THE INFLUENCE OF ALTERING INTERNAL AND EXTERNAL FACTORS CONTRIBUTING TO METATARSOPHALANGEAL JOINT MECHANICS AND THEIR EFFECTS ON RUNNING ECONOMY

by

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DISSERTATION ABSTRACT

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The metatarsophalangeal joint serves as the base of support during running once the heel lifts off the ground. However, it is often neglected as a contributor to forward propulsion. Despite being historically overlooked, advances in footwear technology have shown that the use of stiffer footwear may benefit performance in distance runners. Little is known however about how mechanical function of the metatarsophalangeal joint changes in response to varying external and internal factors contributing to its function. To address this, we had well trained runners run at a range of speeds from perceived easy to mile race pace, strengthen their intrinsic foot muscles, and run in shoes of varying longitudinal bending stiffness. This dissertation was divided into two projects. The first project consisted of running at speeds ranging from 3.69 to 6.11 m/s on an instrumented treadmill. Participants also completed a ramped protocol consisting of three-minute stages while expiratory gases were collected, in order to assess running economy. Half of the participants underwent an intrinsic foot muscle training program for ten weeks while half did not. Participants revisited the lab at five and ten weeks to test for changes in foot strength, gait mechanics, and running economy. Results indicate that metatarsophalangeal

joint moment, range of motion, and dynamic angular resistance change across running speeds. Interestingly though, greater intrinsic foot muscle strength did not alter mechanics of the metatarsophalangeal or ankle joint or running economy. The second study consisted of having well trained runners run in shoes of varying longitudinal bending stiffness at 3.89, 4.70, and 5.56 m/s while joint and gross level mechanics were assessed. Running economy was assessed at 3.89 and 4.70 m/s. Results suggest that changes in joint and gross level mechanics and running economy in response to variable longitudinal bending stiffness of footwear is running speed dependent. Individual subject responses suggest that optimal tuning of the longitudinal bending stiffness of footwear should take speed into consideration. These findings may serve as a guide for how to construct footwear to improve performance in distance runners.

This dissertation includes previously published and unpublished co-authored material.

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CHAPTER I

INTRODUCTION

Background and Significance

The human foot is a complex structure with numerous joints, allowing for many degrees of freedom of movement. Movement of the foot about these joints can be initiated by active or passive mechanisms (Mckeon et al., 2015). The intrinsic foot muscles (IFM) are a group of muscle-tendon units that have origins and insertions within the foot. Primary intrinsic foot muscles are the abductor hallucis, flexor digitorum brevis, and quadratus plantae. The extrinsic foot muscles are a group whose muscle bellies originate in the shank and the tendons cross the ankle joint and insert onto the midfoot region. Primary extrinsic foot muscles are the flexor hallucis longus, peroneus longus, tibialis posterior, and flexor digitorum longus. Altogether, these muscles tend to work together in sequential order during locomotion (Zelik et al., 2015). Passive mechanisms within the foot include the various ligaments or other soft tissues providing resistance to deformation (Ker et al., 1987). The plantar fascia is the largest aponeurosis within the foot, spanning the ventral aspect of the foot with its origin on the calcaneus and crosses the metatarsophalangeal joint (MTPJ) with its insertion on the phalanges.

Within the foot are three different arches (Mckeon et al., 2015). The medial and lateral longitudinal arches and the transverse arch. The longitudinal arch is what is commonly referred to as the spring of the human foot (Ker et al., 1987; McMahon, 1987). When loaded, the longitudinally arch vertically compresses and returns strain energy stored in the various structures of the arch during the unloading phase. The windlass mechanism refers to the plantar fascia being pulled taught by dorsiflexion of the toes about the MTPJ. This mechanism causes the longitudinal arch to stiffen up by increasing tension in the plantar fascia. Dorsiflexion about the MTPJ also influences a repositioning of joints proximal to the metatarsals that result in rotation occurring in all three planes of movement, as opposed to only in the sagittal plane if the MTPJ is plantar flexed (Welte et al., 2018). Elasticity of the plantar fascia helps with weight acceptance during the first half of the stance, while the taught aponeurosis then helps the foot become a stiff lever for push-off during the second half of stance (Ker et al., 1987). Energy stored in the plantar fascia is recoiled proximally to contribute to helping the longitudinal arch shorten during push-off (McDonald et al., 2016; Wager and Challis, 2016). The stiffening of the foot is a primary mechanism of reducing energetic cost during locomotion (Song et al., 2013).

It has been long commonly thought that mechanical function of the longitudinal arch throughout stance was primarily passively modulated by the windlass mechanism of the plantar fascia. There is a growing body of evidence however that the intrinsic foot muscles (IFMs) contribute to longitudinal arch structure and function. The first display of the IFMs being active during the propulsion phase of stance was an investigation of their electromyographic activity during different walking tasks (Mann and Inman, 1964). This was the first evidence of active contributions of the IFMs being coupled with joint movements and overall pronation. A main takeaway, however, was that IFMs were not required to actively support the loaded foot at rest. This suggested that perhaps the passive mechanisms still dominate control of foot stiffness.

More recently the case for the importance of IFMs to foot posture and control has been gaining evidence. Mechanical behavior of the foot is that of a viscous spring-damper, in that its ability to store and return energy is running velocity dependent (Kelly et al., 2018b). Electromyographic activity of the IFMs has been shown to increase with increased static postural demand (Kelly et al., 2012). Standing balance performance is also directly influenced by activity and strength of the IFM (Lynn et al., 2012; Wallace et al., 2018). Additionally, the IFMs play a major role in controlling deformation of the longitudinal arch under static loads greater than body weight (Kelly et al., 2014). This control is primarily a result of isometric contraction of the IFMS to contribute to elastic energy storage and return (Kelly et al., 2018a). During walking and running, the IFMs linearly increase in activation with gait speed (Kelly et al., 2015). This increase in activation of the IFMs with running speed suggests that the IFMs work in parallel with the plantar aponeurosis to facilitate energy transfers within the foot and stiffen the longitudinal arch for push-off. Most importantly, the IFMs have been demonstrated to serve a role in contributing positive work to tasks that require net work performed on the entire body (Riddick et al., 2019). In light of this gaining evidence of the importance of the IFMs, when active control is knocked out via tibial nerve block, there is no change in longitudinal arch compression during the loading phase of walking and running (Farris et al., 2019). Interestingly though, the nerve block results in a decrease in MTPJ moment throughout the push-off phase of stance, resulting in a decrease in MTPJ stiffness. Altogether these results support the notion that while the compression of the longitudinal arch is primarily modulated by passive mechanisms, the IFMs play an important role in the push-off phase of walking and running for the foot to effectively function as a stiff lever to transmit force to the ground (Farris et al., 2019).

Foot morphology may be an important determinant in athletic performance. For sprinting, long toes, whose mechanics are modulated by the IFMs, are advantageous (Lee and Piazza, 2009; Tanaka et al., 2017). Long toes increase the propulsive impulse by increasing the ground reaction force lever arm. Additionally, this change in gear ratio influences ankle plantar flexor force-length and force-velocity operating points (Lee and Piazza, 2009). Contrarily, short toes may be advantageous for distance running. Shorter toes result in less energy dissipation and required mechanical work to run at a given speed (Rolian et al., 2009). Across the duration of an entire run, the small decrease in wasted energy with short toes compared to long toes may be metabolically advantageous (Rolian et al., 2009).

Intrinsic foot muscle volume affects foot morphology. A large flexor digitorum brevis results in the second to fourth metatarsal heads to be raised, affecting transverse arch height (Nakayama et al., 2018). This same study reported that a large abductor hallucis lowers the first metatarsal head (Nakayama et al., 2018). Larger abductor hallucis and abductor digiti minimi muscles are also associated with greater arch height index (Holowka et al., 2018; Miller et al., 2014). Changes in foot morphology are also associated with increased IFM strength as well, notably a shortening of the longitudinal and transverse arch length, most likely due to increases in IFM volume, although not measured (Hashimoto and Sakuraba, 2014).

Training of the IFM can occur via a variety of stimuli. Barefoot locomotion has been extensively supported and argued as the most natural mode of walking and running

(Lieberman et al., 2010). The use of barefoot or minimal footwear substantially increases IFM volume and affects foot structure (Bruggemann and Potthast, 2005; Chen et al., 2016; Hollander et al., 2017; Holowka et al., 2018; Johnson et al., 2016; Miller et al., 2014; Mulligan and Cook, 2013; Potthast et al., 2005; Ridge et al., 2019; Zhang et al., 2018). It is important to note, however, that too quick of a transition to barefoot or minimal footwear can lead to detrimental alterations in mechanics such as increased instability (Ekizos et al., 2017) or loading rates (Willson et al., 2014) during gait. Longer term, a transition to minimal footwear may result in development of bone marrow edema in the foot (Ridge et al., 2013). A successful use of minimal footwear that does not result in injury but does improve IFM strength or fatigue resistance and change arch structure may be beneficial in facilitating a change in kinematics and kinetics associated with injury (Headlee et al., 2008; Williams et al., 2001) or in preventing plantar fasciitis (Cheung et al., 2015). It is worth noting, however, that a transition to minimalist shoes may also be associated with no changes in performance or gross level biomechanics (Fuller et al., 2018).

While it is very well demonstrated that the IFMs can substantially increase in volume and strength in response to use of minimalist of barefoot footwear, strength training is an effective method as well. Use of low-resistance high-repetition (Hashimoto and Sakuraba, 2014; Sulowska et al., 2019) or low-repetition high-resistance (Goldmann et al., 2013) training of the IFMs both substantially increase strength. Increased IFM strength increases sprint performance (Hashimoto and Sakuraba, 2014) and jumping performance (Goldmann et al., 2013; Hashimoto and Sakuraba, 2014). An improvement in functional performance outcomes may be a result of improved energy transfers through the leg (Sulowska et al., 2019). Increased IFM strength also may increase the effective foot length (Endo et al., 2002), defined as the distance the center of pressure travels anteriorly underneath the foot (Hansen et al., 2004). An increase in effective foot length, which may be facilitated by the ability of increased IFM strength to depress the toes into the ground (Endo et al., 2002), will alter gearing about the lower extremity joints and allow for more torque and impulse to be generated due to the increased ground reaction force moment arm (Goldmann and Brüggemann, 2012). Interestingly, however, the MTPJ and ankle moments did not change during walking or running in response to increased IFM strength (Goldmann et al., 2013). While joint level mechanics of the MTPJ and ankle did not change in response to increased IFM strength (Goldmann et al., 2013), little is known about how increased IFM strength may affect metabolic cost of transport or running mechanics across various speeds. Because the MTPJ serves as the base of support once the heel lifts off the ground, strengthening of the IFM may contribute beneficial changes to performance as the IFM contribute to whole body momentum during late stance (Goldmann and Brüggemann, 2012; Miyazaki and Yamamoto, 1993).

Mechanical function of the MTPJ can also be altered via stiffening of footwear. A large amount of energy is dissipated at the MTPJ during running (Stefanyshyn and Nigg, 1997). Stiffening of footwear, often via carbon fiber plates, reduces the energy lost at the MTPJ (Stefanyshyn and Nigg, 2000). Altering the longitudinal bending stiffness (LBS) of footwear influences both sprinting (Smith et al., 2016, 2014; Stefanyshyn, Darren J., Fusco, 2004; Willwacher et al., 2016) and distance running performance (Madden et al., 2015; Roy and Stefanyshyn, 2006). During steady-state running, increased LBS tends to decrease dorsiflexion and angular velocity and increase the joint moment of the MTPJ (Oh and Park, 2017; Willwacher et al., 2013). These results are most likely due to gearing effects about the lower extremity joints, resulting in longer ground reaction force lever arms (Willwacher et al., 2014). A complication of optimizing footwear LBS for performance though is the large amount of variation in individual responses. Factors such as body mass (Kleindienst et al., 2005; Roy and Stefanyshyn, 2006) or ankle plantar flexor strength (Willwacher et al., 2014, 2016) affect how an individual can make use of the increased LBS. Due to this variation, analysis of an optimal bending stiffness tends to divide the participant populations into responders and non-responders (Madden et al., 2015; Willwacher et al., 2014). For responders, an optimal LBS can improve running economy by approximately 1-3% (Madden et al., 2015; Oh and Park, 2017; Roy and Stefanyshyn, 2006). It is theorized that the gearing effects of increased LBS positively influence muscle-tendon properties such as force-velocity and force-length operating points (Madden et al., 2015; Takahashi et al., 2016), similar to that of increased toe-length (Lee and Piazza, 2009). A slowing of ankle plantar flexor velocity reduces the required force for muscle contraction, reducing metabolic cost (Roberts et al., 1998).

There is strong evidence that the IFM can significantly increase in strength and improve performance in tasks such as sprinting and jumping (Goldmann et al., 2013; Hashimoto and Sakuraba, 2014), and that increasing LBS improves positively influences running economy (Madden et al., 2015; Oh and Park, 2017; Roy and Stefanyshyn, 2006). Little is known, however, about how increased IFM strength influences mechanics and running economy across a range of speeds. Additionally, investigation of varying footwear LBS across a range of running speeds would be beneficial for a better understanding of how to tune LBS for optimal performance. Joint mechanics change with running speed (Jin and Hahn, 2018; Stefanyshyn and Nigg, 1998a) but little is known about how function of the MTPJ changes with running speed. Potential changes in mechanics may affect the interaction of the foot and shoe (Oleson et al., 2005).

General and Specific Aims

The overall goal of this dissertation was to investigate the influence of changing internal and external factors that modulate MTPJ mechanics across a range of running speeds. The first objective was to investigate how to quantify forefoot kinetics during stance phase, as there is no commonly accepted method. The second objective was to investigate how MTPJ mechanics change across running speeds. The third objective was to investigate how increased IFM strength affects joint and gross level mechanics and running economy across a range of running speeds. The fourth objective was to investigate how footwear of varying LBS influence joint and gross level mechanics across a range of running speeds. The anticipated outcomes of this project will enhance athletes, coaches, clinicians, and researchers' ability to understand the influence of MTPJ function on performance for distance runners. An investigation of how to quantify MTPJ kinetics would help understand how to compare findings between existing studies and how to most accurately represent *in vivo* mechanics. Findings from mapping out mechanical function of the MTPJ across running speeds would be beneficial in guiding how to construct performance footwear based upon changes in range of motion, joint torque, and stiffness. Results from IFM strengthening will be beneficial for anyone looking to improve distance running performance and understand how increased foot strength affects mechanics and running economy during steady-state running. Lastly, this dissertation seeks to understand if tuning footwear LBS is dependent upon running speed. These findings will be of interest to footwear researchers as well as athletes and coaches seeking to improve performance. The goals of this dissertation are addressed through five specific aims.

Specific Aim 1. To investigate how existing MTPJ joint center locations affect estimated joint moments during running. The secondary goal of this aim was to develop a mathematical model for estimating MTPJ joint moments that did not require use of a fixed joint center location. It was hypothesized that (1) MTPJ joint center definition would affect the onset timing and magnitude of the MTPJ moment; (2) use of non-fixed joint center would result in a smaller MTPJ moment throughout stance and earlier onset than other existing fixed joint center locations.

Specific Aim 2. To investigate how MTPJ mechanics, notably resistance to dorsiflexion, termed critical resistance, change across running speeds in well-trained distance runners. It was hypothesized that as running speed increased so would critical resistance.

Specific Aim 3. To investigate how an increase in IFM strength affects joint and gross level mechanics and running economy. Participants were divided into two groups, an experimental IFM strength training and a control group that maintained normal workout habits. It was hypothesized that (1) the experimental IFM training group would increase maximum IFM strength; (2) increased IFM strength would decrease MTPJ range of motion and ankle plantar flexion and MTPJ dorsiflexion velocity resulting in an increase in MTPJ angular resistance and MTPJ and ankle moments, and contact time; and (3) that running economy would improve for the experimental training group but not the control group.

Specific Aim 4. To identify how varying LBS affects gross and joint level mechanics and running economy across a range of speeds. It was hypothesized that (1) the influence of LBS on stride frequency, stride rate, contact time, and ground reaction forces would be running speed dependent; (2) changes in MTPJ mechanics in response to stiffening plates would be running speed dependent, and (3) the LBS resulting in the best running economy at a faster running speed would be stiffer than that at a slower running speed.

Organization of Dissertation

This dissertation is written in a journal style format, where chapters III-VII have been or will be submitted for publication to peer-reviewed journals. The following explains how these chapters fit together into a coherent body of work. A bridge paragraph explaining the flow of studies is included at the conclusion of Chapters III-VI.

The current chapter (Chapter I) provides existential background information regarding functional anatomy of the foot, the ability for internal structures to respond to training stimuli, and the effect of footwear. This chapter provides the case for the significance of this research and details how the questions were formulated and flow together. Chapter II will detail the methodology utilized for each study. Chapter III describes the creation of a novel mathematical model used to estimated MTPJ kinetics throughout this study.

This dissertation was comprised of two separate projects. Data from the first study were used for chapters III-V. Data from the second study were used for chapters VI and VII. In chapter IV, mechanics of the MTPJ across speeds was investigated. This chapter served two purposes, the first of which was to establish baseline data for which to compare strengthening of the IFM to. Secondly, it served to provide a framework of understanding how kinematics and kinetics of the MTPJ change across running speeds. The increase in MTPJ critical resistance served as rationale for the notion that an optimal footwear LBS may be running speed dependent. Chapter V served to provide a description of how joint and gross level mechanics and running economy changed in response to increased IFM strength. Insights from Chapter IV were used for Chapters VI and VII. Chapter VI investigated how gross level mechanics and running economy changed at two running speeds. Chapter VII served as the joint level mechanics analysis to help further understand the changes in gross level mechanics and running economy from Chapter VI. The final chapter, Chapter VIII, summarizes the notable findings and results of the overall body of work, providing a main take away message from this set of studies while acknowledging limitations and suggesting future directions for work in this area of research.

This dissertation includes co-authored work, some which has already been published in peer-reviewed journals. Chapter III has been published in *Journal of Biomechanics*. Chapter IV is currently under second review in *Human Movement Science*. Chapter V is currently under first review in *Journal of Sports Sciences*. Chapters

VI and VII will be submitted for publication to appropriate journals. For all work in this dissertation, Evan M. Day was the primary investigator, responsible for study design, data collection, analysis, interpretation, and dissemination. Michael E. Hahn, the co-author of all studies, advised on all aspects of this dissertation.

CHAPTER II

GENERAL METHODOLOGY

Subjects

To address Specific Aims 1-3 (Chapters III-V), twenty-three competitive distance runners were recruited, eight of which were female (Table 1). To be included, males had to have a 5000m best of under 18:00 (min:sec) and females under 20:00, no lower extremity injury for the previous six months, and running over 50 km/wk. To address Specific Aims 4-5 (Chapters VI-VII), ten well trained distance runners were recruited study (26 ± 6 years, 1.78 ± 0.04 m, 63.9 ± 4.0 kg, 101 ± 34 km/wk, 15:04 ± 0:38 (min:sec) 5000m personal best). Participants had to able to comfortably fit a male size 10 shoe, have a 5000m best under 16:00, no lower extremity injury for the previous six months, currently running of 50 km/wk, and exhibit steady-state metabolic cost of transport running at 17 km/hr. For Chapters III-VII written informed consent was obtained from subjects and study protocols were approved by the University of Oregon Institutional Review Board (IRB protocol #07272016.025 & 08232018.022).

Table 2.1. Participant characteristics (Mean ± SD) for Specific Aims 1-3.

Sex	Age (yr)	Height (cm)	Mass (kg)	Weekly Mileage (km)	5000m Best (min:sec)
Male	26 ± 10	179 ± 9	66 ± 9	85 ± 24	16:17 ± 0:56
Female	27 ± 7	166 ± 6	56 ± 6	69 ± 16	18:38 ± 1:16

Study Design and Experimental Protocol

Chapter III-IV

Participants ran on an instrumented treadmill (Bertec, Inc., Columbus, OH) at five different running speeds, 3.89, 4.44, 5.00, 5.56, and 6.11 m/s. Speeds were completed in an ascending order and run at for approximately thirty seconds each, with data collected during the last ten strides. Rest between conditions was self-selected. All participants wore the same neutral cushioned footwear (Brooks Launch 3) with window cut outs for direct placement of retro-reflective markers.

Chapter V

Participants visited the lab on three occasions, baseline, five weeks, and ten weeks. Each visit consisted of IFM strength testing, running mechanics analysis, and running economy analysis. A custom-built set-up was used to assess isometric IFM strength in two positions. Running mechanics analyses were completed in the same manner as Chapters III-IV. Running economy assessments were completed on a high-speed treadmill (Woodway, Waukesha, WI) set to 1% grade. Participants completed a ramped protocol beginning at 14 km/hr. Stages lasted three minutes and increased by 2 km/hr until the subjects were near 5000m race pace.

Chapter VI

Data collection occurred on two different days. The first day consisted of running mechanics analysis. Participants ran on an instrumented treadmill at three speeds, 3.89, 4.70, and 5.56 m/s. Footwear of varying LBS were worn, defined as normal, stiff, and very stiff. Speeds were completed in ascending order for each footwear condition. Each speed was run at for thirty seconds with data collected during the last ten seconds. Rest between conditions was self-selected. Order of footwear was randomized between participants.

The second day consisted of running economy assessments. Participants ran on a high-speed treadmill at two speeds, 3.89 and 4.70 m/s, while expiratory gases were analyzed. Each speed was run at six times for five minutes each trial, for a total of twelve trials. There was a five-minute rest in-between trials. The three footwear conditions from the first day were used. Each footwear condition was run in twice at each speed. Order of footwear within speed was randomized. All of the trials at 3.89 m/s were completed before the 4.70 m/s. After the second wearing each at shoe at each speed, participants completed a subjective comfort questionnaire assessing various components of the footwear.

Chapter VII

No additional data collection was performed for chapter VII. Running mechanics data from Chapter VI were used for Chapter VII.

Data Collection

Chapters III-IV

Participants were outfit with a bilateral marker set consisting of 41 retro-reflective markers defining nine segments (forefoot, rearfoot, shank, thigh, pelvis). Three-dimensional marker data were collected at 200 Hz using an 8-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA). Ground reaction force data were collected at 2000 Hz using the force-instrumented treadmill. All participants wore the same neutral cushioned footwear (Brooks Launch 3).

Raw marker coordinate data were filtered with a zero-lag low-pass fourth order Butterworth filter at 20 Hz. Ground reaction force data were filtered in a similar manner.

Chapter V

Participants performed maximum isometric IFM strength testing on a custom-built set-up. Strength data were collected using a strain gauge connected to a SparkFun HX711 (SparkFun Electronics, Niwot, CO) amplifier powered via an Arduino Mega2560 (Arduino, Somerville, MA) microcontroller sampling at 10 Hz.

Running mechanics analyses were completed in the same manner as described for Chapters III-IV.

Measures of VO2 and VCO2 were taken with an open-circuit expired-gas analysis system (Parvomedics TrueOne 2400, Sandy, UT) for running economy assessments on a high-speed treadmill (Woodway, Waukesha, WI). Participants wore the same neutral cushioned footwear as during the running mechanics analyses. A ramped protocol was completed, starting at 14 km/hr and progressing by 2 km/hr every three minutes until participants were at a pace similar to their 5000m race pace. Expired gases were averaged over the last ninety seconds of each three-minute stage. The VO2 values were normalized to body mass (ml/kg/min) to estimate running economy. Metabolic rate (W/kg) was quantified using the Brockway equation (Brockway, 1987).

Chapter VI

Participants were outfit with a unilateral marker set consisting of 26 retro-reflective markers placed on the right forefoot, rearfoot, shank, thigh, and pelvis. Three-dimensional marker coordinate data were collected at 200 Hz using an 8-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA). Ground reaction forces were collected at 1000 Hz while participants ran on a force instrumented treadmill (Bertec Inc., Columbus, OH). Marker coordinate and ground reaction force data were filtered with using a zero-lag low-pass fourth-order Butterworth filter at 20 Hz. For running economy assessments on the second day, participants ran on a high-speed treadmill (Woodway, Waukesha, WI). Expired gases were averaged over the last two minutes of each trial. Average expiratory gas values from each trial were then averaged across the two trials in each footwear condition at each speed. The VO2 values were normalized to body mass (ml/kg/min) to estimate running economy. Metabolic rate was quantified using the Brockway equation (Brockway, 1987).

Chapter VII

Data from Chapter VI were used in a unique analysis for Chapter VII.

Data and Statistical Analysis

Chapter III

To investigate the effect of joint center location on estimated joint moments throughout stance phase, four methods were compared. A two-dimensional (Stefanyshyn and Nigg, 1997), midpoint (Roy and Stefanyshyn, 2006), second metatarsal head

(Willwacher et al., 2013), and non-fixed joint center approach (Rolian et al., 2009) were used. A novel mathematical model to estimate MTPJ moments was created. This model is further detailed in Chapter III. All net joint moments were normalized to body mass.

The magnitude of MTPJ moment throughout stance phase was compared between joint center definitions using Statistical Parametric Mapping (Pataky et al., 2013, 2016) in MATLAB (version 2016b, Mathworks, Natick, MA). Follow up pairwise comparison analyses were conducted using Statistical Parametric Mapping t-tests with a Bonferroni corrected alpha level. Peak moments between joint center definitions were compared using a one-way repeated measures ANOVA ($\alpha = .05$) in SPSS (V23, IBM, Armonk, NY). Pairwise comparisons with Bonferroni adjustments ($\alpha = .05/6 = 0.0083$) were used to further analyze the effect of joint center definition.

Chapter IV

To investigate the change in MTPJ mechanics across running speeds, joint level kinematics, kinetics, and stiffness were estimated. Joint angles were estimated using a two-dimensional model adapted from Goldmann et al. (Goldmann et al., 2013) was used, but with the distal forefoot marker placed on the anterior aspect of the hallux as opposed to second toe. Joint moments were estimated using the sliding joint center method developed in Chapter III. Moments were estimated using an inverse dynamics approach. Two measures of stiffness were estimated, one during the loading phase referred to as active stiffness, and one for the phase up until peak dorsiflexion occurs referred to as critical resistance. The phase consisting of stance until peak dorsiflexion occurs is adopted from previous work (Oh and Park, 2017) but changed to factor in the minimum angle. Further details about stiffness estimations are provided in Chapter IV.

Peak MTPJ moments, range of motion, and stiffness measures were analyzed using one-way repeated measures analysis of variance tests ($\alpha = .05$) to determine the effect of speed. Pairwise comparisons with Bonferroni adjustments ($\alpha = .05/10 = 0.005$) were used post-hoc to analyze main effects. Equal variance in the data was assessed using Mauchly's test of Sphericity ($\alpha < .05$). Greenhouse-Geisser corrections were used when violations of sphericity were detected.

Chapter V

To assess the change in maximum isometric IFM strength across time and between groups, peak force from each trial was extracted. Peak force was averaged across the three maximum contraction trials from each visit for each participant. A linear interaction comparison analysis of variance ($\alpha = .05$) was conducted to determine the change in IFM strength within-groups at baseline, five weeks, and ten weeks.

Mechanics of the MTPJ and ankle and contact time were assessed. Joint angles were estimated using the same two-segment foot model previously described and expanded to include the shank, thigh, and pelvis. Joint level mechanics assessed were: MTPJ maximum moment, range of motion, moment at maximum dorsiflexion, and critical resistance; and ankle maximum moment and peak plantarflexion angular velocity. Contact time was also assessed.

Metabolic data were collected and analyzed during steady-state conditions (RER < 1.0). VO2 consumption, VCO2 production, and RER were averaged over the last ninety seconds of each stage. Normalized VO2 values and metabolic rate were assessed.

To determine the effects of time and group, mixed model analysis of variance ($\alpha = .05$) tests with within-subject factor of time and between-subject factor of group were used to analyze biomechanical and metabolic variables. Equal variance in the data was assessed using Mauchly's test of Sphericity ($\alpha < .05$). Greenhouse-Geisser corrections were used when violations of sphericity were detected. Pairwise comparisons with Bonferroni adjustments ($\alpha = .05/6 = 0.0083$) were used post-hoc to further analyze main effects.

Chapter VI

To investigate the effects of varying footwear LBS on distance running performance across speeds, gross level mechanics and running economy were analyzed. Stride frequency, step length, contact time, braking and propulsive impulses, and peak propulsive force were estimated from ground reaction force data. Metabolic data were collected and analyzed the same manner as described for Chapter V, with the exception being data were averaged over the final two minutes of each stage. Running economy and metabolic rate were assessed. Subjective comfort assessments were analyzed by averaging the score for individual categories as well as an overall shoe comfort score.

