated with ATMA-f and ATMA-te ($p<.001$), and forces derived from ATMA-te were larger than forces from ATMA-f ($p<.001$). SPM revealed differences in Achilles and tibial forces between ATMA

methods between 33-80% and 84-99% of stance ($p < .001$). Conclusion: Results revealed a 21.9% difference in peak tibial loads with a regression equation as compared to ATMA estimated from foot length. ATMA estimation methodology has substantial effects on peak axial forces during midstance of running gait, which may influence the relationship between these forces and BSI.

Introduction

The magnitude of tibial bone force is an important variable for the consideration of etiology and recurrent injury risk of tibial bone stress injuries (BSI) in runners, as the repetitive loads of running have been implicated in the development of BSI.[1–3] Approaches for modeling tibial bone loads often involve identification of participant musculoskeletal geometries using computed tomography or magnetic resonance imaging.[3–8] However, these approaches rely on advanced imaging and modeling techniques, and time investments that are often not practical for larger sample sizes or repeated testing. Therefore, simplified musculoskeletal models are often favored due to decreased imaging needs, increased accessibility of modeling techniques, and reduced time requirements.

Simplified musculoskeletal models are commonly used in running research to estimate distal axial tibial bone force by summing ankle joint reaction force from inverse dynamics with plantarflexor muscle force.[9–14] In contrast to complex models that consider multiple muscle contributions, a simplified distal model assumes that the plantarflexor force, synonymous with the Achilles tendon force, is the primary contributor to the internal force contributions to total tibial load.[14–17] As a result, this force has a substantial influence on total tibial axial force and any parameter related to plantarflexor force will also influence tibial force estimates. One such parameter is the Achilles tendon moment arm (ATMA), defined as the perpendicular distance from the ankle joint center of rotation to the line of action of the Achilles tendon.[18] Because ATMA length influences Achilles force, the method of estimating ATMA has potential to also influence estimated tibial force.

Previous research on participant specific ATMA estimation established that the magnitude of the ATMA depends on the calculation methodology. [9,10,12,19] Additionally, in the absence of advanced imaging, simplified models for tibial force estimation use several different approaches to determine ATMA. First is a participant specific method using ultrasound imaging and the tendon excursion method to calculate ATMA.[9,10,20] While this approach allows for individual measurement through imaging, it requires specialized equipment and processing tools. Second, a fixed moment arm length may be calculated as a function of foot length.[14,21] This approach accounts for some variability between participants, does not require imaging equipment, and data collection processing times are minimal. Finally, a regression equation has been used to estimate moment arm length as a function of ankle angle. The regression approach requires biomechanical processing tools, however, allows ATMA to vary throughout running stance.[12,22,23] The three methods to calculate ATMA have potential to produce substantial variation in Achilles and tibial force estimates.

Participant specific methodologies utilizing magnetic resonance and ultrasound imaging for ATMA estimation have reported differences of up to 27.4% in ATMA length.[10] Additionally, a 20% difference in ATMA length can result in up to 50% error in muscle force estimates highlighting that length differences can be problematic for accurate force estimates.[12,24] However, these observations have only been reported for participant specific methods and differences in ATMA lengths from the three simplified methods commonly used in running studies have yet to be compared. Furthermore, it remains unclear whether and how ATMA length differences propagate to Achilles and tibial forces during running. Thus, the purpose of this study was to compare ATMA lengths from a) the tendon excursion method, b) a fixed estimate based

on foot length, and c) a regression equation based on ankle joint angles, and to investigate how these differences influence plantarflexor and tibial force estimates during running.

Methods

Sixteen healthy adults (M=9, F=7, 23.25 ± 2.32 years, 1.77 ± .08 m, 75.96 ± 12.99 kg, 326.56 ± 158.48 weekly minutes of physical activity) participated in this study. Participants were recruited via word of mouth, social media, and public flyers and informed consent was provided prior to participation. Inclusion criteria were as follows: age 18-35, free of lower extremity injury for the six months prior to data collection, and participation in a minimum of 150 minutes of moderate physical activity per week. This study protocol received approval by the Institutional Review Board at Montana State University (IRB#2024-1661).

Participants completed a five-minute jogging warm-up at a self-selected pace. Following the warm-up, participants ran a length of 15 m over an in-ground force plate (ATMI, Inc, Watertown, MA) at a self-selected pace consistent with an easy run. Approximately 10-12 attempts were required per participant to achieve five successful trials of landing in the center of the force plate without targeting. Successful trials were confirmed by visual inspection of 2D video.[25,26] Markerless motion capture was recorded with eight video cameras (Qualisys AB, Göteborg, Sweden) at 100 Hz while force data was collected at 1000 Hz.

To collect data for the tendon excursion method, each participant was comfortably seated in an isokinetic dynamometer (Biodex Medical Systems, Shirly, NY) with the right hip positioned at 85 degrees, the right knee at 30 degrees, and the right foot positioned so that the axis of ankle rotation was aligned with the lateral malleolus. The right lower extremity was fixed with non-elastic straps to aid in dynamometer control of ankle range of motion (ROM). Neutral

ankle ROM was determined with goniometric measurement after which participants were passively moved through full dorsiflexion (25 degrees) and plantarflexion (30 degrees) to ensure no discomfort within the required range of motion.[28] A 65 mm linear B-mode ultrasound probe was fixed at the level of and perpendicularly to the Achilles musculotendinous junction (AMTJ) with non-elastic bands, while water-based transmission gel was applied between the skin and probe to improve image quality (Figure 3.1). The dynamometer protocol was as follows: isokinetic plantarflexor force was applied throughout passive ankle rotation from neutral to a targeted angle at 10°/sec for three repetitions with a two-second hold at peak ROM and two second rest between repetitions.[11,29] A minimum of 13 Nm torque was required to initiate movement while a 27 Nm threshold was applied in which the protocol was terminated if the threshold was exceeded to avoid discomfort or injury. The protocol was increased by five-degree increments per set and repeated until 25 degrees dorsiflexion and 30 degrees plantarflexion were achieved. AMTJ displacement was recorded with continuous ultrasound imaging (Echowave II, Telemed, Lithuania) at 7 MHz and 30-40 mm depth as described in previous studies.[9–11,20,27] To assess reliability, displacement measurements were collected and measured by two researchers for two participants on multiple occasions separated by five days. Interrater and intrarater ICC estimates for ultrasound measurements were .996 (CI .992-.998) and .994 (CI .985-.998), respectively.

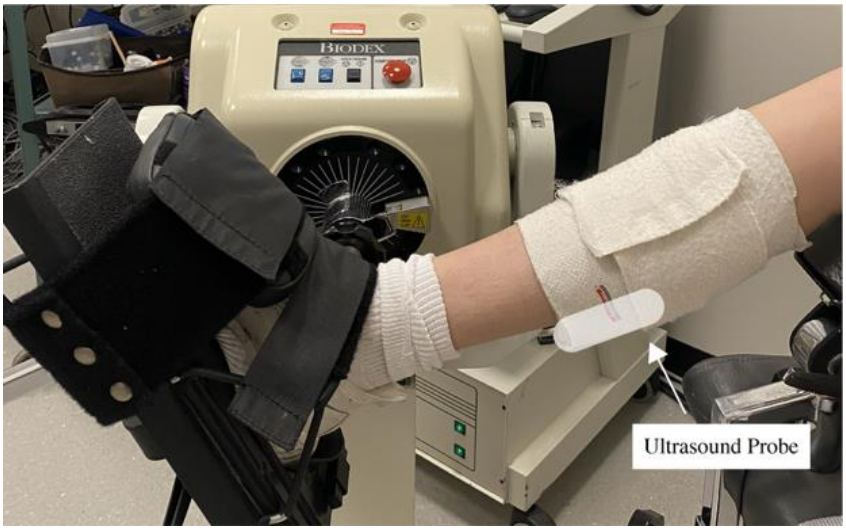


Figure 3.1 Dynamometer and Ultrasound Set-up

Data Processing

Video recordings from running trials were processed in Theia (Theia Markerless, Inc., Kingston, ON, Canada) using synchronized video data and algorithms to identify anatomical landmarks, segment positions, and pose to serve as input to Visual 3D (v2026.01.1, HAS Motion, Germantown, MD, USA).[30,31] Within Visual3D, a seven-segment biomechanical model was created for each participant including torso, pelvis, bilateral femur, bilateral shank, and bilateral foot segments. Ground reaction forces (GRF) were processed with a 20 Hz zero lag lowpass Butterworth filter, and inverse dynamics was used to calculate ankle joint reaction forces and moments, while Cardan XYZ sequencing was used to identify ankle joint angles.

A custom Matlab code (The Mathworks, Inc., Natick, MA) was used to calculate ATMA from ultrasound data, anatomical measurements, kinematic, and kinetic output. The three distinct methodologies used for ATMA estimation were as follows: 1) ATMA estimated from the tendon

excursion method (ATMA-te),[9,10] 2) ATMA estimated as a function of foot length (ATMA-f),[14] and 3) ATMA estimated as a function of a regression equation based on ankle angle (ATMA-r).[12,22,32] The tendon excursion method involved imaging software (Echowave II, Telemed, Lithuania) to manually measure AMTJ displacement on ultrasound video during each repetition of ankle ROM. Displacement was averaged for the three repetitions per set and plotted against ROM, and a second order polynomial was fit for displacement to ankle ROM (Figure 3.2). A fixed ATMA-te was calculated as the ratio of change of the polynomial using large interval differentiation as recommended by Fath, et al.[10] The second method calculated ATMA-f as 20% of foot length. Foot length was defined as the distance from the wall to the end of the first toe when the heel was placed in contact and perpendicularly with the wall. Finally, ATMA-r was estimated using ankle angle (θ) during the stance phase of running as input to a regression equation from Rugg, et al. (Eq. 1)[12]; ATMA-r was averaged across the stance phase for comparison to the previous two methods, however, remained in original format for calculating forces across the stance phase of running gait.

$$\text{ATMA (m)} = (-0.5910 + 0.08297*\theta - 0.0002606*\theta^2) /100. \quad \text{Eq. 1}$$

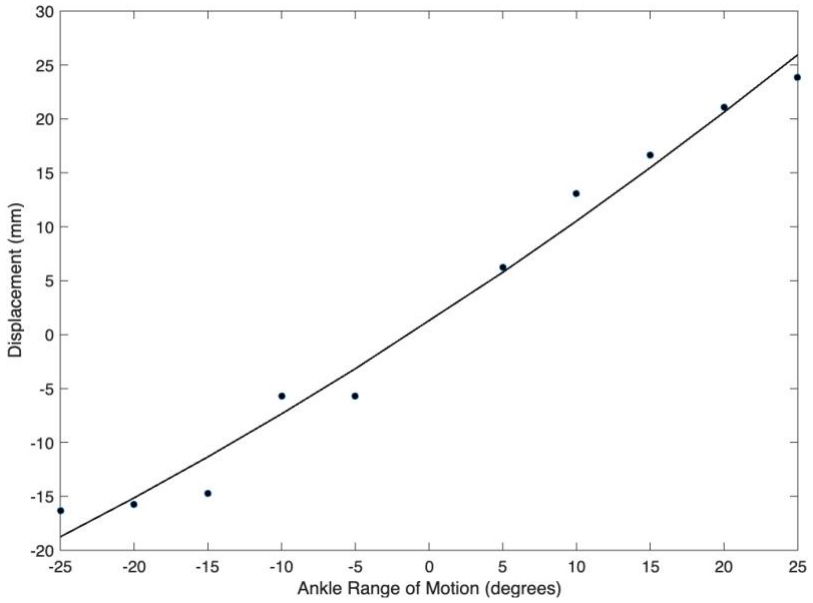


Figure 3.2 2nd Order polynomial for displacement vs ankle ROM for a single random participant. Single dots represent the average displacement across three trials of each set of ROM.

