NEUROMECHANICAL CONTRIBUTIONS TO LOWER EXTREMITY STIFFNESS DURING RUNNING AND HOPPING IN HEALTHY RUNNERS

Jonathan (FJ) Stephen Goodwin

A dissertation submitted to the faculty at the University of North Carolina at Chapel Hill in partial fulfillment of the requirements for the degree of Doctor of Philosophy (Biomechanics) in the Human Movement Science Curriculum in the Department of Allied Health Sciences.

Chapel Hill
2018

Approved by:

J. Troy Blackburn

Darin A. Padua

Eric D. Ryan

Jason R. Franz

Todd A. Schwartz

ABSTRACT

Jonathan (FJ) Stephen Goodwin: Neuromechanical contributions to lower extremity stiffness during running and hopping in healthy runners
(Under the direction of J. Troy Blackburn)

Lower extremity stiffness (K_{Leg}) describes how subjects attenuate load during ground contact while completing dynamic tasks. Alterations in K_{Leg} are associated with increased risk for lower extremity injury. Previous data suggests that lesser mobility during a clinical exam is associated with greater K_{Leg} in healthy runners. The purpose of our study was to analyze the neuromechanical contributions to K_{Leg} during running and hopping in healthy runners. Additionally we analyzed the relationship between running and hopping while also examining the feasibility of utilizing a waist-mounted accelerometer to estimate K_{Leg} in a clinical setting. We analyzed 70 healthy runners with a 2 session cross-sectional study. We collected musculotendinous stiffness of the ankle plantarflexors and knee extensors in session 1. In session 2, we collected K_{Leg} during self-selected running as well as single leg hopping at 3 frequencies (1.5 Hz, self-selected, 3.0 Hz). We also collected waist-mounted accelerations as well as muscle activation of the ankle plantarflexors and knee extensors. We found that at self-selected frequencies and higher, greater K_{Leg} during single leg hopping is significantly associated with greater ankle plantarflexor musculotendinous stiffness, greater ankle plantarflexor muscle activation and greater hopping frequency. Greater K_{Leg} during running is significantly associated with greater knee extensor musculotendinous stiffness, lesser hip internal range of motion and

greater running velocity. We found that subjects who demonstrated greater K_{Leg} during single leg hopping also demonstrated greater K_{Leg} during running however this significant relationship was only minimal. Finally, our waist-mounted accelerometer significantly overestimated K_{Leg} across all hopping frequencies. Out study found that active muscle contraction and greater musculotendinous stiffness of the ankle plantflexors and knee extensors are associated with greater K_{Leg} during hopping and running, respectively. These may serve as rehabilitative targets to alter K_{Leg} in the clinical setting. Additionally, assessing K_{Leg} via hopping and with a waist-mounted accelerometer does not accurately reflect K_{Leg} during running determine via motion capture. Additional studies should be completed to improve the clinical assessment of K_{Leg} to reduce the occurrence of lower extremity injuries.

To my wife, for her unyielding support and commitment
I could not have done this without you.

ACKNOWLEDGEMENTS

There have been numerous mentors who have shaped me throughout my career and this list will certainly not be exhaustive. Firstly, Dr. Ray Moss at Furman University introduced me to research, the scientific method and the joy of being a "lab rat." Additional faculty who fostered me at Furman include Dr. Bill Pierce, Dr. Tony Caterisano, Dr. Matt Feigenbaum, Dr. Julian Reed, Dr. Tim Patrick, & Dr. Scott Murr. Following my undergraduate degree, I pursued my Doctorate of Physical Therapy at East Carolina University. My time at ECU allowed me to develop my abilities with Dr. Blaise Williams and Dr. Walt Jenkins who pushed me to blend the clinical and research worlds to create meaningful, impactful discoveries in the lab. Finally, I would like to acknowledge Dr. Troy Blackburn at the University of North Carolina at Chapel Hill who has gone above and beyond to ensure I become an independent researcher capable of developing and sustaining my own research line. I am forever indebted to him as well as my committee of Dr. Darin Padua, Dr. Eric Ryan, Dr. Jason Franz, & Dr. Todd Schwartz. Their guidance enabled this project along with additional UNC faculty including Dr. Brian Pietrosimone, Dr. Jason Mihalik, and Dr. Erik Wikstrom.

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K_{Leg}	Lower extremity stiffness
K_{Vert}	Vertical stiffness
MTS	Musculotendinous stiffness

Chapter 1: Introduction

Running is an increasingly popular mode of exercise with over 48 million participants in the United States in 2015. [1] Running is associated with reduced mortality and disability [2, 3] but also incurs notable risk for lower extremity injury. Lower extremity injuries occur in up to 79% of individuals who routinely run more than 5 kilometers, [4] and these injuries lead to reduced activity, prolonged recovery, healthcare utilization, and time away from work. [5, 6] A conservative estimate of a 35% lower extremity injury rate in US runners would incur an annual financial burden of over $3.8 billion. [7] Current approaches of strength training, stretching, foot wear selection and training alterations are not consistently effective for reducing lower extremity injury risk. [8] Continued research is necessary to identify individuals at heightened risk of running-related injuries and clinical targets for intervention.

Greater loading rates during dynamic tasks such as running have been linked to greater risk for bony injury, [9-12] while lesser loading rates have been linked to greater soft tissue injury risk. [13, 14] Lower extremity stiffness (K_{Leg}) quantifies resistance to deformation of the lower limbs (e.g. flexion) during the ground contact phase of dynamic tasks, and can be manipulated by the neuromuscular system to alter loading rate. [15] Greater K_{Leg} is associated with a shorter ground contact time, [16] thus resulting in greater loading rate and magnitude [17] given the impulse required to decelerate and subsequently accelerate the center of mass during ground contact. Subjects with a history of Achilles tendinopathy demonstrate lesser K_{Leg} [18] while subjects with a history of tibial stress fracture demonstrate lesser ankle dorsiflexion and talocrural mobility, suggesting greater K_{Leg}.[19, 20] These data suggest that excessive or insufficient K_{Leg} is a modifiable neuromuscular characteristic that is associated with loading rates and subsequent lower extremity injury risk. As such, K_{Leg} could be a clinical target for identifying individuals at heightened risk of lower extremity injury and modifiable variable for reducing injury risk.

K_{Leg} is derived from a multitude of neuromechanical and anatomical factors.[21, 22] We previously demonstrated that less passive foot, ankle, and hip mobility identified during a clinical exam explained 49% of the variance in K_{Leg} during running. However, factors explaining the remaining variance have yet to be elucidated. K_{Leg} is regulated in large part by lower extremity muscle activity,[23] and investigation of neural drive to the musculature and the viscoelastic properties of musculotendinous units warrants further evaluation to provide a more complete understanding of mechanisms of action for changing K_{leg}, thereby insight into targets for clinical intervention.[24] For example, musculotendinous stiffness (MTS) quantifies a muscle's resistance to lengthening and subsequent joint motion,[25] and potentially plays a critical role in determining K_{Leg}.

K_{Leg} is typically evaluated by calculating the ratio of the peak vertical ground reaction force to the downward displacement of the total body center of mass during hopping via motion capture,[26] and is synonymous with vertical stiffness (K_{Vert}) during hopping. However, K_{Leg} during running requires additional information regarding leg length, running velocity, contact time and half the arc swept by the leg.[27] Estimating K_{Leg} during running is impractical in the clinical setting given the associated time, cost and expertise requirements. K_{Leg} is commonly assessed during hopping in the literature, and this approach is potentially more clinically feasible, particularly with the advancement of wearable technology (e.g. waist-mounted accelerometers). [28] However, it is unclear if simple accelerometry can be used to assess the complex behavior of the lower extremity. Previous research demonstrates that accelerations measured on the distal tibia are correlated with loading rate during running, [29, 30] and pelvis-mounted accelerometers are sensitive to center of mass acceleration. [31] Advancements in accelerometer technology may yield real time acceleration, and subsequently loading rate, feedback that would allow runners to monitor and potentially alter lower extremity loading during running. [28, 32]

Additionally, it is unclear if K_{Leg} during hopping is a valid representation of K_{Leg} during running. [17, 33] K_{Leg} calculated during hopping typically assessed within a fixed frequency ranging1.5-3.2 Hz. [16, 34-37] However, running requires forward propulsion along with a greater variability in foot contact, forward velocity, and stride frequency. [38-40] As noted above, calculating K_{Leg} during hopping may represent a clinically-feasible assessment whereas the additional data require to obtain these data during running require dedicated laboratory space and expertise. Therefore, it is imperative to determine if K_{Leg} during hopping reflects K_{Leg} during running to improve identification of individuals with heightened risk for running-related lower extremity injury.

Further understanding of the neuromechanical contributors to K_{Leg} and identifying clinical K_{Leg} assessment strategies is crucial for identifying individuals at greater risk for lower extremity injuries and informing interventions designed to alter K_{Leg}. The ***long-term objective*** of this research is to reduce the risk and financial burden of lower extremity running injuries. The ***overall objective*** of this application is to identify neuromechanical contributors to K_{Leg}. A secondary objective is to evaluate potential clinical indicators of K_{Leg}. Our ***central hypothesis*** is that lesser passive mobility, greater MTS, and greater muscle activation was associated with greater K_{Leg}. We will also assess the role of MTS in mediating the influence of muscle activation on K_{Leg}. We will test our hypotheses by assessing the following specific aims.

Specific Aims:

1. **To identify neuromuscular contributions to K_{Leg} during hopping and running.**

   Eccentric activity of the ankle plantarflexors and knee extensors likely influences lower extremity flexion, and therefore K_{Leg}. We hypothesize that greater K_{Leg} will be associated with greater ankle plantarflexor MTS and greater knee extensor MTS. Greater K_{Leg} would also theoretically be associated with greater ankle plantarflexor and knee extensor activation. However, we hypothesize that MTS would mediate the influence of muscle activation on K_{Leg} such that individuals with greater MTS will require lesser muscle activation to achieve a given level of K_{Leg}. These variables was combined with the aforementioned passive clinical measurements from our previous investigation in a multiple regression model to establish a more robust estimate of K_{Leg}.

2. **To determine the relationship between K_{Leg} during hopping via laboratory (motion capture) and clinical (waist-mounted accelerometer) measurement techniques.**

   Advancements in wearable technology may allow for assessment of K_{Leg} in the clinical setting during hopping via waist-mounted accelerometry. We hypothesize that greater K_{Leg} assessed via laboratory methods during hopping was associated with greater K_{Leg} derived from a waist mounted accelerometer. Additionally, we will determine if K_{Leg} derived from accelerometry is sensitive to changes in K_{Leg} derived from motion capture induced by an increase in hopping frequency. We anticipate that changes in K_{Leg} identified via accelerometry was correlated with changes in K_{Leg} derived via motion capture. Additionally, we hypothesize that loading rate and peak force derived from accelerometry will be positively correlated with loading rate and peak force derived from motion capture.

3. To determine the relationship between K_{Leg} during running and hopping.

K_{Leg} is most commonly assessed during hopping and directly reflects K_{Leg} given the lower extremity's vertical orientation. However, it is unclear if K_{Leg} during hopping reflects K_{Vert} and K_{Leg} during running. We hypothesize that K_{Leg} during single-leg hopping will be correlated with K_{Vert} during running. Additionally, we hypothesize that K_{Leg} during single-leg hopping will be positively correlated with K_{Leg} during running.

It is expected that by obtaining our objectives we will gain a deeper understanding of the contributions to K_{Leg} during dynamic activities. This will allow for an improved clinical assessment and define rehabilitation targets. This will serve to reduce the risk and financial burden of running injuries.