To determine the effect of LBS at each running speed, repeated measures analysis of variance ($\alpha = .05$) tests were used to analyze gross level mechanics and metabolic variables. Equal variance in the data was assessed using Mauchly's test of Sphericity ($\alpha < .05$). Greenhouse-Geisser corrections were used when violations of sphericity were detected. Pairwise comparisons with Bonferroni adjustments ($\alpha = .05/3 = 0.0167$) were used post-hoc to further analyze main effects.

Chapter VII

To further understand the mechanics behind the changes in gross level mechanics and metabolics from Chapter VI, joint level mechanics were assessed. The two-dimensional foot model and expanded marker set previously described were used to calculate joint angle and moment data. Variables assessed for all speeds and footwear conditions were: MTPJ critical resistance, peak moment, range of motion, positive work, negative work; ankle peak moment, range of motion, positive work, negative work; knee positive and negative work; hip positive and negative work.

To determine the effect of LBS at each running speed, repeated measures analysis of variance ($\alpha = .05$) tests were used to analyze gross level mechanics and metabolic variables. Equal variance in the data was assessed using Mauchly's test of Sphericity ($\alpha < .05$). Greenhouse-Geisser corrections were used when violations of sphericity were detected. Pairwise comparisons with Bonferroni adjustments ($\alpha = .05/3 = 0.0167$) were used post-hoc to further analyze main effects.

CHAPTER III

A COMPARISON OF METATARSOPHALANGEAL JOINT CENTER LOCATIONS ON ESTIMATED JOINT MOMENTS DURING RUNNING

This work was published in volume 86 of *Journal of Biomechanics* in January 2019. Evan Day designed this study and collected and analyzed data. Michael E. Hahn provided mentorship including assistance with study design, oversight, and editing and finalizing the final manuscript.

Introduction

The forefoot functions as the base of support during running after the heel lifts off the ground during stance (Miyazaki and Yamamoto, 1993; Stefanyshyn and Nigg, 1997). Rotation about the five metatarsophalangeal joints (MTPJ) occurs about two anatomical axes of rotation; a transverse axis across the first and second MTPJ, and an oblique axis across the second to fifth MTPJ (Bojsen-Moller, 1978). The foot has a complex anatomy with a number of muscle-tendon units crossing the MTPJ axes in addition to the plantar fascia (Mckeon et al., 2015). Insertion points of the muscle-tendon units vary between the second to fifth metatarsals (i.e. flexor digitorum longus, brevis) and the hallux (i.e. abductor hallucis longus, brevis) and the plantar fascia inserts onto all phalanges (Bojsen-Møller and Lamoreux, 1979). However, the MTPJ is commonly modeled as a single oblique axis for running analysis (Smith et al., 2012; Stefanyshyn and Nigg, 1997).

Definition of the MTPJ axis affects estimated moments during sprinting (Smith et al., 2012). A two-dimensional approach using a medial-lateral perpendicular axis from the fifth metatarsal marker (Stefanyshyn and Nigg, 1997, 1998, 2000) results in joint moments two to four times higher than when estimated about an oblique or dual axis (Smith et al., 2012). Further, moments about a dual axis are lower compared to an oblique axis (Smith et al., 2012). During running, most motion occurs about the transverse portion across the first and second metatarsals of the dual axis, as the center of pressure travels medial to the second metatarsal during push-off (De Cock et al., 2008; Smith et al., 2012; Willwacher et al., 2013).

There is no commonly accepted location for where to model the MTPJ joint center. Fixed MTPJ joint center locations have previously included the midpoint of the oblique axis (Hoogkamer et al., 2018; McDonald et al., 2016; Oh and Park, 2017; Oleson et al., 2005; Roy and Stefanyshyn, 2006), in plane with the long axis of the foot in the medial-lateral direction and the first metatarsal head in the anterior-posterior direction, over the second metatarsal (Miyazaki and Yamamoto, 1993; Willwacher et al., 2013), and the fifth metatarsal for two-dimensional analysis (Bezodis et al., 2012; Smith et al., 2012; Stefanyshyn and Nigg, 1997, 1998, 2000). Additionally, a non-fixed joint center method treating the MTPJ as a true hinge joint has been utilized (Rolian et al., 2009; Smith et al., 2014). Use of a non-fixed joint center accounts for inter-subject variability that can arise from toe-out angle, compared to a fixed joint center location (Chang et al., 2007; Rolian et al., 2009). A non-fixed joint center may more accurately represent the anatomy of the forefoot and toe segments, where the primary motion is in the sagittal plane with muscle-tendon units inserting on the metatarsals to modulate flexion and extension (Bojsen-Moller, 1978; Bojsen-Møller and Lamoreux, 1979; Mckeon et al., 2015).

The purpose of this study was to investigate how MTPJ kinetics change with differing joint center definition. We hypothesize that the more posteriorly oriented joint centers will have larger peak moments and that the external dorsiflexion moment will arise earlier in stance phase. It is also hypothesized that the use of a non-fixed joint center will result in the lowest peak moment.

Methods

Recruitment

Nineteen (5 female) competitive runners were recruited for this study (24 ± 6 yr, 63 ± 10 kg, 52 ± 14 mi/wk, 16:43 average 5000m best). Inclusion criteria included: 5000m personal best under 18:00 (males) and 20:00 (females), no lower extremity injury in the previous six months, and currently running over 30 miles/week. Participants provided informed consent prior to data collection. This study was approved by the Institutional Review board at the University of Oregon.

Study Design and Experimental Protocol

Retro-reflective markers were placed on the foot in accordance with Goldmann et al. (2013). The forefoot was defined by markers on the superior distal aspect of the hallux and heads of the first and fifth metatarsals (medial and lateral aspects, respectively). The rearfoot was defined by markers on the medial, lateral, and posterior aspects of the calcaneus. Windows were cut in the shoes to place markers directly on the foot (Bishop et al., 2014). Participants all wore the same standard neutral cushioned footwear (Brooks Launch 3) to eliminate the effects of longitudinal bending stiffness (Stefanyshyn and Nigg, 2000; Willwacher et al., 2013, 2014).

Data Collection

Running trials were conducted on a force instrumented treadmill (Bertec, Inc., Columbus, OH). Kinematic data were collected at 200 Hz and kinetic data were collected at 2000 Hz. Participants ran at five speeds, 3.89, 4.44, 5.00, 5.56, and 6.11 m/s. For the current analysis of MTP endpoint effect, the slowest and fastest speeds, 3.89 and 6.11 m/s, were analyzed.

Data Analysis

A custom MATLAB (version R2016b; MathWorks, Natick, MA) program was used to estimate joint kinematics and kinetics for stance phase only, defined as when the vertical ground reaction force exceed 5% body weight. Marker coordinate data were filtered using a zero-lag, fourth-order low pass Butterworth filter with at 20Hz cutoff frequency (Willwacher et al., 2013). Center of pressure was visually assessed and assured by low-pass filtering force data (cutoff = 20 Hz) to eliminate treadmill vibration noise (Willems and Gosseye, 2013).

The five MTP joints were modeled as a single hinge axis defined by the vector from the first to fifth metatarsal markers (Smith et al., 2012). Three separate methods using fixed joint center locations were defined according to previous studies (Figure 3.1). These methods included joint center locations at the head of the fifth metatarsal (Bezodis et al., 2012; Stefanyshyn and Nigg, 2000, 1998b, 1997), the midpoint between the first and fifth metatarsals (Hoogkamer et al., 2018; McDonald et al., 2016; Oh and Park, 2017; Oleson et al., 2005; Roy and Stefanyshyn, 2006), and at a point that was in the plane of the head of the first metatarsal in the anterior-posterior direction and the long axis of the foot in the medial-lateral direction (Miyazaki and Yamamoto, 1993; Willwacher et al., 2013), referred to henceforth as the second metatarsal head joint center. Kinetic estimations for the fifth metatarsal location were solved in a two-dimensional model.

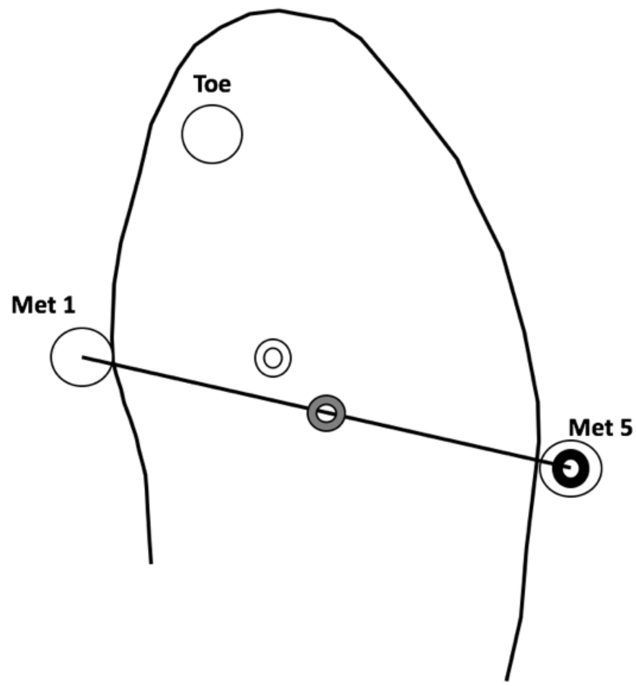

Figure 3.1. Fixed metatarsophalangeal joint center locations at the second metatarsal head (white), midpoint (gray), and fifth metatarsal (black). Black line defines the oblique hinge axis.

In addition to the fixed joint center approaches, we tested a non-fixed joint center method utilizing the perpendicular moment arm from the center of pressure (COP) to the MTP oblique axis. This approach is similar to 1) Rolian et al. (2009) who transferred the COP into the forefoot coordinate system and estimated resultant moments using ground reaction forces and respective moment arms; and 2) Smith et al. (2014) who calculated the horizontal moment arm as the perpendicular distance from the COP to the MTPJ axis.

A novel approach was developed to estimate MTPJ moments using a non-fixed joint center method. A triangle formed by the first and fifth metatarsal markers and COP coordinates and a law of cosines approach was used to calculate the perpendicular moment arm of the COP to the MTP axis (Figure 3.2). The vertical coordinate of the first and fifth metatarsal markers was set to zero to be in plane with the COP and not alter the magnitude of the resultant angle, β. The following equations were utilized to calculate the non-fixed joint center location along the MTP axis at the intersection of the perpendicular moment arm from the COP, referred to as the sliding MTPJ joint center.

$$\beta = \frac{\overrightarrow{C5}^2 + \overrightarrow{met\ axis}^2 - \overrightarrow{C1}^2}{2 * \overrightarrow{C5} * \overrightarrow{met\ axis}}$$

$$joint\ center\ position\ vector = \overrightarrow{C5} * \beta$$

Where β is the angle near the fifth metatarsal marker, $\overrightarrow{C1}$ is the magnitude of the vector from the COP to the first metatarsal, $\overrightarrow{C5}$ is the magnitude of the vector from the COP to the fifth metatarsal, and $\overrightarrow{met\ axis}$ is the magnitude of the vector from the first to fifth metatarsal. The value for β was multiplied by $\overrightarrow{C5}$ to determine the magnitude of the distance from the fifth metatarsal to the location on the MTP oblique axis representative of the perpendicular moment arm intersection, referred to as the joint center position vector.

The vertical components of the first and fifth metatarsal markers were then added back into their respective coordinates. The joint center position vector was then added to the fifth metatarsal marker to obtain the position along the oblique axis representative of the sliding joint center location in global space using the following equation.

$$Sliding\ MTP\ joint\ center = met5 + (FFrm * joint\ center\ position\ vector)$$

Where *met5* is the global coordinate of the fifth metatarsal marker and *FFrm* is the forefoot segment rotation matrix with respect to the global coordinate system.

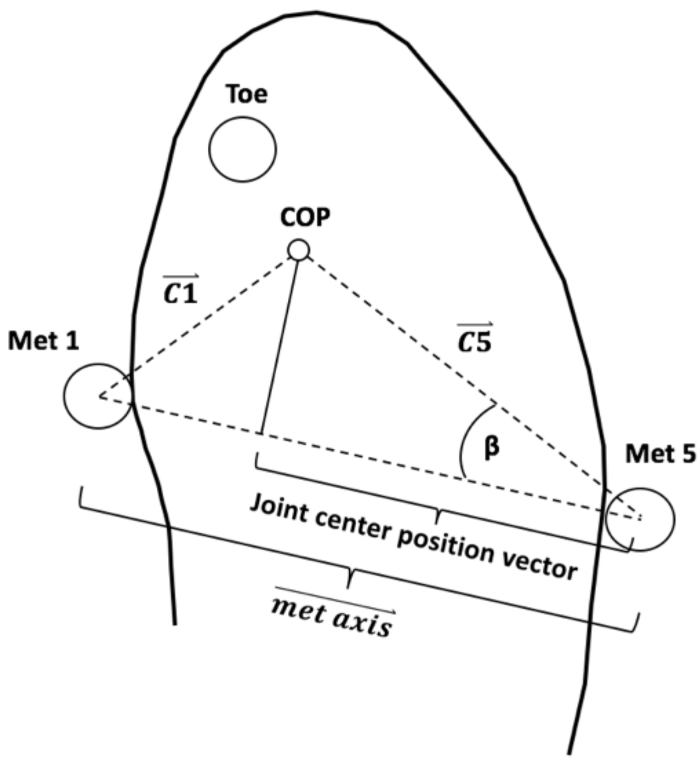

Figure 3.2. Depiction of the methodology for utilizing the law of cosines to define the sliding joint center. Sliding joint center location is the intersection of the moment arm from the center of pressure to the metatarsal joint axis.

An inverse dynamics approach was used to estimate joint moments resolved in the rearfoot coordinate system. The inertial effects of the forefoot were considered negligible (Stefanyshyn and Nigg, 1997). Moments were considered zero until the COP passed anterior to the joint center location.

Statistical Analysis

A one-way repeated measures analysis of variance (ANOVA) was used to compare peak moments between joint center locations for both speeds. Pairwise comparisons with Bonferroni adjustments ($\alpha = .05/6 = .0083$) were used to further analyze main effects. One-dimensional, one-way repeated measures Statistical Parametric

Mapping (SPM) ($\alpha = .05$) was used to assess differences in plantar flexor moment throughout stance (Pataky et al., 2013, 2016). Follow up pairwise comparison analyses were conducted using SPM t-tests with a Bonferroni corrected alpha level.

Results

Peak joint moments were significantly affected by joint center location for both speeds ($p < .001$) (Table 3.1). The trend in change of maximum moment was similar between speeds. More posteriorly oriented fixed joint centers resulted in larger peak moments. The sliding joint center definition resulted in the smallest peak moments.

For the slow speed (3.89 m/s), the sliding ($p < .001$), second metatarsal ($p < .001$), and midpoint ($p = .002$) joint center definitions had significantly lower peak moments than the fifth metatarsal joint center. Peak moments for the midpoint ($p < .001$) and fifth metatarsal ($p < .001$) joint centers were significantly higher than the sliding joint center, but not the second metatarsal joint center ($p = .117$). Peak moments were not significantly different between the second metatarsal and midpoint definitions ($p = .164$).

For the fast speed (6.11 m/s), moments for the sliding ($p < .001$), second metatarsal ($p < .001$), and midpoint ($p = .006$) joint center definitions were all significantly lower than for the fifth metatarsal joint center definition. Peak moments for the second metatarsal ($p = .008$), midpoint ($p < .001$), and fifth metatarsal ($p < .001$) were all significantly higher than for the sliding joint center. Peak moments were not significantly different between the second metatarsal and midpoint joint centers ($p = .114$).

Table 3.1. Metatarsophalangeal joint moments (N-m/kg) as a function of velocity and joint center location; Mean ± SD.

Velocity (m/s)	Sliding	2^{nd} Met. Head	Midpoint	5^{th} Met. Head
3.89	$0.464 \pm .146^{c,d}$	$0.590 \pm .164^{d}$	$0.704 \pm .204^{a,d}$	$0.952 \pm .295^{a,b,c}$
6.11	$0.802 \pm .238^{b,c,d}$	$1.07 \pm .354^{a,d}$	$1.26 \pm .420^{a,d}$	$1.68 \pm .385^{a,b,c}$

a = significantly different from sliding, b = significantly different from 2^{nd} met, c = significantly different from midpoint, d = significantly different from 5^{th} met
Significant difference, $p < 0.05$

Statistical parametric mapping analysis revealed a significant main effect of joint center definition on estimated moments from 22-85% and 88-99% of stance phase when running at 3.89 m/s (Figure 3.3). Post-hoc pairwise comparisons revealed significant difference for the fifth metatarsal definition compared to the other methods (Figure 3.4). The midpoint and second metatarsal joint center definitions exhibited moments that were lower in early stance and higher in late stance, compared to the sliding joint center definition. Post-hoc analysis revealed no significant pairwise comparisons for the main effect from 88-99% of stance.

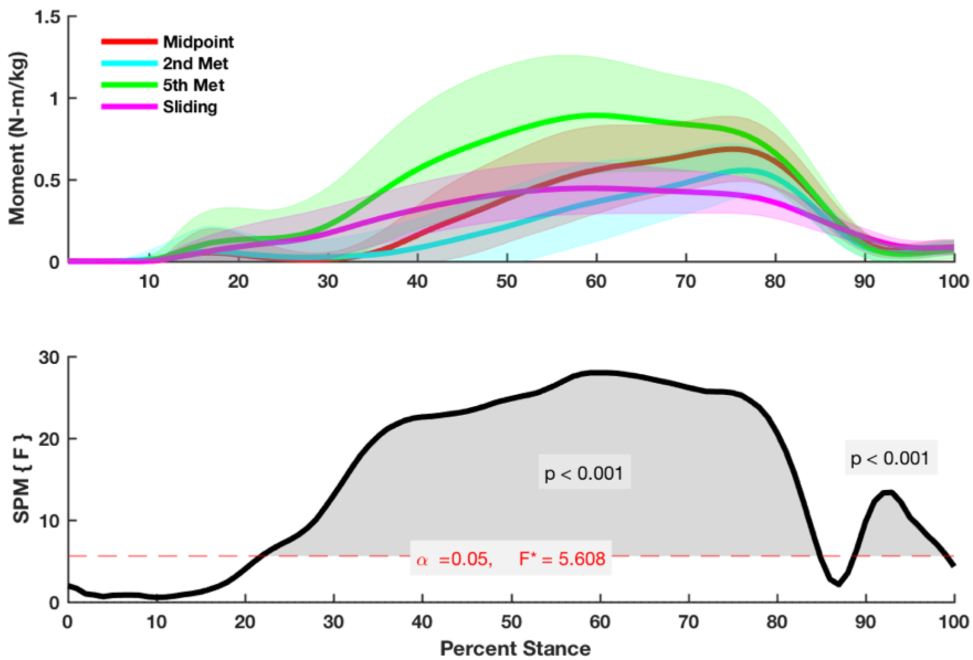

Figure 3.3. Metatarsophalangeal joint moments during stance phase running at 3.89 m/s (top) and running F-critical value (solid line) and F-critical threshold (dashed red line) (bottom). Shaded region represents duration of significant effect of joint center.

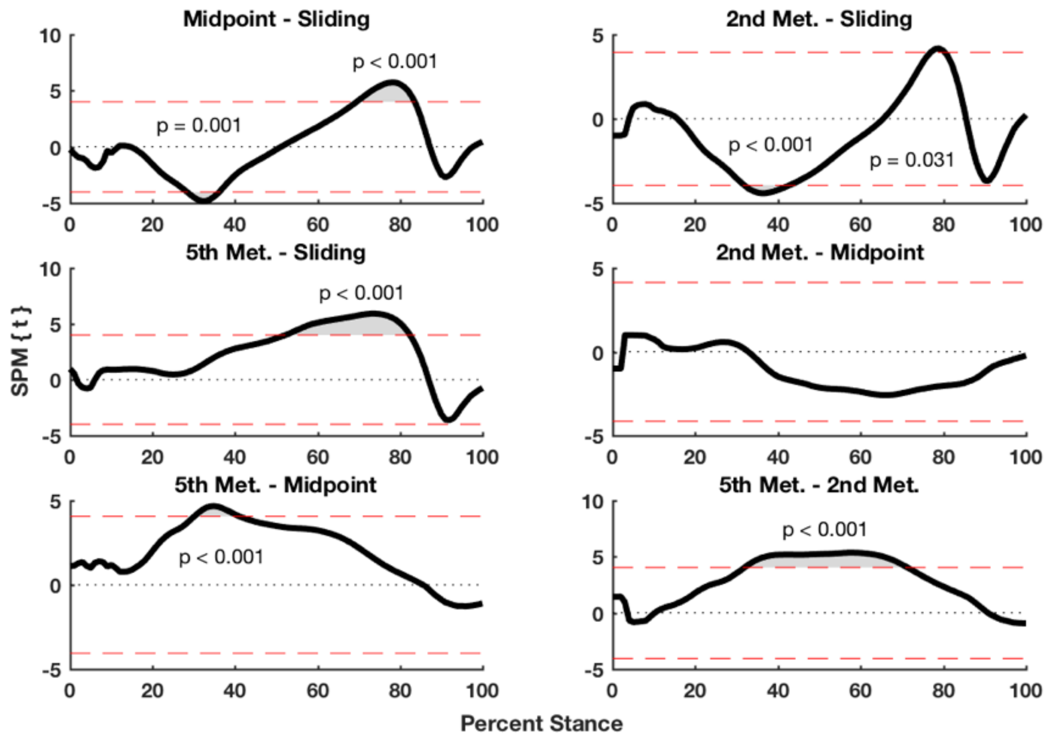

Figure 3.4. Post-hoc pairwise comparisons between individual joint center definitions with running t-statistic (solid line) and t-critical band (dotted red line) and Bonferroni adjusted p-values for running at 3.89 m/s.

For the fast running condition (6.11 m/s) SPM analysis revealed a significant main effect of joint center definition on estimated moments from 22-86% of stance (Figure 3.5). Post-hoc pairwise comparisons revealed significant differences for the fifth metatarsal definition compared to the other joint center conditions (Figure 3.6). Comparison between the midpoint and sliding joint center definitions revealed a significant difference from 60-69% of stance.

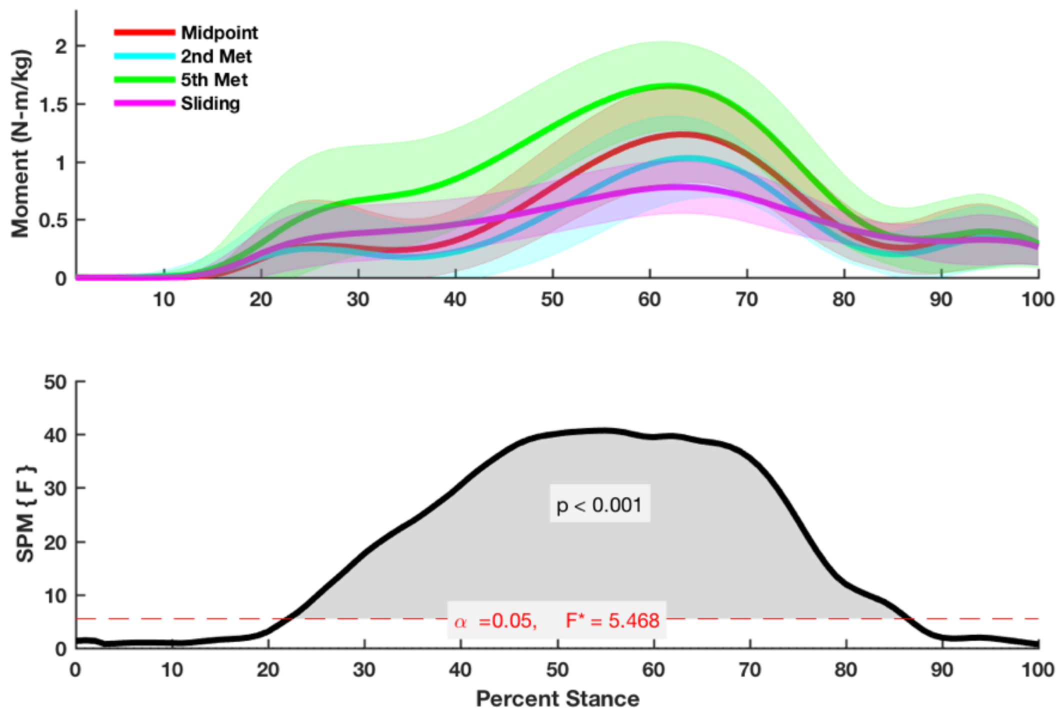

Figure 3.5. Metatarsophalangeal joint moments during stance phase running at 6.11 m/s (top) and running F-critical value (solid line) and F-critical threshold (dashed red line) (bottom). Shaded region represents duration of significant effect of joint center.

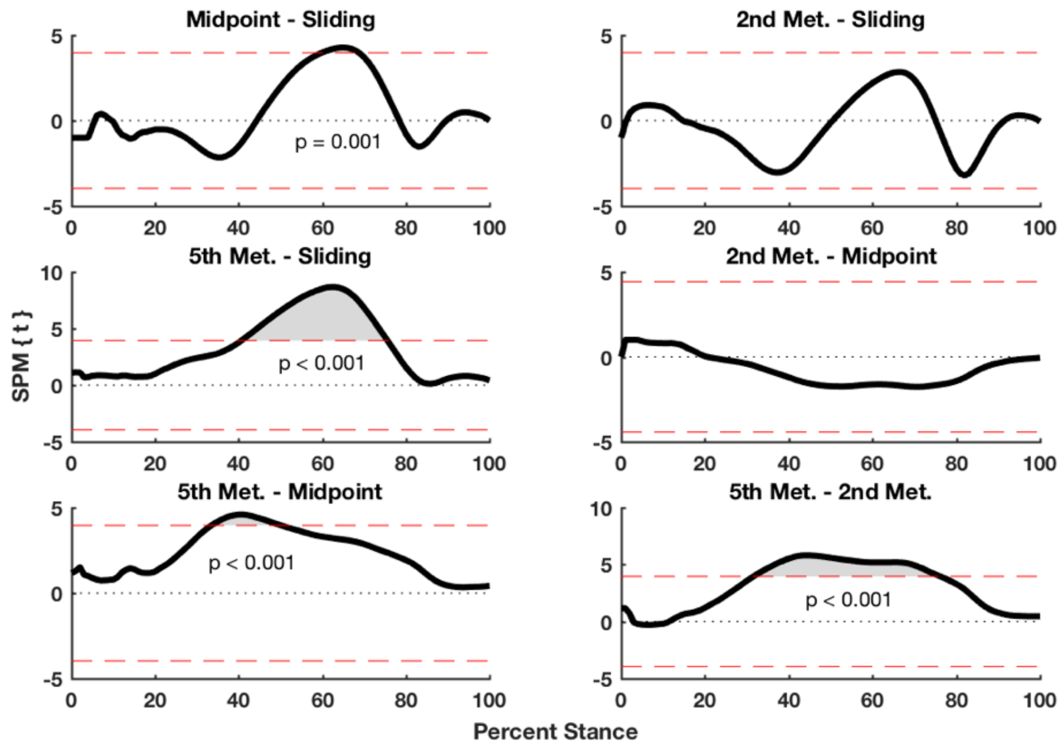

Figure 3.6. Post-hoc pairwise comparisons between individual joint center definitions with running t-statistic (solid line) and t-critical band (dotted red line) and Bonferroni adjusted p-values for running at 6.11 m/s.

Discussion

The purpose of this study was to investigate how MTPJ joint center definition affects estimated MTPJ moments during running. Results suggest that MTPJ joint center definition significantly affects both peak magnitude and magnitude of plantar flexor moment throughout stance. These differences were observed across running speeds.

Previous work comparing MTPJ axis definition demonstrated that estimated moments about a transverse axis from the fifth metatarsal overestimated plantar flexor moments by two to four times compared to an oblique axis (Smith et al., 2012). Our results agree with these findings (Table 3.1). The fifth metatarsal head joint center definition resulted in peak MTPJ moments 105% and 109% greater than the sliding

MTPJ joint center for the slow and fast conditions, respectively. The midpoint joint center resulted in peak moments 52% and 57% greater than the sliding joint center for the slow and fast conditions, respectively. The second metatarsal head approach resulted in peak moments 27% and 33% greater than the sliding joint center for the slow and fast conditions, respectively. These results are intuitive as the moment arms for the fixed joint centers may not be the shortest perpendicular distance from the COP to the MTPJ axis. A fixed joint center can be affected by toe-in or toe-out angle (Chang et al., 2007; Rolian et al., 2009). A toe-out orientation will extend the moment arm from the fixed joint centers to the COP, whereas a toe-in orientation will decrease the moment arm and position the oblique MTP axis into a more transverse orientation with respect to a global coordinate system. Increased variability in our results for the fixed joint center conditions compared to the sliding joint center may be partially explained by variable toe-out angle.