ATMA lengths from the three methods were used to estimate Achilles and tibial axial forces. Achilles force (F_A) was then calculated as the quotient of ankle moment (M_A) to ATMA (Eq. 2) and last, the axial component of ankle joint reaction force (JRF_A) was summed with Achilles force to estimate tibial axial force (F_T) (Eq. 3).[18,23,19] Finally, Achilles and tibial forces were normalized to body weight (BW). A full schematic of the methods and data processing workflow for ATMA estimation from three methods can be visualized in Figure 3.3.

$$F_A\ (N) = \frac{M_A}{ATMA} \qquad \text{Eq. 2}$$

$$F_T\ (N) = F_A + JRF_A \qquad \text{Eq. 3}$$

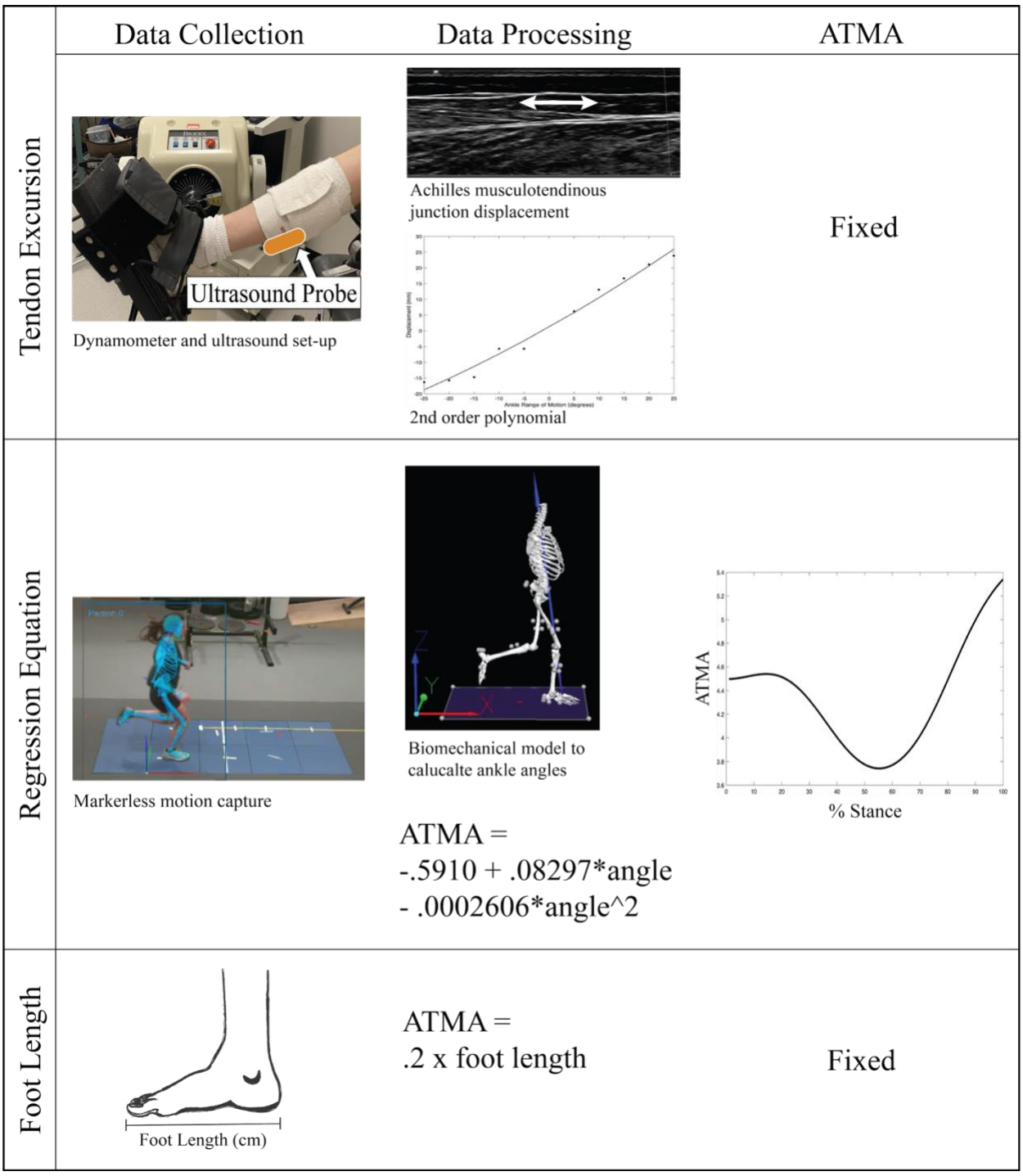


Figure 3.3 Schematic for data collection and processing workflow for three methods for estimating ATMA

Statistical analysis was performed in RStudio (v2025.9.1.401, Posit Software, Boston, MA) using the lme4 package.[34] Linear mixed models tested for a fixed effect of ATMA method on ATMA length, Achilles force, and tibial forces with subject modeled as a random effect. When a significant effect was found, pairwise comparisons with Bonferroni correction evaluated differences in ATMA lengths and peak forces between methods. Effect sizes (Cohen's *d*) were calculated to assess the influence of ATMA estimation method on outcome variables with .2 or less indicating a small effect size, .5 a moderate effect size, and .8 or greater a large effect size.[34] Statistical parametric mapping (SPM) was performed with open-source Matlab code [35] to assess temporal differences in forces between ATMA methods across the stance phase of running. Statistical test significance was set at $\alpha = .05$.

Results

Mean ATMA lengths as well as peak Achilles and tibial forces for each estimation method are listed in Table 2.1. A main effect of ATMA estimation method was found for ATMA length ($F(2,32) = 8.35$, $p = .001$). Post-hoc comparisons found ATMA-f larger than ATMA-te ($p = 0.032$, $d = 0.78$) and ATMA-f larger than ATMA-r ($p = 0.002$, $d = 1.45$), with moderate to large effect sizes. No differences were found between ATMA-te and ATMA-r ($p = 0.762$, $d = 0.28$).

Linear mixed models found a significant group effect of ATMA method on peak Achilles force ($F(2,32) = 22.99$, $p < 0.001$). Pairwise comparisons found that peak Achilles forces from ATMA-te were significantly greater than those derived from the ATMA-f method ($p = 0.012$, $d = 0.71$), and peak Achilles forces derived from ATMA-r were significantly greater than Achilles

forces derived from both ATMA-te ($p = 0.004$, $d = 0.65$) and ATMA-f ($p < 0.001$, $d = 1.55$) methods with moderate to large effect sizes.

Linear mixed models also found a significant group effect of ATMA method on peak tibial force ($F(2,32) = 23.79$, $p < .001$). Pairwise comparisons found peak tibial forces from ATMA-te were significantly greater than forces calculated from the ATMA-f method ($p = 0.012$, $d = 0.61$), and forces derived from ATMA-r were significantly larger than both ATMA-te ($p = 0.003$, $d = 0.61$) and ATMA-f ($p < 0.001$, $d = 1.34$) derived tibial forces.

Table 2.1 Results for ATMA length, Achilles and tibial force

Variable	ATMA-te	ATMA-f	ATMA-r
ATMA (cm)	4.59 ± .93*	5.13 ± .32	4.36 ± .08*
Achilles (BW)	4.88 ± 1.16*	4.22 ± .68	5.62 ± 1.05*†
Tibial (BW)	6.70 ± 1.27*	6.04 ± .85	7.46 ± 1.19*†

*Indicates significant difference from ATMA-f method (p <.05)
†Indicates significant difference from ATMA-te method (p<.05)

SPM results, illustrated in Figure 3.4, revealed a significant difference in both Achilles and tibial forces between the ATMA groups at 34-80% and 83-99% of stance (indicated by vertical grey bands) (df=2,30, $p < 0.001$).

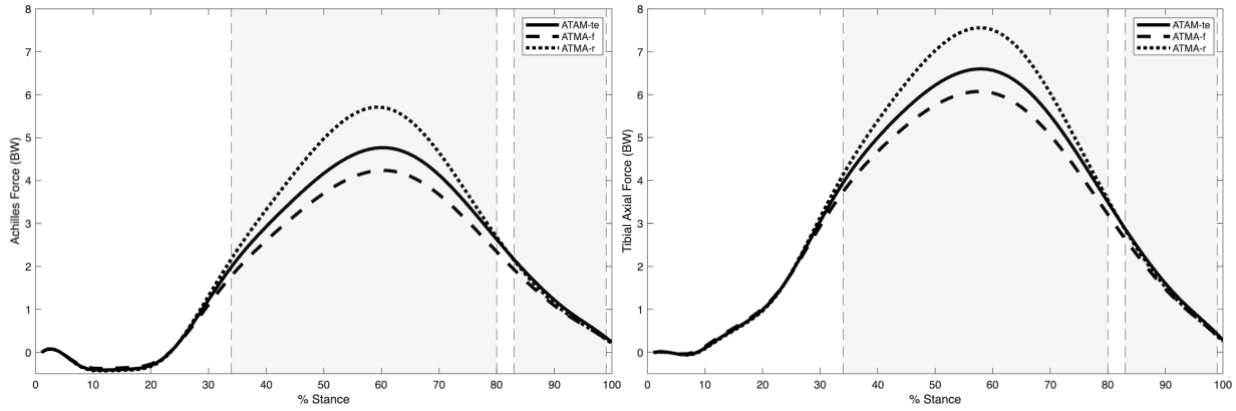


Figure 3.4 SPM for Achilles and tibial force throughout running stance.

Discussion

The purpose of this study was to compare ATMA lengths derived from simplified estimation methods and to examine whether differences in ATMA lengths propagate to Achilles or tibial forces during running. ATMA derived as a function of foot length was the largest while the mean ATMA length derived from a regression equation was the smallest. These length differences propagated to differences in peak Achilles and tibial forces which were largest when calculated with the regression method while the foot-length method resulted in the lowest peak forces.

ATMA lengths ranged from 4.36 - 5.13 cm which is consistent with previously reported data from subject specific magnetic resonance and ultrasound methods for similar ankle angles.[23,36-38] Although lengths from all three methods were consistent with previous reports, slight differences in ATMA lengths between methods substantially influenced subsequent force

estimations. A 16% difference in ATMA length between ATMA-f and ATMA-r methods resulted in a 28.6% difference in Achilles force and 21.1% difference in tibial force. Likewise, an 11% difference in ATMA length between ATMA-te and ATMA-f methods resulted in a 14.9% and 10.5% difference in Achilles and tibial loads, respectively. While the specific relationship of tibial axial force to BSI is unknown, Meardon, et al. (2015) found that posterior compressive tibial stress was increased by 13% in runners with a history of tibial BSI as compared to healthy controls.[3] This highlights that minor differences in tibial load may have clinical utility in differentiating between healthy and injured runners; therefore, methodology that affects estimations of tibial load such as ATMA calculation should be carefully considered.

The results of this study demonstrated that differences in ATMA lengths proliferated to differences in Achilles and tibial loads. The source of these force differences may be attributed to ATMA length differences and therefore estimation method as the moments and force calculations were identical across ATMA estimation methods. This supports the idea that ATMA estimation method is critically important when using a tibial force model that relies on plantarflexor force input. Based on the results of this study, the foot and regression methods resulted in forces that differed meaningfully from the tendon excursion method. When compared to the ATMA-te method, ATMA-r overestimated peak tibial forces by ~10.7% while ATMA-f method underestimated peak tibial forces by ~10.5%. Because these models had similar absolute differences as well as equivalent effect sizes when compared to the ATMA-te method, choosing an appropriate model to calculate ATMA and subsequent forces must be based on assumptions and ability to compare across studies.

A few assumptions and limitations are necessary to discuss when applying the aforementioned methods to estimate ATMA. The ATMA-te method was used within this study as a reference standard and is based on the assumption of virtual work. However, this method assumes constant tendon length, and assumes all excursion is attributable to joint rotation.[11,39] Restraining the isokinetic torque limited variations in muscle force and potential changes in tendon length while still allowing for applied force.[11,28] Another limitation that applied to both ATMA-te and ATMA-f methods was the use of fixed ATMA lengths to estimate Achilles and tibial loads. Previous studies have reported up to a 24.5% increase in moment arm length from plantarflexion to dorsiflexion which has potential to affect estimated forces throughout the running gait cycle. It is possible that a fixed moment arm over- or underestimates forces at extremes of ROM which is especially important during mid-stance when ankle dorsiflexion peaks and coincidentally tibial force also peaks.[12,29] The regression equation allowed for ATMA length to vary based on ankle ROM, however, was originally based on male participants and may not be generalizable to female participants. Last, the three methods for estimating ATMA within this study assumed single-plane sagittal movement of the ankle joint from plantarflexion to dorsiflexion. Cadaver studies have demonstrated minimal contributions to ATMA from frontal or transverse plane movements; therefore, considering only sagittal plane motion is likely appropriate for estimating ATMA.[40]

Finally, choosing a model may be partially dependent on the running variables of interest. Specifically, GRF loading rate and/or peak GRF as outcome variables may influence ATMA model selection. Previous authors have reported greater GRF loading rate and greater increases in loading rate in a fatigued state in runners with a history of BSI compared to healthy

controls.[1,41] The current study found no differences between methods for estimating peak Achilles or tibial loads during the initial contact phase of running. If variables during the initial third of stance are to be surrogates of tibial load and hypothesized to be related to BSI, such as loading rates, then the methodology for estimating ATMA is less important as differences in force curves were only found during mid and terminal stance. However, if force variables during mid or terminal stance are to be used in the consideration of risk factors for tibial BSI, then consideration of ATMA estimation method is warranted.[42]

Conclusion

ATMA length is method-dependent and influenced estimations of Achilles and tibial axial forces during running activity. It remains unclear if a regression-based or foot-length method is optimal for estimating ATMA when compared to a subject specific tendon excursion method. Researchers must evaluate model assumptions and limitations, and consider possible outcome variables prior to selecting an appropriate approach for estimating ATMA.