<u>Potential pitfalls:</u>

K_{Leg}, along with MTS and muscle activation, are influenced by many factors including type of activity, running and hopping velocity, fatigue, age, and current training level. In order to limit the influence of external factors, we will focus our study on adult runners (age 18-40) who run a minimum of 15 miles/week. Our subjects will be free from lower extremity orthopedic injury during the 6 months prior to participation to ensure they are able to complete the study protocol without pain or altered running/hopping biomechanics. These precautions will allow our study cohort to reflect an endurance trained group while minimizing external confounding variables. We will ensure our subjects meet the minimum activity requirements, but there may be an unforeseen effect of greater training distance volume (e.g. Subject A runs 17 miles/week while Subject B runs 75 miles/week). In order to account for this we will record current training volume along with previous best times completed for the preferred race distance (e.g. 5k, 10k, or marathon) for use as potential covariates. Additionally, previous research has indicated that greater K_{Leg} is independently associated with greater MTS and greater lower extremity EMG activation. However, greater MTS with likely mediate and subsequently reduce the amount of EMG required to generate the same resistance to lengthening of the tissue. Therefore, we anticipate an inverse relationship between MTS and EMG when combined in a multiple regression model to predict K_{Leg}

Chapter 2: Literature Review

Running injuries

Running is an increasingly popular mode of exercise with over 64 million participants in formal races in the United States in the spring of 2016. [41] Adults often select running for the positive health benefits [42-44] including reduced mortality and disability. [2, 3] Lee et al. [45] found that runners displayed a 45% lower risk of cardiovascular mortality, the most common cause of death in the United States, compared to non-runners over a 15 year follow-up. However, as with any athletic exposure, there is an increased risk for musculoskeletal injury during running participation, which may serve as major deterrent to continued running participation. [46]

Running related injuries occur in up to 79% of individuals who routinely run more than 5 kilometers, and 84% of novice runners experience a running related injury that results in time lost from training. [4, 47] In conjunction with reduced participation in physical activity, these injuries lead to prolonged recovery, healthcare utilization, and time away from work. [5, 6] Hespanhol et al. [48, 49] estimated that each running related injury incurs $188-196 in costs associated with healthcare utilization and absenteeism from work. [48, 50] The majority (>90%) of injured runners enter the healthcare system for injury management and treatment. [51] Healthcare utilization is often initiated via a general medical practitioner and often involves referral to a specialist (e.g. physical therapist). [52] A conservative estimate of a 35% lower extremity injury rate in US runners with 90% healthcare utilization rate would incur an annual financial burden of over $3.8 billion.

Runners experience a wide variety of injuries in the lower extremities. Over 80% of running related injuries in collegiate cross country athletes occur in the lower extremity, [53] and up to 79% of recreational and competitive runners experience a lower extremity running related injury. [4] There is no standard definition to categorize running related injuries. However the literature typically defines injuries based on 3 domains including a physical compliant, reductions in training/competition, and/or seeking medical attention. [54, 55] Lower extremity injuries can be further categorized as a single, identifiable traumatic event or the more frequent repetitive sub-traumatic injury which occurs gradually over time without appropriate healing. Additionally, running related injuries can be loosely grouped into “hard” tissue (bony/cartilage) injuries and “soft” tissue (ligament/musculotendinous) injuries. [55] The most frequent running related injuries include both bony and soft tissues injuries of the lower extremity: patellofemoral syndrome, iliotibial band syndrome, Achilles tendinopathy, plantar fasciitis, and medial tibial stress syndrome. [56, 57]

Several researchers have attempted to prospectively predict running related injury risk, but with limited success and often conflicting results. Prior history of running related injuries puts runners at increased risk for further injury. [4, 58-60] Therefore, it is imperative to reduce initial injury risk to prevent increased risk of future injury. Total distance run and intensity of miles run are commonly used as markers for cumulative load experienced by the runner. This is supported in the literature which has found that with increasing distance there is an increased risk for overuse injury. [4, 59, 61] Van Gent et al. [4] completed a systemic review demonstrating that greater distance run is associated with a greater risk of running related injuries. However this was only significant in males and found to not be a significant factor in females. Additionally, greater training distance was protective against running related knee injuries. Furthermore, intensity of total distance run is frequently used to assess injury risk. Hespanhol et al. [60] found that runners who completed "speed training" displayed greater risk for injury, but this was not reflected by van Poppel et al. [62] who found that runners who routinely participated in interval training displayed a lesser injury risk compared to those who did not. These discrepancies in the effects of training distance and intensity on injury risk highlight the difficultly of accurately and proactively assessing injury risk. A more tailored approach to assess the runner's internal response to the total external loading experienced during running may provide a more useful injury risk assessment.

Running kinetics and kinematics provide a more informed approach to injury risk for both soft and hard tissue running related injuries. Running subjects the lower extremity to repeated landings that are predominantly mitigated by lower extremity joints and musculature. [23] Mitigating these repeated impact forces appropriately is important for reducing running injury risk. [63] Hreljac et al. [64] found that injury free runners demonstrated significantly lesser peak vertical ground reaction force magnitude and loading rate compared to previously injured runners. This relationship between loading rates and injury risk was further supported by Davis et al. [11] who found female runners who had never been injured displayed significantly lower loading rates compared to those who went on to become injured. Greater loading rate is also associated with bony and cartilage damage in animal models. [9, 10, 65, 66] Mitigating loading rate is predominantly managed by lower extremity articulations and musculature. The kinematic response to loading also plays a role in injury development. Hamstra-Wright et al. [67] found that greater navicular drop, ankle plantarflexion, and hip external range of motion were risk factors for medial tibial stress syndrome. Dudley et al. [68] found prospectively that runners who displayed greater peak knee external adduction moment and greater peak ankle eversion velocity were more likely to become injured over the course of a cross country season. Milner et al. [69] found that runners with a previous history of tibial stress fracture displayed greater knee stiffness during the initial loading phase during running. Collectively, these data suggest that runners who display greater loading rates are at a greater lower extremity injury risk. Clearly, an assessment that accounts for both external load experienced by the runner and the associated kinematic response is imperative to fully quantify injury risk and provide an individualized assessment

Lower extremity stiffness (K_{Leg}) is influenced by both lower extremity kinematics and kinetics, and may provide a more robust indicator of running related injury risk in the lower extremity. K_{Leg} can be examined further based on a joint analysis (e.g. knee stiffness vs ankle stiffness); however, a more incisive analysis of the neuromuscular contributions to K_{Leg} is warranted. This would elucidate the underlying mechanisms of how the lower extremities modulates impact loading during dynamic activities. A deeper understanding of these neuromuscular contributions would allow for more effective, targeted rehabilitation practices and potentially serve as an identifier for preventative interventions.

Lower extremity stiffness

Stiffness describes the ability of an object to resist deformation. [27] More specifically, this property is derived from Hooke's Law which is defined as $F = kx$ where the force required to deform an object (F) is related to a spring constant (k) and the magnitude of linear deformation (x). The spring constant (k) represents the relationship of force divided by displacement (F/x), and is commonly referred to as linear stiffness. K is modeled as a massless constant that is only capable of deformation in a single plane. K_{Leg} represents the stiffness of the lower extremity, and is commonly estimated via a spring mass model describing downward displacement of the lower extremity or total body center of mass on a massless spring during running and hopping, respectively. [70, 71] Despite the relatively simple initial assessment, K_{Leg} is multifactorial, involving underlying biomechanical function, neuromuscular control, and contributions from all involved ligament, tendons, cartilage, bones, and joints. [22, 27]

K_{Leg} is more specifically represented by the ratio of the peak vertical ground reaction force (F_{max}) to the compression /downward displacement of the lower extremity (ΔL) during running ($F_{max}/ \Delta L$) (Figure 2.1). ΔL represents the change in leg length ($\Delta L = \Delta y + L_0 (1-\cos \Theta)$ with $\Theta = \sin^{-1}(ut_c/2L_0)$, where Δy = maximum vertical displacement of the center of mass; L_0 = standing leg length; Θ = half angle of the arc swept by the leg; u = horizontal velocity; and t_c = contact time). Vertical stiffness (K_{Vert}) is a more simplistic model of the lower extremity that is also commonly reported in the literature as the ratio of the peak vertical ground reaction force to the peak vertical displacement of center of mass ($K_{Vert} = F_{max} / \Delta y$) (Figure 2.2). [27, 71] K_{Vert} is synonymous with K_{Leg} during repetitive vertical hopping given there is no horizontal velocity. However, K_{Vert} differs from K_{Leg} during running due to the fact that K_{Leg} accounts for relative leg compression in a

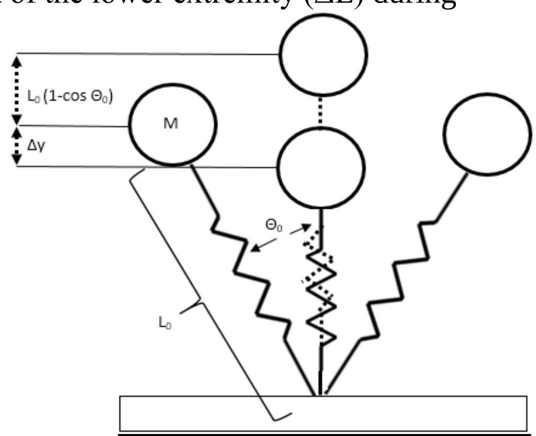


Figure 2.1: Ideal spring mass model for K_{Leg} with the leg orientated in non-vertical position.

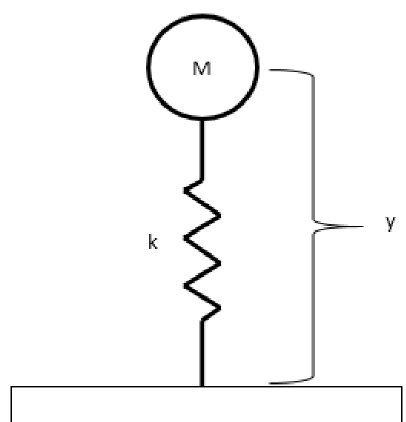


Figure 2.2: Ideal spring mass model for K_{Vert} with the leg

non-vertical alignment along with center of mass displacement whereas K_{Vert} accounts for center of mass vertical displacement only.

K_{Leg} is vital for locomotion and describes the ability of lower extremity to modulate the impulse associated with ground impact. K_{Leg} is highly variable based on task demands and allows the lower extremity to serve as a mobile adapter and allow for weight acceptance and subsequent forward propulsion during running. Additionally, greater K_{Leg} is linked with greater performance parameters. Rogers et al. [72] demonstrated that greater K_{Leg} was associated with faster and more economical running performance in well trained middle distance runners. This is also supported by Nagahara et al. [73] who found that improvements in sprint performance over 6 months of training were associated increases in K_{Vert}. K_{Vert} and K_{Leg} are dependent on and influenced by task, surface, training history, contact time, age, fatigue, and previous injury. [27]

K_{Leg} is task dependent and is most commonly assessed during hopping. [16, 18, 35, 36, 74-78] K_{Leg} can also be obtained during running, however it requires a greater amount of variable control to reproduce the same conditions among subjects (e.g. running velocity, contact time, and arc swept by the leg) whereas the only variable controlled during hopping is frequency. Therefore, K_{Vert} during hopping is reported more commonly in the literature.[38-40, 79] Preferred human hopping occurs at or near 2.2 Hz [21, 77, 80] with ranges for comfortable hopping occurring between 1.5 and 3 Hz. Hopping frequencies outside of these ranges may not accurately represent simple spring mass model characteristics. [77, 80-83]Table 2.1 provides a summary of characteristics from studies during which K_{Vert} was assessed during hopping.