An effect on moment magnitude throughout stance was observed in the slow running condition (Figure 3.4). Post-hoc pairwise comparisons show that the fifth metatarsal joint center definition was significantly larger than from the other joint center definitions during some periods of stance. The fifth metatarsal definition is the most posterior and thus has the longest moment arm to the COP. Comparison to the sliding and second metatarsal joint centers show that the difference is near the timing of maximum moment. The midpoint joint center definition was only different from the fifth metatarsal definition from 30-40% of stance, at onset of the plantar flexor moment.

The midpoint and second metatarsal joint center definitions both exhibit significantly smaller and larger moments than the sliding joint center. From 30-40% stance is when the moments were smaller. The fixed joint center definitions may have later plantar flexor moment onset due to a later passing of the COP across the MTP axis, as the COP tends to move lateral to medial during running (Becker et al., 2014). If the COP crosses the MTP axis lateral to the fixed joint centers, then the sliding and fifth metatarsal joint center definitions will register earlier plantar flexor moment onset due to the oblique nature of the MTPJ axis. The second phase of significant difference is when the midpoint and second metatarsal joint center definitions have significantly larger moments during late stance when their peak moments occur. The larger moments in late stance may be due to supination (Novacheck, 1998) causing lateral movement of the COP (Becker et al., 2014). Though the ground reaction force decreases in late stance, lateral and anterior movement of the COP will lengthen the moment arm with respect to the midpoint and second metatarsal joint centers, resulting in a later peak moment.

Joint center definition primarily only affected the plantar flexor moment during mid-stance for the fast running condition (Figure 3.6). The fifth metatarsal definition displayed significant difference from the other definitions during mid-stance. From 60-69% of stance the midpoint joint center had a significantly larger moment than the sliding joint center, when peak moment occurred. During sprinting the COP tends toward the medial side of the foot, causing the COP to be anterior to the axis across the first and second metatarsals (Smith et al., 2012). A more medial COP passage may explain why there was no difference in time of plantar flexor moment between joint center definitions. The observed higher moments for the fifth metatarsal definition compared to other joint center definitions and for midpoint compared to sliding appears due to the change in moment arm.

At both speeds the COP passed anterior to the MTP axis at approximately 10% of stance and the midpoint and 2nd metatarsal head joint center techniques exhibit a small sinusoidal pattern in plantar flexor moment in the first half of stance before increasing toward the peak moment. This sinusoidal pattern may be attributable to inter-subject differences in the percent of stance phase at which the COP passed anterior to the MTP axis. Foot strike pattern was not controlled for and thus may be the cause of this observation. In a forefoot/midfoot strike pattern the COP moves posteriorly during weight acceptance before moving anteriorly during push-off (Cavanagh and Lafortune, 1980). This may explain the initial increase in plantar flexor moment early in stance before a decrease and subsequent increase once again. The sinusoidal pattern may also be attributable to a bending moment error as a result of the force distribution under the MTPJ axis (Oleson et al., 2005). While the metatarsal heads are in contact with the ground, force is being transmitted anteriorly and posteriorly to the MTPJ. The resultant COP may pass anteriorly and posteriorly across the MTPJ axis when the COP anterior-posterior coordinate is close to the axis. Shod participants in the current study may have further induced this error by distributing force over the larger area of the shoe outsole. In analyzing MTPJ kinetics while the metatarsal heads are in contact with the ground, caution should be used with respect to the potential limitations from COP coordinate calculations.

While this study focused only on MTPJ kinetics, it is intuitive that the effects would carry over into joint energetic calculations as joint work is the integral of joint power, and angular velocity will not change based on MTPJ joint center definition. While these resultant energetic calculations will not affect the consensus that the MTPJ functions primarily as a damper (Stefanyshyn and Nigg, 1997), it will affect net work and power estimations. Negative work estimations will be more severely overestimated due to the toe-flexors primarily acting eccentrically throughout stance phase.

Results from this study may be applicable to running footwear and prosthetic designers. Optimal footwear bending stiffness may improve performance by reducing the energetic cost of running (Hoogkamer et al., 2017b; Roy and Stefanyshyn, 2006). In addition, footwear bending stiffness can alter mechanical function of the MTPJ by shifting the joint moment arm anteriorly (Willwacher et al., 2014) resulting in a reduction of angular deflection and increased plantar flexor moment (Stefanyshyn and Nigg, 2000; Willwacher et al., 2013). Recent efforts have utilized the load-displacement patterns of the MTPJ to tune footwear bending stiffness (Oh and Park, 2017). From the current results, it is evident that such efforts should potentially take into consideration joint center and axis definition (Smith et al., 2012). Larger moments may lead to custom tuned footwear being more stiff. If a shoe is too stiff, gait mechanics and running economy are both negatively affected (Madden et al., 2015; Willwacher et al., 2014). Individuals fit with improperly tuned footwear may exhibit changes in gait mechanics that are potentially detrimental to running performance.

Prosthetic engineers interested in developing a foot with a functional MTP joint based on anatomical limb data should take into consideration how MTPJ joint center definition will affect estimated energy absorption, generation, and stiffness. When tuning dynamic angular stiffness of the prosthetic foot, overestimation of the MTPJ plantar flexor moment may result in an individual adopting a metabolically costly gait pattern due to alteration of load and energy transfer between the lower extremity joints.

Limitations

One limitation of this study is that results can be affected by marker placement error. Any error in the placement of retro-reflective markers on the metatarsal heads to define the MTPJ axis will affect joint moment estimations by altering the axis orientation and joint centers in global reference. Additionally, potential errors in COP calculation may exacerbate differences between joint center definitions.

Future Work

Future work should expand this model to factor in both axes of the MTPJ (Bojsen-Moller, 1978; Smith et al., 2012). Additionally, further work is needed to understand how to partition ground reaction forces between the rearfoot and forefoot, and factor in bending moment error about the MTPJ (Oleson et al., 2005). Further in-depth development of the model will aid in more accurately quantifying joint moments replicative of *in vivo* loading.

Conclusion

In conclusion, this study has demonstrated that MTPJ joint center definition significantly affects estimated moments about the MTPJ. Researchers that include the MTPJ in inverse dynamics analysis and footwear and prosthetic designers that are interested in tuning forefoot bending stiffness should take MTPJ joint center location into consideration, as well as MTPJ axis definition. Due to the observed difference in moments during stance across joint center definitions and the complex anatomy of the foot, it may be more anatomically and mechanically representative for the MTPJ to be modeled as a hinge axis between the first and fifth metatarsal heads (Smith et al., 2012) with a non-fixed joint center where moments are estimated via a perpendicular moment arm from the COP to the MTPJ. Researchers should be aware of differences in timing of moment onset and magnitude that can result from differing MTPJ joint center definitions.

Bridge

The results of this study suggest that MTPJ joint center definition greatly affects estimated moments during the stance phase of running. The sliding joint center method developed in this chapter will be used throughout the remainder of this dissertation as it may most accurately represent joint moments acting about the MTPJ axis. The application of this joint center model will be of importance when assessing changes in MTPJ mechanics across speeds in Chapter IV, and when using insights from Chapter IV to guide tuning of footwear stiffness in Chapters VI and VII.

CHAPTER IV

DYNAMIC ANGULAR RESISTANCE ABOUT THE METATARSOPHALANGEAL JOINT INCREASES WITH RUNNING SPEED

This chapter is currently in review for publication. Evan Day designed the study and collected and analyzed the data. Michael E. Hahn provided mentorship and aided in study design, general oversight, and editing and finalizing the final manuscript.

Introduction

The metatarsophalangeal (MTP) joint serves as the base of support during running once the heel lifts off the ground (Goldmann and Brüggemann, 2012; Rolian et al., 2009; Stefanyshyn and Nigg, 1997). The rearfoot pivots around the MTPJ axis as the whole body center of mass moves anteriorly, resulting in energy being absorbed at the MTPJ and very little being generated (Goldmann and Brüggemann, 2012; Roy and Stefanyshyn, 2006; Stefanyshyn and Nigg, 1997).

In the shod condition the foot and shoe act together to modulate mechanical function of the MTPJ. The foot and shoe act in series to provide linear stiffness to the foot-shoe complex (Kelly et al., 2016) and act in parallel to provide angular stiffness to the foot-shoe complex (Oleson et al., 2005). Changing the longitudinal bending stiffness of footwear via flat carbon fiber plates can affect mechanics of the MTPJ and other lower extremity joints (Stefanyshyn and Nigg, 2000; Willwacher et al., 2013) by altering the lever arm from the joint centers to the resultant ground reaction force vector (Willwacher et al., 2014). Recent efforts have shown that a curved carbon fiber plate in combination with a more compliant foam in the midsole helps reduce the energetic cost of running (Hoogkamer et al., 2017a). Increasing the longitudinal bending stiffness of footwear may influence a shift in muscle force-velocity operating points to a more favorable position (Madden et al., 2015; Takahashi et al., 2016; Willwacher et al., 2013) and improve running economy (Hoogkamer et al., 2017a; Madden et al., 2015; Roy and Stefanyshyn, 2006) to help benefit overall running performance.

Recent efforts have been put forth to determine an optimal bending stiffness of footwear by utilizing the load-displacement plot of the MTPJ to quantify the critical stiffness, defined as the ratio of the change in moment to maximum dorsiflexion (Oh and Park, 2017). It has been proposed that this critical stiffness represents the threshold for the bending stiffness that maximizes the elastic benefit of the shoe without inhibiting natural motion of the MTPJ (Oh and Park, 2017). The net angular impulse of the entire lower limb during running is the product of contributions from the musculoskeletal system and strain energy stored in the shoe. The use of footwear tuned to an optimal bending stiffness to store and return strain energy may reduce the required contribution from the musculoskeletal system to generate angular impulse, leading to a beneficial reduction in metabolic cost (Oh and Park, 2017).

General lower extremity joint stiffness has been shown to increase with running speed (Jin and Hahn, 2018; Stefanyshyn and Nigg, 1998a), but little is known about how MTPJ stiffness changes across running speeds. Because the MTPJ does not exhibit spring like behavior, we describe one measure of joint stiffness and one of resistance in this paper. The purpose of this study was to investigate how these two measures of dynamic angular stiffness and resistance about the MTPJ change across running speeds. We hypothesize that MTPJ stiffness and resistance will increase with running speed.

Methods

Recruitment

Eighteen competitive runners (four female) were recruited for this study (Table 4.1). To be included participants had to have a 5000m personal best under 18:00 (males) and 20:00 (females), no lower extremity injury in the previous six months, and currently running over 30 miles/week. Participants provided informed consent prior to data collection. This study was approved by the Institutional Review Board at the University of Oregon.

Table 4.1. Participant characteristics

Sex	Age (yr)	Height (cm)	Mass (kg)	Weekly Mileage (km)	5000m Best (min:sec)
Male	24 ± 6	180 ± 10	67 ± 8	90 ± 24	16:16 ± 0:57
Female	26 ± 7	165 ± 7	56 ± 8	77 ± 24	17:45 ± 1:07

Study Design and Experimental Protocol

A bilateral lower extremity marker set consisting of 41 retro-reflective markers defining nine segments (forefoot, rearfoot, shank, thigh, pelvis) was used. A two-segment foot model was defined by placing markers on the forefoot and calcaneus (Goldmann et al., 2013). Windows were cut in the shoes to place markers directly on the foot (Bishop et al., 2014). Participants all wore the same footwear (Brooks Launch 3) to control for the effects of longitudinal bending stiffness on MTPJ mechanics (Willwacher et al., 2013). Individual retro-reflective markers were placed on the medial and lateral malleoli, medial and lateral femoral epicondyles, left and right greater trochanter, left and right posterior superior iliac spines, and the sacrum. Quadrad marker clusters were placed on the shank and thigh. Participants performed a static trial after which markers were then removed from the medial malleoli and femoral epicondyles so that they did not interfere with running movements.

Data Collection

Running trials were conducted on a force instrumented treadmill (Bertec Inc., Columbus, OH). Kinematic data were collected at 200 Hz and kinetic data were collected at 2000 Hz. Participants ran at five speeds, 3.89, 4.44, 5.00, 5.56, 6.11 m/s. Data were collected at each speed for ten consecutive strides. Rest between conditions was self-selected. These speeds were chosen as they represent relevant training and racing paces for club level competitive runners.

Data Analysis

A custom MATLAB (version R2016b; MathWorks, Natick, MA) program was used to calculate joint kinematics, kinetics, and stiffness. These metrics were calculated throughout the entire stance phase. Stance phase was defined as the phase when the vertical ground reaction force exceeded 5% of body weight. Raw marker coordinate data were filtered using a zero-lag, fourth-order low pass Butterworth filter with a 20Hz cutoff frequency (Willwacher et al., 2013). The same cut-off frequency was used for force platform data to avoid artifact in joint moment estimations (Bezodis et al., 2013). Joint angles were calculated using an Euler/Cardan rotation order of flexion/extension, abduction/adduction, and internal/external rotation. Sagittal plane MTPJ angles and moments were used for analysis.

Metatarsophalangeal joint moments were estimated using an inverse dynamics approach. The MTPJ was modeled as a hinge axis along the vector connecting the 1st and 5th metatarsal markers (Smith et al., 2012). The ground reaction force moment arm was estimated as the perpendicular distance between the center of pressure and the MTPJ oblique axis (Chapter III, Day and Hahn, 2019). Resultant forces and moments about the MTPJ were considered zero until the center of pressure passed anterior to the MTPJ oblique axis (Stefanyshyn and Nigg, 1997). Inertial effects of the forefoot were considered negligible (Stefanyshyn and Nigg, 1997). Joint moments were resolved in the rearfoot coordinate system for reporting and joint stiffness calculations. Kinematic and kinetic data were re-sampled to 101 data points per stance phase for time-normalized analysis.

Two measures describing the load-displacement relationship of the MTPJ were quantified, active stiffness (K_{active}) and critical resistance (R_{cr}) (Figure 4.1). Load-displacement measures were calculated as follows:

$$K_{active} = \frac{Max\ MTP\ Moment - 0}{\theta_{MTP(at\ max\,moment)} - \theta_{MTP\ min}}$$

$$R_{cr} = \frac{MTP\ moment(at\ max\ dorsiflexion) - 0}{\theta_{MTP\ max} - \theta_{MTP\ min}}$$

Active stiffness represents the phase during which energy is delivered into the forefoot as this phase is when the MTPJ plantar flexor moment is increasing. Critical resistance represents the functional resistance of the MTPJ and represents the amount of strain energy stored in the forefoot that is available for MTPJ plantar flexion. Critical resistance may also be used to identify the stiffness threshold for the elastic benefit of footwear without inhibiting natural motion of the MTPJ (Oh and Park, 2017). Our measure of critical resistance has been previously described as critical stiffness (Oh and Park, 2017). However, as the MTPJ does not behave as a spring mechanism during running, and because a substantial amount of energy has been dissipated up to the occurrence of peak MTPJ dorsiflexion, we believe that this phase is not necessarily a stiffness. Rather, it is a measure of a resistance to dorsiflexion; thus we will use the term critical resistance.

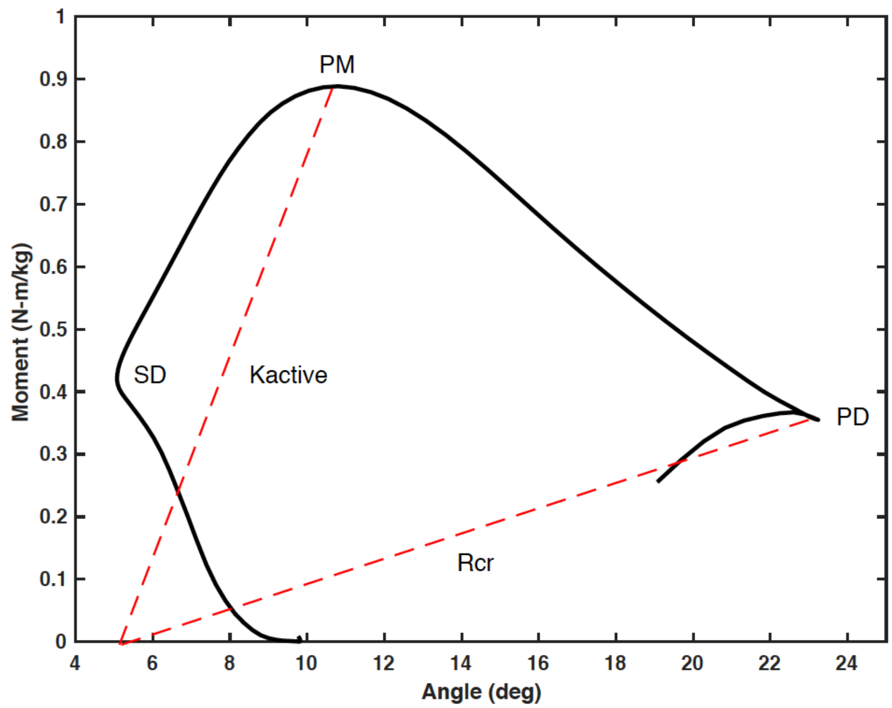

Figure 4.1. Depiction of methodology for metatarsophalangeal joint stiffness calculations. SD = start dorsiflexion, PM = peak moment, PD = peak dorsiflexion.

Instantaneous stiffness was calculated by taking the first derivative of the load-displacement curve. Footwear stiffness was calculated from Instron (Norwood, MA) mechanical testing results provided by the manufacturer (Brooks Sports, Seattle, WA). Passive footwear torque throughout stance was calculated by multiplying the footwear bending stiffness by the MTPJ angle. While the terms joint stiffness and resistance are used throughout this paper, these values represent the behavior of the foot-shoe complex about the MTPJ as modulated by passive and active internal and external structures.

Statistical Analysis

Individual repeated measures univariate analysis of variance (ANOVA, $\alpha < .05$) tests were used to analyze the main effect of running speed on active stiffness (K_{active}), critical resistance (R_{cr}), MTPJ range of motion, maximum MTPJ plantar flexor moment, and MTPJ plantar flexor moment at maximum dorsiflexion. Mauchly's test of Sphericity ($\alpha < .05$) was used to check for equal variance. Greenhouse-Geisser corrections were used when violations of sphericity were detected. Pairwise comparisons with Bonferroni adjustments ($\alpha = .05/10 = 0.005$) were used post-hoc to analyze main effects. Tests of within-subject contrasts were analyzed to determine the linearity of the change in measures across running speeds ($\alpha < .05$).

Results

There was a main effect of speed on critical resistance ($p < .001$), maximum MTPJ plantar flexor moment ($p < .001$), MTPJ plantar flexor moment at maximum dorsiflexion ($p < .001$), and MTPJ range of motion ($p = .013$) (Figures 4.2 and 4.3). Pairwise comparisons are presented in Table 4.1 and 4.2. There was no main effect of speed on active stiffness ($p = .094$).

Table 4.2. Dynamic angular stiffness and resistance (N-m/kg/deg) across running speeds; Mean ± SD.

Velocity (m/s)	$\mathbf{K_{active}}$	$\mathbf{R_{cr}}$
3.89	0.144 ± 0.102	0.009 ± 0.004^{c,d,e}
4.44	0.132 ± 0.114	0.010 ± 0.005^{c,d,e}
5.00	0.116 ± 0.103	0.013 ± 0.007^{a,b,d,e}
5.56	0.097 ± 0.038	0.015 ± 0.008^{a,b,c}
6.11	0.153 ± 0.074	0.017 ± 0.006^{a,b,c}

Pairwise comparisons showing significant differences (p <.05): ^{a} different from 3.89 m/s, ^{b} different from 4.44 m/s, ^{c} different from 5.00 m/s, ^{d} different from 5.56 m/s, ^{e} different from 6.11 m/s

Significant linear trends were detected for critical resistance ($p < .001$), maximum MTP plantar flexor moment ($p < .001$), MTPJ plantar flexor moment at maximum dorsiflexion ($p < .001$), and MTPJ range of motion ($p < .01$). Critical resistance, maximum MTP moment, and MTPJ moment at maximum dorsiflexion all increased as running speed increased.

Instantaneous MTPJ stiffness of the foot-shoe complex was much greater than that of the shoe throughout stance phase (Figure 4.4). The slow (3.89 m/s) and medium (5.00 m/s) speeds displayed plateau regions from approximately 55-70% stance whereas the fastest (6.11 m/s) speed did not exhibit a plateau region. Instantaneous stiffness fluctuated throughout stance at all speeds.

Table 4.3. Metatarsophalangeal joint range of motion and kinetics across running speeds; Mean ± SD.

Velocity (m/s)	Range of Motion (deg)	Maximum Moment (N-m/kg)	Moment at Maximum Dorsiflexion (N-m/kg)
3.89	17.1 ± 4.3[b,c,e]	0.500 ± 0.146[b,c,d,e]	0.143 ± 0.065[c,d,e]
4.44	18.5 ± 5.0[a]	0.570 ± 0.154[a,c,d,e]	0.168 ± 0.067[c,d,e]
5.00	18.7 ± 5.3[a]	0.674 ± 0.181[a,b,d,e]	0.215 ± 0.080[a,b,d,e]
5.56	19.0 ± 5.6	0.741 ± 0.180[a,b,c,e]	0.269 ± 0.096[a,b,c]
6.11	20.0 ± 6.0[a]	0.899 ± 0.210[a,b,c,d]	0.316 ± 0.066[a,b,c]

Pairwise comparisons showing significant differences ($p < .05$): [a] different from 3.89 m/s, [b] different from 4.44 m/s, [c] different from 5.00 m/s, [d] different from 5.56 m/s, [e] different from 6.11 m/s

Discussion

The purpose of this study was to investigate how dynamic angular stiffness and resistance about the MTPJ changes across running speeds. The load-displacement plot of the MTPJ exhibits a notable amount of hysteresis (Figure 4.2), resulting in the peak moment and peak dorsiflexion not occurring simultaneously as observed at the ankle and knee (Hamill et al., 2014; Jin and Hahn, 2018; Stefanyshyn and Nigg, 1998a). Because the peak moment and dorsiflexion occur at different time points during stance, two measures from the load displacement plot were quantified, active stiffness (K_{active}) and critical resistance (R_{cr}).

These results corroborate previous reports suggesting that lower extremity joint stiffness increases with running speed (Jin and Hahn, 2018; Stefanyshyn and Nigg, 1998a). Our hypothesis that MTPJ stiffness and resistance will increase with running speed was partially supported. Critical resistance (R_{cr}) increased with running speed ($p < .001$), but there was no effect of speed on K_{active} ($p = .096$). While the maximum MTPJ moment increased with running speed ($p < .001$), there was no increase in K_{active} due to differences in MTPJ angle excursion between speeds within this phase (Figure 4.2). While the phase of the load-displacement plot defined by K_{active} most closely represents that of quasi-stiffness spring behavior (Latash and Zatsiorsky, 1993), the energy stored during this phase is not quickly returned. Thus, we interpret K_{active} as a representative measure for the rate at which energy is delivered into the forefoot. We speculate that the lack of observed change in K_{active} may be a protective mechanism to maintain a lower loading rate of the bone and soft tissues. It may also be the result of an increase in MTPJ dorsiflexion angular velocity as joint angle excursion increased with running speed. Increased dorsiflexion angular velocity may have been the reason more dorsiflexion occurred by the time of peak MTPJ moment. The increase in R_{cr} with running speed was due to an increase in the MTPJ moment at maximum MTPJ dorsiflexion, as MTPJ range of motion also increased with running speed. The observed ratio of the change in R_{cr} across speeds may be of use to provide a guide for how to tune footwear stiffness to running speed. The linear trends observed for R_{cr}, MTPJ range of motion, and plantar flexor moment at maximum dorsiflexion suggest that effectively tuning footwear structures may be a linear function of speed.

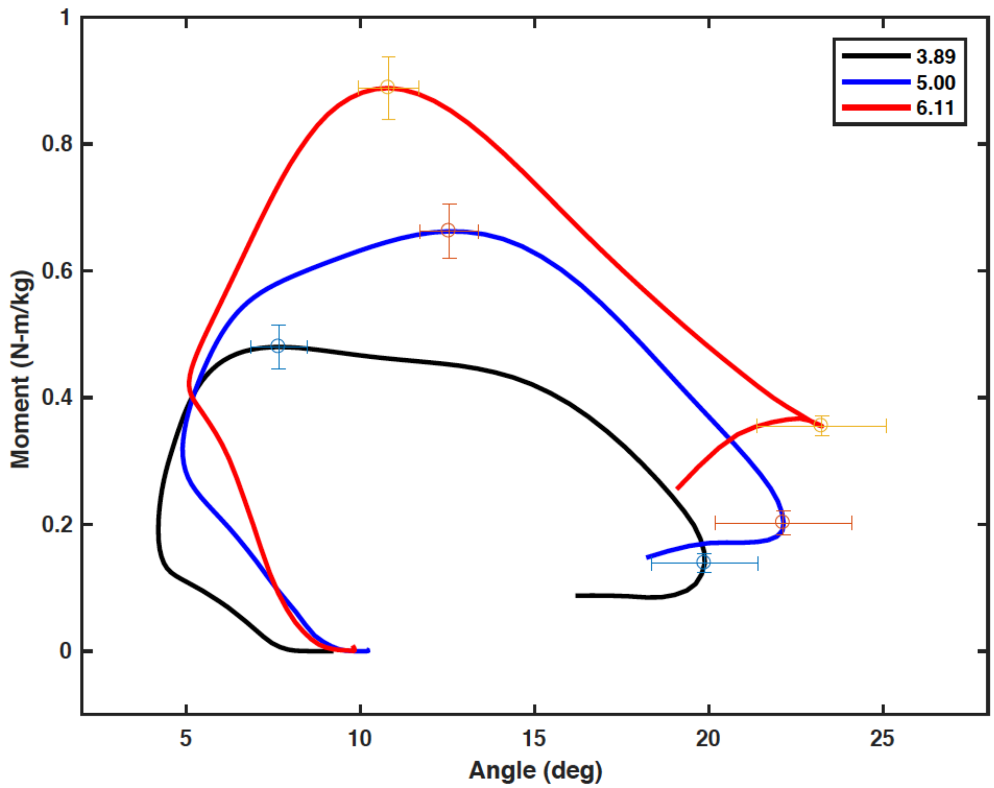

Figure 4.2. Average angular load-displacement of the metatarsophalangeal joint across speeds (m/s).

Metatarsophalangeal joint motion is controlled by the intrinsic and extrinsic toe-flexor musculature (Goldmann and Brüggemann, 2012; Mann and Inman, 1964; Mckeon et al., 2015; Miyazaki and Yamamoto, 1993). Indwelling electromyography studies of the intrinsic foot musculature have shown that their level of activation increases with gait speed (Kelly et al., 2014, 2015). Our results support that intrinsic foot muscle function increases with demand by providing evidence that the toe-flexors increase in mechanical function with running speed. This increase in mechanical function is evidenced by the larger MTPJ plantar flexor moments as running speed increases that have near negligible contribution from the shoe. Additionally, the increase in R_{cr} suggests an increase in resistance to dorsiflexion as running speed increases. Vertical ground reaction force increases with running speed (Clark and Weyand, 2014), resulting in a larger external dorsiflexion moment about the MTPJ that the toe-flexor musculature has to internally counteract (Goldmann and Brüggemann, 2012). As observed in the load-displacement curves across running speeds (Figure 4.2), there are regions of relative plateau during mid-stance at the slowest speed. These plateau regions were not observed at the faster speeds, where active loading and un-loading of the forefoot is occurring. This observation provides further evidence that in addition to larger peak MTPJ moments and R_{cr}, the general shape of the load-displacement plot reveals increased mechanical function of the forefoot.