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Statements and Declarations

The authors have no conflicts of interest to disclose.

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CHAPTER FOUR

A CROSS-SECTIONAL ANALYSIS OF HEALTHY RECREATIONAL RUNNERS TO EXAMINE RELATIONSHIPS BETWEEN CLINICAL MEASURES AND TIBIAL LOAD DURING RUNNING.

Contribution of Authors and Co-Authors

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Contributions: Study conceptualization, data collection and processing, performed statistical analysis, generated graphics, wrote, edited, and revised final manuscript

Co-Author: L. Rosario

Contributions: Assisted with data collection

Co-Author: J. Becker

Contributions: Study conceptualization, methodology development, consulted on statistical analysis, read and revised final manuscript

Abstract

Introduction: Tibial bone load has been implicated in the development of tibial bone stress injury. However, estimating bone load in a clinical setting is challenging; therefore, surrogate measures of tibial bone load are necessary to identify runners who may be at risk for tibial bone stress injury. Purpose: To investigate relationships between strength and balance assessments and tibial bone load during running in a healthy running population. Methods: Forty-one recreational runners (21 F, 20 M, 24.00 ± 4.48 years, BMI 22.86 ± 2.70, 30.38 ± 19.36 km/week) ran at a self-selected pace on an instrumented treadmill for five minutes while kinematic and kinetic data were collected via motion capture. Following the run, clinical measures of ankle strength assessed via hand-held dynamometry (HHD), dynamic balance assessed via Y-Balance Test (YBT), and plantarflexor endurance were performed. Ankle joint angles, reaction forces, and moments were used as inputs to a musculoskeletal model to calculate tibial axial force as the sum of Achilles tendon and vertical ankle joint reaction forces. Instantaneous loading rate (ILR), impulse, and cumulative load over one km were calculated from tibial force and used as outcome variables in addition to peak tibial force. Multiple linear regression was used to determine relationships between strength and balance measures and tibial load variables during running. Results: A regression model containing HHD-dorsiflexion, HHD-plantarflexion, YBT, and SLHR endurance was not predictive of peak tibial force ($F(36) = 2.43$, $p = 0.07$, Adj. $R^2 = 0.12$), ILR ($F(36) = 2.04$, $p = 0.11$, Adj. $R^2 = 0.09$), impulse (($F(36) = 1.75$, $p = 0.16$, Adj. $R^2 = 0.07$) or cumulative load ($F(36) = 2.57$, $p = 0.05$, Adj. $R^2 = 0.14$). Conclusion: Clinical measures of ankle strength, dynamic balance, or plantarflexor endurance cannot be used as surrogates for tibial load variables during running. Continued investigation of running gait mechanics as potential surrogates for tibial load will add to our

understanding of load-related BSI risk factors and clinical interventions to reduce the risk of recurrent injury.

Introduction

Tibial bone stress injuries (BSI) account for up to 46% of all BSI in track and field athletes, and approximately 20% become recurrent injuries.[1,2] One risk factor for tibial BSI is elevated tibial bone load during running due to its contributions to the accumulation of microdamage with prolonged activity.[3–7] While bone load is a broad construct encompassing stress, strain, and force, the latter is commonly calculated for running-related research as it is non-invasive, can be used with large cohorts, and reflects both internal and external forces on the bone. Although often used in a laboratory setting, tibial force is a challenging metric to evaluate clinically due to the equipment requirements and biomechanical modeling techniques often not available to practitioners. This minimizes the practical application of any relationship between elevated tibial bone load and BSI. Therefore, surrogate clinical measures for tibial load during running are necessary to identify individuals with elevated tibial load.

Tibial force is estimated using musculoskeletal models that are reliant on inverse dynamics and muscle force estimation. In a simplified approach, ankle moment and Achilles tendon moment arm are used to calculate the plantarflexor force contribution to internal tibial load during running.[8–11] In this model, plantarflexor force supplies the entirety of the internal force contributions to the tibia. Therefore, clinical assessments intended as surrogate measures of tibial force should ideally target tests of plantarflexor muscle strength or endurance, as disparities in these attributes may correspond to discrepancies in tibial force. [6,12–14]

Common clinical tests that directly assess plantarflexor strength and endurance are hand-held dynamometry (HHD) and the single leg heel raise (SLHR). A hand-held dynamometer is an inexpensive clinical tool used to assess single-plane strength that has been shown to be reliable

for acquiring dorsiflexor, plantarflexor, invertor, and evertor strength.[15] Likewise, the SLHR test requires minimal equipment and is capable of assessing plantarflexor endurance in a weightbearing position, similar to the requirements of running.[16] Additionally, the Y-Balance Test (YBT) is traditionally a measure of dynamic balance that has also been associated with sport-related injuries when balance deficits are present.[17–20] This test requires lower extremity strength and endurance to complete, particularly at the ankle. HHD, YBT, and SLHR each capture specific characteristics of lower limb muscle function that have potential to influence tibial force and therefore provide plausible mechanisms for predicting tibial force. Together, these clinical tests are strong candidates for surrogate measures of tibial axial force during running.

While prospective studies have reported reduced calf girth related to the development of BSI, and muscle fatigue is reported to be related to increased magnitude of bone load, direct relationships between specific clinical assessments and variables of tibial bone load have yet to be established.[3,13,21,22] Therefore, the clinical usefulness of the relationship between elevated tibial bone load and BSI is diminished, highlighting the need to establish surrogate measures of tibial force during running. The purpose of this study was to identify relationships between variables of bone load during running and clinical measures of HHD for ankle strength, YBT, and SLHR endurance in healthy runners.[23–25]

Methods

This cross-sectional study was approved by the Montana State University Institutional Review Board (#2022-137) and all participants provided written informed consent prior to participation. Healthy runners between the ages of 14-35 were solicited to participate in the study via flyers, public announcements, and word of mouth. To be included, participants were required

to run regularly at least twice per week and be free from lower extremity injury for the six months prior to data collection. Participants were excluded if they reported a history of BSI in the previous year or were unable to run continuously for five minutes. An *a priori* power analysis determined a total of 46 participants would be required for a large effect size,[26] α of .05, and β of .80 (G*Power 3.1 Dusseldorf Germany).

Demographic and anthropometric data was collected upon arrival to the laboratory, and each participant was fit with a neutral running shoe (New Balance, Boston, MA) provided by the lab. Retroreflective markers were placed on bilateral acromion, iliac crests, anterior superior iliac spines, posterior superior iliac spines, medial and lateral tibiofemoral joint lines, medial and lateral malleoli, superior and inferior calcaneus, lateral calcaneus, base of 5th metatarsal, dorsum of the foot at the 2nd metatarsal head, and navicular as well as tracking marker sets secured on each thigh and shank. A standing static calibration trial was collected for 30 seconds after which participants performed a five-minute walking warm-up at 1.12 m/s. The warm-up was immediately followed by a five-minute run on an instrumented treadmill (Treadmetrix, Park City, UT) at a self-selected pace (average 2.88 ± 0.47 m/s) defined as an easy 10-k pace. Motion capture data was collected during the final 30 seconds of running using a seven-camera motion capture system (MotionAnalysis, Rohnert Park, CA) at 240 Hz and synchronized ground reaction force data at 1200 Hz.

After the running trial, data was collected from three clinical measures including ankle strength assessed with HHD, dynamic balance with YBT, and calf endurance assessed with a SLHR (Figure 4.1). Isometric ankle strength for dorsiflexion, plantarflexion, inversion, and eversion was assessed with the MicroFet3 HHD (Hoggan Health Industries, West Jordan, UT).

HHD has been reported to have good to excellent reliability and moderate validity for ankle strength measurements.[27–30] Based on previously reported methods, participants were positioned in long sitting with slight knee flexion while a practitioner experienced in HHD stabilized the lower shank with one hand and positioned the HHD in contact with the participant with the opposite hand. Manual resistance was applied as counterforce for each movement without overpowering the participant.[31–33] HHD test positions were assessed in random order, and each position was tested three times. For each test, participants were instructed to increase effort for three seconds followed by a three-second hold of maximum effort. Maximum values were recorded in Newtons, averaged over three trials, and normalized to body mass.

Dynamic balance was assessed with the modified YBT using the Y-Balance Test Kit (Functional Movement Systems, Chatham, VA, USA) which has been found to be a valid tool for dynamic balance assessment with excellent inter- and intrarater reliability in healthy adults in all directions.[34–36] A full description of YBT procedure is described in previous studies.[37,38] Briefly, participants stood on the test leg in socks or barefoot on a central block with hands on hips. Participants were instructed to reach as far as possible with the opposite limb in three directions: anterior, posteromedial, and posterolateral. Two rounds of three trials per direction were completed per limb, recorded to the nearest 0.05 cm, and averaged across trials. A composite score was calculated by summing the average reach of each direction and dividing the sum by 3-times leg length and multiplying the final score by 100. Leg length was defined as the distance in centimeters from the anterior superior iliac spine to the medial malleolus.[38]

The final clinical measure assessed was the SLHR, reported to have good test-retest reliability and reserved for last to avoid any adverse effects of calf fatigue on the other clinical

assessments.[39] SLHR test procedure has been described previously and was as follows.[29,30] Participants stood on a randomized single leg on a 10-degree incline board to begin the test procedure in ankle dorsiflexion and the knee in full extension. Balance was maintained by allowing participants to touch two to three fingertips on a wall with instructions to avoid weight bearing through the hands. A single heel raise was initially performed to familiarize participants with proper technique and to determine calcaneal height at full plantarflexion; height was marked with a thin piece of tape at the calf and on nearby equipment. For the test, single-leg heel raises were performed at the tempo of one beat up and one beat down to a metronome set at 60 beats per minute. Participants were instructed to perform as many repetitions as possible to fatigue which was determined as inability to maintain tempo, inability to achieve full height, attempting to initiation the heel raise with knee flexion, or subjective fatigue. When fatigue was observed, participants were provided a single cue to correct the movement after which the test was stopped once a participant was no longer able to maintain proper technique. After the initial test, the procedure was repeated on the opposite limb.

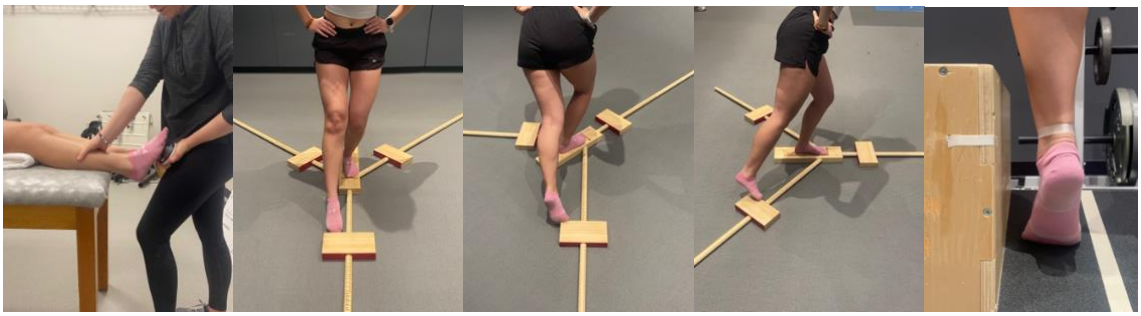


Figure 4.1 Example clinical measures. From left: HHD plantarflexion, YBT anterior, YBT posterolateral, YBT posteromedial, SLHR

Data Processing

Three-dimensional marker trajectories for static and dynamic trials were processed in Cortex (Motion Analysis Corporation, Rohnert Park, California, USA), after which the static trial was used to create an eight-segment rigid body biomechanical model in Visual 3D (v2026.01.1 HAS Motion, Germantown, MD, USA). Kinematic and force data were filtered with a zero-lag fourth-order low pass Butterworth filter at six and 25 Hz, respectively. Segment definitions were as follows: the thorax was defined by markers on right and left acromion and right and left anterior superior iliac spines; the pelvis was identified with markers at the right and left anterior superior iliac spines and a virtual marker midway between right and left posterior superior iliac spines; thigh segments were identified proximally by a virtual hip joint and distally by markers at the medial and lateral tibiofemoral joint line; shank segments were also identified proximally by the tibiofemoral joint line markers and distally by the medial and lateral malleoli; and finally, the foot was identified by markers at the medial and lateral malleoli as well as the 2nd metatarsal head. Segment coordinate systems were defined by the positive x-axis in an anterior orientation, the y-axis in a medial-lateral direction with the positive axis to the left, and the z-axis in an axial orientation with caudal to cranial designated as the positive direction.