Study	Population	Hopping Task/Instructions	Hopping Frequency (Hz)
Farley et al. 1991 [80]	Healthy males and females [N=4 (2M,2F, age = 20-22]	Imposed frequency with metronome; "Preferred frequency"	1.2-3.6; 2.17
Ferris et al. 1997 [34]	Healthy males and females [N=5 (2M,3F, age = 19-26]	Imposed frequency with metronome	2.0, 2.4, 2.8, 3.2
Farley et al. 1998 [21]	Healthy males and females [N=7 (3M,4F, age = 24]	Imposed frequency with metronome	2.2
Voigt et al. 1998 [84]	Healthy males [N=6, age = 31]	Imposed frequency with metronome	2.0
Farley et al. 1999 [85]	Healthy males and females [N=5 (2M,3F, age = 20-23]	Imposed frequency with metronome	2.2
Granata et al. 2002 [81]	Healthy males [N=15, age = 32] vs healthy active females [N=15, age = 32]	Imposed frequency with metronome; "Preferred frequency"	3.0. 2.5; 2.3
Moritz et al. 2004 [86]	Healthy males [N=8, age = 28]	Imposed frequency with metronome	2.2
Padua et al. 2005 [82]	Physically active males [N=11, age = 27] vs physically active females [N=10, age = 24]	Imposed frequency with metronome; "Preferred frequency"	3.0; 2.3
Hobara et al. 2007 [76]	Physically active males [N=7, age = 23]	"Preferred vs short ground contact time"	2.12, 2.11
Hobara et al. 2008 [36]	Power trained males [N=7, age = 20] vs male runners [N=7, age = 20]	Imposed frequency with metronome	3.0, 1.5
Lloyd et al. 2009 [87]	Physically active young males [N=18, age = 13]	Imposed frequency with metronome	2.0, 2.5
Hobara et al. 2010 [16]	Physically active males [N=10, age = 22]	Imposed frequency with metronome	1.5, 2.2, 3.0
Hobara et al. 2010 [75]	Untrained males [N=8, age = 24] vs male runners [N=8, age = 19]	Imposed frequency with metronome	2.2
Oliver et al. 2010[77]	Physically active males [N=8, age = 19-30] vs physically active boys [N=11, age =11-12]	Imposed frequency with metronome; "Preferred frequency"	1.5, 3.0; 1.8-2.0
Kuitunen et al. 2011 [74]	Physically active males [N=8, age = 29]	"Hop with the shortest ground contact time possible"	2.9 – 1.8
Hobara et al. 2014 [83]	Physically active males [N=10, age = 28]	Imposed frequency with metronome	2.2. 2.6, 3.0, 3.4
Hobara et al. 2015 [35]	Sedentary males and females [N=11 (5M,6F), age = 29] vs sedentary old males and females [N=11 (5M,6F), age = 67]	Imposed frequency with metronome	2.2, 2.6, 3.0

Table 2.1: Representative sample of hopping studies with respective populations and hopping frequencies

Greater K_{Leg} is associated with greater hopping frequency and running velocity. Hobara et al. [16, 35] found significant increases in K_{Vert} with increased hopping frequency from 1.5 Hz to 3.0 Hz and from 2.2 Hz to 3.0 Hz. By increasing the hopping frequency, there is a reduction in ground contact time which limits the amount of center of mass displacement (Δy) during ground contact and increases K_{Vert}. Greater K_{Leg} induced by increased hopping frequency is supported by Granata et al. [81] who found greater K_{Leg} when subjects increased their hopping frequency to 3 Hz from their preferred hopping frequency of 2.34 Hz. This change in stiffness was predominantly influenced by a reduction of center of mass displacement (Δy). Aramapatzis et al. [88] had subjects complete drop jump landings from 20 cm, 40 cm, and 60 cm height. Subjects were then instructed to obtain maximum jump height with the shortest amount of contact time following initial landing. The subjects were then stratified into 5 groups with group 1 having the longest contact time and group 5 having the shortest contact time. Across all drop jump landing heights, group 5 displayed significantly lesser Δy and significantly greater F_{max} which resulted in significantly greater K_{Leg}. Greater K_{Leg} is also reflected during running with greater running velocity and greater stride frequency. [89] Arampatzis et al. [90] had subjects run at five different running velocities between 2.5 and 6.5 m/s and reported greater K_{Leg} with increasing velocity primarily due to an increase in F_{max}. Farley and Gonzalez [38] had subjects run at a variety of stride frequencies at 2.5 m/s on a treadmill that ranged from a 26% reduction in stride frequency to a 35% increase in stride frequency relative to the preferred frequency, and reported a 2.3x increase in K_{Leg} between the lowest and highest stride frequencies despite the fixed running velocity.

K_{Leg} can be modulated within the first step of landing of a new surface with unexpected stiffness. [15, 91] Ferris et al. [15] found that runners are able to decrease K_{Leg} by 29% when transitioning from running on a compliant surface to a hard surface. This reduction in K_{Leg} allowed subjects' center of mass displacement to remain consistent despite a reduction in surface compression from 6 cm to 0.25 cm. [15] Similarly, K_{Leg} can be increased to compensate for reductions in surface stiffness. [91] This increase in K_{Leg} allows the runner to minimize center of mass disturbances to maintain overall trajectory and subsequent forward propulsion. This maintenance of total overall stiffness is reflected during hopping as well. Ferris et al. [34] found that healthy young adults modulate their K_{Leg} to maintain a similar overall total system stiffness when hopping on a more compliant surface.

K_{Leg} is predominantly influenced by active contributions from the musculature of the lower extremity that alter individual joint stiffnesses. [22, 27, 88, 90] There is discrepancy in the literature regarding which joint musculature has the greatest influence on K_{Leg} during dynamic tasks. Several studies have demonstrated that knee stiffness has the greatest influence on $K_{Leg.}$ Hobara et al. [92] found that knee joint stiffness serves as the primary determinant of K_{Leg} during maximal vertical hopping with smaller contributions from the hip and ankle joints. This is also supported by Arampatzis et al. [93] who found that increases in K_{Leg} during drop jump landings from 20 cm and 40 cm were driven by increases in both ankle and knee stiffness. However, there was a stronger correlation between K_{Leg} and knee stiffness compared to ankle stiffness at both 20 cm ($K_{Ankle} = 0.52$, $K_{Knee} = 0.69$) and 40 cm ($K_{Ankle} = 0.18$, $K_{Knee} = 0.74$). Hobara et al. [16] also reported a significant increase in knee stiffness associated with increased K_{Leg} when increasing hopping frequency from 1.5 Hz to 3.0 Hz. This dominance of knee stiffness in determining lower extremity stiffness is also observed during running. Kuitunen et al. [94] reported that change in knee stiffness was the primary determinant of change in K_{Leg} during running. Subjects ran at 70%, 80%, 90% and 100% maximum velocity, and while ankle stiffness remained unchanged across velocities, knee stiffness increased with increasing velocity was associated with the increase in K_{Leg}.

The ankle joint has also been found to play an important role in K_{Leg} modulation. Farley et al. [21] had subjects complete bilateral hopping on surfaces with different stiffnesses, and found that as subjects moved from a stiff surface to a compliant surface, ankle stiffness increased 1.75x to correspond with the increase K_{Leg} while knee and hip stiffnesses did not significantly change. This finding is supported by a follow up study by Farley et al. [85] in which subjects completed bilateral hopping at preferred and maximal heights, and found that increasing ankle stiffness by 1.9x caused a 2x increase in $K_{Leg.}$ When knee stiffness was increased by 1.7x there was a minimal effect on K_{Leg}. Muller et al. [95] reported ankle stiffness dominance as the determinant of K_{Leg} during running during overground running over obstacles.

Other studies have found K_{Leg} is modulated through different interactions across multiple joints including significant changes in ankle and knee [75] and the knee and hip. [16] The literature suggests that K_{Leg} is predominantly influenced by knee and ankle joint stiffness working in concert to provide resistance to deformation and subsequent propulsion during ground contact. Given the literature is equivocal regarding whether the knee or ankle joint musculature is the most critical determinant to $K_{Leg,}$ the focus of this investigation will involve rigorous analysis of both knee extensor and ankle plantarflexor musculature.

K_{Leg} is also be influenced by age. Oliver et al. [77] found that men (19-30 y.o.) had greater stiffness compared to young boys (11-12 y.o.) hopping at higher frequencies (3 Hz) but not at lower frequencies (1.5 Hz). Similar results have been reported in 15 y.o. boys compared to 9 y.o. and 12 y.o boys. [96] However, stiffness appears to decline with continued aging, as healthy elderly adults display lesser K_{Leg} compared to young adults during hopping. Older adults also display greater EMG activation despite displaying lesser $K_{Leg.}$ [97] Contrastingly, this is not supported by Hobara et al. [35] who found that K_{Leg} did not differ between older and younger adults during hopping. Given these variations in K_{Leg} across age ranges, our investigation will focus on adults who have reached a similar level of musculoskeletal maturity between the ages of 18-40.

K_{Leg} is also dependent on the level of training, with more highly trained subjects displaying greater K_{Leg} during dynamic activities. Power training athletes have greater K_{Leg} during hopping compared to endurance athletes. [36] Additionally endurance trained athletes display greater K_{Leg} during hopping vs untrained subjects.[75] Greater K_{Leg} is also associated with improved performance; subjects who demonstrate greater vertical jump height and achieve greater running velocities display greater K_{Leg} during jumping and running. [73, 98, 99] This indicates that athletes may have an "optimal K_{Leg}" for performance that increases with performance demands [88]. To ensure similar training levels throughout our study, we will recruit only runners who have been running a minimum of 15 miles/week for at least 3 months. [100]

Several studies have examined the effects of fatigue on lower extremity stiffness. With repeated 40m sprints, K_{Vert} was significantly decreased, but K_{Leg} remained constant throughout 12 repetitions. [101] An analysis was repeated over 100m repetitions to fatigue in which decreases in K_{Vert}, step frequency, and contact time were noticed as the trials progressed. However, K_{Leg} again remained relatively constant despite a decrease in mean and maximum running velocity. [102] Evidence suggests that there may be both K_{Vert} and K_{Leg} decrease with longer runs to exhaustion. [79, 103, 104] However, this contrasts with Degache et al. [105] who found runners maintained K_{Leg} and K_{Vert} following a 5 hour trail run. However this may simplify the response to fatigue. There is also evidence that skilled runners do not alter motor patterns in response to fatigue. During 4 repetitions of 100 m to fatigue, skilled sprinters were able to maintain K_{Leg} and even increase it on the fourth repetition. [106] The skilled sprinters showed minimal change in stride frequency throughout the trials which is different from novice sprinters. They were also able to maintain consistent contact time intervals whereas novice sprinters demonstrated increased contact time and subsequently decreased K_{Leg}. In order to minimize fatigue, subjects in our investigation was instructed to maintain normal training volume during the time frame of the data collection and refrain from running on the collection day.

K_{Leg} is a relatively simple measurement to describe the complex interaction and modulation of the lower extremity during dynamic activities. As described by Latash et al. [22] the modulation of the lower extremity requires regulated neuromuscular control of all lower extremity to generate appropriate stiffness and subsequent propulsion however may not adequately be modulated to reduce injury risk.