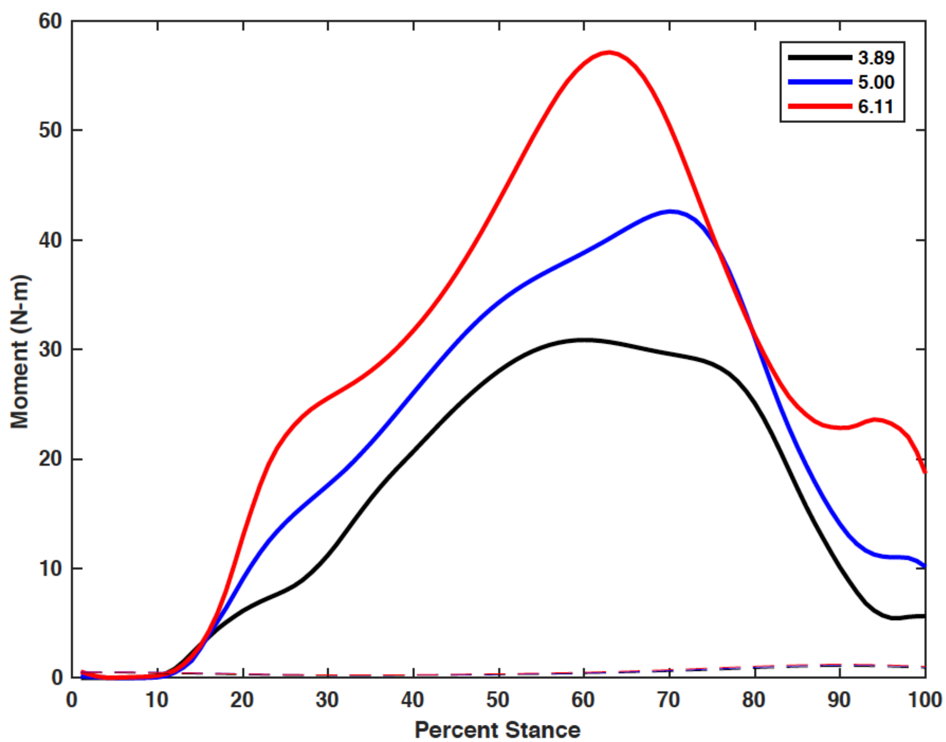

Figure 4.3. Average metatarsophalangeal joint moment across speeds (m/s). Dashed lines represent passive elastic torque of the footwear.

While R_{cr} did increase as a result of larger MTPJ moment, MTPJ range of motion also increased by 2.9° from the slowest to fastest running speeds. This increase in range of motion may be due to an increase in hip extension and ankle plantarflexion with running speed (Orendurff et al., 2018). A more extended hip will position the foot further posterior relative to the pelvis, requiring an increase in MTPJ dorsiflexion to maintain a surface contact area by which to transmit force. Secondly, force generating capacity of the toe-flexor musculature decreases when the ankle is plantar-flexed but increases with MTPJ dorsiflexion (Goldmann and Brüggemann, 2012). Thus the increase in MTPJ dorsiflexion with running speed may also be a mechanism to retain force-generating capacity of the toe-flexors (Goldmann and Brüggemann, 2012). On the contrary, the increase in dorsiflexion may also be due to the toe-flexor musculature being unable to eccentrically generate enough force and subsequent internal joint moment to resist MTPJ dorsiflexion. The intrinsic foot muscles can only generate approximately 6 N-m of torque as a result of their small pennation angles, muscle volume, and muscle moment arms to the MTPJ axis of rotation (Farris et al., 2019; Ledoux et al., 2001). In addition, the function of individual intrinsic foot muscles may vary between modulating MTPJ motion and stabilizing the medial and longitudinal arches of the foot (Tosovic et al., 2012). If the primary function of the intrinsic foot muscles is to stabilize the arches of the foot and aid in the transfer of force from the leg to the ground (Kelly et al., 2014, 2015), then this may explain the observed increase in MTPJ plantar flexor moment and joint range of motion.

While the precise physiological mechanism behind the increase in range of motion remains unknown, this kinematic change may have application to the design of footwear midsoles. An increase in joint range of motion will affect muscle-tendon unit properties such as force-length and force-velocity operating points. Recent footwear design efforts that make use of a curved carbon fiber plate in the midsole (Hoogkamer et al., 2017a) may also benefit by taking changes in MTP dorsiflexion range of motion into consideration. The observed increase in MTPJ range of motion with speed may serve as a framework for how to shape the curvature of an embedded plate if tuning footwear for optimal performance at a specific running speed. It has been demonstrated that the carbon fiber plates in footwear serves as a lever, not a spring (Hoogkamer et al., 2018). A plate shaped to the curvature of the foot will theoretically not store as much energy through bending, but rather serve as a lever and potentially facilitate a difference in perceived 'ride' of the shoe, defined as the peak anterior-posterior velocity of the center of pressure (Lam et al., 2018). A curved plate may result in a lower peak center of pressure velocity, resulting in a roll-through feeling from weight acceptance to push-off phases of stance. A curvature too concave or too shallow compared to natural MTPJ dorsiflexion range of motion may result in a user not perceiving as smooth of a ride in the shoe as they possibly could. Additionally, a curved plate may affect force-length and force-velocity operating points of the musculature about the MTPJ and ankle joints based upon gearing effects of a stiffened shoe (Takahashi et al., 2016; Willwacher et al., 2014). Due to the carbon fiber plate primarily acting as a lever and not a spring, it may be of relevance to methodically consider the curvature of the plate in footwear to match natural range of motion of the foot, and how these kinematics change with running speed.

The MTPJ moment is the net product of passive torque contribution from the shoe and active contribution from the musculoskeletal system (Figure 4.3). The musculoskeletal system appears to be the dominant contributor to the MTPJ moment, as the passive torque contribution from the shoe in this study did not exceed 2 N-m. The near negligible passive torque contributions from the shoe observed across speeds suggest that the use of a stiffer shoe that increases the amount of elastic energy stored and returned may be beneficial to running performance by increasing the passive contribution to the net angular impulse (Oh and Park, 2017). However, when increasing shoe stiffness it is important to note that restriction of MTPJ dorsiflexion may lead to altered joint mechanics, soft tissue function within the foot, and power generating capacity of the lower extremity (Bojsen-Møller and Lamoreux, 1979; Goldmann and Brüggemann, 2012; Oh and Park, 2017; Thewlis et al., 2012; Willwacher et al., 2013). A shoe that is too stiff may also increase the energetic cost of running and potentially facilitate detrimental biomechanical parameters such as increased trunk lean or contact time (Madden et al., 2015; Oh and Park, 2017; Roy and Stefanyshyn, 2006; Willwacher et al., 2014). While there appears to be merit to increasing the stiffness of footwear to increase the passive torque contribution of the shoe to the net angular impulse, care should be taken in ensuring that natural motion of the MTPJ is not inhibited. It may be of greater importance to tune the stiffness to running velocity. While the passive torque contribution is small, when seeking to optimize running performance we should not neglect to maximize contribution of the shoe to forward propulsion. Additionally, a larger torque will dorsiflex the shoe more. It may be important to maintain the shoe functioning as a lever by which a stiffer plate will resist the greater MTPJ moment as speed increases. The ratio of the increase in R_{cr}, reflective of the increase in MTPJ moment at maximum dorsiflexion, may serve as a framework for how to increase stiffness of a shoe with running speed.

Instantaneous MTPJ stiffness of the foot-shoe complex fluctuated throughout stance phase and was of greater magnitude than that of the shoe (Figure 4.4). The fluctuating instantaneous stiffness suggests that the foot dominates the stiffness of the foot-shoe complex and is also time dependent, as the shoe exhibits constant stiffness when dorsiflexed less than thirty degrees. These results are in agreement with previous efforts that have investigated contributions of the foot and shoe to forefoot stiffness (Oleson et al., 2005). The time dependent nature of the instantaneous MTPJ stiffness may be of use to footwear designers to improve energy storage and return dynamics in the midsole. The positive regions during the loading phase are when the MTPJ moment is increasing at a greater rate than the dorsiflexion angle, indicative of when the foot is performing work on the midsole. The negative regions are when the MTPJ moment is decreasing while dorsiflexion is still occurring, indicative of when energy should be returned. We speculate that these negative regions are when the shoe midsole is expanding after being compressed during the energy absorption phase over the first half of stance. Thus, the energy returned to the foot during these phases is likely to be oriented normal to the midsole. Combining the instantaneous stiffness data with center of pressure or pressure insole data may then provide insight as to where within the midsole to place compliant or resilient materials. An improved understanding of where within the midsole to position varying materials or how to guide the foot within the shoe based upon the location of where work is being performed on the midsole or energy is being returned may be of use to improve dynamics of the foot-shoe interaction.

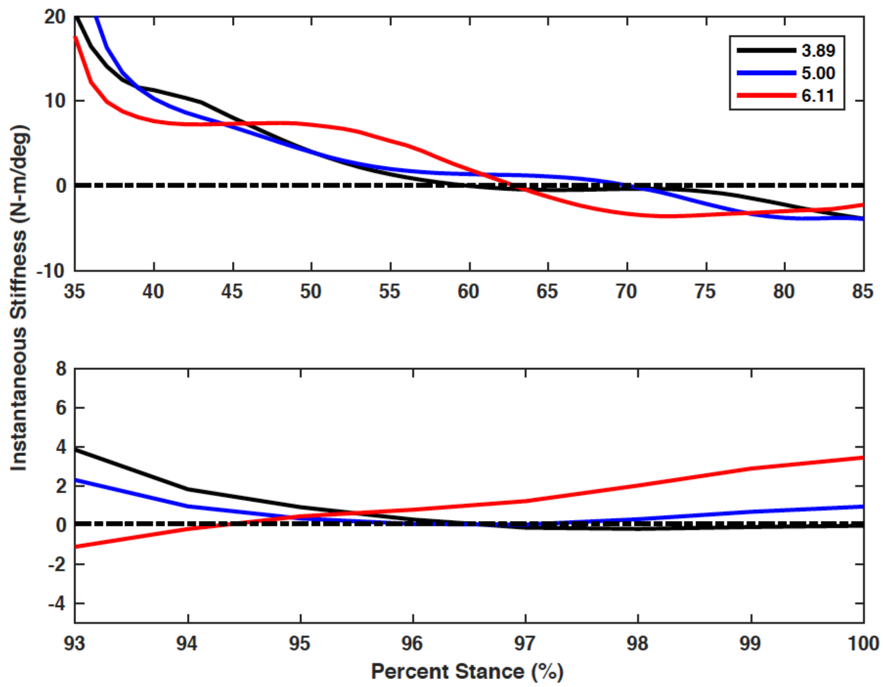

Figure 4.4. Instantaneous metatarsophalangeal joint stiffness during loading (top) and push-off (bottom). Horizontal line represents footwear stiffness.

As exhibited by the large amount of hysteresis present in the MTP joint load-displacement plot (Figure 4.2), the use of the term joint stiffness does not represent the same behavior as stiffness estimations at the ankle and knee. Stiffness implies that the system is in equilibrium and that elastic energy is being stored and returned (Latash and Zatsiorsky, 1993). Joint stiffness is often referred to as a quasi-stiffness behavior, yet the damper function of the MTPJ suggests that there is little to no spring behavior and that negligible strain energy stored during energy absorption is utilized for push-off. Thus, we adopted the term critical resistance to describe the resistance to external dorsiflexion comprised of contributions from the shoe, musculoskeletal system, and passive soft tissues such as the plantar aponeurosis. Individual contributions of the foot and shoe to this stiffness may also vary with speed as footwear and soft tissue elicit viscoelastic behavior.

Limitations

One limitation to this study is that the testing speed of the Instron (Norwood, MA) used to determine the bending stiffness of the footwear was unable to match the MTPJ plantarflexion velocity observed during running. Because running shoe midsoles are often comprised of viscoelastic materials, their resistance to deformation will be greater with the higher loading rates of running gait. In addition, the midsole is compressed in a linear fashion by the human body during running, whereas the shoe is unloaded during bending stiffness tests. It is possible that the shoe is dynamically stiffer than mechanical testing data would suggest and may contribute more to the stiffness of the foot-shoe complex and MTPJ plantar flexor moment than described.

Future work

Continued work in the field of exploring MTPJ kinetics should further in-depth analyze the effect of differences in foot strength, body mass, trunk lean, foot anthropometrics, ankle plantar flexor strength, etc. More work is also needed in understanding how the across speed relationship may change using footwear of varying LBS. Additionally, because the MTPJ mechanics are intrinsically controlled by the IFM, it would be of interest to investigate how increased IFM strength affects this relationship across speeds.

Conclusion

The present study provides evidence that the dynamic angular resistance about the MTPJ increases across running speeds. These findings provide a framework for how to potentially tune the longitudinal bending stiffness and shape of midsole structures in footwear to improve running performance. It is suggested that the bending stiffness of footwear should be increased as the passive torque contribution of the footwear was minimal. Further, it appears that an optimal footwear bending stiffness may be dependent upon running speed.

Bridge

This chapter established a framework for how MTPJ mechanics change with running speed. Additionally, they support that they foot is dominant modulator of mechanics of the foot-shoe complex. This chapter suggests that, (1) if an increase in IFM strength changes mechanics about the MTPJ, the expression should be not be masked by footwear, and (2) an argument that footwear LBS should be tuned to running speed. Chapter V will investigate how an increase in IFM strength affects joint mechanics and running economy. Chapters VI-VII will investigate the change in joint and gross level mechanics and running economy in response to varying footwear LBS across a range of speeds.

CHAPTER V

INCREASED TOE-FLEXOR MUSCLE STRENGTH DOES NOT ALTER METATARSOPHALANGEAL AND ANKLE JOINT MECHANICS OR RUNNING ECONOMY

This chapter is currently in review for publication. Evan Day designed the study and collected and analyzed the data. Michael E. Hahn provided mentorship and aided in study design, general oversight, and editing and finalizing the final manuscript.

Introduction

The metatarsophalangeal joint (MTPJ) contributes to lower limb power output and postural stability once the heel lifts off the ground when running (Goldmann and Brüggemann, 2012; Stefanyshyn and Nigg, 1997). During push-off, an external dorsiflexion moment about the MTPJ axis is counteracted internally by the intrinsic and extrinsic foot muscles (Goldmann et al., 2013; Goldmann and Brüggemann, 2012; Mann and Inman, 1964; Miyazaki and Yamamoto, 1993). The intrinsic foot muscles (IFM) also help control deformation of the longitudinal arch under loads much greater than body weight (Kelly et al., 2012, 2014, 2015). The IFM work alongside the plantar aponeurosis to modulate vertical linear stiffness of the foot to store and return strain energy contributing to metabolic savings during running (Kelly et al., 2015a, 2016, 2018a; Ker et al., 1987; McDonald et al., 2016; Stearne et al., 2016; Wager and Challis, 2016) and resist MTP dorsiflexion (Bojsen-Møller and Lamoreux, 1979). The IFM functionally work together with the extrinsic foot muscles, which have tendon insertions on the foot

(Mckeon et al., 2015). Altogether, the IFM contribute to control of whole body momentum during late stance (Goldmann and Brüggemann, 2012; Miyazaki and Yamamoto, 1993).

Training the foot through use of minimal footwear (Chen et al., 2016; Goldmann et al., 2013; Johnson et al., 2016; Miller et al., 2014; Potthast et al., 2005; Ridge et al., 2019) or resistance exercises (Goldmann et al., 2013; Hashimoto and Sakuraba, 2014) has been shown to increase strength and cross-sectional area of the IFM. Two previous studies have investigated the effects of increased IFM strength on athletic performance (Goldmann et al., 2013; Hashimoto and Sakuraba, 2014). Both studies reported an improvement in horizontal jump length, and Hashimoto & Sakuraba (Hashimoto and Sakuraba, 2014) reported a decrease in 50m dash time. However, Goldmann et al. (Goldmann et al., 2013) reported no change in ankle or MTPJ moments during running. While muscle hypertrophy occurs in response to resistance training for all modes of muscle contraction (Adams et al., 2004), the aforementioned studies included only one mode of muscle contraction; isometric or concentric. A more holistic program involving concentric, eccentric, and isometric contractions may result in more beneficial adaptations.

Altering the longitudinal bending stiffness (LBS) of footwear via carbon fiber plates can affect gait mechanics and running economy (Hoogkamer et al., 2018; Madden et al., 2015; Oh and Park, 2017; Roy and Stefanyshyn, 2006; Stefanyshyn and Nigg, 2000; Takahashi et al., 2016; Willwacher et al., 2013, 2014). Carbon fiber plates act in parallel with the IFM to stiffen the MTPJ complex (Hoogkamer et al., 2018; Oleson et al., 2005) resulting in reduced peak MTPJ dorsiflexion (Hoogkamer et al., 2018;

Kleindienst et al., 2005; Madden et al., 2015; Oh and Park, 2017; Smith et al., 2016; Stefanyshyn and Nigg, 2000; Willwacher et al., 2013), larger MTPJ plantar flexor moments (Kleindienst et al., 2005; Oh and Park, 2017; Smith et al., 2016; Willwacher et al., 2013), decreased ankle plantar flexion velocity (Madden et al., 2015; Takahashi et al., 2016), and a larger lever arm from the center of pressure to the MTPJ axis and other lower extremity joints (Takahashi et al., 2016; Willwacher et al., 2014). Oh and Park (Oh and Park, 2017) determined that the optimal LBS of footwear is similar to the natural MTPJ rotational stiffness, defined as the ratio of the MTPJ moment to maximum MTPJ dorsiflexion. Stiffer footwear has also been shown to improve running economy (Hoogkamer et al., 2017a; Madden et al., 2015; Oh and Park, 2017; Roy and Stefanyshyn, 2006), an indicator of distance running performance (Hoogkamer et al., 2016). However, improvements in running economy appear to be subject-specific and may be attributable to individual differences in ankle plantar flexion strength, body mass, or foot morphology (Madden et al., 2015; Takahashi et al., 2016; Willwacher et al., 2014).

While alterations in LBS can elicit mechanical changes, it has been postulated that the foot dominates the overall stiffness of the foot-shoe complex about the MTPJ (Oleson et al., 2005). Effective foot length, defined as the anterior displacement of the center of pressure under the foot (Hansen et al., 2004), can be modulated by altered MTPJ kinematics via carbon fiber plates (Oh and Park, 2017; Willwacher et al., 2014) or increased IFM strength (Endo et al., 2002; Goldmann and Brüggemann, 2012). Competitive sprinters and distance runners tend to have longer toes, allowing for generation of a larger propulsive impulse by increasing the ground reaction force moment arm and contact time (Baxter et al., 2012; Lee and Piazza, 2009; Ueno et al., 2018). Increased toe joint stiffness of a prosthetic foot has also been shown to increase center of mass push-off work, further supporting the importance of toe mechanics for performance (Honert et al., 2018). Increasing effective foot length alters the gear ratio of the foot (Carrier et al., 1994), which decreases shortening velocity of the ankle plantar flexors (Lee and Piazza, 2009; Madden et al., 2015; Takahashi et al., 2016), requiring less metabolic energy for muscle contraction (Fletcher et al., 2013; Fletcher and MacIntosh, 2017). Despite the apparent benefits of a longer effective foot length, it has been argued that the short toes of humans are an evolutionary adaptation that decreases negative work about the forefoot, potentially improving metabolic cost and allowing humans to run long distances more economically (Rolian et al., 2009). However, performing negative work is not as energetically costly as positive work (Abbot et al., 1952), and the metabolic savings from altered gearing may outweigh the savings from performing less negative MTPJ work.

Despite the large amount of research on IFM strengthening, it remains uninvestigated how increased IFM strength affects MTPJ and ankle joint mechanics and running economy. Previous work suggests that strengthening of the IFM increases the effective foot length (Endo et al., 2002; Goldmann and Brüggemann, 2012), which may result in similar changes in gait mechanics as observed with increased prosthetic toe joint stiffness (Honert et al., 2018), longer toes (Baxter et al., 2012; Lee and Piazza, 2009; Ueno et al., 2018), or carbon fiber plates in footwear (Takahashi et al., 2016; Willwacher et al., 2014). We hypothesized that increased toe-flexor strength will decrease MTPJ dorsiflexion range of motion and ankle plantar flexion velocity, and will increase angular

MTPJ stiffness, MTPJ and ankle moments, and contact time. Secondly, we hypothesized that these changes will result in decreased VO2 and metabolic rate at representative training and racing paces.

Methods

Recruitment

Twenty-three competitive distance runners (8 female) were recruited for this study (Table 5.1). Inclusion criteria consisted of a 5000m personal best under 18:00 (male) and 20:00 (female), no lower extremity injury in the previous six months, and currently running over 30 miles/week. Participants provided informed consent prior to data collection. This study was approved by the Institutional Review Board at the University of Oregon.

Table 5.1. Participant characteristics (Mean ± SD)

Sex	Age (yr)	Height (cm)	Mass (kg)	Weekly Mileage (km)	5000m Best (min:sec)
Male	26 ± 10	179 ± 9	66 ± 9	85 ± 24	16:17 ± 0:56
Female	27 ± 7	166 ± 6	56 ± 6	69 ± 16	18:38 ± 1:16

Study Design and Experimental Protocol

Participants visited the lab on three-occasions; at baseline, five weeks, and ten weeks; and were randomly assigned to one of two groups; IFM training (experimental) or the control group. The control group was not prescribed any additional strengthening protocol to augment their current training. Each visit involved assessment of IFM strength, running biomechanics, and running economy.

Data Collection

Strength Assessment. Intrinsic foot muscle isometric strength was assessed for the right leg using a novel testing apparatus (Figure 5.1). Participants sat in a Biodex dynamometer (Biodex Medical Systems, Shirley, NY) with their right leg positioned as if performing ankle strength assessments. A custom hinge-plate with a mounted protractor was attached securely to the Biodex foot plate for the participant to position their right foot with the transverse axis of the MTPJ aligned over the hinge. An inelastic strap attached to a strain gauge mounted in the wall behind the Biodex chair was wrapped around the participants' toes to provide resistance during isometric strength testing. The strain gauge voltage was amplified using a SparkFun HX711 (SparkFun Electronics, Niwot, CO) load cell amplifier and powered via an Arduino Mega2560 microcontroller sampling at 25 Hz. Test-retest of the set-up for five individuals showed strong reproducibility ($r = .997$). Due to the influence of extrinsic toe-flexors, relative ankle angle was controlled for (Goldmann and Brüggemann, 2012). Toe-flexor muscle strength was assessed in two positions: ankle 0° with the MTP 45° dorsiflexed, and ankle 20° plantar flexed with the MTP 25° dorsiflexed. The first position was chosen as it is the orientation in which the IFM can generate the largest isometric moment about the MTPJ (Goldmann and Brüggemann, 2012). The second position was chosen to mimic the average position of the ankle and MTPJ near terminal stance when MTPJ plantar flexion is about to begin, and thus when the IFM will contribute positive work (Hamill et al., 2014; Roy and Stefanyshyn, 2006). Participants were instructed to push as hard as possible into the band with their toes. If there was noticeable change in ankle angle or a large variance in force output, the trial was thrown out and repeated. Maximum voluntary contractions were three seconds in duration and performed three times with five seconds rest between each contraction.

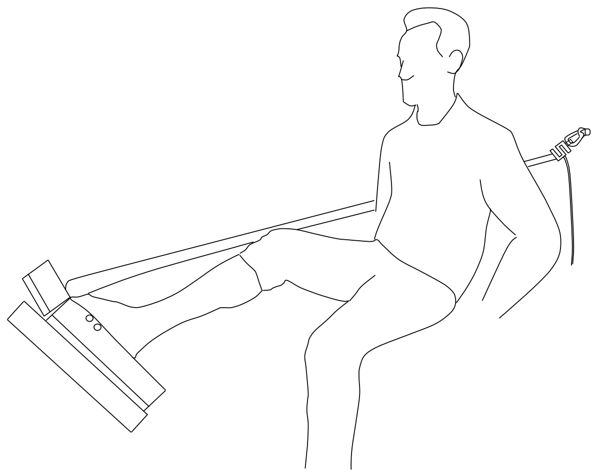

Figure 5.1. Novel IFM strength testing apparatus. The participant is seated on a Biodex.

Running Biomechanics. Upon completion of IFM strength assessments, participants underwent running biomechanics assessment at 14, 16, 18, 20, and 22 km/hr (3.89, 4.44, 5.00, 5.56, 6.11 m/s). These speeds were chosen as they represent relevant training and racing paces for competitive distance runners. Three-dimensional marker coordinate and ground reaction force data were collected using an eight-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA) and force instrumented treadmill (Bertec Corp., Columbus, OH). Motion data were collected at 200 Hz and ground reaction force data were collected at 2000 Hz. Running speed conditions were completed in ascending order. Data were extracted for ten strides at each velocity. Rest between trials was self-selected. A bilateral marker set consisting of 41 retro-reflective markers defining nine segments (forefoot, rearfoot, shank, thigh, pelvis) was used. A two-segment foot model was defined by placing markers on the forefoot and calcaneus (Goldmann et al., 2013). Windows were cut in the shoe uppers to place markers directly on the foot (Bishop et al., 2014). Individual retro-reflective markers were placed on the medial and lateral malleoli, medial and lateral femoral epicondyles, left and right greater trochanter, left and right posterior superior iliac spines, and the sacrum. Quadrad marker clusters were placed on the lateral aspects of the shank and thigh. Participants performed a static trial after which markers were then removed from the medial malleoli and femoral epicondyles so that they did not interfere with running movements. Participants all wore the same neutral running footwear (Brooks Launch 3) to eliminate the potential effects of varied LBS on MTP joint mechanics.

Metabolics. Upon completion of running biomechanics assessment, participants underwent a ramped stage metabolic assessment on a high speed treadmill (Woodway, Waukesha, WI) set to 1% grade (Jones and Doust, 1996). The protocol began at 14 km/hr and increased by 2 km/hr every three minutes to the individuals approximate 5000m race pace. While such running velocities would likely result in participants not remaining at sub-maximal levels (RER < 1.0), these faster velocities were included to assess potential improvements in running economy in response to the IFM training protocol. Measures of VO2 and VCO2 were taken with an open-circuit expired-gas analysis system (Parvomedics TrueOne 2400, Sandy, UT). All participants wore the same footwear as in the biomechanics assessment to eliminate potential effects of varied LBS on running economy.

Dynamic Exercise Training Protocol. Experimental group participants were provided a Hacky sack and set of resistance bands with written instructions for specific exercises. The lead researcher demonstrated the exercises at the first visit. Experimental group participants completed the prescribed protocol consisting of isometric, concentric, and eccentric exercises three times per week (Table 5.2). The short foot exercise, band curl, and foot curl were adopted from previous studies (Goldmann et al., 2013; Hashimoto and Sakuraba, 2014; Lynn et al., 2012); however, equipment was modified to help with ease of home-use. Participants were instructed to change the relative position of their ankle angle each training session between a plantar flexed, neutral, and dorsiflexed position. Training logs were collected from the experimental and control groups and analyzed on a weekly basis to account for any varying amount of training stimuli between groups, and to address participant feedback if necessary.

Table 5.2. Dynamic exercise training protocol utilized by the experimental training group

Exercise	**Repetitions**	**Equipment**
Range of motion	30	None
Foot curl	3 x 10	Hacky sac
Eccentric band curl	3 x 10	Resistance band
Short foot	3 x 10	None
Concentric band curl	3 x 10	Resistance band

Data Analysis

Maximum isometric strength was calculated by averaging the peak force across the three maximum voluntary contraction trials. Data were output from the strain gauge in pounds, then converted to Newtons and normalized to body mass (N/kg).

A custom MATLAB (version R2016b; MathWorks, Natick, MA) program was used to calculate joint kinematics and kinetics. Variables were calculated for stance phase, defined as when the vertical ground reaction force exceeded 5% body mass. Raw marker coordinate data were filtered using a zero-lag, fourth-order low pass Butterworth filter with a 20Hz cutoff frequency (Willwacher et al., 2013). The same cut-off frequency was used for ground reaction force data. Joint angles were calculated using an Euler/Cardan rotation order of flexion/extension, abduction/adduction, and internal/external rotation. Sagittal plane joint angles and moments were used for subsequent analyses.