A flexion/extension-abduction/adduction-axial rotation Cardan sequence was used to identify ankle joint angles while inverse dynamics was used to calculate ankle joint reaction forces and moments, reported in the shank coordinate system. Further analysis was performed only on the dominant limb which was identified as the extremity in which the participant would be most likely to kick a ball. Stance phase was determined using a 40-N threshold to identify initial contact and toe-off[12]; and was normalized to 101 data points. Ten gait cycles from the

dominant limb were input into a custom Matlab code (R2025b, The Mathworks, Inc., Natick, Massachusetts, USA) to calculate tibial load outcome variables.

Tibial load outcome variables included peak tibial axial force, tibial force instantaneous loading rate (ILR), tibial force impulse, and the cumulative tibial force over one kilometer. These variables were chosen as they describe both magnitude and duration of tibial load during running. Prior to force calculations, ankle angle (θ) was used as input to a regression equation to estimated Achilles tendon moment arm (ATMA) as reported in previous studies (Eq. 1).[40–43] Achilles force (F_A) was then calculated as the quotient of ankle moment (M_A) to ATMA (Eq. 2) and last, the axial component of ankle joint reaction force (JRF_A) was summed with Achilles force to estimate tibial axial force (F_T) (Eq. 3).[8–11,44]

$$ATMA\ (m) = (-0.5910 + 0.08297*\theta - 0.0002606*\theta^2) /100. \quad \text{Eq. 1}$$

$$F_A\ (N) = \frac{M_A}{ATMA} \quad \text{Eq. 2}$$

$$F_T\ (N) = F_A + JRF_A \quad \text{Eq. 3}$$

Tibial force was normalized to body weight (BW) and applied to calculations of vertical instantaneous loading rate (ILR), impulse, and cumulative load over one kilometer (CL). Vertical ILR was defined as the maximum slope from 20-80% of the most linear portion of the tibial force curve from the initial rise to the main peak.[12,45] Impulse was calculated at the total area under the tibial axial force curve using a trapezoid method to calculate the approximate integral

of tibial force across stance.[46] Cumulative load was then calculated as the impulse multiplied by the number of steps per kilometer.[8] Steps per kilometer was determine by first calculating step time as the time from strike to the next strike, multiplying treadmill speed by step time to gain step length, and finally calculating the number of steps per kilometer and multiplying the steps per kilometer by tibial force impulse.

Statistical Analysis

Statistical analysis was conducted in RStudio (v2026.01.0.+39, Posit team, Boston, MA). Interquartile ranges and leverage were used to assess for independent variable outliers. Individual observations that that fell outside of 1.5x the interquartile range and demonstrated leverage greater than three times the average leverage were considered for removal.

Four multiple linear regression models were used to identify predictors of tibial load from the independent variables of HHD, YBT, and SLHR assessments. Participants with a Cook's distance value of more than 4/n were flagged as potentially influential.[47,48] Participants were considered for removal if the Cook's distance was greater than .09 *and* the dependent variable was found to be an extreme outlier as determined by 1.5x interquartile range. If necessary, models were re-fit with removed points; coefficients, adjusted R^2, F-statistics, and *p*-values were compared between models with and without influential points to determine if removing influential points improved model fit. Model assumptions were assessed with visual inspection of histogram and Q-Q plot residuals to verify normal distribution, and homoscedasticity was assessed with visual inspection of a scatterplot of residuals vs. predicted values. Multicollinearity was determined by variance inflation factor values (VIF) and correlation matrices. Variables with VIF values greater than five and Pearson correlation coefficient values >.8 were considered

multicollinear and removed from the model. Model fit was determined by model p-values, and Pearson's correlation coefficient (r) between actual and predicted tibial load variables. Finally, model effect sizes using Cohen's f^2 were determined with cutoffs of 0.02, 0.15, and 0.35 for small, medium, and large effect sizes respectively.[49,50] Statistical significance was set at $\alpha = 0.05$.

Results

Data for 46 runners was collected, however, data from three participants was omitted from analysis due data loss, and two participants were removed after they were determined to be highly influential. Participant demographics can be found in Table 3.1. Distribution of independent variables are found in Figure 4.2 with mean ± SD as follows: HHD dorsiflexion 1.70 ± 0.40 N/kg, HHD plantarflexion 2.13 ± 0.75 N/kg, HHD inversion 1.23 ± 0.38 N/kg, HHD eversion 1.24 ± 0.40 N/kg, YBT composite score of 95 ± 12% leg length, SLHR 25.17 ± 5.77 repetitions. Distribution of outcome variables can also be found in Figure 3.1. Mean ± SD of outcome variables were as follows: peak tibial axial force 8.16 ± 1.20 BW, tibial ILR 105.92 ± 22.56 BW/s, impulse 1.14 ± 0.17 BW/step, and cumulative load 1100.60 ± 216.90 BW/km.

Table 3.1 Anthropometrics and Demographics

Participants (n)	Total = 41 (20 M 21 F)
Age (years)	24.00 ± 4.48
Height (m)	1.75 ± 9.80
Mass (kg)	71.20 ± 13.25
BMI (kg/m^2)	22.86 ± 2.70
Run Volume (km/week)	30.38 ± 19.36

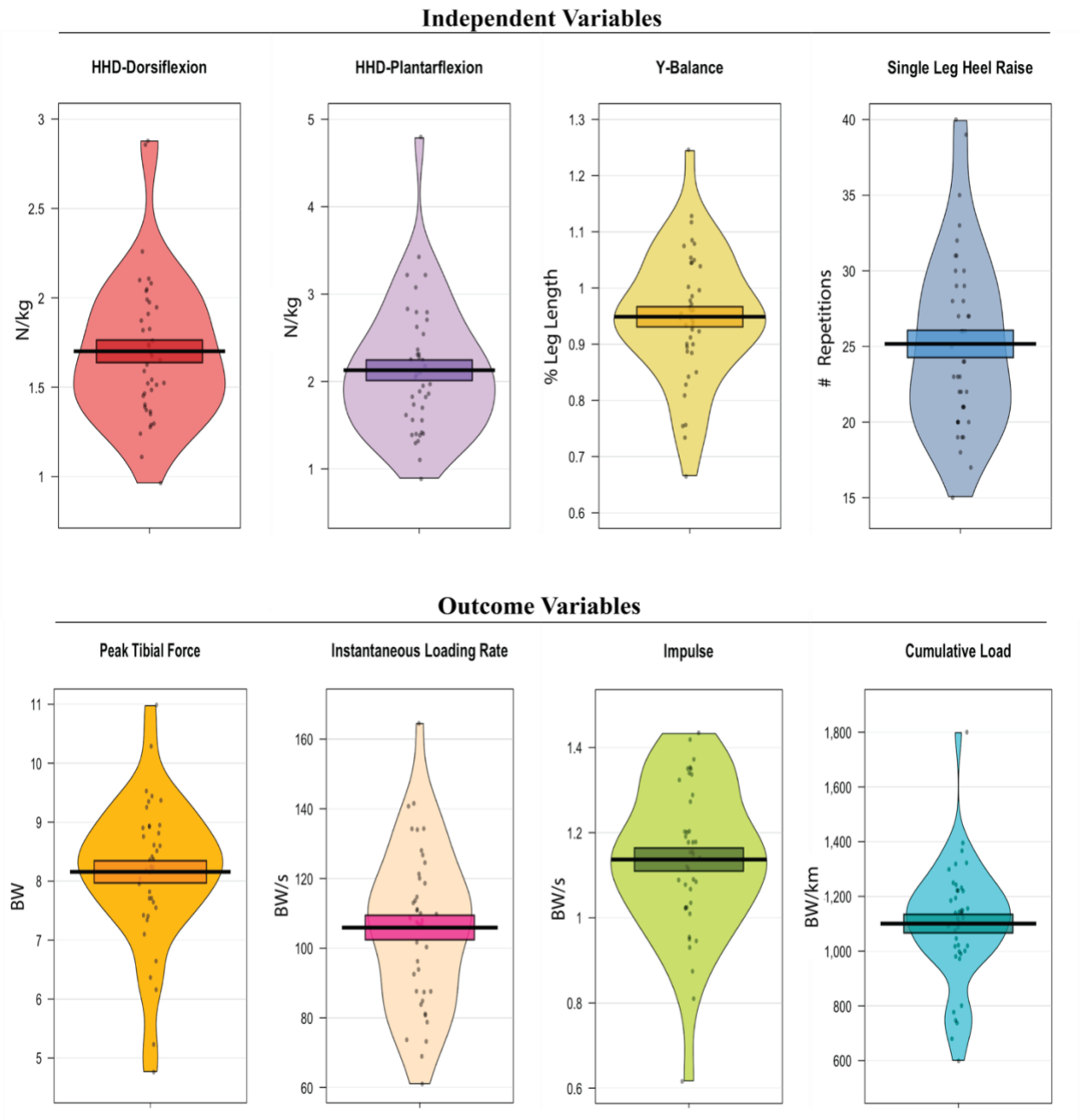


Figure 4.2 Distributions of independent and outcome variables.

Analysis of boxplots revealed up to two outliers per independent variable. However, no individual observations were found to be extreme outliers outside of a 1.5x IRQ, therefore, none were removed. Per Cook's distance, one participant (A) was highly influential in all four models of tibial load, and a second participant (B) was highly influential in three tibial load models. A secondary analysis of quartile ranges revealed all tibial load measures for participant A were extreme outliers, while peak tibial force and cumulative load were extreme outliers for participant B; therefore, participants A and B were removed from further analysis.

Model residuals were approximately normally distributed, and assumptions of homoscedasticity were met (See Appendix for residual plots). A VIF of >5 was found for HHD-inversion and HHD-eversion for all models. HHD-inversion and HHD-eversion were highly correlated (Pearson's $r = 0.89$) with moderate correlation of HHD-inversion (Pearson's $r = 0.75$) and HHD-eversion (Pearson's $r = 0.74$) to HHD-plantarflexion (Figure 4.3). Therefore, HHD-inversion and HHD-eversion were removed from the original model. [26]