Stiffness and injury risk

Both hopping and running entail an eccentric loading phase to control center of mass descent followed by a concentric propulsive phase to propel the center of mass forward or upward, respectively. This interaction is derived from Newton's second law ($F\Delta t=m\Delta v$) in which the impulse ($F\Delta t$) is equal to the force (F) applied multiplied by the change in time (Δt) and momentum ($m\Delta v$) is equal to the mass of the individual (m) multiplied by the change in velocity (Δv). In order to achieve faster running velocities, there is a greater change in velocity along with a shorter ground contact time. Since the individual's mass remains constant, there must be an increase in the applied force which increases the impulse.

This impulse during eccentric loading affects musculoskeletal tissues differently due to differences in their viscoelastic properties. The elastic modulus is the ratio of stress, the amount of external force per cross sectional area placed on the tissue, to strain, which represents the amount of relative deformation the tissue subsequently undergoes compared to original length of the tissue. This ratio is the material analog of stiffness after accounting for the size of the tissue. The elastic modulus also influences the strain the tissue can undergo prior to reaching its yield point which further stress causes tissue damage. For instance, compression of skeletal bone and articular cartilage demonstrate relatively high elastic modulus which permits a large amount of stress to be applied while demonstrating minimal strain. However, lengthening of skeletal muscle, tendon and ligaments display a relatively smaller elastic modulus, and consequently experience relatively large deformation relative to the stress placed upon them along with a smaller amount of stress the tissue can withstand prior to incurring damage (Figure 2.3).

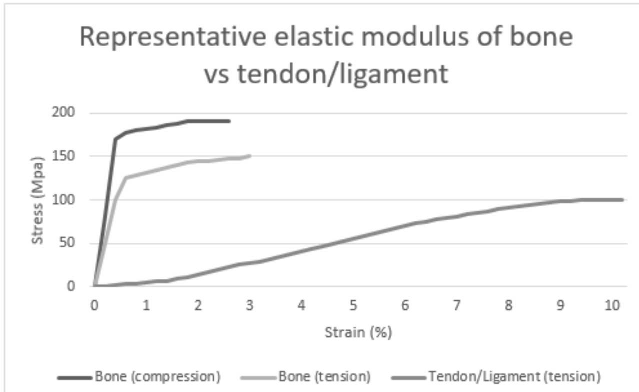


Figure 2.3: Representative elastic modulus for selected musculoskeletal tissues.

The amount of force applied to a tissue along with the rate it is applied is crucial for understanding injury risk. Greater loading rate has been linked with potential injury in both animal and human models. [9-11, 66, 107, 108] This impulsive, or high rate, loading has negative effects on both bone and cartilage health. Radin et al. [9] found biomarkers of both bone and cartilage breakdown after cyclical loading rabbit’s knees at 1.5x body weight at 40 cycles/min for 20-40 minutes/day over periods of 9, 15 and 20 days. Subchondral bone metabolic breakdown and stiffening was observed after 9 days of loading that was exacerbated at 20 days. After a rest period during which subchondral bone biomarkers returned to baseline levels, metabolic biomarkers of cartilage breakdown remained elevated. This study served as a groundwork for a later study by Radin et al. [10] which repeated 1.5x body weight cyclical loading of rabbit's knees over periods of 1, 2 and 3 days as well as 1, 3 and 6 weeks. A similar protocol of 40 minutes of loading of 60 cycles/min was completed daily with the load delivered across a 50 ms window. This greater, or impulsive, loading rate was deleterious for bone and cartilage health. Subchondral bone stiffening was observed, particularly at the bone/cartilage interface. The stiffened subchondral bone caused increased stress and eventual fracture of the cartilage matrix with cartilage fibrillation observed after 3 weeks of loading. The influence of loading rate compared to overall load was analyzed by Radin et al. [66] who found that severe cartilage fibrillation occurred more frequently at higher loading rates compared to lower loading rates. This damage occurred even when the joint exposed to lower loading rates demonstrated greater overall load. The importance of rate of loading was further examined by Ewers et al. [109] who found that the same magnitude of load applied over 5 ms compared to 50 ms generated more surface fissuring in rabbit retropatellar cartilage. This is supported in human subjects by Radin et al. [110] who found that subjects with mild knee pain demonstrated significantly greater loading rate and subsequent tissue damage during walking as compared to asymptomatic controls. This impulsive loading occurred within the first 25 ms following ground contact which has been shown to be potentially damaging to bone and cartilage including joint effusion, fibrillation of the articular surface, loss of cartilage, and proliferation of osteophytes. [65, 66]

Impulse is induced to the lower extremity during ground contact through the vertical ground reaction force (Figure 2.4). The vertical ground reaction force in certain populations can mirror impulsive loading in animal models. The peak loading magnitude and rate of loading are associated with injury risk. Zadpoor et al. [12] completed a meta-analysis and found that greater average and instantaneous loading rate plays a significant role in tibial and metatarsal stress fracture injury risk. Davis et al. [11] found retrospectively that runners who displayed greater vertical average loading rate were more likely to experience a tibial stress fracture. This relationship between loading rate and injury risk was also found prospectively by Bredeweg et al. [111] who found that novice male runners who went on to become injured during a running training program displayed greater loading rates and shorter ground contact times compared to the male runners that remained uninjured. However, this relationship was equivocal since there were no significant relationships between loading rate and contact time in the female runners. This demonstrates that the loading experienced by the lower extremity can contribute to injury risk; however, the response to this loading can be mitigated by altering lower extremity joint angle position and "tuning" the lower extremity muscles to dampen the peak vertical impact forces. [112] This demonstrates the need for further examination of the ability of the lower extremity musculature to alter the response to external loads during dynamic tasks.

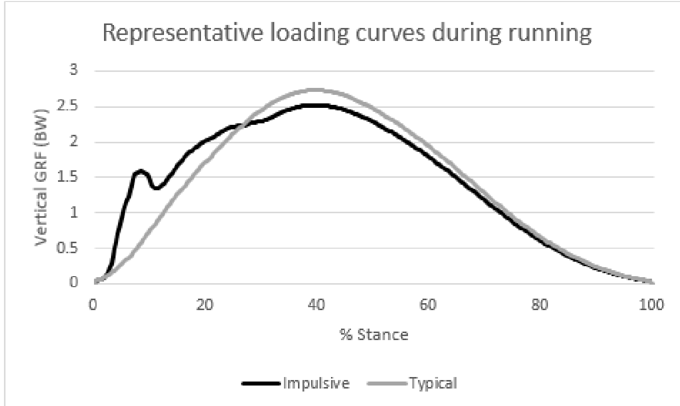


Figure 2.4: Representative vertical ground reaction forces between an impulsive runner and a non-impulsive runner.

K_{Leg} is influenced by both running kinematics and kinetics, and is associated with injury risk. Williams et al. [13, 113] found that runners with higher, stiffer arches exhibited a greater number of ankle and bony injuries whereas runners with lower, more compliant arches exhibited more knee and soft tissue injuries. Lorimer et al. [114] found that greater K_{Leg} is associated with higher risk for Achilles tendon injury in runners. Retrospectively, Maquirriain [18] found that subjects with Achilles tendinopathy demonstrated lesser K_{Leg} during single leg hopping on the affected limb compared to the unaffected limb. Pruyn et al. [115] also found that inter-limb differences in K_{Leg} were associated with a greater number soft tissue injuries in Australian rules football players. Prospectively, greater K_{Leg} also resulted in greater risk for hamstring injury in Australian rules football players during a season. [116] Both excessive and insufficient K_{Leg} appear to increase lower extremity injury risk to musculotendinous and bony tissues. As such, K_{Leg} may serve as a proxy for lower extremity injury risk in runners. This study will examine the neuromuscular contributions to K_{Leg}, thereby elucidating the specific mechanisms for altering K_{Leg} and potentially allow for more targeted preventative and rehabilitative exercise programs.

Neuromuscular contributions to lower extremity stiffness

Skeletal muscle activity and elastic energy return from tendons are the primary method for producing and transmitting force in the human body. The basic muscle model was first proposed by A.V. Hill and then improved upon by Huxley with the development of the cross-bridge theory. [117-119] These developments in modeling demonstrate that both muscle and tendon are viscoelastic tissues with a contractile element (i.e. muscle), a series elastic element (i.e. tendon) and an additional parallel elastic component associated with the total passive tissues embedded in the muscle fibers (i.e. aponeurosis). Nigg et al. [112] found that muscle stiffness through muscle tuning is capable of dampening the magnitude of vertical impact forces during running This muscle tuning can be influenced by active muscle activation as well as passive musculotendinous stiffness.

Muscle Activity

K_{Leg} is influenced by muscle activation measured by electromyography (EMG) prior to and during ground contact. EMG measures the relative neural drive to muscles and yields information regarding muscle on/off times and duration as well as relative muscle activation. EMG during running and hopping is typically analyzed 200 ms prior to ground contact (EMG_{PRE}) and up to 200 ms after ground contact (EMG_{POST}). [16, 36, 76, 95, 100] Generally, EMG_{POST} can be broken into eccentric and concentric phases, with the instant of the lowest vertical position of the center of mass serving as the transition between the eccentric loading phase and the concentric propulsive phase. The EMG_{POST} signal can be further broken down to analyze the acute neuromuscular response during ground contact. Specifically, the first 0-30 ms is often termed background activity (EMG_{BG}) prior to subject response. From 30-60 ms (EMG_{M1}), the EMG signal represents the voluntary supraspinal muscle activity as well as a short-latency stretch reflex. Lastly, from 60-90ms (EMG_{M2}) represents voluntary muscle activity and a long-latency stretch reflex. Responses occurring 100 ms after ground contact are characterized as a strictly voluntary motor response. [16, 36, 76, 84, 120-124]

However, EMG activation does not have a direct association with K_{Leg}. Inducing greater or lesser K_{Leg} (e.g. changing ground contact time, verbal performance instructions, etc.) can yield varied EMG response patterns of the lower extremity musculature based on the interaction between muscular demands and lower extremity touchdown angles. Several studies suggest that greater EMG activation results in greater K_{Leg}. Hobara et al. [76] found greater ankle plantarflexor

EMG activation during hopping with shorter contact times (i.e. greater K_{Leg}) compared to preferred contact times. Greater running velocity is also associated with greater EMG activity and K_{Leg}. Kuitunen et al. [94] reported increases in soleus, rectus femoris and vastus lateralis EMG amplitudes as subjects ran at progressively faster speeds (70%, 80%, 90%, and 100% maximal velocity). This is also reflected by Castro et al. [125] who demonstrated increases in rectus femoris, vastus lateralis, vastus medialis and lateral gastrocnemius activation at 100% vs. 60% maximal running velocity. Greater activation of the ankle plantarflexors and knee extensors as noted in these previous investigations is associated with greater $K_{Leg.}$ [126] Collectively, these data indicate that individuals with greater lower extremity stiffness likely display greater activity in lower extremity muscle groups during hopping and running.