Metatarsophalangeal joint moments were estimated using an inverse dynamics approach. The MTP joint was modeled as a hinge axis defined by the vector connecting the 1st and 5th metatarsal markers (Smith et al., 2012). Ground reaction force moment arm was estimated as the perpendicular distance between the center of pressure and the MTPJ oblique axis (Chapter III, Day and Hahn, 2019). Resultant forces and moments about the MTPJ were considered zero until the center of pressure passed anterior to the MTPJ axis (Stefanyshyn and Nigg, 1997). Inertial effects of the forefoot were considered negligible (Stefanyshyn and Nigg, 1997) while inertial parameters of the foot were modeled accordingly (de Leva, 1996). Joint moments were resolved in the proximal segment coordinate system. Kinematic and kinetic data were re-sampled to 101 data points per stance phase for time-normalized analysis. Joint excursion data were analyzed as opposed to peak angles to improve between-day reliability (Ferber et al., 2002). Maximum MTPJ and ankle moments and MTPJ moment at peak MTPJ dorsiflexion were analyzed. Metatarsophalangeal joint rotational stiffness was defined from the MTPJ load-displacement plot (Figure 2). Angular resistance (R_{cr}) was calculated as follows (Chapter IV, Oh and Park, 2017).

$$R_{cr} = \frac{MTP\ moment\ (at\ max\ dorsiflexion) - 0}{\theta_{MTP\ max} - \theta_{MTP\ min}}$$

Metabolic data were analyzed by averaging submaximal (RER < 1.0) VO2 consumption, VCO2 production, and RER over the last ninety seconds of each stage. The VO2 values were normalized to body mass (ml/kg/min). Metabolic rate (W/kg) was quantified using the Brockway equation (Brockway, 1987). While running economy is traditionally reported as a mass-standardized VO2, metabolic rate was included as it accounts for substrate utilization (Fletcher et al., 2009).

Statistical Analysis

A linear interaction comparison analysis of variance (ANOVA, $\alpha < 0.05$) was conducted in SPSS (version 23, IBM, Armonk, NY) to determine the change in IFM strength within-groups at baseline, five weeks, and ten weeks. Mixed-model analysis of variance (ANOVA, $\alpha < 0.05$) tests with within-subject factor of time and between-subject factor of group were used to analyze biomechanical and metabolic variables. Pairwise comparisons with Bonferroni adjustments ($\alpha = .05/6 = .0083$) were used *post-hoc* to further analyze main effects. Gait mechanics at 14, 18, and 22 km/hr and running economy and metabolic rate at 14 km/hr were chosen for reporting. Partial eta squared (η^2) was used to determine effect sizes, defined as small ($\eta^2 = .01$), medium ($\eta^2 = .06$), or large ($\eta^2 = 0.14$) (Cohen, 1988).

Results

Twenty-three participants were initially recruited into this longitudinal study. Three participants dropped out after reporting non-running related injury or sickness that resulted in missing more than one week of training, leaving eleven participants in the experimental group and nine in the control group. Due to equipment failures, gait mechanics data analyzed at 14 and 18km/hr were from nine participants in the experimental group and seven in the control group. Gait mechanics data at 22 km/hr were analyzed on eight participants in the experimental group and five in the control group due to some of the participants not feeling comfortable running on a treadmill at such a high speed. Running economy at 14 km/hr was analyzed on nine participants in the experimental group and seven in the control group that exhibited sub-maximal RER values. Throughout the remainder of this chapter, toe-flexor muscles and IFM will be used interchangeably to describe the muscles that produce MTPJ plantar flexion.

The linear interaction comparison revealed a significant group*time interaction (p = .002, η^2 = .285). Toe-flexor strength significantly increased (*p* = .006, η^2 = .350) in the experimental group across time (Figure 5.2). Experimental group toe-flexor strength was 16% greater at week five, and 27% greater at week ten, compared to baseline. Toe-flexor strength in the control group did not change across time (*p* = .607, η^2 = .074).

Metatarsophalangeal joint angular resistance (R_{cr}) did not change across time in the experimental or control group for running at 14 km/hr (*p* = .493, η^2 = 0.042), 18 km/hr (*p* = .404, η^2 = 0.063), or 22 km/hr (*p* = .083, η^2 = 0.203) (Tables 5.3, 5.4, 5.5). A main effect of time on MTPJ range of motion was detected (*p* = .027, η^2 = 0.227) at 14 km/hr. Pairwise comparisons revealed significant MTPJ range of motion differences in the control group from baseline to week 10 (*p* = .048, η^2 = 0.425) but no change in MTPJ range of motion in the experimental group (*p* = .514, η^2 = .056). There was no main effect of time on MTPJ range of motion at 18 km/hr (*p* = .396, η^2 = 0.064) or 22 km/hr (*p* = .429, η^2 = 0.074). Maximum MTPJ moment did not change across time in either group at 14 km/hr (*p* = .099, η^2 = 0.152), 18 km/hr (*p* = .582, η^2 = 0.038), or 22 km/hr (*p* = .780,

$\eta^2 = 0.022$), nor did MTPJ moment at maximum dorsiflexion at 14 km/hr ($p = .455$, $\eta^2 = 0.046$), 18 km/hr ($p = .464$, $\eta^2 = 0.053$), or 22 km/hr ($p = .191$, $\eta^2 = 0.140$).

Peak ankle plantar flexor moment did not change across time in either group at 14km/hr ($p = .754$, $\eta^2 = 0.020$), 18 km/hr ($p = .537$, $\eta^2 = 0.036$), or 22 km/hr ($p = .611$, $\eta^2 = 0.044$). Peak ankle plantar flexion velocity did not change across time in either group at 14 km/hr ($p = .459$, $\eta^2 = 0.042$), 18 km/hr ($p = .279$, $\eta^2 = 0.087$), or 22 km/hr ($p = .748$, $\eta^2 = 0.026$). A main effect of time on contact time was detected at 18 km/hr ($p < .001$, $\eta^2 = 0.484$). Pairwise comparisons revealed significant contact time differences in the control group, with a mean 0.01 second difference ($p < .001$, $\eta^2 = 0.866$). Contact time did not change across testing sessions in either group at 14 km/hr ($p = .249$, $\eta^2 = 0.095$) or 22 km/hr ($p = .583$, $\eta^2 = 0.048$). Running economy (VO2/ml/kg) did not change across time in either group ($p = .340$, $\eta^2 = 0.062$), nor did metabolic rate ($p = .725$, $\eta^2 = 0.069$).

Table 5.3. Changes in gait mechanics across time and speeds at 14 km/hr. P value represents within-subjects main effect of time. Exp = experimental group, Con. = control group

	Group	Baseline	Week 5	Week 10	P
MTPJ ROM (deg)	Exp.	19.3 ± 3.4	19.3 ± 4.0	20.0 ± 5.4	**.027**
	Con.	16.4 ± 4.5[c]	16.0 ± 5.2[c]	19.0 ± 4.0[a,b]	
Max MTP Mom. (N-m/kg)	Exp.	0.56 ± .14	0.64 ± .19	0.59 ± .23	.099
	Con.	0.47 ± .12	0.52 ± .13	0.55 ± .12	
MTP mom. at max dorsi. (N-m/kg)	Exp.	0.15 ± .06	0.16 ± .08	0.16 ± .08	.455
	Con.	0.17 ± .06	0.19 ± .05	0.22 ± .08	
Rcr (N-m/kg/deg)	Exp.	.008 ± .004	.009 ± .006	.010 ± .008	.493
	Con.	.010 ± .004	.013 ± .006	.012 ± .007	
Max Ankle Mom. (N-m/kg)	Exp.	3.40 ± .39	3.21 ± .67	3.26 ± .51	.754
	Con.	3.33 ± .59	3.44 ± .58	3.54 ± .38	
Peak Ankle Pflex Vel. (rad/sec)	Exp.	13.9 ± 1.6	14.0 ± 1.4	14.1 ± 1.5	.455
	Con.	14.4 ± 3.9	13.1 ± 0.9	13.4 ± 2.5	
Contact time (sec)	Exp.	0.21 ± .02	0.21 ± .02	0.21 ± .02	.249
	Con.	0.20 ± .01	0.20 ± .01	0.20 ± .01	

Pairwise comparisons showing significant differences: [a] different from baseline, [b] different from week 5, [c] different from week 10

Table 5.4. Changes in gait mechanics across time and speeds at 18 km/hr. P value represents within-subjects main effect of time. Exp = experimental group, Con. = control group

	Group	Baseline	Week 5	Week 10	P
MTPJ ROM (deg)	Exp.	21.1 ± 4.7	22.4 ± 5.8	22.7 ± 5.6	.396
	Con.	17.1 ± 5.4	17.5 ± 6.5	18.6 ± 2.7	
Max MTP Mom. (N-m/kg)	Exp.	0.77 ± .17	0.83 ± .22	0.77 ± .20	.582
	Con.	0.63 ± .15	0.61 ± .17	0.60 ± .10	
MTP mom. at max dorsi. (N-m/kg)	Exp.	0.22 ± .09	0.27 ± .10	0.23 ± .10	.464
	Con.	0.20 ± 0.08	0.24 ± .09	0.21 ± .08	
Rcr (N-m/kg/deg)	Exp.	.011 ± .006	.011 ± .008	.012 ± .009	.564
	Con.	.013 ± .007	.017 ± .010	.012 ± .005	
Max Ankle Mom. (N-m/kg)	Exp.	3.60 ± .59	3.35 ± .81	3.70 ± .50	.537
	Con.	3.55 ± .55	3.54 ± .36	3.55 ± .44	
Peak Ankle Pflex Vel. (rad/sec)	Exp.	17.9 ± 2.1	17.8 ± 1.3	18.3 ± 1.3	.279
	Con.	16.1 ± 1.1	16.0 ± 1.2	14.7 ± 1.7	
Contact time (sec)	Exp.	0.18 ± .01	0.18 ± .01	0.17 ± .01	**< .001**
	Con.	0.17 ± .01[c]	0.17 ± .01[c]	0.19 ± .01[a,b]	

Pairwise comparisons showing significant differences: [a] different from baseline, [b] different from week 5, [c] different from week 10

Table 5.5. Changes in gait mechanics across time and speeds at 22 km/hr. P value represents within-subjects main effect of time. Exp = experimental group, Con. = control group

	Group	Baseline	Week 5	Week 10	P
MTPJ ROM (deg)	Exp.	25.5 ± 4.1	24.1 ± 2.1	25.1 ± 4.9	.429
	Con.	17.9 ± 6.6	16.4 ± 4.3	18.7 ± 5.4	
Max MTP Mom. (N-m/kg)	Exp.	0.99 ± .18	1.03 ± .25	0.99 ± .26	.780
	Con.	0.86 ± .28	0.90 ± .18	0.91 ± .16	
MTP mom. at max dorsi. (N-m/kg)	Exp.	0.32 ± .05	0.32 ± .04	0.34 ± .04	.191
	Con.	0.32 ± .08	0.43 ± .12	0.36 ± .12	
Rcr (N-m/kg/deg)	Exp.	.014 ± .003	.014 ± .001	.014 ± .004	.083
	Con.	.020 ± .006	.028 ± .011	.020 ± .008	
Max Ankle Mom. (N-m/kg)	Exp.	4.21 ± .63	3.92 ± .51	4.51 ± .54	.611
	Con.	4.29 ± .74	4.35 ± .84	4.47 ± .55	
Peak Ankle Pflex Vel. (rad/sec)	Exp.	21.1 ± 2.4	21.1 ± 1.7	21.5 ± 1.5	.748
	Con.	17.5 ± 0.6	18.1 ± 0.6	17.0 ± 2.5	
Contact time (sec)	Exp.	0.15 ± .01	0.15 ± .01	0.15 ± .01	.583
	Con.	0.15 ± .01	0.15 ± .01	0.15 ± .01	

Pairwise comparisons showing significant differences: [a] different from baseline, [b] different from week 5, [c] different from week 10

Discussion

The purpose of this study was to investigate how increased toe-flexor strength affects MTPJ and ankle joint mechanics and running economy. We hypothesized that increased toe-flexor strength would result in decreased MTPJ range of motion, larger peak MTPJ moment and angular resistance, larger peak ankle plantar flexor moment, decreased ankle plantar flexion velocity, and lower steady-state VO2 and metabolic rate. While toe-flexor strength did significantly increase in the experimental group (Figure 5.2), there were no changes in gait mechanics or metabolic measures.

Participants performed isometric, concentric, and eccentric exercises (Table 5.2), and improved toe-flexor strength by 27% on average over the 10-week study. Compared to previous studies utilizing low-resistance concentric or isometric exercise programs that improved toe-flexor strength by 50% and 60-70% (Goldmann et al., 2013; Hashimoto and Sakuraba, 2014), our participants showed smaller increases in toe-flexor strength. One possible explanation for our observed lack of change in gait mechanics could be due to the increase in toe-flexor strength not being large enough to elicit changes to the mechanics of the foot and ankle. However, Goldmann et al. also reported no change in MTPJ or ankle joint moments during running (Goldmann et al., 2013). Compared to horizontal jumping where the trunk leans anteriorly over the forefoot and stronger IFM can depress the toes to increase the ground reaction force moment arm to increase propulsive impulse, such changes may not be natural during running due to a more upright trunk posture (Goldmann et al., 2013).

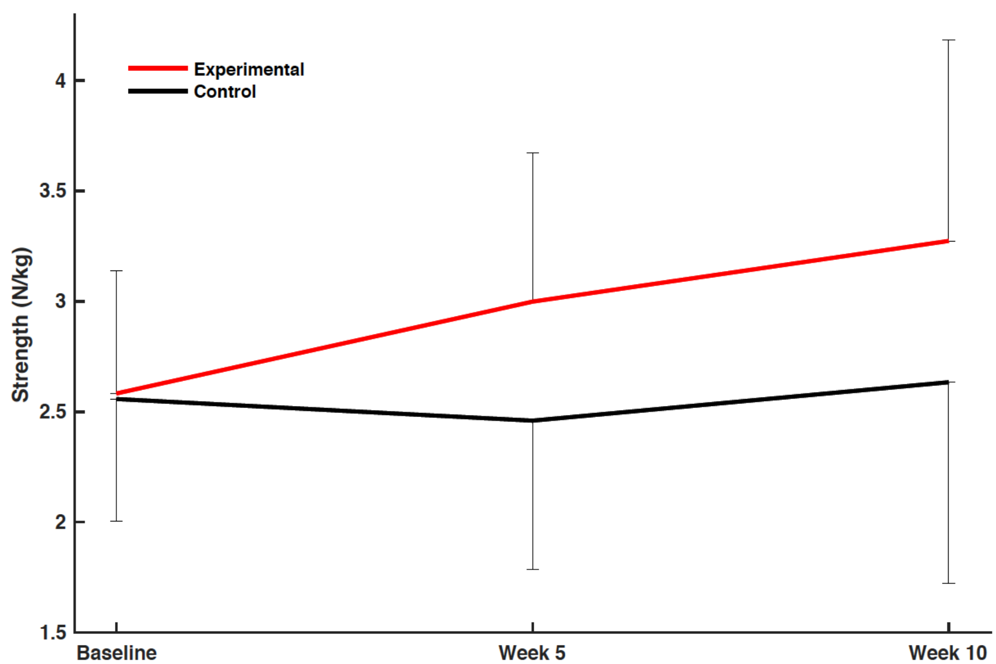

Figure 5.2. Toe-flexor strength across time (Mean ± SD).

Our hypothesis that increased foot muscle strength would elicit similar changes in MTPJ and ankle mechanics to that of using carbon fiber plated footwear was not supported. One explanation for the inability of the IFM to reduce MTPJ dorsiflexion or increase MTPJ plantar flexor moments may be that these muscles do not have enough torque generating capability about the MTPJ to do so. While the IFM do have a low fiber to muscle length ratio (Kura et al., 1997), they have small physiological cross-sectional areas (Kura et al., 1997), small pennation angles (Ledoux et al., 2001), and a small moment arm from the MTPJ axis (Rolian et al., 2009). Altogether, these characteristics result in the IFM being able to generate approximately 6 N-m of torque about the MTPJ, a much smaller torque than the external MTPJ dorsiflexion moment they counteract (Farris et al., 2019). The IFM architecture and lack of mechanical advantage about the MTPJ may partially explain why we did not observe an increase in maximum MTPJ moment as a result of stronger IFM.

A recent study demonstrated that knockout of active regulation of the IFM results in a decrease in MTPJ stiffness due to a decrease in maximum MTPJ moment, defined as the slope of the moment-angle curve from the occurrence of maximum MTPJ moment to maximum MTPJ dorsiflexion (Farris et al., 2019). This approach demonstrates the ability of the forefoot to act as a stiff lever for push-off. We quantified MTPJ stiffness (referred to as angular resistance in this dissertation) in a different manner and observed no change in maximum MTPJ moment or range of motion. Thus, we conclude that stronger IFM are unable to effectively stiffen the forefoot to act as an enhanced lever during running.

The foot acts as a viscous spring-damper during running, contributing large amounts of recoiled strain energy during push-off (Kelly et al., 2018b; Ker et al., 1987).

A large majority of strain energy is stored in structures supporting the longitudinal arch, notably the IFM and plantar aponeurosis (Hicks, 1954; Kelly et al., 2014, 2015). Increasing MTPJ dorsiflexion increases arch elongation and subsequent energy storage and return via the windlass mechanism (Welte et al., 2018). Another potential reason there was no decrease in MTPJ range of motion could be that reducing MTPJ dorsiflexion is not energetically beneficial due to a reduced effectiveness of the windlass mechanism (Welte et al., 2018).

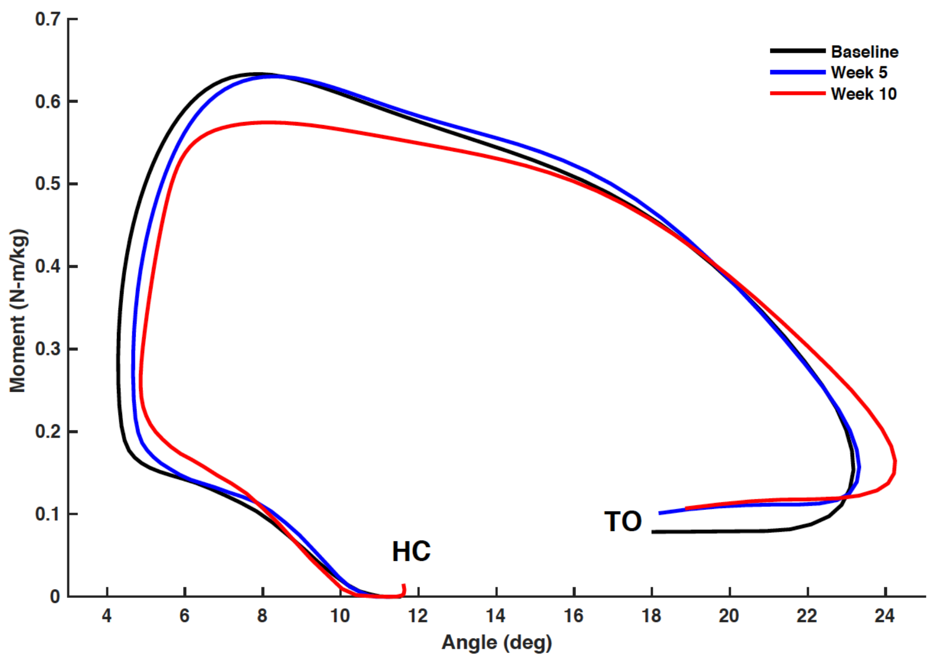

Figure 5.3. Average angular MTP load-displacement across time for running at 14 km/hr. HC = heel contact; TO = toe-off

The dearth of change in gait mechanics was accompanied by no change in VO2 or metabolic rate. Knockout of active regulation of the IFM can cause alterations in stride frequency and hip joint work, but no change in metabolic rate (Farris et al., 2019). Our results compliment these findings by showing that increased IFM strength does not affect metabolic rate either. This may be explained by the lack of observed changes in gait mechanics, or smaller contribution of the foot to net mechanical work performed on the center of mass, compared to the other primary lower extremity joints (Riddick et al., 2019). Additionally, longitudinal arch compression is not primarily supported by IFM contraction, thus it is likely that energy storage and return dynamics in the longitudinal arch were unaffected by an increase in IFM strength.

Table 5.6. Changes in metabolic outcomes across time and group at 14 km/hr. P value represents within-subjects main effect of time.

	Group	Baseline	Week 5	Week 10	P
VO2	Exp.	44.6 ± 2.7	45.0 ± 3.1	44.9 ± 2.9	.340
(ml/kg/min)	Con.	44.6 ± 1.6	45.3 ± 2.0	45.0 ± 1.9	
Metabolic	Exp.	14.4 ± 0.8	14.6 ± 0.9	14.5 ± 0.9	.294
Rate (W/kg)	Con.	14.5 ± 0.5	14.7 ± 0.6	14.6 ± 0.6	
RER	Exp.	0.85 ± .04	0.85 ± .04	0.85 ± .05	.725
	Con.	0.87 ± .06	0.87 ± .08	0.86 ± .07	

Stiffening of the foot-shoe complex via carbon fiber plates has been shown to improve running economy by altering ankle joint energetics (Hoogkamer et al., 2017a; Madden et al., 2015). Another reason for the lack of change in metabolics may be due to our observation of no reduction in peak ankle joint plantar flexion velocity, as greater muscle shortening velocities increase the energetic cost of muscle contraction (Fletcher et al., 2013; Fletcher and MacIntosh, 2017). We observed no change in MTPJ range of motion and no increase in contact time, which most likely influenced the lack of change in ankle plantar flexion velocity, due to a reduced gearing effect on the ground reaction force moment arm. Our findings of increased muscle strength not influencing a change in gait mechanics corroborate previous reports of no change in ankle kinematics in response to increased ankle plantar flexor strength and Achilles tendon stiffness (Albracht and Arampatzis, 2013). Runners may prefer to stay within their preferred movement path even after a change in muscle strength or muscle-tendon unit structure, similar to the response to wearing different footwear (Nigg et al., 2017). If an increase in IFM strength could alter MTPJ mechanics, it would be speculative to estimate the increase necessary to elicit such a response.

One potential method for intrinsically altering MTPJ mechanics could be stiffening of the plantar aponeurosis or IFM tendons. Greater MTPJ passive stiffness has been shown to correlate with improved running economy (Man et al., 2016). Kelly et al. (Kelly et al., 2018a) demonstrated that the flexor digitorum brevis contracts isometrically during arch compression, suggesting that energy storage and return in the IFM is primarily a function of the tendon structures and not muscle tissue. If the other IFM behave similarly, then perhaps increasing IFM strength is not an effective mechanism to alter MTPJ mechanics. Increasing Achilles tendon stiffness has also been shown to improve running economy (Albracht and Arampatzis, 2013), suggesting that tendon remodeling may be as or more important than muscle strengthening. At a minimum, tendon stiffness must increase with increased muscle strength, or else a faster, less efficient shortening of the muscle fascicles may occur (Albracht and Arampatzis, 2013; Lichtwark and Wilson, 2008). The IFM have a low fiber length to muscle length ratio, effectively reducing their natural excursion (Kura et al., 1997). Thus, the interplay between IFM tendon and plantar aponeurosis stiffness and the potential to passively alter MTPJ mechanics requires further investigation.

Limitations

This study was not without limitations. Foot length, arch height index, and IFM volume were not assessed. Information about structural changes of the foot and IFM would improve our ability to explain the lack of changes in gait mechanics. Due to the quick increase in strength and relatively low training load (compared to maximum voluntary contraction force for our participants), we assume that strength improvements were due to neurological adaptations and that structural changes may not have occurred (Goldmann et al., 2013; Hashimoto and Sakuraba, 2014; Komi et al., 1978). Secondly, there is inherent variability in assessing within-subject changes in running economy across time (Morgan et al., 1991). Participants were asked to not eat within two hours prior to data collection, and efforts were made for visits to occur at the same time of day to help eliminate confounding factors that can affect running economy. Lastly, isolation of the IFM from other extrinsic foot muscles for strength assessment is difficult, so we adopted the terminology of toe-flexor strength (Soysa et al., 2012). While toe-flexor strength increased in our participants, we cannot be sure that the recorded strength values were entirely due to IFM contributions.

Future Work

Future work investigating the influence of increased IFM strength on running performance should include a fatigue component. Additional insights regarding arch structure, intrinsic muscle volume, and navicular drop during stance phase would be of benefit to further understand the factors contributing to foot-ankle mechanics. While sagittal plane joint level mechanics and running economy did not change, variables associated with injury such as prolonged rearfoot eversion (Becker et al., 2018) may be of interest to investigate.

Conclusion

In conclusion, from our observations it appears that increased toe-flexor strength does not alter MTPJ and ankle joint mechanics or running economy. We speculate that this may be due to the lack of torque generating capability by the IFM and low mechanical advantage about the MTPJ, and limitations in IFM muscle architecture. While our results do not support the notion that increased toe-flexor strength may improve running economy and subsequent distance running performance, further research may be warranted in investigating the interplay between foot strengthening and footwear construction, fatigue resistance, as well as in injury prevention or rehabilitation.

Bridge

Participants in this study successfully increased their IFM strength. Results from this chapter addressed the question of how mechanics and metabolics change across running speeds in response to increased IFM strength. The next two chapters will now address how mechanics and metabolics change across speeds in response to varying footwear LBS.

CHAPTER VI

OPTIMAL FOOTWEAR BENDING STIFFNESS TO REDUCE METABOLIC COST IS RUNNING VELOCITY DEPENDENT

This chapter is currently unpublished. Evan Day designed the study and collected and analyzed the data. Michael E. Hahn provided mentorship and aided in study design, general oversight, and editing and finalizing the final manuscript.

Introduction

Recent advances in footwear engineering have been redefining the potential for footwear to influence distance running performance (Hoogkamer et al., 2017a, 2017b). Changes in shoe weight (Franz et al., 2012; Frederick, 1984; Hoogkamer et al., 2016), midsole material (Worobets et al., 2014), longitudinal bending stiffness (LBS) (Madden et al., 2015; Roy and Stefanyshyn, 2006), and subjective comfort (Luo et al., 2009) have all been demonstrated to influence metabolic cost during running, an indicator of performance (Hoogkamer et al., 2016).

Increased LBS alters mechanics of the metatarsophalangeal (MTPJ) and ankle joint during running (Willwacher et al., 2013, 2014). A newly developed shoe with a novel midsole material that stores and returns 87% energy and contains a stiff carbon fiber plate has been shown to reduce the energetic cost of running by 4% (Hoogkamer et al., 2017a). Joint level mechanical analysis reveals that the carbon fiber plate within the shoe acts as a lever rather than spring (Hoogkamer et al., 2018). Interpretation remains difficult however because with the interaction of the very high energy return foam and carbon fiber plate it is unknown how either characteristic behaves and contributes to mechanical or energetic changes in isolation. Flores et al. sought to address this gap (Flores et al., 2018), however a primary limitation to their study is that the foam used in the footwear only returned 62% energy and participants only ran at an average 10 km/hr, compared to the 14-18 km/hr for energetic cost analysis and 16 km/hr for joint level mechanics analysis previously reported by Hoogkamer and colleagues (Hoogkamer et al., 2017a, 2018).

No study has investigated the effect of altering only LBS at a range of running velocities. Previous findings have shown that sagittal plane MTPJ moment and stiffness increase linearly with running speed, which will influence the ability of the foot to bend a shoe about the forefoot break point (Chapter IV). Increasing LBS of a shoe in proportion to the increase in MTPJ moment or stiffness may be a mechanism to maximize the contribution of footwear to improving distance running performance (Chapter IV). Natural changes in mechanics of the foot and leg across speeds, notably an increase in MTPJ and ankle moments (Chaper IV, Kelly et al., 2018; Schache et al., 2011), may affect an optimal interaction between the biological limb and shoe.

The purpose of this study was to investigate the effects of varied LBS across a range of speeds. We investigated changes in spatiotemporal parameters, ground reaction forces, subjective comfort, and metabolic cost. Based upon previous literature (Flores et al., 2018; Hoogkamer et al., 2017a; Willwacher et al., 2013), we hypothesized that an increase in LBS would cause an increase in contact time and stride length, and a decrease in stride frequency and peak propulsive force across speeds. For comfort related metrics we hypothesized that total shoe comfort, measured by questionnaire responses regarding individual components of footwear, would be different across speeds, and that an increase in LBS would be more preferable at faster running speeds. Lastly, we hypothesized that at 14 km/hr a less stiff shoe would result in the lowest metabolic cost, whereas at 17 km/hr a stiffer shoe would exhibit the lowest metabolic cost.

Methods

Recruitment

Ten competitive male runners were recruited for this study (26 ± 6 years, 1.78 ± 0.04 m, 63.9 ± 4.0 kg, 101 ± 34 km/wk, 15:04 ± 0:38 (min:sec) 5000m personal best). To be included participants had to have a 5000m personal best under 16:00, no lower extremity injury in the previous six months, and currently running over 50 km/week. Participants provided written informed consent prior to data collection. This study was approved by the Institutional Review Board at the University of Oregon.

Study Design and Experimental Protocol

Participants visited the lab on two occasions. A screening test was completed on the first day to ensure that participants exhibited steady-state metrics running at 17 km/hr (RER < 1.0, VO2 and VCO2 deviate less than 10% during last two minutes) (McClave et al., 2003; Reeves et al., 2004). If participants exhibited being at steady-state, they were included in the study.