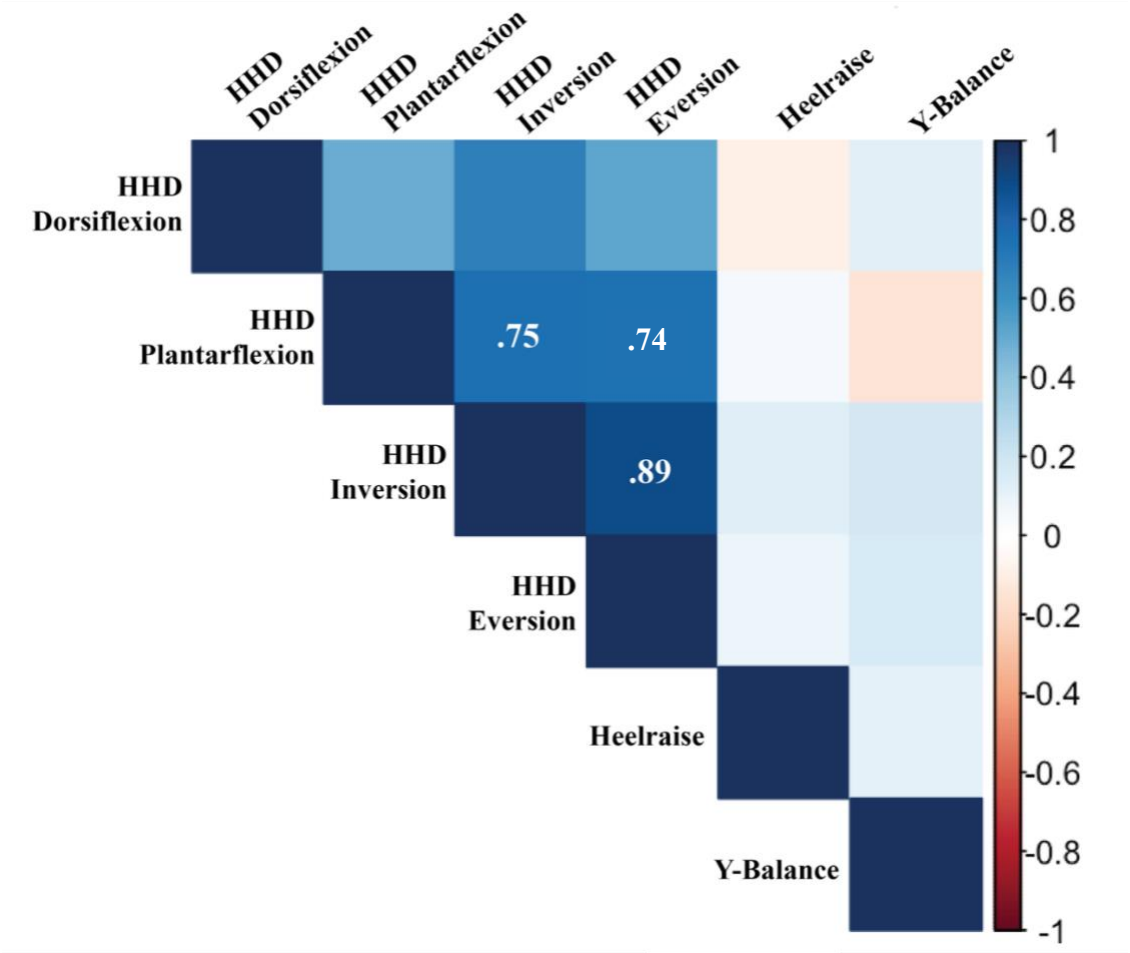


Figure 4.3 Correlation matrix and Pearson's *r* values >.70 for clinical measures

Clinical measures of HHD, YBT, and SLHR were not predictive of peak tibial force ($F(36) = 2.43$, $p = 0.07$, Adj. $R^2 = 0.12$, RMSE = 1.05, $f^2 = 0.27$), tibial ILR ($F(36) = 2.04$, $p = 0.11$, Adj. $R^2 = 0.09$, RMSE = 20.12, $f^2 = 0.23$), impulse ($F(36) = 1.75$, $p = 0.16$, Adj. $R^2 = 0.07$, RMSE = 0.16, $f^2 = 0.20$), nor cumulative load ($F(36) = 2.57$, $p = 0.05$, Adj. $R^2 = 0.14$, RMSE = 188.92, $f^2 = 0.32$). Scatterplots of predicted versus actual outcomes are illustrated in Figure 4.4.

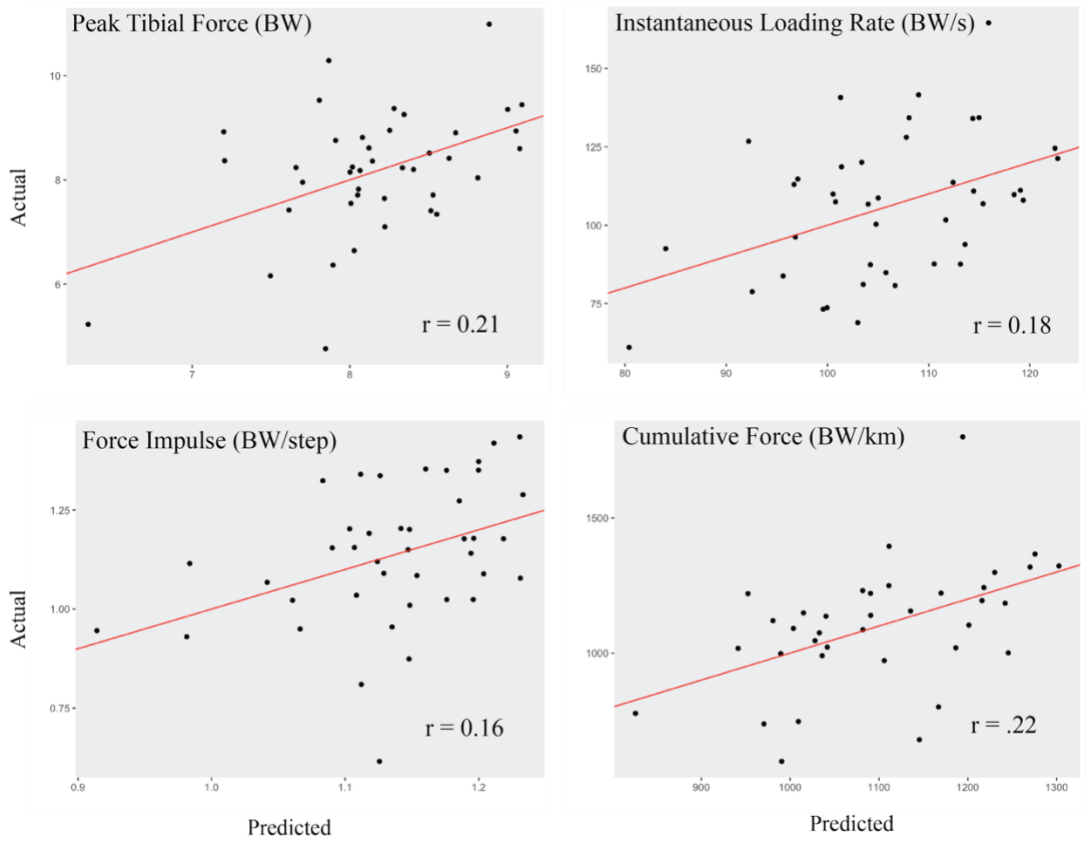


Figure 4.4 Predicted versus actual values of tibial bone load during running with regression line (red), and Pearson's *r* indicated on plots.

Discussion

This study investigated clinical measures of lower extremity strength, balance, and endurance as potential surrogates of mechanical tibial load during running. HHD, YBT, and SLHR were not predictors of peak tibial force, tibial ILR, impulse, or cumulative force per kilometer of running.

To add context to findings, peak tibial load estimates and clinical measures were compared with previously reported literature. Studies using similar modeling approaches have

reported tibial forces ranging between 6.74 - 9.80 BW, consistent with our mean peak force of 8.16 BW.[8–11] Similarly, this cohort averaged 25 SLHR repetitions and a YBT composite score of 95% leg length, consistent with reports from healthy adult cohorts of similar age and activity levels.[16,38] In contrast, HHD outcomes were lower compared to previously reported data: non-normalized plantarflexion strength averaged 152 N in our study, whereas previous reports ranged from 209-277 N.[15,27,32] Non-normalized dorsiflexion strength (~123 N) in this cohort was also lower than previously reported values of ~162 to 209 N. [27,32] These discrepancies may reflect differences in study populations, as prior studies included general healthy adults while our sample included only runners. Additionally, HHD measurements in prior studies were often non-normalized which may inflate values relative to the current study. While these comparisons suggest acceptable reliability of results in general, they do not explain the observed absence of association between clinical measures and tibial load variables.

The lack of association is somewhat surprising given that the selection of clinical measures specifically assesses plantarflexor function, which is known to strongly influence tibial force.[40,44] While no association is plausible, these results may be influenced by model assumptions – particularly the assumption that plantarflexor muscle activity is the primary contributing factor to internal tibial bone load, while contributions from other musculature are ignored.[40] Models that include proximal muscle groups such as quadriceps and hamstrings[51–53] or models that rely on static optimization to estimate a complete set of distal muscle contributions[26] typically report higher magnitudes of peak tibial force. In this study, clinical measures were chosen specifically because of the model's simplistic reliance on plantarflexor force; including

additional musculature would likely complicate the rationale for use of particular clinical measures.

Although direct associations were not observed, measures of strength, balance, or endurance may still influence tibial load indirectly through gait mechanics. Ankle moments are directly related to Achilles and tibial forces, therefore any variable that influences ankle moments during running may also influence tibial force. Consistent with this, peak ankle moments were highly correlated with peak tibial force (Pearson's $r = 0.91$). Running gait kinematics or mechanics such as stride length or strike pattern, may alter ankle moments and therefore tibial force. Previous studies have shown that manipulating step length, strike pattern, or cadence have resulted in changes in tibial load, supporting the idea that gait mechanics may provide alternative yet clinically accessible measures as surrogates for tibial force.[26,53,54]

Results from this study should be interpreted in context of several methodological considerations. First, the study was slightly underpowered due to data loss and removal of influential points. However, a *post-hoc* power analysis determined a minimal detectable effect size of 0.33 ($\alpha = 0.05$, $\beta = 0.80$) was possible after removal of the two predictors and five participants. Therefore, the final sample size remained large enough to capture a moderate to large effect size if it existed. Alternatively, a moderate or small effect size is possible however would not have been detected with the current sample size. Another limitation is the use of a generic model. While our results were consistent with previously reported results on peak tibial forces, a participant-specific model may provide further insight on the associations between tibial bone load and clinical measures. Additionally, validation of tibial force is challenging as *in*

vivo methods for the direct measurement of tibial force have been conducted in populations inconsistent with a healthy, young, running population.[55,56]

Finally, speed and footstrike can contribute to variability in tibial load outcomes yet were not adjusted for within this study. Higher running speeds have been consistently shown to increase plantarflexor muscle activity and tibial bone force.[8,11,57–59] In a *post-hoc* analysis, adding speed to the regression model for peak tibial force did not substantially influence the model significance ($F(35) = 2.38$, $p = 0.06$, Adj. $R^2 – 0.15$). Strike pattern may also influence joint moments and, consequently, Achilles and tibial forces. Almonroeder, et al. (2013) reported similar values between midfoot and rearfoot strikers supporting the inclusion of both in a single cohort.[12,40] However, other reports suggest that non-rearfoot strikers experience higher Achilles loads and calf muscle activation, making strike pattern a potential limitation in interpreting results.[60–62]

Conclusion

Clinical measures of ankle strength, dynamic balance, or plantarflexor endurance cannot be used as surrogates for tibial bone force or other metrics of tibial load during running. Future studies should investigate gait mechanics that influence ankle moments and/or plantarflexor muscle activity during running as potential surrogates of internal tibial loads. Continued investigation of clinically assessable measures as potential surrogates for tibial load will add to our understanding of BSI risk factors and clinical interventions to reduce the risk of recurrent injury.

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Statements and Declarations

The authors have no conflicts of interest to disclose.

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CHAPTER FIVE

ASSOCIATIONS BETWEEN TIBIAL BONE LOAD AND CLINICAL MEASURES IN THE INITIAL RETURN TO RUN PHASE FOLLOWING TIBIAL BONE STRESS INJURY

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Abstract

Introduction: Tibial bone stress injuries (BSI) are common running related injuries with a one in five recurrence rate. A suggested factor for recurrent BSI is elevated tibial loads in the early return to run phase. Altered strength, balance, and endurance post injury have potential to influence running mechanics and therefore contribute to variations in tibial load during running. Purpose: The purpose of this study was to compare tibial forces and clinical measures between runners who have recently returned to running following tibial BSI and healthy controls. Methods: Five tibial BSI (4F, 1M, 24.20 ± 7.00 years, 63.53 ± 10.53 kg, 1.70 ± 8.38 m, speed 2.88 ± .60 m/s) with an average of 2.60 weeks since returning to running were matched for sex, mass, height, and self-selected treadmill speed to three healthy controls. Participants ran on an instrumented treadmill for five minutes while kinematic and kinetic data were collected. Clinical measures of ankle strength via hand-held dynamometry (HHD), dynamic balance via the Y-Balance Test (YBT), and plantarflexor endurance via the single leg heel raise (SLHR) were also collected. Ankle joint angles, reaction forces, and moments were used as inputs to a musculoskeletal model to calculate tibial axial force as the sum of Achilles tendon and vertical ankle joint reaction forces. Instantaneous loading rate (ILR), impulse, and cumulative load over 1 km were calculated from tibial force and used as outcome variables. Independent and outcome variables between healthy and injured groups were compared with Welch's t-test or Wilcoxon Signed-Rank test. Results: Injured runners demonstrated less HHD-plantarflexion (p = 0.02) and plantarflexor endurance (p = 0.04) and greater impulse (p = 0.02), and cumulative load (p = 0.03) as compared to healthy runners. Conclusion: Greater tibial load and decreased measures of

strength and endurance were observed in runners who have recently returned to running after tibial BSI compared to healthy controls.