However, these previous studies conflict with other research demonstrating that lower EMG activation is associated with greater K_{Leg} during dynamic tasks. Hobara et al. [36] reported that power trained athletes displayed lesser EMG activation of the ankle plantarflexors and knee extensors but greater K_{Leg} during bilateral hopping at 3 Hz compared to endurance trained athletes. Additionally, both power trained and endurance trained athletes demonstrated lesser EMG amplitudes and greater K_{Leg} when hopping at 3 Hz compared 1.5 Hz. This is also supported by Hobara et al. [16] who found that healthy subjects displayed lesser EMG_{M1} activation of the knee extensors when hopping at 3 Hz compared to 1.5 Hz, as well as lesser EMG_{M2} activation of the ankle plantarflexors and knee extensors when hopping at 3 Hz compared to 1.5 Hz. These differing results could be attributed to methodical differences. Previous research that found greater EMG activation associated with greater K_{Leg} had subjects complete a submaximal task first (e.g. hopping at preferred frequency, submaximal running velocity) then alter their movement pattern by instructing them to hop with minimal contact time [76] or increase running velocity maximally. [94, 125] Hobara et al. [76] found that even though subjects increased their K_{Leg} with reduced contact time during hopping, they displayed similar hopping frequencies (short contact time: 2.11 Hz, preferred contact time: 2.12 Hz). These data suggest further analysis of muscle activity of the ankle plantarflexors and knee extensors during hopping and running are warranted to understand neuromuscular contributors to K_{Leg}.

Training status also effects K_{Vert}, with endurance trained athletes displaying greater lower extremity EMG activity with lesser K_{Vert} and power trained athletes demonstrating lesser lower extremity EMG activity with greater K_{Vert} during hopping at 2.2 Hz.[36] Baur et al. [127] found that weekly training volume influenced lower extremity muscle activation patterns in male runners with lower training runners (< 30 km/week) displaying greater gastrocnemius activation during the weight acceptance phase of running compared to both intermediate (30 -45 km/week) and high (>45 km/week) training runners. However this group difference was reversed during push-off in which the lower training runners displayed significantly less gastrocnemius activation compared to intermediate and high training runners. Alterations in EMG activation are also seen in subjects who complete resistance training. Jenkins et al. [128] had male subjects complete a 6 week resistance training protocol utilizing high load (80% 1-rep max) vs low load (30% 1-rep max) for the leg extensors. Both groups displayed improvements in strength; however the high load group displayed greater leg extensor EMG activation during maximal contraction compared to the low load group. However, the high load group displayed lesser EMG activation during submaximal contractions. To ensure our subjects minimize the influence of training on EMG activation, subjects was required to run 15 miles/week for the previous 3 months and complete lower extremity musculature resistance training <3 times/week. Additionally we will track average weekly training distance to account for any neuromuscular changes associated with greater training volume.

These studies demonstrate the importance of understanding the interplay of EMG activation patterns in modulating both K_{Vert} and K_{Leg}. Training status of the subjects must also be accounted for, as endurance trained subjects appear to utilize relatively less EMG activation compared resistance trained subjects during submaximal tasks. Collectively, these data suggest that endurance trained subjects demonstrate greater EMG activation of the ankle plantarflexors and knee extensors during preactivation and weight acceptance compared to strength trained individuals during hopping. This greater activation is associated with lesser lower extremity compression, lesser center of mass displacement, and greater K_{Leg} during dynamic activities. Therefore the focus of this study will analyze similarly trained runners to ensure the effect of training is minimized and subsequently reduce a potential confounding factor of training.

Musculotendinous stiffness

Musculotendinous stiffness (MTS) describes the ability of the muscle-tendon unit to resist changes in its length. This measure takes into consideration contributions from passive structures, the active muscle fibers generating force and the tendon which serves as a power amplifier and energy redistributor to the skeletal system. [129]

MTS plays a role in muscle performance and subsequent overall dynamic performance. Greater MTS is associated with a greater rate of force development, [130] which may serve as a positive adaption from training along with increases in overall muscular strength to improve athletic performance. MTS is modifiable by training and increases with both isometric and plyometric training. [131-133] Spurrs et al. [131] had long distance runners complete a 6 week course of plyometric training in conjunction with their normal training load after which the MTS of the ankle plantarflexors increased bilaterally. Additionally, the runners demonstrated lesser metabolic demand during running at fixed velocities compared to pre-test and a 2.7% reduction in time during a 3 km time trial. The improvements in performance during the 3 km run were without a subsequent increase in metabolic demand. This demonstrates non-oxygen dependent muscular changes that may be associated with greater elastic energy return allowed for improvements in running velocity and endurance.

It is well known that training improves muscle size and strength,[134] however the tendon is also modifiable and demonstrates unique stiffness independent of training and sex. Females demonstrate lower tendon stiffness compared to males.[135] Tendon structures display greater stiffness after exposure to training driven by changes in elastic modulus. Bayliss et al. [136] found that the Achilles tendon of the take-off leg of a jumping athletes displayed greater stiffness compared to the flight leg. However, tendon stiffness can also decrease with training. Kubo et al. [137] found that long distance runners displayed lesser knee extensor and plantarflexor tendon stiffness during a road racing season compared to a track season. Tendon stiffness is also variable across muscle groups in trained populations. Sprinters display more compliant tendons for the knee extensors, but not for the plantarflexors compared to healthy controls. [138] Long distance runners display less extensible knee extensor tendons compared to controls, but not for the plantarflexors. [139] These tendon values are also associated with functional performance. Kubo et al. [140] found that runners with faster 5k times displayed stiffer tendons in the knee extensors but more compliant ankle plantarflexor tendons. Given this influence on running distance, we will collect weekly running distance, weekly training pace along with best performance in a race (e.g. 5k, 10k, or marathon) within the previous year.

MTS also potentially plays a role in injury prevention. Blackburn et al. [141-143] found that greater hamstring MTS may be associated lesser ACL loading. Greater hamstring MTS associated with greater knee flexion during peak anterior tibial shear force and peak internal knee-extension and –varus moments during landing. This demonstrates that greater MTS may be protective against excessive or abnormal joint loading, but may increase musculotendinous and overuse injury risk. Runners with a previous history of tibial stress fracture displayed greater ankle plantarflexor MTS compared to healthy runners. [144] Watsford et al. [116] prospectively found that Australian rules football players who demonstrated greater hamstring MTS were more likely to sustain a non-contact hamstring injury during the season. Given K_{Leg} is a combination of individual stiffnesses of joints, muscles, tendons and ligaments along with the neuromuscular control of contractile muscular tissue [22] there is likely an optimal K_{Leg} for performance for maximal vertical jump performance, [88] but optimal values for both K_{Leg} and MTS for minimizing injury risk have yet to be determined.

Greater EMG activation is associated with greater muscle fiber force. However absolute EMG activation is not synonymous with absolute muscle force. [145] The muscle tendon unit can be tuned to provide appropriate muscle force for the desired total muscle-tendon action [146]. The muscle-tendon unit can perform as a motor providing positive work, a brake generating negative work, or the muscle fibers can activate isometrically effectively serving as a strut while the tendon deforms and recoils generating elastic force. [147] This highlights the interplay between EMG activation and MTS and their respective ability to alter the muscle-tendon unit. The ankle plantarflexors and knee extensors are associated with eccentrically controlling center of mass descent during landing. Therefore greater MTS would require greater external tensile force to lengthen the muscle-tendon unit and subsequently induce joint angle changes (e.g. a stiffer knee extensor muscle group would provide greater resistance to knee flexion). Since K_{Leg} is influenced by the characteristics of all involved tissues, we hypothesize that greater stiffness values in the muscle-tendon unit via MTS was associated with greater K_{Leg} during hopping. We anticipate that MTS will mediate the influence of EMG activation on K_{Leg}.

As previously stated, K_{Leg} is predominantly modulated through the knee and ankle joints. [16, 21, 75, 85, 90, 92-95, 148] Man et al. [149] found that metatarsophalangeal stiffness was positively correlated with K_{Vert} and K_{Leg} during running indicating that restrictions in one of the lower extremity joints is associated with overall global stiffness of the lower extremity. This is supported by Nagahara et al. [73] who found that sprinters increased their K_{Vert} by increasing ankle stiffness via reduced ankle dorsiflexion displacement during running. By increasing the stiffness of the individual joints of the lower extremity, there is a reduced capacity of the lower extremity to compress during ground contact. This hypothesis is supported by previous research demonstrating that lesser passive range of motion at multiple joints of the lower extremity observed during a clinical exam was associated with greater K_{Leg} (Goodwin et al., In Review). Subjects who displayed lesser hip internal rotation, ankle dorsiflexion, and lesser first ray mobility exhibited greater K_{Leg} during overground running. This supports the previous research indicating that limited lower extremity motion via active contributions of muscular tissue or reduced capacity of the lower extremity to deform (e.g. landing in an extended posture, joint capsule and musculotendinous restrictions) would be associated with greater K_{Leg} during dynamic tasks.

Previous literature demonstrates that training affects MTS levels, and endurance trained runners likely display greater MTS. [131, 132, 139, 140] Additionally, elite level runners demonstrate lesser EMG activation of the lower extremity musculature during dynamic activities compared to national level runners. [150] Spurrs et al [131] reported increases in lower extremity MTS following a plyometric training protocol. This increase in MTS was associated with improvements in 3k running performance, but in the absence of an increase in metabolic demand. This supports the notion that there is improved energy production, via stored elastic return of passive tissues, which is not associated with active muscle contributions which would be represented in metabolic demand. Independently, greater MTS and greater EMG activation are associated with greater $K_{Leg.}$ However, EMG activation contributes to the active component of MTS and, therefore, MTS may mediate the amount of required muscle activation. Specifically, individuals with greater MTS likely require a lower level of muscle activation to achieve the same level of resistance to lengthening. This greater MTS with lesser EMG activation is due in part to the greater amount of passive stiffness components of the muscle tendon unit (e.g. stiffer tendon and aponeurosis). For example, a subject with greater MTS will require lesser EMG activation to obtain the overall resistance the tissue lengthening. Given this relationship, we anticipate that MTS and EMG will display an inverse relationship when compared to K_{Leg} during hopping and running.

Aim 1 of this project is to identify neuromuscular contributions to K_{Leg} during hopping and running. We hypothesize that greater ankle plantarflexor MTS combined with lesser ankle plantarflexor EMG activation was associated with greater K_{Leg}. Additionally we hypothesize that this inverse relationship will also be reflected with greater knee extensor EMG activation and lesser knee extensor MTS associated with greater K_{Leg}. Limited ankle and knee joint displacement during hopping and running induced by restrictions in both passive and active muscular tissue extensibility will reduce lower extremity compression. These variables was combined with the aforementioned passive clinical measurements from our previous investigation in a multiple regression model to establish a more robust estimate of K_{Leg}.

MTS was assessed by calculating linear stiffness of the ankle plantarflexors and knee extensors. Linear stiffness (k) was calculated utilizing the damped oscillatory method using the equation $k = 4\pi^2 mf^2$ where (m) represents total mass of the system, (f) represents the damped frequency of the oscillation calculated as the inverse of the first two successive oscillatory time peaks. [151, 152] The total mass of the system includes the total mass of the foot and shank [153] along with the applied load. This load is typically 30-50% of previously obtained maximal voluntary contraction force.