Day one consisted of running biomechanics assessments on an instrumented treadmill at three speeds using footwear of three different LBS values. Day two consisted of metabolic assessments on a high-speed treadmill at two speeds using footwear of three different LBS values. Subjective comfort questionnaires were completed by the participants for each speed by shoe combination.

Data Collection

Footwear Bending Stiffness Tests. Three shoe conditions (normal, stiff, very stiff) were tested in this study (Table 6.1). Longitudinal bending stiffness of the shoes was modified by inserting full length 3-D printed plates (Nylon 11) underneath the insole, matching the shape of the shoe foot bed (Epic React Flyknit size US10, Nike, Beaverton, USA). The normal condition was defined by the use of no plate, stiff condition had one plate, and the very stiff condition had two plates. Individual plates weighed approximately 50 grams.

Table 6.1. Physical properties of footwear used.

Shoe	Plates (n)	Plate Thickness (mm)	Shoe Mass (g)	Stiffness (N-m/rad)
Normal	0	0	239	5.9
Stiff	1	3	292	10.5
Very Stiff	2	6	346	17.0

Longitudinal bending stiffness was quantified using a custom set-up (Figure 6.1). The testing set-up was designed to best mimic a previously described mechanical testing procedure (Hoogkamer et al., 2018). The shoe was anchored upside down on a table using a c-clamp, with the forefoot bending axis aligned on the edge of the table. A carabiner was used to connect a strain gauge to the heel loop of the shoe. The strain gauge voltage was amplified using a SparkFun HX711 (SparkFun Electronics, Niwot, CO) load cell amplifier and powered via an Arduino Mega2560 microcontroller (Arduino, Somerville, MA) sampling at 10 Hz. Linear displacements of the heel point and heel loop where the strain gauge was attached were measured using standard tape measures affixed to the table structure. One test cycle consisted of pulling vertically downward on the strain gauge attached to the shoe. Three test cycles were done for each shoe condition. The loading phases of each test cycle were all completed in the same time duration, approximately five seconds. Video of the test cycles was recorded on an iPhone 7s (Apple, Cupertino, CA) for the purpose of assessing linear displacements of the heel point and heel loop.

Figure 6.1. Set-up for longitudinal bending stiffness test

Force data were output from the strain gauge in pounds and converted to Newtons. Angular displacement of the shoe was quantified from linear displacement data (Figure 6.2). The horizontal and vertical displacements of the heel tip of the shoe were recorded at the beginning and end of each loading cycle. These linear displacements were then used to quantify angular displacement using the following equation:

$$\theta_{bend} = tan^{-1}\ \frac{\Delta y}{L - \Delta x}$$

Where ‘L’ is the distance from the heel tip to the forefoot rotation axis of the shoe, and Δx and Δy are the respective linear displacements of the heel tip. The horizontal displacement of the heel loop throughout the loading phase was quantified to account for the change in moment arm of the pulley about the forefoot rotation axis. Because start and end coordinates of the heel tip and pulley were recorded, it was assumed that translations occurred in a linear fashion.

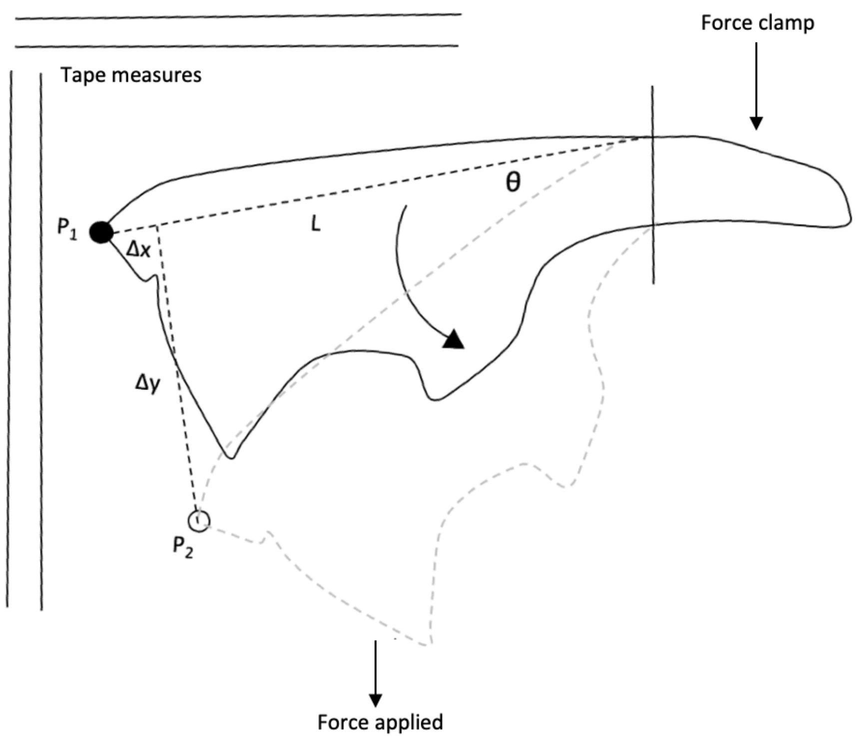

Figure 6.2. Depiction of linear distances used to quantify flexion angle and longitudinal bending stiffness

Bending angle, dynamic moment arm, and force data were extrapolated to 101 data points per loading cycle, and averaged across test cycles. Strain gauge force was multiplied by the dynamic moment arm to estimate shoe extension torque throughout the test cycles. The average LBS of each shoe was quantified based on the slope of the moment-angle curve (Oh and Park, 2017)

Running Biomechanics. Running trials assessing spatiotemporal parameters and ground reaction forces were conducted on a force instrumented treadmill (Bertec Inc., Columbus, OH). Ground reaction force data were collected at 1000 Hz. Participants ran at three speeds, 3.89, 4.70, and 5.56 m/s (14, 17, 20 km/hr), in three different footwear conditions of varying LBS (Table 1). Data were collected at each speed for approximately ten strides. Order of footwear was randomized between participants. Each participant completed the three speeds in ascending order and then switched shoes. Rest between conditions was self-selected.

Metabolics. Participants visited the lab a second time to perform a series of metabolic analyses after a self-selected warm-up. Twelve five-minute running trials were completed on a high speed treadmill (Woodway, Waukesha, WI) set to 1% grade (Jones and Doust, 1996). Six trials were run at 14 km/hr and six at 17 km/hr. The three shoe conditions from the first day were used again. Each shoe was worn for two trials at each speed. Order of shoe testing within each speed was randomized. The 14 km/hr trials were completed before the 17 km/hr trials. Measures of VO2 and VCO2 were taken with an open-circuit expired-gas analysis system (Parvomedics TrueOne 2400, Sandy, UT). Rest between trials was approximately five minutes. Participants body mass was assessed between each trial while wearing shoes to account for changes in hydration and shoe weight.

Comfort Assessments. Subjective dynamic comfort assessments were completed for each shoe at each speed immediately after the participants finished the second trial in each shoe. The following factors were rated: forefoot cushioning, rearfoot cushioning, forefoot flexibility, stability, heel-to-toe transition, and weight (Luo et al., 2009). A five-point scale was used for rating: 5 meaning "not acceptable", 3 meaning "not great but acceptable", and a 1 meaning "just right." Ratings were averaged across all factors to determine an overall comfort score for each shoe at each speed. Participants were blinded to their previous assessments of the other shoes.

Data Analysis

A custom MATLAB (version 2016b; Mathworks, Natick, MA) program was used to calculate contact time, stride length, stride frequency, and anterior-posterior ground reaction force impulses. Ground reaction force data were filtered using a low-pass zero-lag 4th order Butterworth with a 20Hz cut-off and down-sampled to 200Hz (fs). Stance phase was defined as the phase when the vertical ground reaction force exceeded 5% of body weight. Braking and propulsive impulses were quantified by integrating the area under the positive and negative regions of the anterior-posterior ground reaction force. Contact time, stride frequency, and stride length were calculated from filtered ground reaction force data using the following equations:

(1) Contact time (s) = # frames stance phase * fs

(2) Step frequency (Hz) = 1 / (# frames between ipsilateral foot contacts/fs)

(3) Stride length (m) = (# frames between ipsilateral foot contacts/fs) * treadmill belt speed (m/s)

Metabolic data were analyzed by averaging submaximal (RER < 1.0) VO2 and VCO2 consumption over the last two minutes of each stage. The VO2 values were normalized to body mass (ml/kg/min). Metabolic rate (W/kg) was quantified using the Brockway equation (Brockway, 1987).

Statistical Analysis

Repeated measures analysis of variance (ANOVA, $\alpha < 0.05$) tests were used to analyze braking and propulsive impulses, peak propulsive force, stride frequency, stride length, contact time, oxygen consumption, metabolic rate, and subjective comfort between footwear conditions at each speed. For comfort assessments, overall shoe score and individual factors were analyzed. Greenhouse-Geisser adjustments were used when Mauchley's test of Sphericity was significant (<.05). Pairwise comparisons with Bonferroni adjustments ($\alpha = .05/3 = .0167$) were used *post-hoc* to further analyze a significant effect of shoe. Effect sizes (partial eta squared, η^2) were calculated and defined as small ($\eta^2 = 0.01$), medium ($\eta^2 = 0.06$), or large ($\eta^2 = 0.14$) (Cohen, 1988).

Results

Altering LBS resulted in changes in spatiotemporal variables across speeds (Table 6.2). There was a significant effect of shoe on contact time at 14 km/hr ($p = .01$, $\eta^2 = 0.402$) and 20 km/hr ($p = .001$, $\eta^2 = 0.566$), but not at 17 km/hr ($p = .057$, $\eta^2 = 0.272$). There was a significant effect of shoe on stride frequency at 14 ($p < .001$, $\eta^2 = 0.636$), 17 ($p = .005$, $\eta^2 = 0.445$) and 20 km/hr ($p = .048$, $\eta^2 = 0.287$), but not at 20 km/hr. There was a significant effect of shoe on stride length at 14 ($p < .001$, $\eta^2 = 0.616$) and 17 km/hr ($p = .016$, $\eta^2 = 0.368$), but not at 20 km/hr ($p = .051$, $\eta^2 = 0.282$).

Horizontal ground reaction forces were affected by a change in LBS (Table 6.2, Figure 6.4). There was a significant effect of shoe on peak propulsive force at 14 ($p < .001$, $\eta^2 = 0.712$), 17 ($p = .002$, $\eta^2 = 0.505$), and 20 km/hr ($p < .001$, $\eta^2 = 0.701$). Braking impulse was not significantly different between shoe conditions at 14 ($p = .465$, $\eta^2 =$

0.082), 17 (p = .782, η^2 = 0.035) or 20 km/hr (p = .767, η^2 = 0.029). Likewise, propulsive impulse was not significantly different between shoe conditions at 14 (p = .249, η^2 = 0.145), 17 (p = .746, η^2 = 0.032), or 20 km/hr (p = .071, η^2 = 0.255).

Table 6.2. Spatiotemporal and ground reaction force variables across shoe conditions at three running speeds

		Normal	**Stiff**	**Very stiff**	**P**
Spatiotemporal variables					
14 km/hr	Contact time (s)	0.207 ± .009[c]	0.210 ± .011	0.212 ± .011[a]	**.010**
	Stride length (m)	2.62 ± .10[c]	2.66 ± .10	2.67 ± .10[a]	**<.001**
	Stride frequency (Hz)	1.49 ± .06[b,c]	1.47 ± .06[a]	1.46 ± .06[a]	**<.001**
17 km/hr	Contact time (s)	0.188 ± .010	0.188 ± .010	0.189 ± .011	.057
	Stride length (m)	3.02 ± .15[c]	3.05 ± .15	3.05 ± .15[a]	**.016**
	Stride frequency (Hz)	1.56 ± .08[c]	1.54 ± .08	1.54 ± .08[a]	**.005**
20 km/hr	Contact time (s)	0.165 ± .009[b,c]	0.169 ± .010[a]	0.170 ± .010[a]	**.001**
	Stride length (m)	3.34 ± .18	3.38 ± .17	3.34 ± .16	.051
	Stride frequency (Hz)	1.67 ± .09	1.66 ± .09	1.67 ± .09	**.048**
Horizontal ground reaction force					
14 km/hr	Braking impulse (N-s)	14.9 ± 2.0	15.0 ± 1.9	15.2 ± 2.1	.465
	Prop. Impulse (N-s)	13.7 ± 1.3	13.8 ± 1.2	13.6 ± 1.3	.240
	Peak prop. Force (N)	230 ± 21[b,c]	224 ± 21[a,c]	215 ± 21[a,b]	**<.001**
17 km/hr	Braking impulse (N-s)	16.3 ± 2.1	16.4 ± 2.5	16.2 ± 2.3	.729
	Prop. Impulse (N-s)	15.2 ± 1.5	15.2 ± 1.3	15.1 ± 1.4	.746
	Peak prop. Force (N)	281 ± 27[c]	276 ± 26	266 ± 24[a]	**.002**
20 km/hr	Braking impulse (N-s)	17.2 ± 2.6	17.2 ± 2.7	17.1 ± 2.4	.767
	Prop. Impulse (N-s)	16.0 ± 1.7	16.1 ± 1.7	15.8 ± 1.6	.071
	Peak prop. Force (N)	334 ± 33[b,c]	326 ± 32[a,c]	316 ± 31[a,b]	**<.001**

Pairwise comparisons showing significant (p < .05) differences: [a] = different from normal, [b] = different from stiff, [c] = different from very stiff

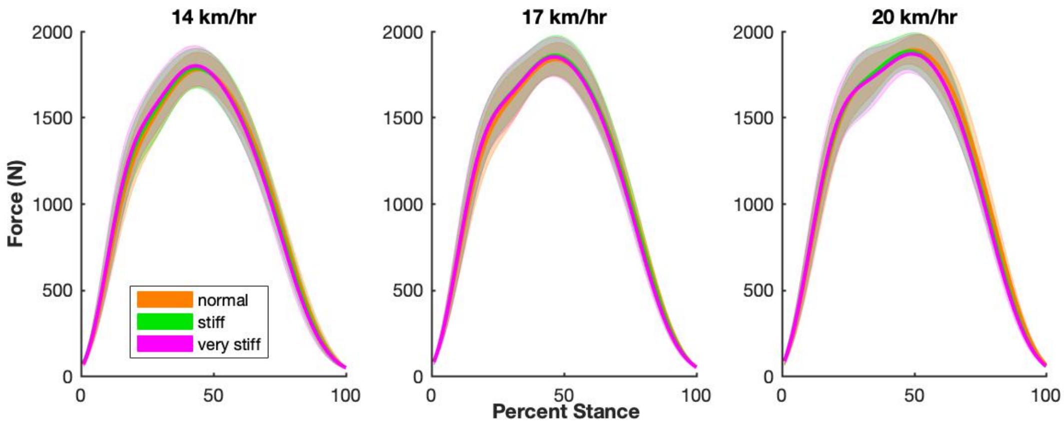

Figure 6.3. Vertical ground reaction forces across shoe conditions at three speeds

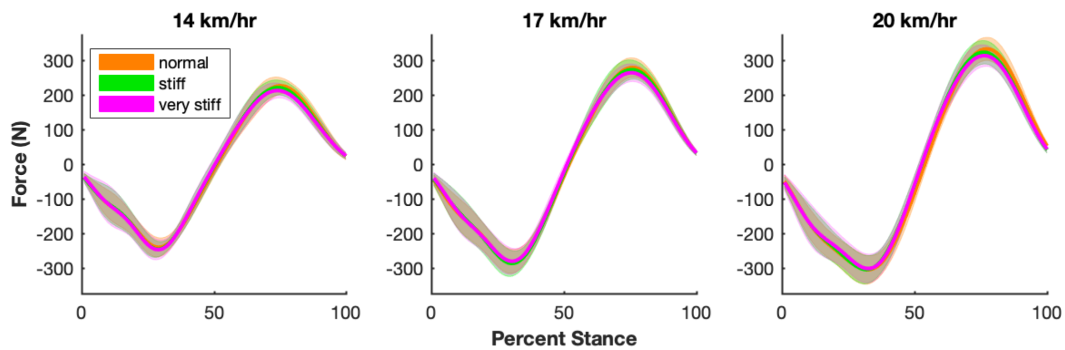

Figure 6.4. Anterior-posterior ground reaction forces across shoe conditions at three speeds

There was an effect of shoe on metabolic variables at both 14 and 17 km/hr (Table 6.3). Normalized VO2 (ml/kg/min) was significantly different between shoes at 14 (p = .005, η^2 = 0.479) and 17 km/hr (p = .006, η^2 = 0.565), as was metabolic rate at 14 (p = .002, η^2 = 0.532) and 17 km/hr (p = .002, η^2 = 0.652). One participant had issues with their nose clip staying on during the 14 km/hr trials, resulting in analysis for 14 km/hr metabolic data on only nine of the ten participants.

Table 6.3. Normalized oxygen uptake and metabolic rate across shoe conditions at two running speeds

		Normal	**Stiff**	**Very stiff**	**P**
14 km/hr	VO2 (ml/kg/min)	45.33 ± 3.26[c]	45.86 ± 3.48	46.24 ± 3.31[a]	**.005**
	Metabolic Rate (W/kg)	14.42 ± 1.06[c]	14.61 ± 1.08	14.76 ± 1.07[a]	**.002**
	VO2 (L/min)	2.86 ± 0.28[c]	2.89 ± 0.30	2.92 ± 0.27[a]	**.002**
	VCO2 (L/min)	2.46 ± 0.26[c]	2.49 ± 0.28	2.51 ± 0.27[a]	**.026**
17 km/hr	VO2 (ml/kg/min)	57.30 ± 3.43[c]	57.13 ± 3.84[c]	58.58 ± 3.57[a,b]	**.006**
	Metabolic Rate (W/kg)	18.21 ± 1.14[c]	18.22 ± 1.15[c]	18.75 ± 1.21[a,b]	**.002**
	VO2 (L/min)	3.61 ± 0.31[c]	3.60 ± 0.30[c]	3.70 ± 0.32[a,b]	**.001**
	VCO2 (L/min)	3.20 ± 0.33[c]	3.21 ± 0.31[c]	3.33 ± 0.35[a,b]	**.001**

Pairwise comparisons showing significant ($p < .05$) differences: [a] = different from normal, [b] = different from stiff, [c] = different from very stiff

Overall subjective comfort was significantly different between shoes at 14 (p = .013, $\eta^2 = 0.384$) and 17 km/hr (p = .001, $\eta^2 = 0.515$) (Table 6.4). Individual factors that were significantly different in comfort at 14 km/hr were forefoot cushioning ($p < .001$, $\eta^2 = 0.555$) and flexibility ($p = .005$, $\eta^2 = 0.450$). Rearfoot cushioning, stability, heel-toe transition, and weight were not significantly different in perceived comfort between shoes at 14 km/hr. At 17 km/hr, forefoot cushioning ($p = .031$, $\eta^2 = 0.320$), rearfoot cushioning ($p = .037$, $\eta^2 = 0.307$), flexibility ($p < .001$, $\eta^2 = 0.585$), and weight ($p = .003$, $\eta^2 = 0.484$) were significantly different between shoes. However, there were no significant pairwise comparisons for rearfoot cushioning at 17 km/hr. Stability and heel-toe transition were not significantly different in perceived comfort between shoes at 17 km/hr.

Table 6.4. Subjective dynamic comfort assessments across shoes at two running speeds

		Normal	Stiff	Very stiff	P
14 km/hr	Total	2.2 ± 0.8	2.0^{c} ± 0.8	3.1^{b} ± 1.0	**.013**
	Forefoot cushioning	2.6 ± 1.3^{c}	2.4 ± 1.3^{c}	4.6 ± $0.8^{a,b}$	**.001**
	Rearfoot cushioning	1.8 ± 1.0	2.0 ± 1.4	2.8 ± 1.8	.120
	Flexibility	2.2 ± 1.0^{c}	2.2 ± 1.4^{c}	4.0 ± $1.1^{a,b}$	**.005**
	Heel-toe transition	2.2 ± 1.4	2.0 ± 1.1	2.6 ± 1.8	.439
	Stability	2.8 ± 1.5	1.6 ± 1.3	1.8 ± 1.4	.068
	Weight	1.8 ± 1.4	2.0 ± 1.1	2.6 ± 1.3	.286
17 km/hr	Total	2.1 ± 0.7^{c}	1.9 ± 0.5^{c}	3.2 ± $1.1^{a,b}$	**.001**
	Forefoot cushioning	2.0 ± 1.1^{c}	2.0 ± 1.4	3.6 ± 1.6^{a}	**.031**
	Rearfoot cushioning	1.8 ± 1.0	2.0 ± 1.1	3.2 ± 1.8	**.037**
	Flexibility	2.8 ± 1.5^{c}	2.0 ± 1.1^{c}	4.0 ± $1.1^{a,b}$	**<.001**
	Heel-toe transition	2.0 ± 1.1	2.0 ± 1.1	2.4 ± 1.6	.717
	Stability	2.4 ± 1.6	1.6 ± 1.0	2.2 ± 1.7	.442
	Weight	1.6 ± 1.3	1.6 ± 1.0^{c}	3.8 ± 1.7^{b}	**.003**

Pairwise comparisons showing significant ($p < .05$) differences: ^{a} = different from normal, ^{b} = different from stiff, ^{c} = different from very stiff

Discussion

The purpose of this study was to investigate the effects of varying LBS on spatiotemporal variables, ground reaction forces, subjective comfort, and metabolic cost. In support of our hypotheses, we observed significant changes in all variables across the range of speeds tested in response to altered LBS (Tables 6.2, 6.4, 6.4).

Increasing LBS systematically increased contact time at 14 and 20 km/hr, matching previous results (Flores et al., 2018; Willwacher et al., 2013, 2014). Interestingly, there was no change in contact time at 17 km/hr. A potential reason for this may be that the stiffness of the shoe is more aligned to that of the MTPJ, which increases with running speed (Chapter IV). A shoe that is too stiff prolongs contact time either by increasing the ground reaction force moment arm about the MTPJ axis or by creating a necessity to increase joint torque impulse due to not being able to overcome the stiffer shoe with a greater internal moment (Willwacher et al., 2014). In contrast, the assumed larger MTPJ moment at 17 km/hr (Chapter IV) may be enough to overcome the increased LBS of the stiff shoe and maintain natural MTPJ dorsiflexion. This may help runners stay within their preferred movement path (Nigg et al., 2017) and maintain normal contact time.

There was a significant decrease in stride frequency as LBS increased at all three speeds (Table 2). The observed decrease may primarily be due to increased contact time. This contradicts Flores et al. who reported no change in stride frequency with increased LBS (Flores et al., 2018). However, their testing speed was only 10 km/hr, compared to the 14, 17, and 20 km/hr speeds tested in the current study. A decreased stride frequency has been previously observed in footwear that reduces energetic cost by 4% (Hoogkamer et al., 2017a). It is unknown however if the carbon fiber plate or soft midsole material of increased thickness is the mechanism behind the decreased stride frequency. In support of midsole thickness affecting stride frequency, Law et al. observed a systematic decrease in stride frequency with increased midsole thickness (Law et al., 2018). Thus, the decrease in stride frequency may primarily be due to the increased contact time. However, it is unknown how increased LBS affects swing phase kinematics due to any potential alterations in late-stance mechanics that may be at play.

Stride length increased with the use of stiffening plates at all three speeds tested (Table 6.2). At 14 km/hr there was an average 4cm longer stride using the stiff shoe, and an additional 1cm longer using the very stiff shoe. At 17 km/hr, stride length was the same for the stiff and very stiff shoe, but 3cm longer than the normal shoe. We speculate that the increase in stride length may be due to the increase in the percentage of stance that the foot is in propulsion due to the increased LBS (Willwacher et al., 2013), and the subsequent effect on center-of-mass dynamics. Increased prosthetic toe-joint stiffness has been shown to increase center-of-mass push off work in amputees during walking (Honert et al., 2018). Perhaps an increase in push-time affects how far the center-of-mass travels anteriorly during stance and its subsequent take-off velocity. Increased push-phase duration during contact would also cause the foot to be positioned slightly more posterior with respect to the pelvis, requiring that it be recovered further during swing phase to the next foot contact. Both factors may contribute to the observed 3-5cm increase in stride length at 14 and 17 km/hr.

Interestingly, at 20 km/hr, average stride length was the same for the normal and very stiff shoe, but 4cm longer using the stiff shoe. While there were no changes in horizontal ground reaction force impulses between shoe conditions to explain the differences in stride length, there may be changes in joint level angular impulses in response to varying LBS (Oh and Park, 2017). During running the ankle plantar flexors tend to act in a nearly isometric range, resulting in the Achilles tendon generating the majority of positive work (Lai et al., 2014). With increased footwear stiffness though, greater muscle force and ankle plantar flexor moment is required to overcome the stiff plates (Willwacher et al., 2014). At the fastest running speed concentric contributions of the ankle plantar flexors in addition to the Achilles tendon contributions may have been required to overcome the very stiff shoe. Ankle plantar flexor fascicle shortening velocity may have been fast enough that it resulted in a large enough decrease in force generation capability that they were unable to generate enough force to overcome the very stiff shoe compared to the normal and stiff shoe. This may explain why contact time for the very stiff shoe was longer than the normal shoe, yet participants elicited the same stride length. The increased contact time may have been necessary to maintain propulsive impulse due to a compromised ability to generate a large enough ankle plantar flexor moment (Willwacher et al., 2014).

Peak propulsive ground reaction force decreased at all three speeds with increased LBS, in support of previous observations (Flores et al., 2018). The decrease in peak propulsive force is most likely due to the interaction between reduced MTPJ dorsiflexion and having to overcome the increased LBS. Both of these factors may influence a more vertical orientation of the resultant ground reaction force. There were no changes in braking or propulsive impulses however, most likely due to the increased contact time that accompanied the decreased peak propulsive force. Horizontal forces are costly to generate in metabolic terms, and thus if running velocity can be maintained by decreasing the required posteriorly oriented force, this could be a mechanism which could contribute to metabolic savings (Chang and Kram, 1999). However, the magnitude of difference in peak propulsive force is relatively small (15-20N) and thus may have a negligible effect. If, after being deflected due to MTPJ dorsiflexion, the stiff plate snapping back to maintain shape in late stance can offload a large enough magnitude of the required horizontal propulsive forces, there may be potential to mechanistically influence a decrease in metabolic cost.

Increasing LBS affected running economy and metabolic rate at both speeds. One characteristic that is most likely influencing this observation was the difference in shoe weight as a result of the stiff plates. The stiff and very stiff shoes weighed 50 and 100g more than the normal shoe (Table 6.1), most likely influencing a difference in energy cost by approximately 0.5 to 1.0% (Franz et al., 2012; Frederick, 1984; Hoogkamer et al., 2016). Because modern marathon racing shoes weigh approximately 200-250g (Hoogkamer et al., 2017a) we felt it unreasonable to add mass to make all shoes 350g. These differences in shoe mass should be explored in further detail at the paces examined in the present study.

At 14 km/hr, seven of nine participants were most economical in the normal shoe (Figures 6.5 & 6.6), similar to previous studies showing that at a pace reflective of a normal training run metabolic cost increases in an overly stiff shoe (Oh and Park, 2017; Roy and Stefanyshyn, 2006). This trend changed at 17 km/hr where four of ten participants exhibited lower VO2 and metabolic rate wearing the stiff shoe. For those participants that did not show a minimum energetic cost in the stiff shoe at 17 km/hr, the percentage difference in VO2 between the stiff and normal shoes was less at 17 km/hr than at 14 km/hr. The same trends were observed for metabolic rate.