Introduction

Bone stress injuries (BSI) are common overuse injuries in endurance athletes with up to 21% yearly incidence in competitive track and field athletes.[1] Unfortunately, one in five runners diagnosed with BSI suffer from a subsequent BSI.[1,2] The elevated risk for recurrent injury after initial diagnosis suggests that risk factors related to the original BSI are not properly addressed prior to returning to sport. Additionally, the return to run decision is often predicated on symptom presentation rather than objective assessment.[3,4] Identifying risk factors that can be objectively measured may inform readiness to return to run, and overlooking persistent deficits may perpetuate the risk of recurrent BSI.

Biomechanical variables such as loading rate, tibial shock, free moment, and tibial bone load have been reported as risk factors for tibial BSI.[5–7] Specifically, Meardon et al. (2015) reported an association between elevated peak tibial bone load and a history of tibial BSI in runners. [5] Although this measure is objective, estimations of tibial bone load during running are clinically inaccessible due to equipment and modeling requirements. This challenges the practical utility of the relationship between bone load and BSI risk unless a surrogate measure for tibial bone load is available.

Surrogate measures for tibial bone load must be established to identify runners with elevated tibial load who may be at increased risk for recurrent injury. Measures that specifically assess plantarflexor activity are particularly important as the plantarflexors contribute up to 80% of internal tibial bone load during running.[8] Clinical assessments of plantarflexor strength and endurance include hand-held dynamometry (HHD) and the single leg heel raise (SLHR), both of which are direct assessments of plantarflexor function.[9,10] The plantarflexors are also an integral

component of dynamic balance, commonly assessed with the Y-Balance Test (YBT).[11,12] While these measures all have potential as surrogates of tibial bone load, their relationship to elevated tibial load in an injured population remains unknown.

Clinical assessments may serve an additional role in determining readiness for return to running after BSI diagnosis. Few objective tests or measures exist to guide the decision-making process to return to running.[3,4] Yet, deficits in strength, balance, or endurance have potential to contribute to tissue fatigue and elevated tibial loads during running.[13–15] Identifying runners with deficits in strength, endurance, or dynamic balance is critical as these deficits may ultimately increase the risk of recurrent BSI.

Finally, it is unknown if elevated tibial bone load is present in the early return to run phase or if it develops over time. The early return to run period may be a critical time frame in which to identify elevated loads or functional deficits that may contribute to elevated loads. However, biomechanical studies on load variables as risk factors report an average of 2-5 years from tibial BSI diagnosis to data collection.[6,7,16] Additionally, strength, endurance, and balance metrics have yet to be described in the early return to run phase after tibial BSI. Because tibial bone load and clinical measures of strength, endurance, and dynamic balance have a plausible relationship, identifying any associations in the early return to run phase would be a logical step to reduce the risk of recurrent BSI.

Clinical assessments of plantarflexor strength and endurance and dynamic balance have potential as surrogate measures of tibial bone load. However, no studies have examined these clinical measures or tibial bone loads between healthy runners and previously injured runners in the early return to run phase, nor have studies examined associations between clinical measure

and bone load across cohorts. Therefore, the purpose of this study is threefold: 1) to compare clinical measures of strength, dynamic balance, and endurance; 2) to identify differences in tibial bone load variables; and 3) to investigate relationships between the aforementioned clinical measures and tibial bone load between runners who have recently returned to running after diagnosis of tibial BSI and a healthy cohort of runners.

Methods

This cross-sectional study was approved by the Montana State University Institutional Review Board (#2022-137) and all participants provided written informed consent prior to participation. Local medical and rehabilitation clinics, high school and collegiate cross-country teams, and local running clubs were informed of the study to solicit for potential participants. Runners between the ages of 14-35 who had been diagnosed with a tibial BSI by an athletic trainer, physiotherapist, or physician *and* was within the initial four weeks of returning to running after diagnosis were invited to participant in the study. Healthy runners between the ages of 14-35 who were free of lower extremity injury for the six months prior to data collection, and who ran at least twice per week were also solicited for the study. Healthy participants were excluded if they reported a history of BSI in the previous year and any participant was excluded if unable to run continuously for five minutes. Healthy runners were matched to injured runners based on sex, height, mass, and treadmill running speed.

Anthropometric data was collected upon arrival to the laboratory, and each participant was fit with a neutral running shoe (New Balance, Boston, MA) provided by the lab. Retroreflective markers were placed on bilateral acromion, iliac crests, anterior superior iliac spines, posterior superior iliac spines, medial and lateral tibiofemoral joint lines, medial and

lateral malleoli, superior and inferior calcaneus, lateral calcaneus, base of 5th metatarsal, dorsum of the foot at the 2nd metatarsal head, and navicular as well as tracking marker sets secured on each thigh and shank. A standing static calibration trial was collected for 30 seconds after which participants performed a five-minute walking warm-up at 1.12 m/s. The warm-up was immediately followed by a five-minute run on an instrumented treadmill (Treadmetrix, Park City, UT) at a self-selected pace (average 2.87 ± 0.47 m/s) defined as an easy 10-k pace. Motion capture data was collected during the final 30 seconds of running using a seven-camera motion capture system (MotionAnalysis, Rohnert Park, CA) at 240 Hz and synchronized ground reaction force data at 1200 Hz.

After the running trial, data was collected from three clinical measures including ankle strength assessed with HHD, dynamic balance assessed with YBT, and plantarflexor endurance assessed with a SLHR (Figure 5.1). Isometric ankle strength for dorsiflexion, plantarflexion, inversion, and eversion was assessed with the MicroFet3 HHD (Hoggan Health Industries, West Jordan, UT). HHD has been reported to have good to excellent reliability and moderate validity for ankle strength measurements.[17,18] Based on previously reported methods, participants were positioned in long sitting with slight knee flexion while a practitioner experienced in HHD stabilized the lower shank with one hand and positioned the HHD in contact with the participant with the opposite hand. Manual resistance was applied as counterforce for each movement without overpowering the participant.[19–21] HHD test positions were assessed in random order, and each position was tested three times. For each test, participants were instructed to increase effort for three seconds followed by a three-second hold of maximum effort. Maximum values were recorded in Newtons, averaged over three trials, and normalized to body mass.

The modified YBT was used to assess dynamic balance using the Y-Balance Test Kit (Functional Movement Systems, Chatham, VA, USA) which has been found to be a valid tool for dynamic balance assessment with excellent inter- and intrarater reliability in healthy adults in all directions.[22–24] A full description of YBT procedure is described in previous studies.[25,26] Briefly, participants stood on the test leg in socks or barefoot on a central block with hands on hips. Participants were instructed to reach as far as possible with the opposite limb in three directions: anterior, posteromedial, and posterolateral. Two rounds of three trials per direction were completed per limb, recorded to the nearest 0.05 cm, and averaged across trials and rounds. A composite score was calculated by summing the average reach of each direction and dividing the sum by 3-times leg length and multiplying the final score by 100. Leg length was defined as the distance in centimeters from the anterior superior iliac spine to the medial malleolus.[26]

The final clinical measure assessed was the SLHR, reported to have good test-retest reliability and reserved for last to avoid any adverse effects of calf fatigue on the other clinical assessments.[27] SLHR test procedure has been described previously and was as follows.[10,27,28] Participants stood on a randomized single leg on a 10-degree incline board to begin the test procedure in ankle dorsiflexion and the knee in full extension. Balance was maintained by allowing participants to touch two to three fingertips on a wall with instructions to avoid weight bearing through the hands. A single heel raise was initially performed to familiarize participants with proper technique and to determine calcaneal height at full plantarflexion; height was marked with a thin piece of tape at the calf and on nearby equipment. For the test, single-leg heel raises were performed at the tempo of one beat up and one beat down to a metronome set at 60 beats per minute. Participants were instructed to perform as many repetitions as possible to

fatigue which was determined as inability to maintain tempo, inability to achieve full height, attempting to initiation the heel raise with knee flexion, or subjective fatigue. When fatigue was observed, participants were provided a single cue to correct the movement after which the test was stopped once a participant was no longer able to maintain proper technique. After the initial test, the procedure was repeated on the opposite limb.

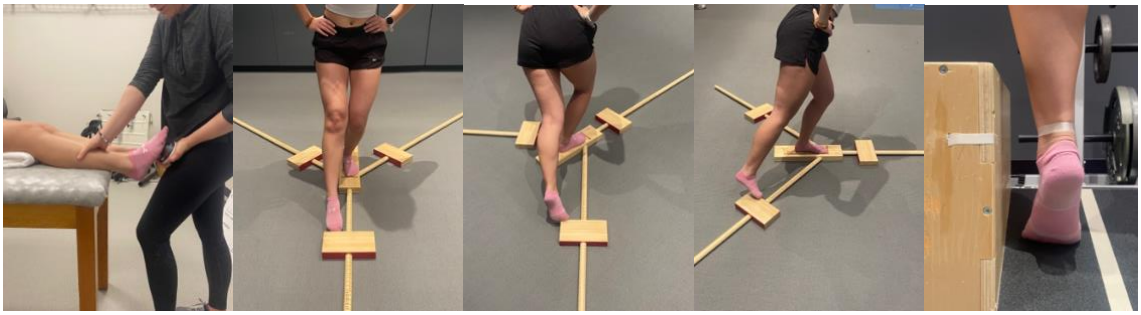


Figure 5.1 Examples of clinical measures from left to right: HHD plantarflexion, YBT-anterior, YBT-posterolateral, YBT-posteromedial, SLHR

Data Processing

Three-dimensional marker trajectories for static and dynamic trials were processed in Cortex (Motion Analysis Corporation, Rohnert Park, California, USA), after which the static trial was used to create an eight-segment rigid body biomechanical model in Visual 3D (v2026.01.1 HAS Motion, Germantown, MD, USA). Kinematic and force data were filtered with a zero-lag fourth-order low pass Butterworth filter at six and 25 Hz, respectively. Segment definitions were as follows: the thorax was defined by markers on right and left acromion as well as right and left anterior superior iliac spines; the pelvis was identified with markers at the right and left anterior superior iliac spines and a virtual marker midway between right and left posterior superior iliac

spines; thigh segments were identified proximally by a virtual hip joint and distally by markers at the medial and lateral tibiofemoral joint line; shank segments were also identified proximally by the tibiofemoral joint line markers and distally by markers at the medial and lateral malleoli; and finally, the foot was identified by markers at the medial and lateral malleoli and the 2nd metatarsal head. Segment coordinate systems were defined by the +x-axis in an anterior orientation, the y-axis in a medial-lateral direction with the positive axis to the left, and the z-axis in an axial orientation with caudal to cranial designated as the positive direction.

A flexion/extension-abduction/adduction-axial rotation Cardan sequence was used to identify ankle joint angles while inverse dynamics was used to calculate ankle joint reaction forces and moments, reported in the shank coordinate system. Further analysis was performed only on the injured limb in the injured cohort and the matching limb in the healthy cohort. Stance phase was determined using a 40-N threshold to identify initial contact and toe-off;[15] and was normalized to 101 data points. Ten gait cycles from the dominant limb were input into a custom Matlab code (R2025b, The Mathworks, Inc., Natick, Massachusetts, USA) to calculate tibial load outcome variables.

Tibial load outcome variables included tibial axial force, tibial force instantaneous loading rate (ILR), tibial force impulse, and the cumulative tibial force over one kilometer were calculated as follows. Achilles tendon moment arm (ATMA) was estimated as a function of ankle angle (θ) used as input to a regression equation as reported in previous studies (Eq. 1).[8,29–33] Achilles force (F_A) was then calculated as the quotient of ankle moment (M_A) to ATMA (Eq. 2) and last, the axial component of ankle joint reaction force (JRF_A) was summed with Achilles force to estimate tibial axial force (F_T) (Eq. 3).[34–37]

$$ATMA\ (m) = (-0.5910 + 0.08297*\theta - 0.0002606*\theta^2) /100. \quad \text{Eq. 1}$$

$$F_A\ (N) = \frac{M_A}{ATMA} \quad \text{Eq. 2}$$

$$F_T\ (N) = F_A + JRF_A \quad \text{Eq. 3}$$

Tibial force was normalized to body weight (BW) and applied to calculations of vertical instantaneous loading rate (ILR), impulse, and cumulative load over one kilometer (CL). Vertical ILR was defined as the maximum slope from 20-80% of the most linear portion of the tibial force curve from the initial rise to the main peak.[15,38] Impulse was calculated at the total area under the tibial axial force curve using a trapezoid method to calculate the approximate integral of tibial force across stance.[39] CL was then calculated as the impulse multiplied by the number of steps per kilometer.[34] Steps per kilometer was determine by first calculating step time as the time from strike to the next strike, multiplying treadmill speed by step time to gain step length, and finally calculating the number of steps per kilometer and multiplying the steps per kilometer by tibial force impulse.