EMG activation normalized to standing muscle activity of the knee extensors and ankle plantarflexors was averaged 75 ms prior to and 75 ms after ground contact during hopping. EMG activation was average as one value for knee extensor activation and one value for ankle plantarflexors. This will provide a gross assessment of preparatory lower extremity muscle activity with anticipation to ground contact, short-stretch cycle activity during ground contact, and subsequent concentric muscle action to prepare for the next hopping aerial phase. [36, 76]

Clinical Application

K_{Leg} is a global measure assessing the lower extremity's ability to decelerate the center of mass and reaccelerate it to the next step during gait, and is linked with potential injury risk. Assessing K_{Leg} with motion capture is ideal in a dedicated laboratory setting with the capacity to analyze lower extremity kinematic and kinetic variables. However, given the increased cost for motion capture equipment and time for personnel training, this set up does not reflect the majority of current clinical capacities. Estimating K_{Leg} in the clinical setting requires at minimum a force plate to measure ground reaction forces. [87, 154] However, portable body-worn accelerometers have emerged as novel method to track activity level, and may permit an estimate of $K_{Leg.}$ [155, 156]

Accelerometers

Accelerometers can be utilized to estimate both position data and loading during dynamic activities. Uni-axial acceleration data is reflective of the kinetic loading response during ground contact by multiplying the acceleration curve by the subject's mass to obtain the vertical force curve. During vertical hopping the predominant acceleration experience by a subject is in the vertical direction. Tri-axial accelerometers could be used to obtain resultant acceleration vector to account for landing in a non-vertical position (e.g. running). K_{Leg} may also be calculated via analysis of the vertical force curve during ground contact. [157]The peak of this curve can serve as F_{max}. Additionally the acceleration curve can be double integrated to obtain an estimate of the center of mass vertical displacement during ground contact (Δy). Thus K_{Leg} was estimated via information obtained via body worn accelerometer.

Body-worn accelerometers have traditionally been used to track overall physical activity. [158] However, recent advancements in technology have allowed for reduced cost and improved function of portable body-worn accelerometers for estimating accelerations experienced by the lower extremity. [28, 159, 160] Lim et al. [161] found that accelerometers were sensitive enough to detect step and stride time during walking. Accelerometers are also a valid and reliable measurement of characteristics associated with more dynamic activities (e.g. jumping). Choukou et al. [159] found that body-worn accelerometers were a valid and reliable method to measure vertical jump height. Setuian et al. [162] found that body-worn accelerometers were capable of detecting different phases of vertical jump performance including take-off, peak vertical height and subsequent landing ground contact. These data suggest that body-worn accelerometers are capable of detecting temporal and performance parameters during running and hopping including take-off, flight phase and ground contact.

In addition to temporal parameters, accelerometers are capable of providing information regarding loading magnitude and rate during running and hopping. Elvin et al. [163] found that tibial accelerations were positively correlated with peak vertical ground reaction force during vertical jumping. Willy et al. [28] utilized waist mounted accelerometers to assess loading rate during running, and Zhang et al. [29] found that peak acceleration measured at the lateral malleolus was correlated with average and instantaneous loading rate during running. This demonstrates that peak vertical acceleration assessed via body-worn accelerometers on the leg are correlated with loading rate which has previously shown to be associated with increased injury risk. [11, 12, 164] However, our study will utilize waist mounted accelerometers to measure the acceleration experience by the center of mass. This demonstrates the potential for portable commercially available accelerometers (i.e. cell phones [165]) to be used clinically to estimate K_{Leg}, loading rate and subsequently injury risk.

Aim 2 of our study is to determine the relationship between K_{Leg} during single leg hopping determined via laboratory (motion capture) and clinical (wearable technology) measurement techniques. We hypothesize that greater K_{Leg} assessed via motion capture during hopping at self-selected frequency was associated with greater K_{Leg}, peak force, and peak loading rate derived from a waist mounted accelerometer. We will also have subjects complete single leg hopping under two additional conditions (1.5 Hz and 3 Hz) prescribed by metronome while sampling both laboratory kinematic and kinetic data as well as a waist mounted accelerometer data. [21, 34, 87] By utilizing three conditions, we will determine the relationship between K_{Leg} via motion capture and K_{Leg} via accelerometer data. By comparing the change in K_{Leg} between 1.5 Hz and 3 Hz via both motion capture and accelerometer we will determine if the accelerometer is sensitive to changes in K_{Leg} induced by increasing hopping frequency. [34]

<u>Comparing running vs hopping</u>

K_{Vert} is often measured during hopping and applied to populations that complete long distance running given its relative ease to calculate compared to K_{Leg}. [27, 36, 75] However, K_{Vert} during hopping may not be a valid representation of load experienced by the lower extremities during running. While both measures utilize peak vertical ground reaction force (F_{max}) and vertical displacement (Δy) of the center of mass, K_{Leg} also accounts for contact time, the arc swept by the leg during ground contact and forward velocity. K_{Leg} likely provides a more robust description of the lower extremity due to the fact that it accounts for the variability of running velocities and relative leg compression.

Studies that have examined both K_{Vert} and K_{Leg} during running have found they are not synonymous. K_{Leg} appears to be modified to maintain K_{Vert} when running at a fixed velocity on surfaces with variable stiffnesses. Ferris et al. [91] found that K_{Vert} during running on surfaces of differing stiffness was maintained at a constant value even though K_{Leg} increased to maintain running velocity. These authors conducted a similar study in which subjects transitioned from a soft surface with low stiffness (21.3 KN/m) to a hard surface with high stiffness (533 KN/m) while running, and found that subjects reduced K_{Leg} by 29% when transitioning to the hard surface, yet vertical displacement of the center of mass remained similar across the conditions [15]. Kerdok et al. [166] reported similar results when subjects completed treadmill running on a progressively stiffer surface (74-945 KN/m).

Both K_{Vert} and K_{Leg} change differently with changes in stride frequency and running velocity. Farley and Gonzalez [38] measured K_{Vert} and K_{Leg} during running at varied stride frequencies between -30% and +40% of preferred frequency, and found that both K_{Vert} and K_{Leg} increased when moving from the lowest frequency to the highest frequencies. However, there was a 3.5x increase in K_{Vert} (15.1 KN/m to 52.4 KN/m) compared to only a 2x increase in K_{Leg} (7.03 KN/m to 16.34 KN/m). Similar results were reported by Arampatzis et al. [90] for running at increasing velocities between 2.5 and 6.5 m/s during which K_{Vert} increased with speed but K_{Leg} remained relatively stable. These data suggest that K_{Vert} and K_{Leg} describe overall kinetic and kinematic behavior of the center of mass and leg respectively during running; however, K_{Leg} and K_{Vert} can be altered independently based on surface stiffness, stride frequency and running velocity. This relationship warrants further understanding to elucidate the role of both K_{Vert} and K_{Leg} during running.

The influence of fatigue on K_{Vert} and K_{Leg} is equivocal during running. Several studies have found that runners demonstrate reductions in either K_{Vert} or K_{Leg} when fatigued. Morin et al. [102] found that K_{Vert} decreased throughout repeated 100m sprints to fatigue while K_{Leg} remained unchanged despite reductions in running velocity. This reduction in K_{Vert} is supported by Dutto et al. [79] who found subjects significantly reduced K_{Vert} and K_{Leg} during a treadmill run to voluntary exhaustion between 31-90 minutes. Rabita et al. [104] found that K_{Leg} decreased during overground running to exhaustion while K_{Vert} remained constant in elite triathletes. These studies disagree with Hunter et al. [40] who found that experienced runners maintained K_{Vert} and K_{Leg} during the beginning and near the end of a 1 hour treadmill run to exhaustion despite a significant decrease in stride frequency. This maintenance of stiffness may be due to the experienced nature of the runners in this study and long-duration activity compared to the study by Morin et al. involving a primarily anaerobic task. Given the influence of training and experience on the maintenance of K_{Vert} and K_{Leg} during running, we will assess only runners during submaximal running and hopping to ensure we minimize the effect of both training and fatigue on K_{Vert} and K_{Leg}. Additionally, subjects was instructed to refrain from strenuous exercise 24 hours prior to the data collections.

K_{Vert} during running and hopping is calculated via the same equation ($F_{max}/\Delta y$). Therefore, we expect this variable to be highly associated during these tasks. Additionally, many of the same variables used to calculate K_{Leg} are also used to calculate K_{Vert} during running (i.e. F_{max} serves as the numerator in both equations, Δy is utilized as part of the equation to determine ΔL), we expect greater K_{Vert} during running to be associated with greater K_{Leg} during running.[38, 167, 168] However, we do not expect K_{Leg} during single limb hopping, K_{Vert} during running, or K_{Leg} during running to be synonymous. Additionally, K_{Vert} during double leg hopping may not be reflective of single leg stiffness characteristics during running. Brauner et al.[169] found that single limb hopping K_{Vert} was 24% lower than K_{Vert} during double limb hopping. However, single limb hopping explained 76% of the variance in double limb hopping. Single limb hopping more closely represents running gait given the single leg push-off, flight phase and subsequent landing on one foot. This demonstrates that landing behaviors during hopping are similar and K_{Leg} during hopping may be able to estimate both K_{Vert} and K_{Leg} during running. By comparing hopping K_{Leg} to K_{Vert} and K_{Leg} during running, we was able to more closely describe and understand the relationship between a proposed clinical screening task (i.e. hopping in place) and running biomechanics. This understanding between hopping and running may potentially allow for clinical identification of runners who display specific K_{Leg} patterns during running and may consequently be at an elevated injury risk.

Aim 3 of this study is to determine the relationships between K_{Leg} during hopping and K_{Vert} and K_{Leg} during running. K_{Leg} during hopping was assessed during single limb hopping. Single limb hopping was completed on the dominant limb determined by the subject's limb preference to kick a soccer ball for maximum distance. We hypothesize that K_{Leg} during single leg hopping was positively correlated with K_{Vert} during running. Additionally, we hypothesize that K_{Leg} during single leg hopping was positively correlated with K_{Leg} during running. This will provide us with a more robust understanding of the relationship between mechanical energy absorption strategies utilized during hopping and running.

Chapter 3: Methods

Experimental Design:

This cross-sectional study consisted of two data collection sessions separated by 1 week. Session 1 involved assessments of strength and musculotendinous stiffness of the knee extensors and ankle plantarflexors. Session 2 involved assessments of kinematics, kinetics, and EMG of the lower extremity during running and hopping. Given ease of accessibility to the location of Session 1, Session 1 preceded Session 2 to improve subject adherence to testing protocol and subject retention.

Subjects:

Seventy young (age 18-40 years) healthy subjects were recruited for participation in this study. Subjects were required to be free from lower extremity orthopedic injury for the 6 months prior to participation, have no history of lower extremity reconstructive surgery, not be currently pregnant, have no history of neurological disorder, and currently running a minimum of 15 miles/week over the previous 3 months, and were instructed to maintain their typical training volume between testing sessions and to refrain from running on testing days. Subjects filled out a brief survey on typical weekly running distance, typical training pace and injury history. All subjects provided written informed consent to participate.

Session 1

The first data collection took place in the Neuromuscular Research Laboratory. Initially, subject height and mass were recorded. Subjects completed a dynamic 5 minute shod running warm-up on a treadmill that was used to determine comfortable running speed for the running protocol in Session 2. A pace between 2.5-4 m/s [15, 38, 103, 170] was required, as this range represents a pace subjects could "comfortably" sustain for 20-30 minutes (10:43-6:42 minutes/mile).

Figure 3.1: Experimental set-up for ankle plantarflexion MVIC with ankle joint in neutral position

Subjects then completed knee extensor and ankle plantarflexion strength and musculotendinous stiffness assessments in a counterbalanced order.