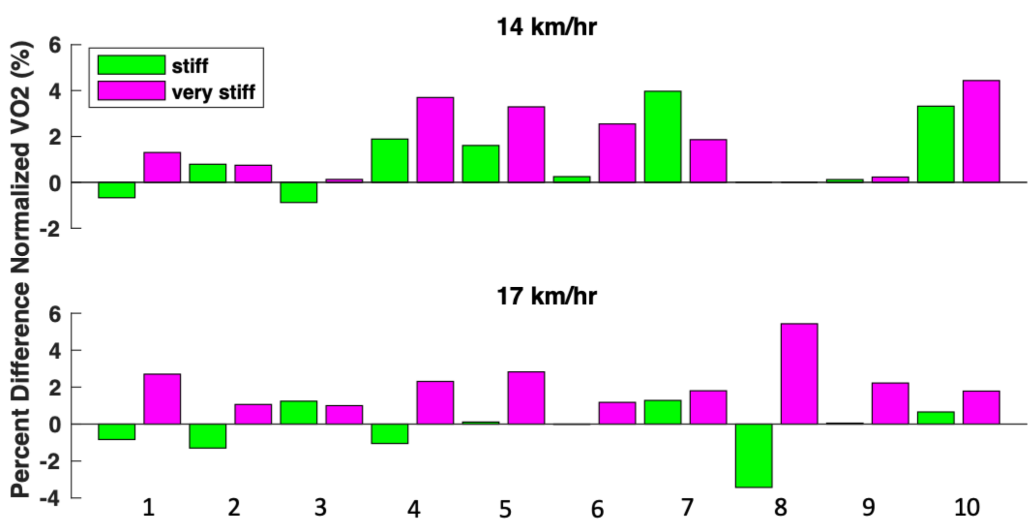

Figure 6.5. Individual participant differences in running economy (VO2 ml/mg/min) expressed as percent difference relative to normal shoe condition. Top is at 14 km/hr (n=9), bottom is at 17 km/hr (n=10). Participant number denoted on the x-axis.

Participants' body mass was measured in-between each running trial to account for any change due to sweat loss or difference in shoe mass. While the VO2 values were normalized to total mass, which fluctuated approximately 0.1kg between wearing the normal and very stiff shoe, this change in weight was due to non-metabolically functioning mass. Thus, based upon the 1% increase per added 100-gram theory (Frederick, 1984), it may be reasonable to extrapolate that the mechanism of increased LBS at 17 km/hr may be more influential than the results show. Specifically, four participants already exhibited a minimum energetic cost in the stiff shoes at 17 km/hr despite the extra mass. Four other participants only exhibited a higher metabolic cost compared to the normal shoe by a half percent difference or less in the stiff shoes (Figure 6.5 & 6.6). If mass were equalized between shoes, or a lighter material such as carbon fiber were used to make the stiffening plate, then perhaps the group mean improvement using the stiff shoes at 17km/hr would be even more pronounced. While changes in energetic cost are linear with increased weight, a limitation to this speculation is that the smallest change in weight tested in previous studies is 100g. Thus, it is unknown if a 50g increase in mass truly results in a 0.5% increase in energetic cost.

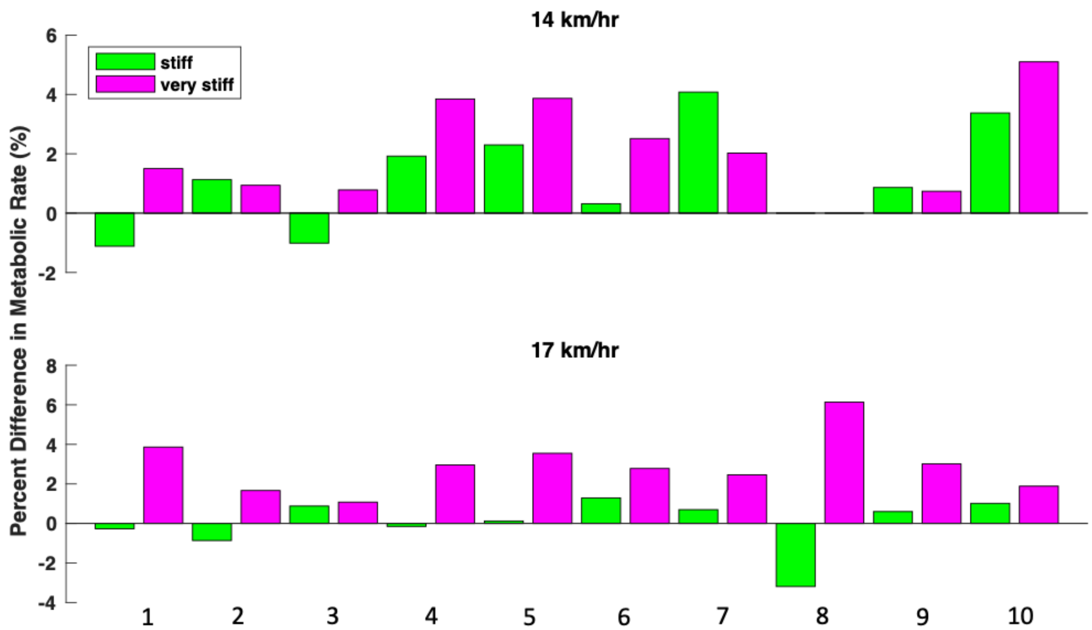

Figure 6.6. Individual participant differences in metabolic rate (W/kg) expressed as percent difference relative to normal shoe condition. Top is at 14 km/hr (n=9), bottom is at 17 km/hr (n=10). Participant number denoted on the x-axis.

Subjective comfort appears to be velocity dependent (Table 6.4). The main factors contributing to perceived comfort differences were forefoot cushioning, flexibility, and weight. The plates were placed directly underneath the insole, as opposed to being embedded in the midsole (Hoogkamer et al., 2017a). This is most likely the reason why the stiff plates were not preferred in regard to forefoot cushioning comfort. The very stiff condition was the worst rated for perceived flexibility comfort at both 14 and 17 km/hr. This likely influenced the greater metabolic cost using this shoe (Luo et al., 2009). While across-speed comparisons were not statistically analyzed, average comfort for flexibility of the normal shoe was slightly worse at 17 km/hr, increasing in score from 2.2 to 2.8, and slightly improved for the stiff shoe from 2.2 to 2.0. On an individual basis, four participants thought the flexibility of the stiff shoe was more comfortable at 17 km/hr, four reported no change, and only two perceived it as worse. For the normal shoe, five participants thought the flexibility was worse at 17 km/hr, three reported no change, and only two participants thought it was better. These subjective comfort results suggest that when designing shoes for faster running, they should be lightweight, soft in the forefoot, and stiffer than a conventional trainer. We speculate that the reason for the preferred increase in stiffness is related to the increase in MTPJ moment and stiffness with running speed (Chapter IV). It is likely that runners desire a stable or propulsive feeling from the shoe during push-off, as opposed to an overly flexible shoe that feels like it may be dissipating energy or not contributing as much torque to push-off as it could, without inhibiting the runner's natural or preferred movement path.

Limitations

A limitation to this study is that it is unknown how individuals will respond to greater LBS under fatigued conditions. Increasing LBS results in greater stress on the ankle plantar flexor muscle-tendon units (Willwacher et al., 2014). Without adequate strength or fatigue-resistance to maintain mechanics similar to those experienced in a non-fatigued state, detrimental compensations such as increased contact time or forward lean may occur (Willwacher et al., 2014). Secondly, in this study there was not a longitudinal adaptation period. Because this study used very well trained distance runners (average 5000m personal best 15:04), and it has been demonstrated that runners quickly adjust their mechanics to changes in surface stiffness (Ferris et al., 1999), we feel that this may not necessarily be a limitation but is worth bringing to attention. Lastly, due to the constraint of needing to find participants who were sub-maximal at 17 km/hr and able to fit a male size 10 shoe, our participant numbers were limited to ten. A *post-hoc* power analysis (G*Power) using the effect size from the VO2 results ($\eta^2 = 0.565$) as input revealed a statistical power of 0.997 and a total required sample size of 6, ensuring our participant population of ten resulted in adequate statistical power.

Future work

More participants of varying foot strike patterns, anthropometrics, etc. would allow for potential advanced statistical methods such as cluster or principal component analysis to identify characteristics that influence whether someone is a responder or a non-responder to the increased LBS footwear. Most importantly, specifically tuning stiffness of the shoe to an individuals' natural change in MTPJ mechanics would be interesting. This may eliminate the risk of individuals being a responder or non-responder.

Conclusion

Altogether, this study provides evidence that altering LBS elicits changes in spatiotemporal variables, metabolic cost, horizontal ground reaction forces, and subjective comfort that are running velocity dependent. Our results demonstrate that changes in LBS influence a reorganization of stride length and stride frequency, highlighting the complex interaction between internal and external factors that contribute to optimization of energetic cost during running (Cavanagh and Kram, 1985; Cavanagh and Williams, 1982). We suggest that if footwear is engineered for athletes with specific goal race paces, LBS should be heavily weighted both from a metabolic and perceived comfort standpoint.

Bridge

This chapter investigated how gross level mechanics and metabolics changed in response to footwear of varying LBS at a range of speeds. Chapter VII will use this same data set to understand the changes in joint level mechanics to varying LBS across speeds.

CHAPTER VII

MECHANICS OF THE ANKLE BUT NOT METATARSOPHALANGEAL JOINT ELICIT A RUNNING SPEED DEPENDENT RESPONSE TO VARIABLE STIFFNESS FOOTWEAR

This chapter is currently unpublished. Evan Day designed the study and collected and analyzed the data. Michael E. Hahn provided mentorship and aided in study design, general oversight, and editing and finalizing the final manuscript.

Introduction

The metatarsophalangeal joint (MTPJ) is unique in that it generates near negligible positive work during steady-state running, unlike the ankle, knee, and hip (Stefanyshyn and Nigg, 1997). Despite its minimal contribution to push-off, manipulation of MTPJ function with modification of the longitudinal bending stiffness (LBS) of footwear has been gaining notoriety in sport performance settings (Stefanyshyn and Wannop, 2016).

The first demonstrated relationship between joint level mechanical changes in response to varying LBS and metabolic cost was when Roy & Stefanyshyn reported a 1% improvement in running economy using a stiff shoe (Roy and Stefanyshyn, 2006). This improvement was replicated by Madden et al., but it is worth noting that they further divided their participant population into responders and non-responders (Madden et al., 2015). A similar division of responder and non-responder has also been done in classifying how individuals alter ankle mechanics in response to varying LBS (Willwacher et al., 2014). The conclusion across studies suggests that there are multiple factors at play affecting how an individual responds to varying LBS, such as body mass, ankle plantar flexor strength, or foot anthropometrics (Madden et al., 2015; Roy and Stefanyshyn, 2006; Willwacher et al., 2014). Altogether, these studies provide evidence that increasing footwear LBS may elicit beneficial mechanical changes to performance.

It is commonly accepted that inserting stiff plates in footwear reduces MTPJ dorsiflexion, increases MTPJ moments, and reduces energy dissipation (Willwacher et al., 2013). However, contrary results have been shown where participants displayed no change in MTPJ kinematics or kinetics in response to stiff plates (Roy and Stefanyshyn, 2006). The footwear used by Roy & Stefanyshyn was stiffer than that used by Willwacher et al. though, making cross-study comparisons difficult. On influencing joint kinetics, stiff footwear tend to accentuate anterior displacement of the center of pressure, and thus lengthen the ground reaction force lever arms about the MTPJ and ankle joints (Willwacher et al., 2013, 2014). This may influence muscle-tendon unit force-length and force-velocity operating points by influencing changes in angular velocity of the MTPJ and ankle joint (Madden et al., 2015; Roy and Stefanyshyn, 2006; Takahashi et al., 2016; Willwacher et al., 2013). A shift in force-velocity operating point of the ankle plantar flexors is metabolically beneficial due to the reduction in force generated during muscle contraction (Roberts et al., 1998). Energy storage and return capacity of the ankle plantar flexor complex and the subsequent influence on metabolic cost may also be affected by a change in ankle plantar flexor moments in response to increased LBS (Willwacher et al., 2014).

While changes in joint level mechanics in response to varying LBS at a single speed during running has been fairly well investigated, there is a paucity in the knowledge of within-participant mechanical responses to varying LBS across running speeds. Multiple studies have investigated only distance runners at one speed (Oh and Park, 2017; Roy and Stefanyshyn, 2006; Willwacher et al., 2013, 2014) or sprinters (Smith et al., 2016; Stefanyshyn, Darren J., Fusco, 2004). Only one study reported within-participant changes across speeds, but it involved using cleated footwear on an artificial turf surface (Wannop et al., 2017). Because MTPJ kinematics and kinetics change with running speed (Chapter IV) as well as ankle kinematics and kinetics (Jin and Hahn, 2018; Orendurff et al., 2018; Stefanyshyn and Nigg, 1998a), there is a possibility that joint level mechanical responses to footwear of varying LBS may be running speed dependent.

The purpose of this study was to investigate if the change in joint level mechanics and energetics to footwear of varying LBS in well trained distance runners is running speed dependent. We hypothesized that MTPJ dorsiflexion would be reduced at 14 km/hr for both the stiff and very stiff shoes, at 17 km/hr for just the very stiff shoe, and at 20 km/hr there would be no dorsiflexion restriction, due to the greater ankle and MTPJ moments encountered with high running speed. We also hypothesized that at all speeds MTPJ and ankle moments and angular resistance of the MTPJ would systematically increase with LBS. Lastly, we hypothesized that there would be a decrease in positive and negative ankle joint work, but no change in knee or hip energetics with increasing LBS at all speeds.

Methods

Recruitment

Ten competitive male runners were recruited for this study (26 ± 6 years, 1.78 ± 0.04 m, 63.9 ± 4.0 kg, 101 ± 34 km/wk, 15:04 ± 0:38 (min:sec) 5000m personal best). To be included participants had to have a 5000m personal best under 16:00, no lower extremity injury in the previous six months, and currently running over 50 km/week. Participants provided written informed consent prior to data collection. This study was approved by the Institutional Review Board at the University of Oregon.

Study Design and Experimental Protocol

Running trials were conducted on a force-instrumented treadmill (Bertec Inc., Columbus, OH) after a self-selected warm up. Kinematic data were collected at 200 Hz and kinetic data at 1000 Hz. Participants ran at three speeds, 3.89, 4.70, and 5.56 m/s (14, 17, 20 km/hr). Data were collected at each speed for approximately ten strides. Order of footwear was randomized between participants. Each participant completed the three speeds in ascending order and then switched shoes. Rest between conditions was self-selected.

Data Collection

A unilateral lower extremity marker set consisting of 26 retro-reflective markers defining the forefoot, rearfoot, shank, and thigh of the right leg and pelvis was used. A two-segment foot model was defined by placing markers on the forefoot and calcaneus (Chapter III, Day and Hahn, 2019; Goldmann et al., 2013). Individual markers were placed on the medial and lateral malleoli and femoral epicondyles, left and right greater trochanters, anterior superior iliac spines, posterior iliac spines, sacrum, and over the seventh thoracic vertebrae. Marker quadrad clusters were placed over the lateral aspect of the shank and thigh and secured with bandage tape. Participants wore the same footwear as previously described in chapter IV (Nike Epic React Flyknit, Nike, Beaverton, USA) and used the same stiffening plates. Footwear LBS was 0.10, 0.18, and 0.30 N-m/deg for the normal, stiff, and very stiff shoes (Figure 7.1). After completion of a static trial the medial malleoli and femoral epicondyle markers were removed so as to not interfere with running movements.

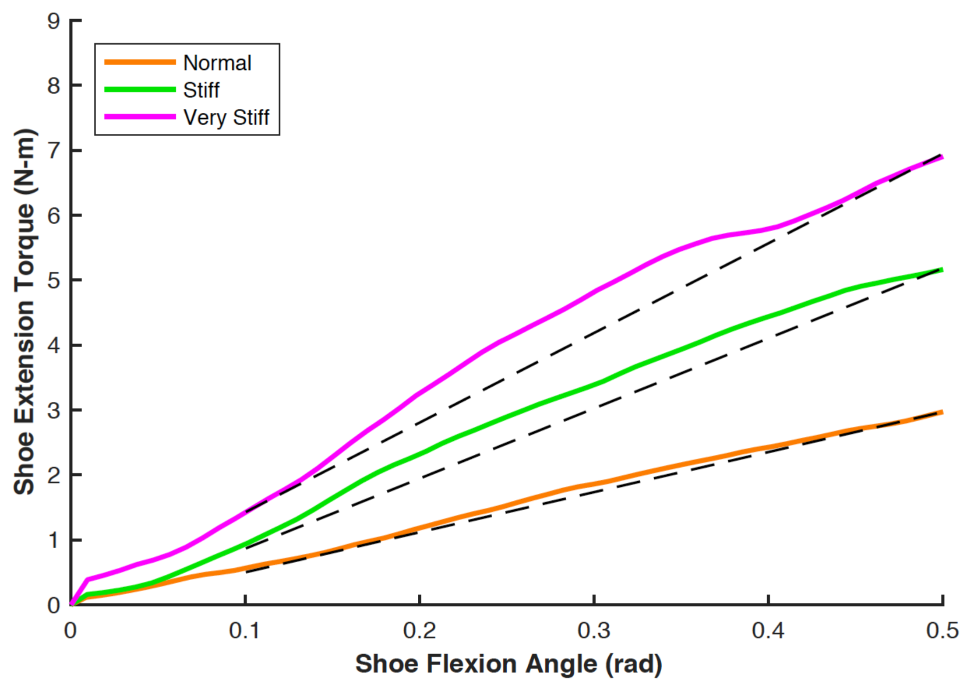

Figure 7.1. Load-displacement curves for the three shoes, data were obtained using a custom bending stiffness testing set-up described in Chapter IV.

Data Analysis

A custom MATLAB (version 2016b; MathWorks, Natick, MA) program was used to calculate joint kinematics, kinetics, and energetics during stance phase. Stance phase was defined as the phase when the vertical ground reaction force exceeded 5% of body weight. Raw marker coordinate data were filtered with a fourth-order low-pass

Butterworth filter with a 20Hz cutoff frequency (Day and Hahn, 2019; Willwacher et al., 2013). Force platform data were filtered in a similar manner to eliminate treadmill vibration noise (Willems and Gosseye, 2013). Joint angles were calculated using an Euler/Cardan rotation order of flexion/extension, abduction/adduction, and internal/external rotation. Sagittal plane angles, moments, and energetics were used for analysis.

The MTPJ was modeled as a hinge axis along the vector connecting the 1st and 5th metatarsal markers (Smith et al., 2012). The moment arm of the ground reaction force was estimated as the perpendicular distance between the center of pressure and the MTPJ axis. A non-fixed location joint center was used to model the MTPJ (Chapter III, Day and Hahn, 2019). Joint centers for the ankle and knee were defined by the midpoint between the medial and lateral malleoli and femoral epicondyle markers. Hip joint center was defined as 25% distance between the left and right greater trochanter markers (Weinhandl and O'Connor, 2010). All forces and moments about the MTPJ were considered zero until the center of pressure passed anterior to the MPTJ axis (Stefanyshyn and Nigg, 1997). Inertial effects of the forefoot were considered negligible (Stefanyshyn and Nigg, 1997) while the inertial parameters of the foot, shank, and thigh were modeled accordingly (de Leva, 1996). Joint moments were estimated using an inverse dynamics approach and resolved in the coordinate system of the proximal segment. Passive torque contributions of the shoe throughout stance phase were calculated as the product of the MTPJ dorsiflexion angle and the shoe bending stiffness. Ground reaction force lever arm about the MTPJ sliding joint center was estimated according to Willwacher et al. (Willwacher et al., 2014). Kinematic and kinetic data were re-sampled to 101 data points per stance phase for time normalized analysis. Angular resistance (R_{cr}) of the MTPJ was quantified from the load-displacement plot as previously described (Chapter IV). Joint power was calculated as the product of the joint moment and the joint angular velocity using the following equation:

$$P_j = M_j(\omega_p - \omega_d)$$

Where P_j and M_j represent power and moment of the respective joint, and ω_p and ω_d represent the angular velocities of the respective proximal and distal segments. Positive and negative joint work were quantified as the time integral of the positive and negative regions of the joint power curve.

Statistical Analysis

Repeated measures analysis of variance (ANOVA, $\alpha < 0.05$) tests were used to analyze peak MTPJ and ankle moments, MTPJ and ankle sagittal plane range of motion, MTPJ angular resistance, and positive and negative work of the MTPJ, ankle, knee, and hip, between the three footwear LBS conditions at each speed. Greenhouse-Geisser adjustments were used when Mauchley's test of Sphericity was significant (<.05). Pairwise comparisons with Bonferroni adjustments ($\alpha = .05/3 = .0167$) were used *post-hoc* to further analyze a significant effect of LBS. Effect sizes (partial eta squared, η^2) were calculated and defined as small ($.01 < \eta^2 < .05$), medium ($.06 < \eta^2 < 0.14$), or large ($\eta^2 > 0.14$) (Cohen, 1988).

Results

Altering LBS affected MTPJ mechanics at all speeds (Tables 7.1 & 7.2). There was a significant effect of LBS increasing Rcr (N-m/kg/deg) at 14 ($p < .001$, $\eta^2 = 0.719$), 17 (p = .001, η^2 = 0.648), and 20 km/hr (p < .001, η^2 = 0.756). Maximum MTPJ moment (N-m/kg) did not significantly change across footwear at 14 (p = .482, η^2 = 0.061), 17 (p = .644, η^2 = 0.032), and 20 km/hr (p = .966, η^2 = 0.004). Sagittal plane range of motion of the MTPJ was significantly reduced with increased LBS at 14 (p < .001, η^2 = 0.850), 17 (p < .001, η^2 = 0.774), and 20 km/hr (p < .001, η^2 = 0.800). Negative MTPJ work (W/kg) was reduced with increasing LBS at 14 (p = .012, η^2 = 0.480), 17 (p = .015, η^2 = 0.449), and 20 km/hr (p = .003, η^2 = 0.486), (Table 2). Positive MTPJ work (W/kg) did not differ between footwear at 14 (p = .362, η^2 = 0.107), 17 (p = .584, η^2 = 0.058), and 20 km/hr (p = .180, η^2 = 0.173).

Ankle range of motion did not change between footwear at 14 (p = .671, η^2 = 0.043), 17 (p = .083, η^2 = 0.272), or 20 km/hr (p = .232, η^2 = 0.127), (Table 7.1). Maximum ankle moment did not change between footwear at 14 (p = .304, η^2 = 0.119), 17 (p = .179, η^2 = 0.185), or 20 km/hr (p = .453, η^2 = 0.084). Negative ankle work did not change at 14 (p = .071, η^2 = 0.255) or 20 km/hr (p = .505, η^2 = 0.059) but did significantly change at 17 km/hr(p = .036, η^2 = 0.310), eliciting a ‘u-shaped’ response (Table 7.2). Ankle positive work did not change between LBS conditions at 14 (p = .280, η^2 = 0.131), 17 (p = .879, η^2 = 0.014), or 20 km/hr (p = .065, η^2 = 0.262).

Table 7.1. Metatarsophalangeal and ankle joint kinematics, kinetics, and stiffness across shoe conditions at three running speeds.

Speed (km/hr)		Normal	Stiff	Very Stiff	P
Metatarsophalangeal Joint					
14	Rcr (N-m/kg/deg)	0.010 ± 0.004^{b,c}	0.012 ± 0.005^{a,c}	0.014 ± 0.005^{a,b}	**< .001**
	Range of motion (°)	26.5 ± 3.6^{b,c}	23.7 ± 3.7^{a,c}	21.2 ± 3.2^{a,b}	**< .001**
	Max Mom. (N-m/kg)	0.91 ± 0.29	0.90 ± 0.29	0.87 ± 0.28	.482
17	Rcr (N-m/kg/deg)	0.011 ± 0.003^{b,c}	0.013 ± 0.005^{a,c}	0.017 ± 0.007^{a,b}	**.001**
	Range of motion (°)	29.9 ± 3.2^{b,c}	25.5 ± 3.9^{a,c}	23.6 ± 3.4^{a,b}	**< .001**
	Max Mom. (N-m/kg)	1.06 ± 0.29	1.03 ± 0.33	1.02 ± 0.25	.644
20	Rcr (N-m/kg/deg)	0.013 ± 0.003^{c}	0.015 ± 0.005^{c}	0.019 ± 0.006^{a,b}	**< .001**
	Range of motion (°)	30.0 ± 2.7^{b,c}	27.1 ± 4.0^{a,c}	25.1 ± 3.5^{a,b}	**< .001**
	Max Mom. (N-m/kg)	1.23 ± 0.33	1.23 ± 0.40	1.22 ± 0.36	.996
Ankle joint					
14	Range of motion (°)	38.2 ± 3.8	37.8 ± 3.1	37.3 ± 4.1	.671
	Max Mom. (N-m/kg)	3.56 ± 0.52	3.72 ± 0.45	3.70 ± 0.49	.304
17	Range of motion (°)	37.7 ± 3.9	36.9 ± 3.4	36.3 ± 4.1	.083
	Max Mom. (N-m/kg)	3.91 ± 0.51	3.89 ± 0.53	4.0 ± 0.61	.179
20	Range of motion (°)	34.1 ± 5.6	35.7 ± 4.1	34.1 ± 4.4	.294
	Max Mom. (N-m/kg)	4.21 ± 0.56	4.14 ± 0.53	4.13 ± 0.46	.453

Pairwise comparisons showing significant ($p < .05$) differences: ^{a} = different from normal, ^{b} = different from stiff, ^{c} = different from very stiff

Energetics at the knee and hip did not significantly change between footwear at all speeds except for positive knee work at 14 km/hr ($p = .011$, $\eta^2 = 0.396$), (Table 7.2). However, no significant pairwise comparisons between footwear were detected. Positive knee work was not different between footwear at 17 ($p = .448$, $\eta^2 = 0.085$) or 20 km/hr ($p = .253$, $\eta^2 = 0.142$). Negative knee work did not significantly change between footwear at 14 ($p = .226$, $\eta^2 = 0.152$), 17 ($p = .444$, $\eta^2 = 0.072$), or 20 km/hr ($p = .498$, $\eta^2 = 0.075$). Negative hip work did not change between footwear at 14 ($p = .061$, $\eta^2 = 0.267$), 17 ($p = .917$, $\eta^2 = 0.010$), or 20 km/hr ($p = .331$, $\eta^2 = 0.110$). Positive hip work did not change between footwear at 14 ($p = .389$, $\eta^2 = 0.099$), 17 ($p = .378$, $\eta^2 = 0.089$), and 20 km/hr ($p = .314$, $\eta^2 = 0.113$).

Table 7.2. Energetics of the metatarsophalangeal, ankle, knee, and hip joints across shoe conditions at three running speeds

Speed (km/hr)		Normal	Stiff	Very Stiff	P
14	MTP positive (J/kg)	0.075 ± 0.040	0.076 ± 0.036	0.083 ± 0.045	.362
	MTP negative (J/kg)	-0.382 ± 0.131[b,c]	-0.343 ± 0.111[a]	-0.308 ± 0.110[a]	**.012**
	Ankle positive (J/kg)	1.21 ± 0.217	1.283 ± 0.157	1.228 ± 0.187	.280
	Ankle negative (J/kg)	-0.988 ± 0.226	-1.043 ± 0.263	-1.085 ± 0.286	.071
	Knee positive (J/kg)	0.549 ± 0.152	0.515 ± 0.132	0.619 ± 0.165	**.011**
	Knee negative (J/kg)	-0.215 ± 0.106	-0.203 ± 0.109	-0.224 ± 0.116	.226
	Hip positive (J/kg)	0.488 ± 0.198	0.586 ± 0.342	0.522 ± 0.212	.389
	Hip negative (J/kg)	-0.044 ± 0.053	-0.017 ± 0.024	-0.021 ± 0.023	.061
17	MTP positive (J/kg)	0.101 ± 0.041	0.094 ± 0.041	0.101 ± 0.039	.584
	MTP negative (J/kg)	-0.504 ± 0.139[b]	-0.439 ± 0.146[a]	-0.412 ± 0.145	**.015**
	Ankle positive (J/kg)	1.392 ± 0.167	1.377 ± 0.180	1.373 ± 0.266	.879
	Ankle negative (J/kg)	-1.190 ± 0.315	-1.163 ± 0.323[c]	-1.244 ± 0.346[b]	**.036**
	Knee positive (J/kg)	0.636 ± 0.179	0.612 ± 0.150	0.658 ± 0.203	.448
	Knee negative (J/kg)	-0.210 ± 0.114	-0.220 ± 0.113	-0.231 ± 0.126	.444
	Hip positive (J/kg)	0.658 ± 0.235	0.801 ± 0.531	0.675 ± 0.256	.378
	Hip negative (J/kg)	-0.031 ± 0.026	-0.034 ± 0.058	-0.036 ± 0.039	.917
20	MTP positive (J/kg)	0.139 ± 0.057	0.124 ± 0.051	0.117 ± 0.039	.180
	MTP negative (J/kg)	-0.601 ± 0.156[c]	-0.568 ± 0.186	-0.522 ± 0.150[a]	**.003**
	Ankle positive (J/kg)	1.492 ± 0.251	1.458 ± 0.171	1.389 ± 0.165	.065
	Ankle negative (J/kg)	-1.409 ± 0.328	-1.372 ± 0.359	-1.367 ± 0.327	.505
	Knee positive (J/kg)	0.719 ± 0.221	0.673 ± 0.192	0.729 ± 0.228	.253
	Knee negative (J/kg)	-0.181 ± 0.117	-0.185 ± 0.133	-0.195 ± 0.133	.498
	Hip positive (J/kg)	0.803 ± 0.254	0.881 ± 0.511	0.733 ± 0.241	.314
	Hip negative (J/kg)	-0.031 ± 0.022	-0.042 ± 0.049	-0.050 ± 0.043	.331

Pairwise comparisons showing significant ($p < .05$) differences: [a] = different from normal, [b] = different from stiff, [c] = different from very stiff

Discussion

The purpose of this study was to investigate if biomechanical adaptations to footwear of varying LBS change across running speeds. Our hypothesis that changes in MTPJ dorsiflexion would be running speed dependent was not supported. At all three speeds the stiffer plates reduced MTPJ dorsiflexion (Table 7.1). Our hypothesis that maximum MTPJ moment would increase with the stiffer plates was also not supported. Maximum MTPJ moment did not change across LBS conditions at all three speeds. Our hypothesis that MTPJ angular resistance would increase with stiffer plates was supported at all three speeds. We also hypothesized that ankle moments would increase, which was not supported. Lastly, our hypothesis that energetics of the knee and hip would not change was supported (Table 7.2).