Statistical Analysis

Statistical analysis was conducted in RStudio (v2026.01.01.+39, Posit team, Boston, MA). Shapiro Wilks test for normality was performed for anthropometrics, independent, and outcome variables. Welch's t-test was used to identify differences between group means when normal distribution was present and under the assumption of unequal variance, whereas

Wilcoxon Signed-Rank test was used when normal distribution was not evident. Hedge's *g* described effect sizes between groups with .2 or less indicating a small effect size, .5 a moderate effect size, and .8 or greater a large effect size.[40] Statistical parametric mapping (SPM) was performed with open-source Matlab code [41] to assess temporal differences in forces between injured and healthy runners across the stance phase of running. Statistical test significance was set at $\alpha = .05$.

Results

No violations of normality for demographic variables nor any significant differences in demographics were found between injured and healthy groups. Mean and standard deviation of healthy and injured group demographics are described in Table 4.1. The average time from initiating a return to run program to data collection for injured participants was 2.6 weeks (range 1-4 weeks).

Results for independent and outcome variables are described in Table 4.2 and Figure 5.2. All variables were normally distributed except for peak tibial force; therefore, Wilcoxon's Signed Rank Test statistic (W) was reported for the comparison of peak tibial force values between groups. HHD-plantarflexion was significantly larger in the healthy group by .60 N/kg (95% CI: 0.18-1.02) as was number of SLHR repetitions by 3.6 repetitions (95% CI: 0.25-6.95). For the outcome variables, significantly increased impulse was observed in injured runners by 0.24 BW/s (95% CI: 0.06-0.41) and CL by 218.51 BW/km (95% CI: 26.89-410.12). No other comparisons demonstrated significance between healthy and injured groups.

SPM results are illustrated in Figure 5.3. Mean tibial force curves during running differed between 0-2.3% and 60-97% of stance between injured and healthy groups.

Table 4.1 Mean demographics ± SD

	Healthy (n = 5)	Injured (n = 5)	Welch's *p* value
Sex	1 M, 4 F	1 M, 4 F	-
Age (years)	25.50 ± 5.57	24.20 ± 7.00	0.77
Height (m)	1.73 ± 5.72	1.70 ± 8.38	0.71
Mass (kg)	65.21 ± 11.24	63.53 ± 10.53	0.85
BMI (kg/m^2)	21.74 ± 2.47	21.66 ± 1.81	0.99
Run volume (km/week)	28.97 ± 13.07	19.31 ± 14.34	0.25
Treadmill speed (m/s)	2.78 ± 0.29	2.88 ± 0.60	0.91

*Indicates a significant difference between groups ($p < .05$)

Table 4.2 Mean ± SD for clinical measures and tibial load variables

	Injured	Healthy			
	Mean ± SD	Mean ± SD	T	*p*	*g*
HHD Dorsiflexion (N/kg)	1.60 ± 0.40	1.72 ± 0.34	0.50	0.63	0.32
HHD Plantarflexion (N/kg)	2.11 ± 0.60	2.71 ± 0.63	3.99	0.02*	0.98
HHD Inversion (N/kg)	1.21 ± 0.13	1.38 ± 0.30	1.24	0.28	0.73
HHD Eversion (N/kg)	1.30 ± 0.14	1.41 ± 0.35	0.84	0.45	0.41
Y-Balance (% leg length)	0.89 ± 0.06	0.95 ± 0.11	1.64	0.18	0.67
Single leg heel raise (# reps)	20.40 ± 3.20	24.00 ± 4.64	2.98	0.04*	0.90
Tibial loading rate (BW/s)	101.40 ±13.57	104.57 ± 48.91	0.15	0.89	0.09
Impulse (BW/s)	1.19 ± 0.12	0.95 ± 0.03	3.79	0.02*	2.74
Cumulative load (BW/km)	1219.09 ± 233.67	1000.58 ± 101.06	3.17	0.03*	1.21
			W		
Peak tibial axial force (BW)	8.60 ±1.22	7.23 ± 0.95	0	0.06	1.25

*Indicates a significant difference between groups ($p < .05$)

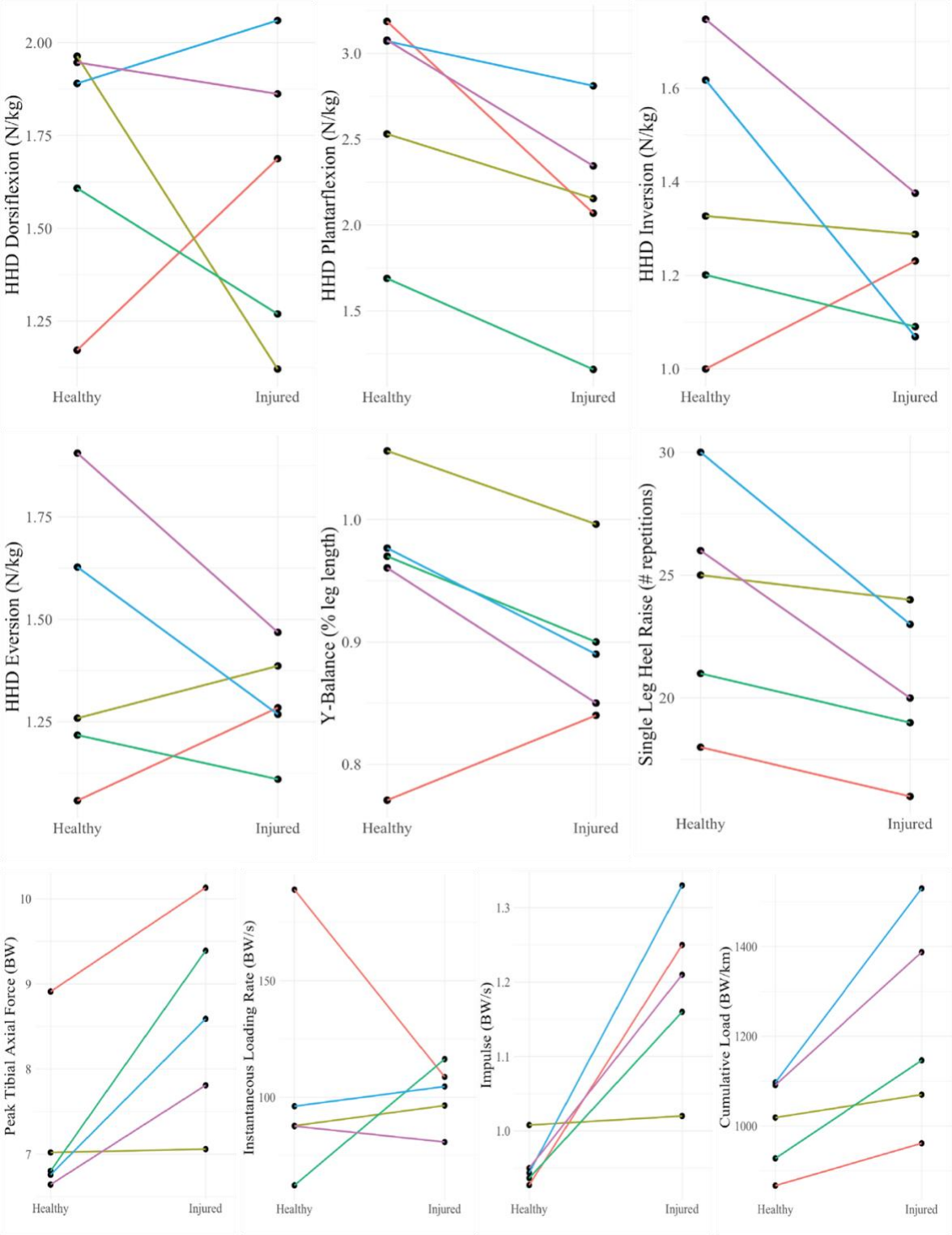


Figure 5.2 Clinical measures and tibial load variables for each injured participant and the healthy control. *Indicates significant differences between groups ($p < .05$).

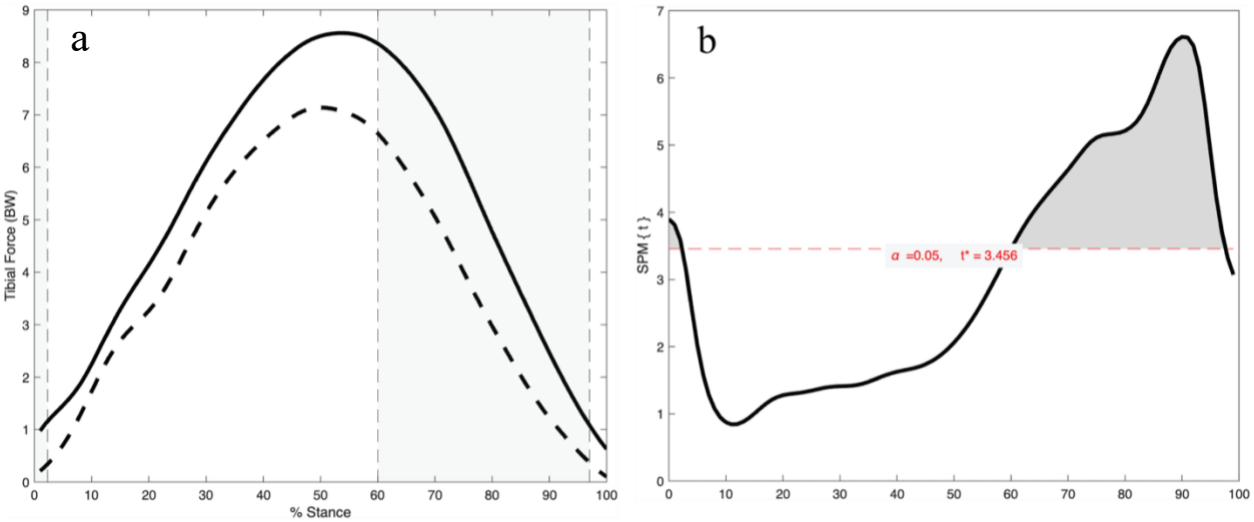


Figure 5.3 Statistical parametric mapping results for tibial force between injured and healthy groups a) Mean trajectories for tibial force during the stance phase of running in injured (solid) and healthy (dashed) runners. Time frames during which the curves were significantly different is noted in a vertical grey band. b) A paired t-test statistics {t}. The critical threshold 3.46 (red dashed line) was exceeded from 0-2.3% and 60-97% stance with a probability values of p=0.05 indicating significantly greater tibial force.

Discussion

Ankle strength, dynamic balance, plantarflexor endurance and tibial bone load in the early return to run phase following diagnosis of tibial BSI are not well characterized. Preliminary findings indicate that runners who have recently returned to running after tibial BSI demonstrate reduced plantarflexor strength and endurance, along with elevated tibial impulse and cumulative load, compared to healthy runners.

Deficits in plantarflexor strength and endurance among injured runners may have important implications for tibial bone load during running. Healthy runners produced ~0.6 N/kg greater plantarflexor force and completed, on average, four more SLHR repetitions compared to

injured runners. These differences suggest a potential mechanism for the relationship between elevated tibial load and history of BSI. Although plantarflexor function is not directly related to tibial load within the biomechanical model, these deficits may alter gait mechanics, creating an indirect relationship between clinical measures and tibial load variables.

The biomechanical model used within this study defines a direct relationship between ankle moments, plantarflexor force, and tibial force. The observed increases in tibial impulse and cumulative load and a trend toward elevated tibial force seen in injured runners would be expected to result from greater ankle joint moments, joint reaction forces, or a combination of both. A *post-hoc* paired t-test revealed no difference in peak ankle joint reaction forces between groups ($T(4) = 0.02$, $p = 0.98$, $g = 0$) suggesting that joint reaction forces did not contribute to between-group differences in tibial load. In contrast, peak ankle moments trended higher in injured runners, with a large effect size ($T(4) = 2.38$, $p = 0.07$, $g = 1.06$), indicating that increased ankle moments may underlie elevated tibial loads.