Plantarflexor Strength and Musculotendinous Stiffness Assessments

Subjects identified their dominant limb by answering which leg would they use to kick a ball for maximum distance. The dominant limb was analyzed for all analyses in Session 1. Subjects were seated in a custom chair which places the hip, knee, and ankle in 90° of flexion. The metatarsal heads of the test limb was placed on a custom wood block anchored on top of a force plate (Bertec, Columbus, OH, USA). A second wooden block of equal height was placed under the heel to ensure the ankle is maintained in 90° of flexion. A custom loading device was placed on the superior aspect of the distal thigh (Figure 3.1) and secured to the floor to fix the limb in the testing position and prevent ankle joint motion. Subjects then were instructed to plantarflex maximally while the vertical ground reaction force is sampled at 1000 Hz from the force plate. Force plate data was lowpass filtered at 75 Hz with a 4th order Butterworth filter. [171] Subjects completed 1 practice trial to ensure proper adherence with testing protocol followed by 3 collection trials separated by 1 minute of rest to reduce the likelihood of fatigue. The largest peak vertical ground reaction force of the three trials were utilized as the peak plantarflexion maximal isometric voluntary contraction (MVIC) force and used to represent plantarflexion strength.

Figure 3.2: Experimental set-up for ankle plantarflexion MTS. Downward perturbation noted by arrow.

After obtaining plantarflexion strength, subjects completed the ankle plantarflexion musculotendinous stiffness (MTS_{Ankle}) assessment. The strap was removed from the loading device to permit superior and inferior movement and sagittal plane movement of the ankle (Figure 3.2). A load representing 30% of the subject's plantarflexion MVIC force was placed on the loading device in the form of weighted plates including the weight of the shank and foot segment (6.1% of body weight). [153, 172] Subjects donned headphones playing white noise and a blindfold to occlude visual and auditory feedback. The wooden block was removed from under the subject's heel, and the subject was required to maintain isometric ankle plantarflexion to support the applied load with the ankle in 90° of flexion. A downward manual perturbation was then be applied to the loading device within 10 seconds following the isometric contraction. The subject was instructed to attempt to maintain this position without volitionally resisting the perturbation. The perturbation will induce a series of

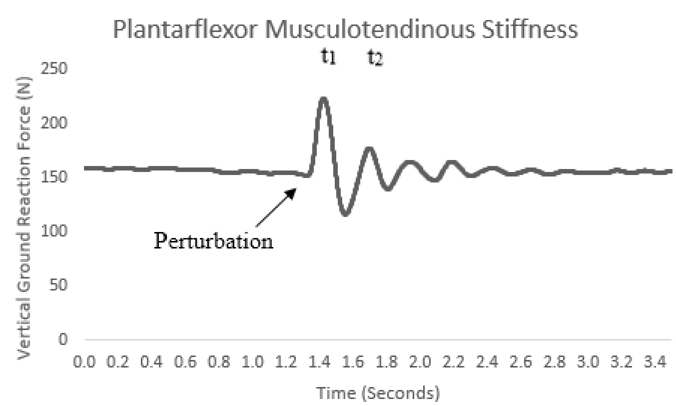


Figure 3.3: Vertical ground reaction force oscillations following downward manual perturbation. Damped oscillation (f) = 1/ (t_2-t_1)

plantarflexion/dorsiflexion oscillations that was characterized in the vertical ground reaction force (Figure 3.3). Ankle plantarflexor linear stiffness (k) was estimated utilizing the damped oscillatory method in which damped frequency of oscillation (f), and total mass of the system (m) are utilized in the equation $k = 4\pi^2 m f^2$. [144, 171, 173] The total mass of the system is equal to the summed masses of the foot and shank (6.1% body mass) and the applied load. [153, 172, 174] The damped frequency of oscillation was calculated from the inverse of time between the first two peaks of the vertical ground reaction force oscillations [$f = 1/(t_2 - t_1)$]. Subjects was allowed to rest one minute between each of 5 trials for the dominant limb. We normalized MTS_{Ankle} by dividing by body mass to reduce the influence of subject anthropometrics. During pilot testing, we collected data from 8 healthy adults who completed bilateral MTS_{Ankle} assessment described above with 3 trials collected on each limb for MVIC and MTS. Mean dominant limb (MTS_{Ankle} = 184.12 ± 52.3 N/m/kg) demonstrated high intra-session reliability (ICC (2,1) = 0.91) and precision (standard error of the mean = 18.52 N/m/kg).

Knee Extensor Strength and Musculotendinous Stiffness Assessment

Subjects was seated in the HUMAC (CSMI USA, MA, USA) dynamometer with the knee in 30° of flexion, and will perform maximal isometric knee extension against a bolster fixed at the distal shank. Torque data was sampled at 600 Hz and low pass filtered at 50 Hz. Subjects were instructed to extend maximally for 5 seconds to obtain knee extension MVIC (Figure 3.4). This was completed a minimum of three times on the dominant limb to ensure maximal strength is obtained in the form of peak knee extension torque.

Following the knee extension MVIC testing, subjects completed knee extension musculotendinous stiffness (MTS_{Knee}) testing. Subjects were supine on a table with the thigh supported on a wooden wedge in 30° of hip flexion (Figure 3.5). The knee and shank was not be supported to allow for free sagittal plane movement of the knee. A rigid Orthoplast splint was attached to the subject's foot and shank to maintain the ankle in a neutral position. A load equal to 30% MVIC was applied via ankle weights including the weight of the shank and foot. The subjects then activated the quadriceps isometrically to support the applied load with the knee in 30° of flexion. Subjects were again blindfolded and wearing headphones playing white noise to reduce anticipation to the perturbation. A downward manual perturbation was applied to the distal shank to induce knee flexion within 10 seconds after initial knee contraction. Subjects were instructed to attempt to maintain the initial position without volitionally resisting the perturbation.

Figure 3.4: Experimental set-up for knee extensor MVIC

The perturbation induced oscillations of the shank about the knee into flexion and extension that were captured in the tangential acceleration derived from an accelerometer (352C65, PCB Piezotronics, Depew, NY, USA) attached to the splint sampled at 1000 Hz.

Figure 3.5: Experimental set-up for knee extensor MTS. Downward perturbation noted by arrow.

(Figure 3.6). Accelerometer data was low passed filtered at 50 Hz with a 4th order zero-phase-lag Butterworth filter.

Active knee extensor linear stiffness (k) was calculated utilizing the damped oscillatory method in which damped frequency of oscillation (f), and total mass of the system (m) are utilized in the equation $k = 4\pi^2 m f^2$.[171, 173] The total mass of the system is equal to the summed masses of the foot and shank (6.1% body mass) and the applied load.[153, 172, 174] The damped frequency of oscillation is calculated as the inverse of the period between the first two oscillatory peaks. Again, MTS_{Knee} was normalized to body mass to account for subject anthropometrics. During pilot testing we collected data from 8 physically active adults who completed the MTS_{Knee} assessment described above with 3 trials for MVIC and MTS on the dominant limb. Mean dominant limb MTS_{Knee} = 21.13 ± 2.18 N/m/kg. MTS_{Knee} demonstrated high intra-session reliability (ICC (2,1) = 0.83) and precision (standard error of the mean = 0.77 N/m/kg).

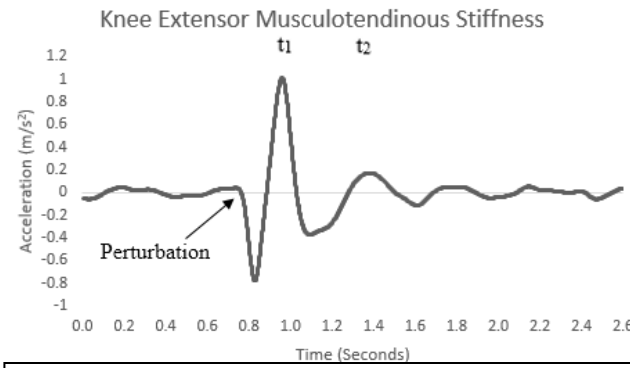


Figure 3.6: Foot and shank acceleration oscillations following downward manual perturbation. Damped oscillation (f) = 1/ $(t_2 - t_1)$

To screen for excessive muscular coactivation during the MTS assessments, mean EMG values for the ankle plantarflexors and knee extensors during the MTS_{Ankle} and MTS_{Knee} trials respectively were required to be ≤ 30% MVIC for a minimum of 1 second prior to delivery of the manual perturbation.[175] Trials with excessive antagonist activation (≥5% MVIC value), tibialis anterior and medial and lateral hamstring respectively for MTS_{Ankle} and MTS_{Knee} we excluded from our analysis.

Session 2

Subjects will report to the Applied Biomechanics Laboratory where they will first completed a 5 minute dynamic warm-up on cycle ergometer. Subjects were then be fit with retroreflective markers on the C7 spinous process and sacrum, and bilaterally on the anterior superior iliac spines, posterior superior iliac spines, greater trochanters, medial and lateral knee epicondyles, medial and lateral ankle malleoli, calcaneus, and 1st and 5th metatarsal heads (Figure 3.7). Additionally, plates with retroreflective markers were placed bilaterally on the thighs and shanks. This marker set was utilized to create a link segment model of the torso, pelvis, thighs, shanks and feet. Subjects were also fit with wireless electromyography (EMG) sensors (Trigno, Delsys, Inc, MA, USA) bilaterally on the vastus lateralis, vastus medialis, medial gastrocnemius, lateral gastrocnemius, and soleus. Proper EMG sensor placement and signal was verified by simple manual muscle tests. Subjects were also fit with a waist-mounted triaxial accelerometer (Trigno, Delsys, Inc, MA, USA) medial to the anterior superior iliac spine and secured with double sided tape (Figure 3.7).

Figure 3.7: Experimental kinematic marker and electromyography sensors marker set-up for running and hopping protocol. Waist mounted accelerometer noted in yellow circle.

Subjects then completed shod running and hopping protocols in a counterbalanced order. Initially subjects completed a standing calibration trial where they stood motionless for 5 seconds. The calibration trial allowed for determination of foot, shank, thigh and pelvis segments along with baseline muscle activity of the lower extremity which was used for kinematic model generation and EMG normalization. The hip joint centers were estimated utilizing the Bell method [176] following the digitization of the right and left anterior superior iliac spines. The knee and ankle joints center were estimated as the midpoint between the medial and lateral epicondyles and medial and lateral malleoli, respectively. Angle conventions were defined via the joint coordinate system. [177] Following the calibration trial, the medial malleolus and medial epicondyle markers were removed for ease of mobility during running. For the running protocol subjects ran at the self-selected, comfortable pace identified in Session 1 for 2 minutes on a dual belted instrumented treadmill (Bertec, Columbus, OH, USA).

Additionally, subjects completed a shod single-leg hopping protocol on the dominant limb at 3 counterbalanced frequencies controlled by metronome for 30 seconds per frequency: 1.5 Hz, self-selected frequency (anticipated near resonant frequency), and 3 Hz. [16, 178, 179] During both running and hopping ground reaction forces were sampled at 1000 Hz and whole body kinematic data was sampled at 100 Hz via a 14 camera Cortex motion capture system (Motion Analysis, Santa Rosa, CA, USA). EMG data was sampled at 1000 Hz with a signal bandwidth of 20-450 Hz. The Trigno accelerometer sensors have a sensitivity of ± 6g with an intrinsic sampling rate of 148 Hz. Kinematic, kinetic, EMG and acceleration data was collected by the same data collection system. Subjects were given a 5 minute rest between the hopping and running protocols and 2 minutes of rest between hopping frequencies to minimize the likelihood of fatigue.

Data Processing

The main outcome of this study is K_{Leg} during single-limb hopping at self-selected frequency [16, 35, 75] This frequency would most likely be reproduced in a clinical population instructed to hop at a comfortable rate given this is near resonant hopping frequency. [16, 75] Given healthy subjects often display asymptomatic interlimb asymmetries, [180] we only assessed the dominant limb.