Torque about the MTPJ increases with running speed, and there should be a greater capacity to dorsiflex footwear as speed increases (Chapter IV). Previous data from Chapter VI showed that contact times at 17 km/hr were not different between shoes of varying LBS, as they were at 14 and 20 km/hr. It was speculated that perhaps the internal MTPJ moment at 17 km/hr would be large enough to overcome the stiff shoes and facilitate runners staying within their preferred range of MTPJ dorsiflexion (Chapter IV). The current study's findings suggest the contrary, that at all speeds the use of stiff plates systematically reduced MTPJ dorsiflexion by approximately five degrees (Table 7.1, Figure 7.2). Dorsiflexion of footwear occurs when the forefoot is depressed into the shoe to serve as the base of support as the ankle plantar flexor complex pulls the rearfoot superiorly and anteriorly. Previous work has shown that restriction of MTPJ dorsiflexion results in compensatory effects at proximal joints, and increases overall metabolic cost during running (Oh and Park, 2017). While metabolic cost has been shown to increase at 14 km/hr using stiff shoes, it did not increase at 17 km/hr (Chapter VI). These differences in observations between studies are most likely due to changes in lower limb mechanics between the running speeds tested (Chapter IV, Jin & Hahn, 2018; Orendurff et al., 2018). Oh and Park participants ran at speeds between 2.22 and 2.78 m/s (Oh and Park, 2017), compared to the 3.89 and 4.70 m/s speeds in our study (Chapter VI).

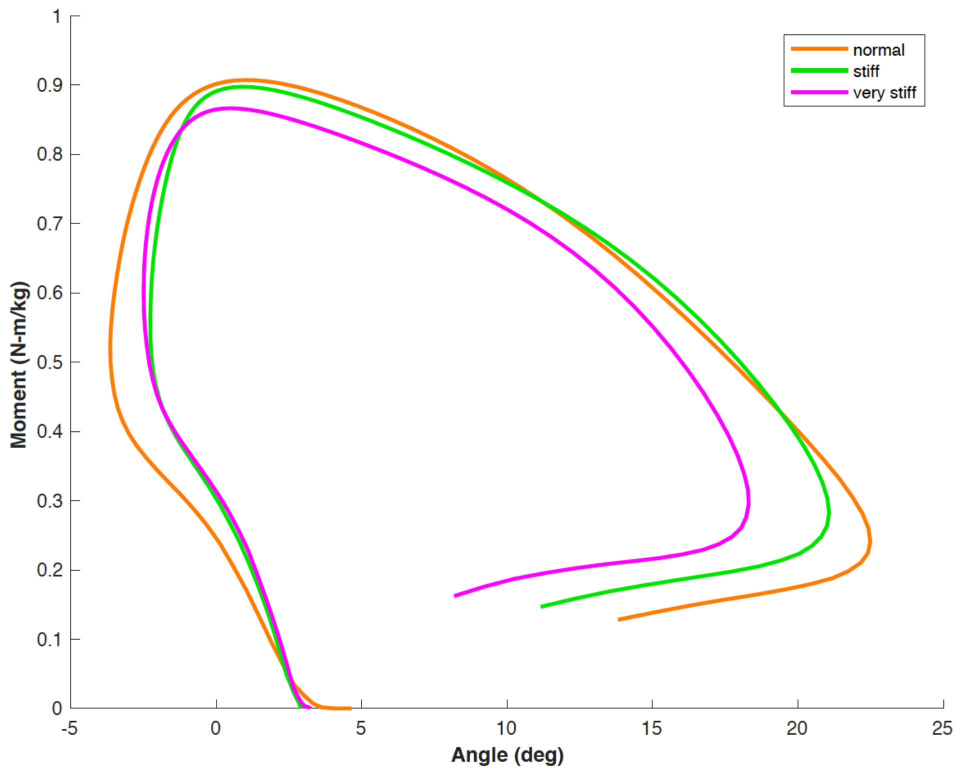

Figure 7.2. Average load-displacement curve of the metatarsophalangeal joint at 14 km/hr for three shoes of varying. Load-displacement curves showed similar trends at 17 and 20 km/hr

Maximum MTPJ moment did not change across footwear conditions at all three speeds (Table 7.1). Changes in maximum MTPJ moment in response to varying LBS have been mixed in the literature (Oh and Park, 2017; Roy and Stefanyshyn, 2006; Willwacher et al., 2013). The lack of increase in maximum MTPJ moment may be due to multiple mechanisms, including the capacity of the toe-flexor muscles to actively generate a plantar flexor moment about the MTPJ axis during stance phase (Farris et al., 2019). To adequately translate the center of pressure anteriorly, the toes need to be depressed into the surface they interact with (Endo et al., 2002). The ability to do so is directly related to toe-flexor strength (Endo et al., 2002), but the toe-flexor muscles are only able to actively generate between approximately 6 to 15 N-m of torque about the

MTOJ axis (Farris et al., 2019; Goldmann and Brüggemann, 2012). Thus, the remaining torque contribution about the MTPJ is likely due to passive structures. While our previous results demonstrate that the moment about the MTPJ increases in response to greater external demand (Chapter IV), it was also demonstrated that increased toe-flexor strength does not translate to increased MTPJ moments (Chapter V). It may be that since there is no external demand for a greater internal moment, the body maintains a naturally preferred kinetic profile.

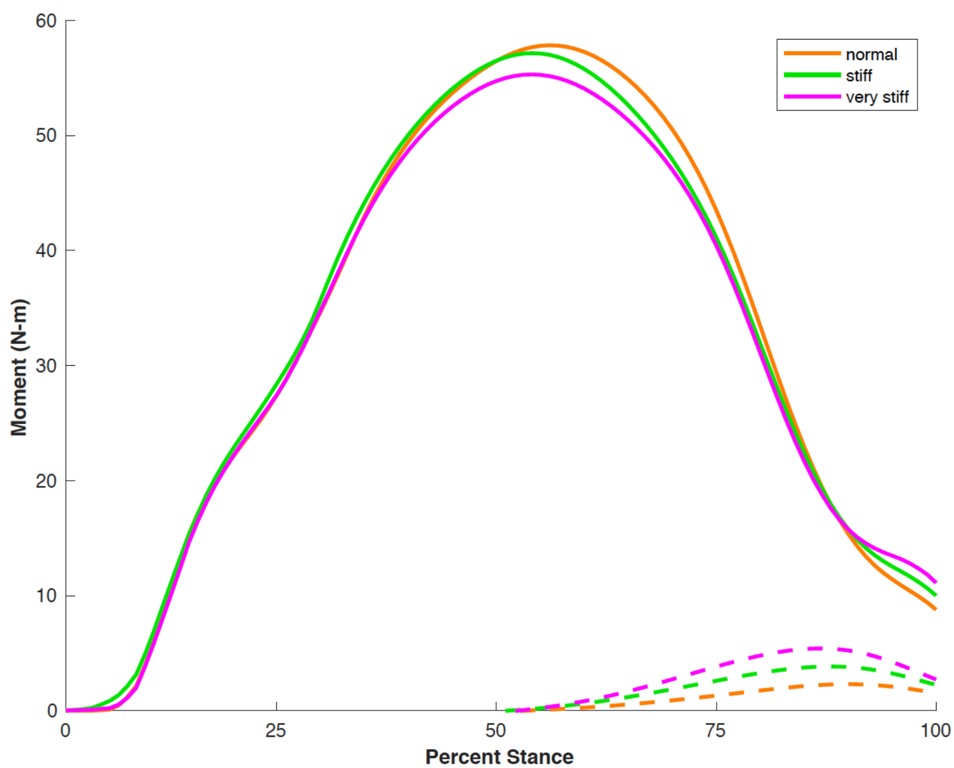

Figure 7.3. Mean metatarsophalangeal joint moments at 14 km/hr for the foot-shoe complex (solid lines) and shoe only (dashed lines). Joint moment profiles showed similar trends at 17 and 20 km/hr

Interestingly, the lever arm between the center of pressure and the MTPJ axis remained relatively constant throughout the majority of stance, affecting moment generating capacity (Figure 7.4). This is in contrast to Willwacher et al. who reported an increasing lever arm about the MTPJ and ankle joint during stance phase (Willwacher et al., 2014). We speculate that our observed difference could be due to differences between treadmill and overground running. While no studies to date have examined changes in MTPJ mechanics between running on a treadmill and overground, it has been observed that treadmill running is associated with a flatter foot landing position (Nigg et al., 1995), which may affect anterior progression onto the forefoot and subsequent MTPJ mechanics. There may also be differences in MTPJ and forefoot function due to the foot progressing posteriorly underneath the center of mass during treadmill running, in contrast to overground running where the foot plants on the ground and the center of mass travels anteriorly over the foot.

The observed magnitude of difference in lever arms between footwear conditions at all speeds was within a few millimeters (Figure 7.4). This suggests that LBS does not have a large influence on the capacity of an insole plate to function as a lever during treadmill running (Figure 7.4). The magnitude of the lever arm maintained a value of around 4cm throughout stance, until the very end where it sharply progressed, most likely due to final roll off of the forefoot. Interestingly though, the normal shoe condition had the steepest roll off whereas the very stiff shoe had the least steep roll off (Figure 7.4). This may indicate that when using the very stiff plates runners prefer to pick their foot up off the ground as opposed to roll forward off the toe, most likely due to the increased angular resistance of the plate. In a comparison of marathon racing shoes of variable LBS and midsole material during treadmill running, a shoe that was twice as stiff as the others displayed no difference in ground reaction force lever arms about the MTPJ or ankle joint (Hoogkamer et al., 2018). Our observed lever arm magnitudes somewhat contradict

Willwacher et al. who reported a steady progression to approximately 6 – 7.5cm using stiff plates (Willwacher et al., 2014). Average toe length of our participants, quantified as the distance between reflective markers placed on the lateral aspect of the first metatarsal head and the distal anterior aspect of the hallux, was on average 7-8.5cm. Altogether, these data suggest that rather than anteriorly progressing the center of pressure throughout stance during treadmill running, it may be preferred to move onto the forefoot and maintain the center of pressure closer to the MTPJ axis to utilize force distribution across the entire forefoot for push-off. The end result of this is a potential decrease in the ability of the forefoot and stiffening plates to act as a spring or lever during the stance phase of running; at least while running on a treadmill.

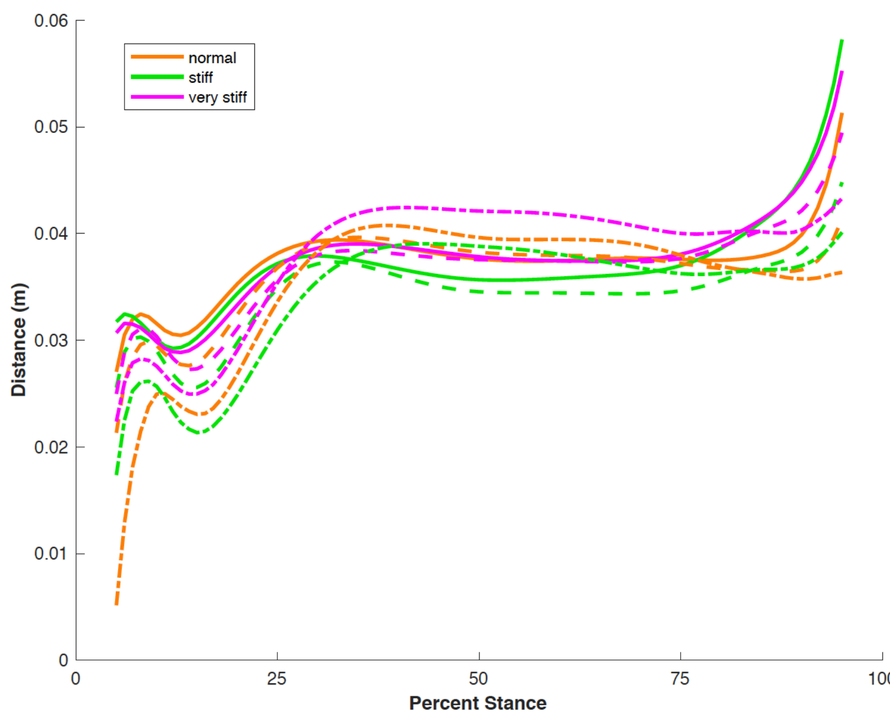

Figure 7.4. Average ground reaction force lever arm about the metatarsophalangeal joint axis in the sagittal plane. Solid lines represent 14 km/hr, dashed lines represent 17 km/hr, and dashed-dotted lines represent 20 km/hr.

Angular resistance (R_{cr}) about the MTPJ increased with LBS at all speeds (Table 7.1, Figure 7.2). This increase in angular resistance is predominantly due to restriction of MTPJ dorsiflexion. While the MTPJ moment at maximum dorsiflexion was not analyzed, visual assessment of the load-displacement plot at 14 km/hr reveals very little change in moment at that discrete time point. Of note, however, is that even though MTPJ dorsiflexion was restricted, there was still an increase in shoe extension torque with the stiffer plates (Figure 7.3). Because the foot and shoe act in parallel in regard to angular deformation about the MTPJ axis (Oleson et al., 2005), another potential reason for the lack of observed increase in MTPJ moment may be that the body is actually resolving optimal angular stiffness of the foot-shoe complex in an in-series manner as it does with linear deformation of the longitudinal arch and footwear midsole (Kelly et al., 2016). Once the shoe starts undergoing significant bending during the second half of stance, the shoe extension torque at the timing of maximum dorsiflexion accounts for approximately one-third to half of the total MTPJ moment (Figure 7.3). If the body is modulating torque output in an in-series manner, this would result in a decrease in the internally generated MTPJ plantar flexor moment due to the increase in torque contribution from the shoe. This would also help further explain the decrease in MTPJ dorsiflexion. Another potential reason may be that because the lever arm remained relatively constant throughout stance (Figure 7.4), the increase in joint moment would require extra muscular force production, which would be metabolically costly (Roberts et al., 1998).

Negative work at the MTPJ was systematically reduced by use of stiffer plates at all three speeds, in agreement with previous findings (Stefanyshyn and Nigg, 2000; Willwacher et al., 2013), (Table 7.2). This is primarily due to a reduction of MTPJ angular velocity, as there was a reduction in MTPJ range of motion and no increase in MTPJ moment. Positive work did not change in response to varying LBS at any speed (Table 7.2). If the stiff plates acted as a spring, there would be a proportional amount of positive to negative work about the MTPJ. The current data suggest that the plates do not serve as a spring or a lever during treadmill running. Their primary purpose may then to be to serve as a stiffening mechanism improving how the user perceives the shoe (Chapter VI).

In agreement with previous studies, we observed a decrease in negative ankle work (Roy and Stefanyshyn, 2006; Stefanyshyn and Nigg, 2000) (Table 7.2). This observed trend only occurred at 17 km/hr, however. Because there was no change in ankle range of motion or moment, the decrease in energy absorption must be due to a decrease in ankle dorsiflexion velocity. There was no change in positive ankle work, but a *post-hoc* investigation of peak plantar flexion velocity at 17 km/hr show an apparent reduction with increasing LBS (normal: 830 ± 74, stiff: 776 ± 92, very stiff: 749 ± 74 deg/sec). A decrease in ankle plantar flexion velocity is metabolically beneficial by shifting the force-velocity operating point of the ankle plantar flexors (Madden et al., 2015) and thus requiring less force (Roberts et al., 1998). The lack of observed change in positive ankle joint work is likely due to the increased duration of contact time, notably during the push phase of stance, when using footwear of increased LBS (Chapter VI, Willwacher et al., 2013). This decrease in plantar flexion velocity may partially explain the observed decrease in metabolic cost compared to the normal shoes in Chapter IV. In agreement with the aforementioned studies, we observed no change in knee or hip joint energetics at any speed (Roy and Stefanyshyn, 2006; Stefanyshyn and Nigg, 2000).

Limitations

A limitation of this study was that there was no longitudinal adaptation period to the new shoes. While Willwacher et al. utilized a short 200m warm-up to adapt to shoes of varying LBS, other studies have involved participants running in the shoes for a week prior to data collection (Stefanyshyn and Nigg, 2000; Willwacher et al., 2013). It should be noted however that runners have been reported to optimize stiffness of their lower extremity as quick as their first step on a new surface (Ferris et al., 1999). Secondly, there are a plethora of methods to assess and report footwear LBS. A novel method previously described (Chapter VI) was used for quantification of footwear LBS to determine shoe plantar flexion torque during stance. Our footwear had LBS values of 0.10, 0.18, and 0.30 N-m/deg, which is similar to those reported by Stefanyshyn et al. of 0.04, 0.25, and 0.38 and Hoogkamer et al. of 0.12, 0.16, and 0.32 N-m/deg, which were all tested in a fashion that included applying a point load that induced rotation of the footwear (Hoogkamer et al., 2018; Stefanyshyn and Nigg, 2000). Other testing procedures, such as the three-point bending test, make it near impossible to compare footwear LBS to ours to adequately compare results across studies. Lastly, it is unknown how layering the plates directly under the insole impacts the foot-shoe interaction as opposed to embedding the plates within the midsole. The difference in the foot interacting directly with a stiff plate and not cushioned midsole may affect subjective comfort of the shoe, which could in turn affect joint level mechanics.

Future work

Future work should further investigate the changes in ankle plantar flexor muscle-tendon unit dynamics and electromyographic activation in response to varying LBS at different running speeds. Because changes in ankle mechanics appear to be the reason behind the observed changes in metabolics in Chapter VI, a further understanding of the mechanism behind the improved running economy would be beneficial. Future work should also investigate the differences in MTPJ mechanics between treadmill and overground running, notably the use of the forefoot as a lever. Such findings would improve the ability to translate results from this study to overground running.

Conclusion

In conclusion, shoes of varying LBS directly affected MTPJ mechanics at all speeds. These effects washed out at the ankle, knee, and hip joints, with the exception of negative ankle work at 17 km/hr. It seems that stiffening plates do not function primarily as a spring or lever during steady-state treadmill running, but rather to generally stiffen and increase dynamic angular resistance of the MTPJ by restricting dorsiflexion. These stiffening effects may affect muscle force-velocity operating points of the muscle-tendon units that cross the MTP and ankle joints and subsequently impact running economy.

CHAPTER VIII

CONCLUSION

Summary of Results and Findings

This dissertation set out to further develop our understanding of mechanical function of the MTPJ during running and the factors that influence it. The roadmap by which this was accomplished started with developing a kinetic model of the MTPJ in Chapter III. Next, mechanical function of the MTPJ across a range of speeds relevant to training and racing for competitive distance runners was assessed in Chapter IV. Next, to change an internal factor contributing to MTPJ mechanics, it was investigated how increasing strength of the IFM affects joint level mechanics and running economy across a range of speeds in Chapter V. To change an external factor contributing to MTPJ mechanics, it was investigated how changing footwear LBS affects gross level mechanics and running economy at a range of speeds in Chapter VI. The same dataset was used to analyze changes in joint level mechanics in chapter VII.

The methodology developed in Chapter III will be useful for researchers interested in quantifying MTPJ moments, or including them in an inverse dynamics link-chain analysis. There is currently no agreed upon MTPJ joint center in the literature, and thus the comparison of the existing methods will also aid in readers ability to interpret results between studies. The developed model reduces variability in estimated MTPJ moments, and may more accurately represent the functional anatomy of the MTPJ axis by truly treating it as a hinge that is rotated about as opposed to a fixed joint center.

Chapter IV demonstrated how MTPJ mechanics change across running speeds. The main takeaway was that the maximum moment and the moment at maximum dorsiflexion increased with running speed, resulting in an increase in dynamic angular resistance about the MTPJ. Analysis of instantaneous resistance of the MTPJ throughout stance revealed that the foot is the primary modulator of MTPJ mechanics. Because control of the foot-shoe complex is dominated via the foot, it was hypothesized that perhaps MTPJ function could change with increased IFM strength. This hypothesis was investigated in Chapter V. Secondly, because the increase in dynamic angular resistance with running speed was due to an increase in joint moment, it was hypothesized that an increase in LBS of footwear proportional to the increase in dynamic angular resistance may serve as a framework for how to tune the LBS of running shoes. This hypothesis was investigated in Chapters VI-VII.

The results of Chapter V indicate that increased IFM strength does not alter MTPJ or ankle mechanics, contact time, or running economy. The purpose of this investigation was to examine if an increase in IFM strength may result in similar gait adaptations to that of using stiffened footwear. Despite an average increase in maximum isometric IFM strength of 30%, no one elicited any mechanical changes. There were a multitude of reasons to physiologically explain the findings. The main takeaway is that the function of the IFM may be to support the medial longitudinal arch as opposed to modulate MTPJ mechanics. Secondly, increasing IFM strength may not be a reasonable manner to increase the ability of the IFM to bend stiffer shoe about the MTPJ axis.

Chapters VI and VII examined the effect of varying LBS on gross and joint level mechanics, metabolics, and subjective comfort across a range of speeds. The main takeaway from these two chapters is that optimal tuning of footwear LBS may be running speed dependent. There were speed dependent effects of LBS on stride frequency, stride length, and contact time. These results further strengthen the importance of the relationship between external factors such as footwear and the bodies' ability to self-optimize movement parameters to minimize energy cost during running (Cavanagh and Kram, 1985). Subject-by-subject analysis revealed that some participants responded more strongly than others to the use of stiff footwear at the faster running speed. Analysis of joint level mechanics reveal that these responses may be due to changes in ankle mechanics, as MTPJ mechanics responded similarly to increased LBS at all speeds. Overall, findings from Chapters VI-VII suggest that a general increase in footwear LBS is preferred as running speed increases, both from a comfort and performance standpoint.

As a whole, these studies provide a method for analyzing MTPJ moments, a framework for understanding MTPJ function across running speeds, furthered our knowledge of the role of the IFM, and highlighted the importance of footwear effects on MTPJ function for distance running performance. Very broadly, this work will help academic and footwear researchers, athletes, and coaches. As such, future work should further investigate how to optimally tune footwear LBS on an individual by individual level. Such a method may further improve the changes in running economy observed in Chapter VI. It would also be of interest to assess increases in IFM strength and the effect on injury related variables. Overall, this work demonstrates that altering function about the smallest primary joint in the lower-limb affects gross level performance. However, some limitations may have affected the results of this study and more work is needed to further understand implementation in the athletic performance and academic research realms.

Limitations

One major limitation affecting the translation of these findings to sports performance or product development is the numerous factors that can introduce variability into the system. For Chapters III-V, participants were recruited across both sexes and ranging in body mass, foot anthropometrics, etc. One limitation to this, however, was the inability to recruit enough participants to functionally group individuals. Chapters VI-VII sought to reduce sources of variability by limiting data collection to males that can wear a size 10 shoe, but differences such as muscular strength, foot strike, and body mass still introduce variability into the system. Additionally, findings from this limited participant pool may not apply to other demographics, such as females. Across both studies, more participants spanning a greater variety of descriptors would be beneficial to fully understand how findings from this set of studies translates in the real world.

A second limitation was that all data collections were performed on a treadmill. There are inherent differences between treadmill and overground running, such as differences in plantar load distribution (Hong et al., 2012), ankle plantar flexor moments (Willy et al., 2016), and foot angle at contact (Nigg et al., 1995). Little is known about how MTPJ mechanics differ between treadmill and overground running, however. Because of exhibited differences in ankle and foot mechanics, there may be underlying unknown differences in MTPJ mechanics. If differences do occur, then translation between findings in this dissertation and overground running may be limited.

For Chapter V, a limitation was not recording data regarding changes in IFM muscle volume, tendon thickness, or arch structure. Additionally, recording electrical activity of the IFM would have been beneficial to understand if neural changes associated with the strength training protocol were present. Information about structural changes of the foot across time would have been beneficial in explaining the lack of observed mechanical changes, as opposed to being largely speculative.

The method for manipulating the footwear used in Chapters VI-VII required insertion of Nylon plates onto the foot-bed. These plates weighed approximately 50g each, and thus added substantial mass to the shoe. Equalizing the mass between all conditions was not feasible though, as 350g is too heavy for participants to run comfortably at 17 km/hr during the metabolic tests. Additionally, the Nylon plates inserted by being placed on the foot-bed of the shoe were uncomfortable to all participants, as rated by the low 'forefoot cushioning' subjective comfort scores. If custom footwear more representative of a market-available racing flat that included lighter stiffening plates such as carbon fiber and embedded within the midsole were used, then participant responses to the varying LBS would be more externally valid.

Lastly, there was no fatigue component associated with any protocol in this study. While running economy is a direct indicator of performance (Hoogkamer et al., 2016), joint level kinematics have been reported to change throughout the course of a run (Weir et al., 2018). Thus, it remains unknown how increased IFM strength or varying LBS will affect mechanics and metabolics when in a fatigued state.

Recommendations for Future Work

This dissertation has opened up a variety of potential research paths involving MTPJ mechanics and the role of the IFM and footwear in distance running performance.

First, a large database of running mechanics analyses using at minimum a two-segment foot model with a defined MTPJ axis should be established. Existing literature investigating changes in LBS has divided participants in responders and non-responders (Madden et al., 2015; Willwacher et al., 2014) or hypothesized body mass may be a primary determining factor on the effectiveness to benefit from increased LBS (Roy and Stefanyshyn, 2006). A large database involving individuals of both sexes and a full spectrum of body masses, muscular strength, anthropometric differences, etc. would allow for further in-depth analysis of understanding what constitutes individuals being a responder or non-responder to the use of increased LBS. Statistical methods such as cluster analysis or principal component analysis may reveal the largest contributing factor to variation in the data, or successfully group individuals into responder or non-responder groups. Further understanding of the mechanism behind being a responder or non-responder will aid in the ability for more individuals to utilize increased LBS to improve performance.

Future work should investigate the effect of increased IFM strength or varying LBS under fatiguing conditions. Because joint level mechanics change even during a non-fatiguing run (Weir et al., 2018), MTPJ mechanics and the foot-shoe complex interaction may change throughout a run as well, especially under fatiguing conditions. Further investigation on this front will aid in the understanding of how either factor directly contributes to performance.

A primary takeaway from this dissertation was that increased IFM strength does not affect joint level mechanics or running economy. Of notable interest is that the IFM primarily act isometrically during longitudinal arch compression, and total muscle-tendon unit excursion is undertaken by the tendon (Kelly et al., 2018a). Training paradigms targeting tendon remodeling may be a successful method for how to internally influence MTPJ mechanics, especially because passive MTPJ stiffness is related to running economy (Man et al., 2016).

Lastly, an improved understanding of alterations in ankle plantar flexor muscle-tendon unit mechanics would be of benefit. Understanding differences in contractile versus series-elastic contributions to plantar flexor moments, changes in electromyographic activation, or changes in force-length or force-length velocity operating points in response to footwear of varying LBS would greatly enhance our ability to understand the mechanisms behind the observed physiological changes in Chapter VI.

This entire dissertation was focused on improving performance and distance runners, and for the most part largely neglected the increased potential risk of injury. Prior to translating findings from this dissertation to application, notably the use of footwear of increased LBS, it is advised that research is undertaken investigating the effects of the associated changes in joint mechanics on loading redistributions throughout the foot and leg and their potential influence on development of injury.

With these areas of work pursued, a more holistic understanding of the influence of changing MTPJ mechanics via internal or external factors will help both researchers, clinicians, and footwear developers improve performance and understand any potential associated injury risk in doing so.

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