Peak moments occurred during mid-stance, indicating that any gait deviations contributing to elevated moments and tibial force would logically occur during this phase. During early to mid-stance, the plantarflexors act eccentrically to control forward rotation of the tibia over the ankle joint.[42,43] Deficits in the plantarflexor strength or endurance may impair this control, resulting in greater anterior tibial progression. This alteration would shift the center of mass anteriorly and consequently the center of pressure forward, increasing the external dorsiflexion moment by the time the runner reaches mid-stance.[44] This increase in dorsiflexion moment would result in an elevated plantaflexor moment and therefore increased peak tibial forces during mid-stance.[49.50] Additionally, reduced plantarflexor capacity may delay heel rise,

prolonging elevated moments into late stance. This is supported by SPM findings demonstrating greater tibial forces in injured runners from mid to late stance.

An alternative explanation is that injured runners modify gait mechanics in an attempt to offload symptomatic tissue. A common strategy to offload tissue while running is to transition from a rearfoot strike to a non-rearfoot strike, which can reduce the impact peak and loading rate of ground reaction forces.[45–47] Similar to the mechanism described in the previous example, a non-rearfoot strike shifts the center of pressure anteriorly on the foot, inadvertently increasing the external dorsiflexion moment at the ankle.[48] This results in a concomitant increase in internal plantarflexor moment and, consequently, increased tibial bone force.[49,50] This mechanism is consistent with recent findings from Han, et al. (2025) who reported increased muscle activity and resultant tibial load with a forefoot strike during running as compared to a rearfoot strike.[51]

Both mechanisms would result in increased ankle moments, and consequently elevated Achilles and tibial forces. While each explanation is plausible, the initial hypothesis is grounded in findings of strength and endurance deficits and is more directly supported by the findings of this study. Further investigation with additional participants is necessary to confirm these mechanisms and to better characterize the influence of gait mechanics on tibial load.

Results of this study are preliminary and must be interpreted with extreme caution due to limitations with respect to a small sample size and ensuing variability. Definitive conclusions regarding surrogate measures of tibial bone load in this injured cohort cannot yet be identified as the sample size is prohibitive for statistical analysis of clinical measures as predictors of tibial load. While early results generate some important considerations, the small sample size will likely inflate variability which may contribute to erroneous conclusions. Additionally, the

retrospective nature of this study limits the ability to determine whether observed deficits preceded or resulted from the development of BSI.

The results of this study will be refined as additional data is collected. Assessing clinical measures of strength and endurance prior to the return to run phase have potential to identify key areas of focus for rehabilitation programs and reduce the risk of recurrent injury. Observed deficits in injured runners suggest a plausible mechanism for elevated tibial bone loads and if these patterns persist, they may also increase the risk of recurrent BSI. Further data collection is needed to clarify the relationship between clinical measures and tibial load during the early return to run phase following BSI.

Conclusion

Preliminary results suggest that runners who have recently returned to running after tibial BSI exhibit decreased plantarflexor strength and endurance, along with increased impulse and cumulative load while running as compared to healthy runners. These findings should be interpreted with caution due to the small sample size; however, observed deficits in clinical measures may help inform our understanding on gait deviations that contribute to elevated tibial load.

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Statements and Declarations

The authors have no conflicts of interest to disclose.

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CHAPTER SIX

DISCUSSION AND CONCLUSIONS

Bone stress injuries (BSI) are common overuse injuries in an endurance running population with a high risk of recurrence.[1–3] Elevated magnitude of mechanical load on the tibia associated with repetitive impact has been cited as a risk factor for tibial BSI.[4–7] Mechanical load cannot be quantified in a clinical setting; therefore, clinically accessible surrogate measures are needed to identify runners who may be at risk for BSI. Modeling approaches are sensitive to methodology and demonstrate high variability in outcomes of tibial load. This dissertation addresses these challenges by comparing tibial force outcomes across different modeling techniques and investigates the influence of Achilles tendon moment arm estimation techniques on tibial loads. Additionally, this dissertation addresses clinical measures as potential surrogates for tibial load in healthy recreational runners, and investigates differences in clinical measures and tibial loads during running among healthy and injured runners in the early return to run phase after diagnosis of tibial BSI.

In Chapter Two, a systematic review and meta-analysis of fifteen moderate- to high-quality articles was completed to identify differences in tibial force estimates across three modeling methodologies. Based on the findings, model method accounted for 50% of the variability in tibial force estimates. Tibial axial force was approximately 3.5 BW lower when estimated with a distal model compared to a model using static optimization. This review also confirmed substantial variability between studies, likely influenced by factors such as running

experience, speed, step frequency, surface, grade, and load carriage. Based on these findings, future studies should compare results only to those obtained with similar modeling methodologies and should also account for relevant confounding variables.

Additional methodology was explored in Chapter Three to identify differences in Achilles tendon moment arm lengths from three estimation approaches. Results from this study indicate that Achilles tendon moment arm estimates are method-dependent and have meaningful impact on Achilles and tibial forces. Differences in forces between methods were observed from mid- to terminal stance of running gait, and peak tibial forces differed by up to 22% across moment arm estimation techniques. This difference was meaningful, as previous studies have found as little as 13% difference in peak tibial load between runners with and without a history of tibial BSI.[8] What remained unclear was which method was most favorable for estimating Achilles tendon moment arm. Therefore, the assumptions and limitations of each method must be carefully considered when selecting an approach for estimating Achilles tendon moment arm.

Chapter Four transitioned from a methodological focus to a translational research focus. The purpose of this chapter was to address the inaccessibility of tibial bone load in a clinical setting by identifying relationships between clinical measures and tibial load. Measures of ankle strength, dynamic balance, or plantarflexor endurance were identified as potential surrogates for tibial bone load during running, as they are directly related to plantarflexor activity. However, results from this study indicate that these clinical measures cannot be used as surrogates for tibial force or other metrics of tibial load during running.

Chapter Five continued the investigation of clinical measures as potential surrogates for tibial bone load in an injured population. The novel aspect of this study was that the injured

runners were, on average, 2.6 weeks into their return to running program following diagnosis of a tibial BSI - a time frame which has not yet been addressed in the literature with respect to tibial load or clinical measures. The sample size, unfortunately, was too small to establish relationships between clinical measures and tibial load metrics. However, preliminary results suggested that runners who have recently returned to running after tibial BSI have measurable deficits in plantarflexor strength and endurance, as well as elevated impulse and cumulative loads. These deficits may provide insight into potential mechanisms for elevated tibial loads through alterations in gait mechanics during running. While these findings must be interpreted with caution, results may inform future studies investigating clinical measures associated with tibial loads.

This dissertation provides insight into tibial load modeling methodologies and their influence on peak force estimates and related tibial load metrics. Results from Chapters Two and Three highlight the need for transparent reporting of methodology, particularly for variables such as tibial force that are difficult to quantify *in vivo.* It is also important to acknowledge that variability in tibial bone load is not only influenced by the model but also by numerous confounding variables. The majority of studies examining the influence of these variables on tibial bone load have been conducted in healthy populations; therefore, additional work is necessary to understand their role in injured populations.

The lack of association between clinical measures and tibial load variables implies that measures of strength, balance, and endurance should not be used as surrogates of tibial bone load. Additionally, the heterogeneity identified in Chapter Two may have also influenced the results observed. If the participant population had been more homogenous with respect to sex,

running volume, running experience, speed, or footstrike, the results may have differed. Despite a lack of association among clinical measures and tibial load in a healthy cohort, preliminary results from the injured cohort suggest that clinical and tibial load variables may differ between healthy and injured populations in the early return to run phase. Therefore, further investigation is warranted to explore the relationship between clinical measures and tibial bone load in the early return to run period to identify runners at risk of recurrent BSI.

This dissertation highlights the complexities of modeling bone load to obtain clinically relevant outcomes, with the overarching goal of contributing to our understanding of risk factors for BSI. Future studies should continue to address tibial load estimation methodologies and relationships between clinical measures and tibial load variables in injured running populations. A direct comparison of tibial force estimation methods is warranted to further our understanding of the influence of model method on outcomes. Additionally, continued data collection in injured participants will refine the results on clinical measures in the early return to run phase, highlight potential mechanisms for recurrent BSI, and provide insight on areas of focus for interventions to reduce recurrent injury risk. Future studies should also investigate gait mechanics during the early return to run phase to better understand their relationship with elevated tibial loads. By integrating tibial load modeling and clinically accessible tools, future research can enhance our understanding of BSI, guide strategies to identify runners at risk for BSI, and reduce the likelihood of recurrent injury.

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APPENDIX

MODEL RESIDUAL ASSUMPTIONS

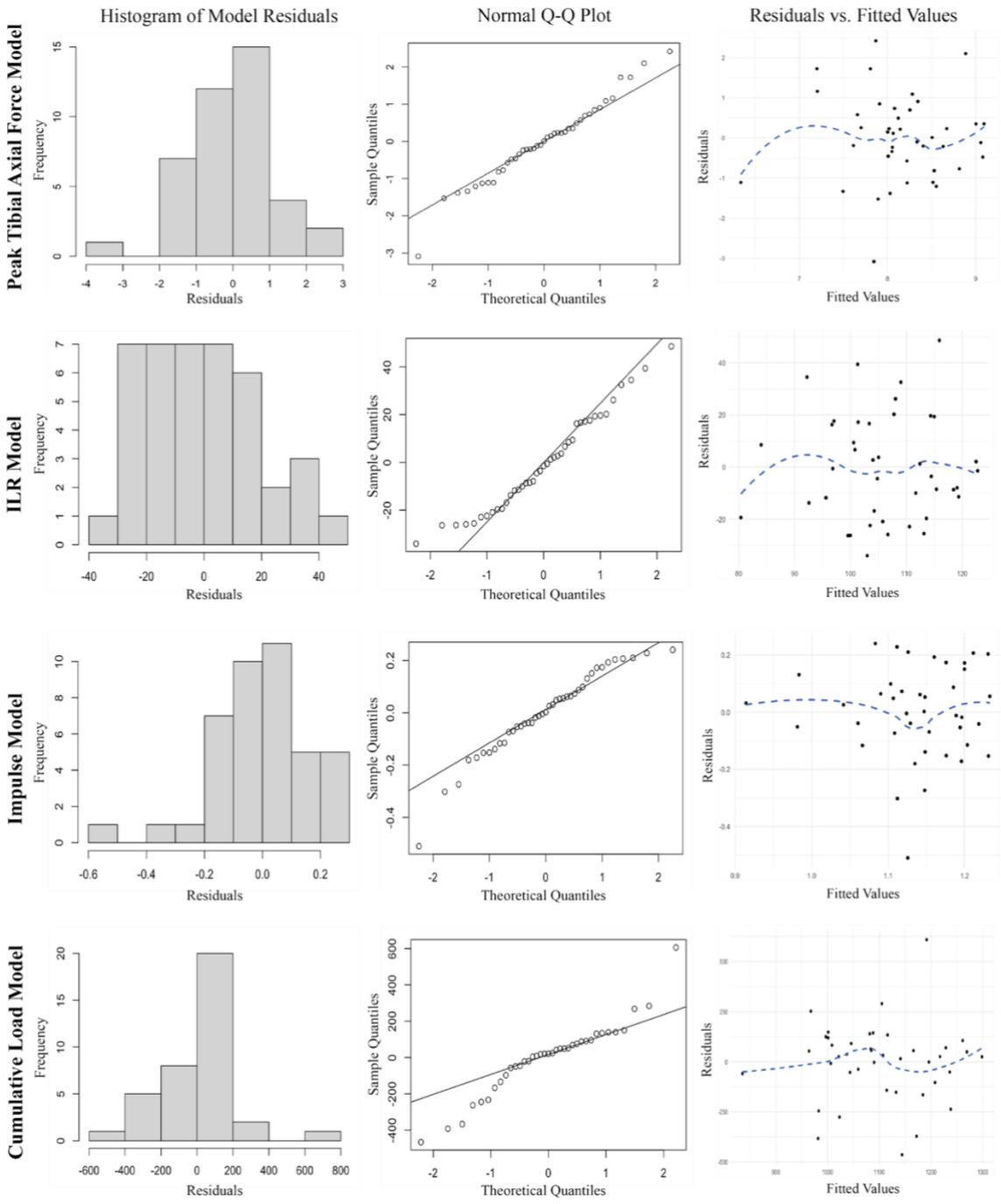


Figure 1. Histograms, Normal Q-Q plots, and Residuals vs. Fitten Values plots for individual models.