All data were reduced and analyzed using a custom LabView program (National Instruments, Austin, TX, USA). K_{Leg} was calculated as the ratio of the peak vertical ground reaction force (F_{max}) to the downward displacement of the lower extremity (ΔL) during running ($F_{max}/ \Delta L$). ΔL is equal to the change in vertical leg length ($\Delta L = \Delta y + L_0 (1\text{-}\cos \Theta)$ where $\Theta = \sin^{-1}(ut_c/2L_0)$, Δy = maximum vertical displacement of center of mass; L_0 = standing leg length; Θ = half angle of the arc swept by the leg; u = horizontal velocity; and t_c = contact time). K_{Vert} was also assessed during running as the ratio of F_{max} to the downward displacement of the center of mass ($F_{max}/\Delta y$). Center of mass location was estimated from the sacrum marker and standing leg length was estimated from the height of the greater trochanter relative to the floor during the standing calibration. Maximal vertical displacement was measured as the maximum inferior displacement of the sacrum marker following initial ground contact. Contact time defined as the interval from initial ground contact (vertical ground reaction force >20N) to toe off (vertical ground reaction force <20 N). Horizontal running velocity was determined from the treadmill velocity.[27, 38, 71] Stiffness was assessed during single-leg hopping (1.5 Hz, self-selected, 3.0 Hz) as the ratio of F_{max} to the downward displacement of the center of mass ($F_{max}/\Delta y$).We will normalize K_{Leg} and K_{Vert} to reduce the influence of subject anthropometrics [181]. Specifically, F_{max}

was normalized to body weight (BW) and Δy was normalized to standing leg length (L_0) [$K_{Vert} = (Fmax/BW)/(\Delta y/L_0)$] [181].

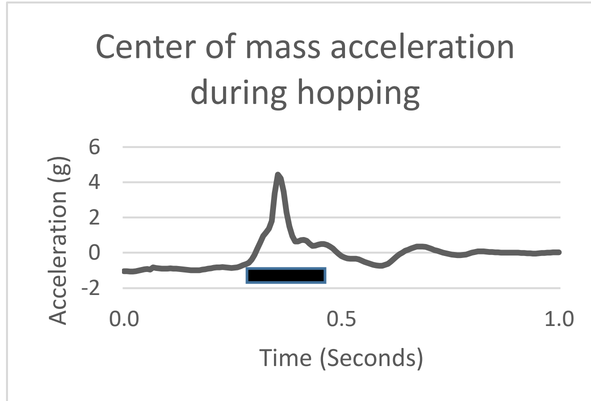


Figure 3.8: Center of mass acceleration during hopping. Stance phase denoted by solid black bar.

The vertical ground reaction force for running and hopping values determined via motion capture was calculated from the signal from the force plate. For accelerometer derived parameters, vertical ground reaction force was obtained by multiplying the center of mass acceleration by the subject's mass following initial ground contact during hopping (Figure 3.8). The acceleration curve was lowpass filtered at 50 Hz. The acceleration curve was then double integrated to obtain the estimate of center of mass position. The acceleration and position curves were time synced with the vertical ground reaction force from the force plate to establish ground contact. This allowed K_{Leg} during hopping to be estimated from data obtained from the waist-mounted accelerometer. Additionally, we utilized the vertical force curve determined via both accelerometer data and force plate data to assess both peak vertical force and loading rate. Loading rate was determined as the slope of the vertical ground reaction force curve from 0 to peak force following initial ground contact (vertical

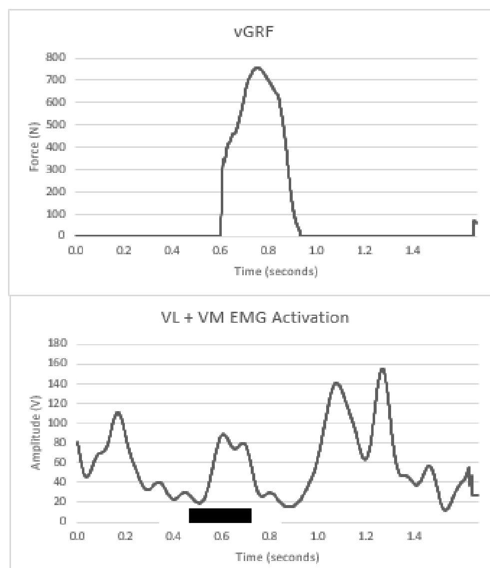


Figure 3.9: A) Vertical ground reaction force during stance phase of running B) Summed rectified EMG signal vastus lateralis and vastus medialis signal. Signal centered ± 75 ms about ground contact noted with black bar.

ground reaction force >20 N) [182]. The vertical force curve from the accelerometer was time synced with the force curve from the force plate to indicate ground contact.

Individual muscle raw EMG data was corrected for DC bias, rectified, bandpass (20-350 Hz) and notch (59.5-60.5 Hz) filtered with a 4th order Butterworth filter. Individual muscle EMG data was normalized to the mean EMG activity of the respective lower extremity musculature from the quiet standing calibration trial and expressed as a percentage. Mean knee extensor EMG (EMG_{EXT}) was averaged across the vastus lateralis and vastus medialis. Mean ankle plantarflexor EMG (EMG_{PF}) was averaged across the soleus, medial gastrocnemius and lateral gastrocnemius. EMG_{Ext} and EMG_{PF} amplitudes was averaged ±75 ms centered about ground contact (vertical ground reaction > 20N (Figure 3.9))

Prior research has indicated that passive mobility assessed during a clinical exam influences dynamic lower extremity stiffness. We will also complete a unilateral assessment of dominant limb passive lower extremity range of motion in the form of four tests that we previously demonstrated were associated with K_{Leg} during running (i.e. lesser range of motion was associated with greater K_{Leg}) [183]. For our clinical exam, we will obtain first ray mobility in a seated position on an exam table with the feet hanging freely (Figure 3.10A). The examiner clutched the base of the first metatarsal with one hand and the

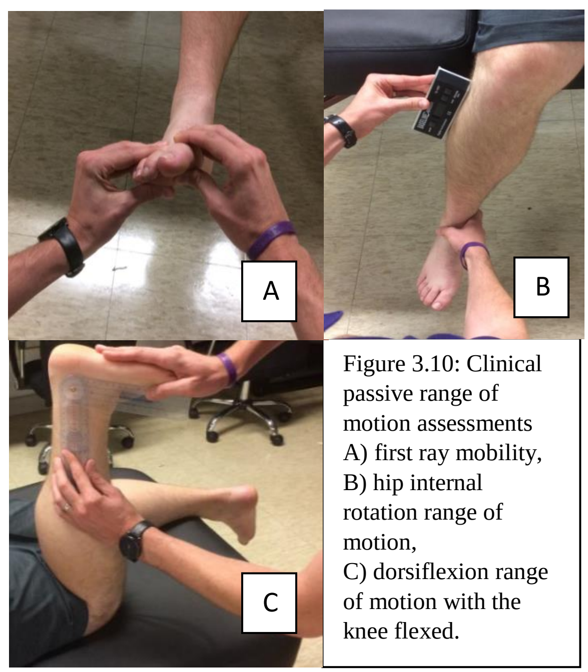


Figure 3.10: Clinical passive range of motion assessments A) first ray mobility, B) hip internal rotation range of motion, C) dorsiflexion range of motion with the knee flexed.

other four lateral metatarsal heads with the other hand. The examiner then moved the base of the first metatarsal superiorly and inferiorly while the lateral four lateral metatarsal heads are stabilized. First ray mobility was graded on an ordinal scale of hypomobile, normal mobile, or hypermobile. Hip inversion range of motion was assessed in the seated position on an exam table with the foot and shank allowed to move freely (Figure 3.10B). The examiner moved the foot laterally to induce hip rotation in the transverse plane and visually and physically track the subject's ipsilateral iliac crest. Once a firm joint end feel at the hip and superior movement of the iliac crest are noted, the examiner maintained the position of the shank. The angle of the shank relative to a superior/inferior axis was noted for the amount of hip internal rotation as measured by a digital inclinometer. The next clinical test was assessed with the subject prone on the exam table with the feet off the end of the table. The subject's knee was flexed 90° in the prone position so that the shank was directed towards the ceiling. While the knee is held in flexion, the subject's ankle was moved into maximum dorsiflexion. Ankle angle was measured via goniometer between the foot and long axis of the shank. By having the knee flexed, this will put the gastrocnemius in relatively slack compared to the soleus which is unaffected by knee flexion angle (Figure 3.10C).

Statistical Analysis & Hypothesis

All statistical analysis were analyzed in JMP Pro v13.0 statistical software (SAS Institute Inc., Cary, NC), and significance was established *a priori* as $\alpha = 0.05$

<u>Aim 1:</u> To examine the role of both passive and active contributions to K_{Leg} during single-limb hopping and running, we completed a series of multiple regression analyses. We have previously demonstrated that a parsimonious model of three passive clinical variables and body mass

explains 49% of the variance in lower extremity stiffness during running. We hypothesized that greater K_{Leg} with be associated with lesser passive mobility assessed via clinical exam.

We completed forward stepwise model selection with four "active" variables, $MTS_{Ankle,}$ $MTS_{Knee,}$ $EMG_{Knee,}$ and $EMG_{Ankle,}$ to explain the variance in normalized K_{Leg} during hopping and running along with 3 "passive" clinical variables. We included all predictor variables for K_{Leg} given the expected mediating effect of MTS on EMG during hopping. We expected that MTS will mediate the influence of EMG activation on K_{Leg} (e.g. greater MTS_{Ankle} was associated with lesser EMG_{Ankle}). We assessed variable inflation factors to account for collinearity. Our regression model candidates will include 7 total variables (3 "passive" variables, 4 "active" variables) and hopping frequency or running velocity for the self-selected conditions. We hypothesized that K_{Leg} would be associated with greater MTS, lesser EMG activation, and lesser passive mobility.

Aim 2: We completed a correlational analysis to evaluate associations between laboratory based measurement of K_{Leg} utilizing motion capture and K_{Leg} assessed via the waist-mounted accelerometer during hopping at self-selected frequency. We hypothesized that greater K_{Leg} assessed via motion capture was positively associated with K_{Leg} derived from a waist-mounted accelerometer. We also completed an additional analysis to determine if the waist-mounted accelerometer is sensitive to changes in lower extremity stiffness induced by changes in hopping frequency. We computed change scores between 1.5 and 3 Hz for both K_{Leg} assessed via motion capture and waist-mounted accelerometry and evaluate the relationship between these change scores. To ensure K_{Leg} assessed via motion capture changes with the change in frequency, we conducted a paired t-test between K_{Leg} for hopping at 1.5 Hz and 3.0 Hz.

A 2(method) x 3(hopping frequency) mixed-model repeated measures analysis of variance (ANOVA) was used to compare K_{Leg} between hopping frequencies using both the accelerometer data and motion capture data to determine if the waist-mounted accelerometer was sensitive to changes in lower extremity stiffness induced by changes in hopping frequency. Bonferroni-adjusted p-values were generated for post-hoc tests to evaluate significant ANOVA models. We will also compute change scores between 1.5 and 3 Hz for K_{Leg} assessed via motion capture and waist-mounted accelerometry, and evaluate the relationship between these change scores using Pearson product-moment correlation.

Pearson product-moment correlations were calculated to determine the relationship between loading rate and peak vertical ground reaction force determined via accelerometry and motion capture during hopping at the self-selected frequency. Additionally, we will utilize separate paired t-tests to compare each of loading rate and peak vertical force determined via accelerometry versus via motion capture.

We expected K_{Leg} at 3 Hz to be significantly higher compared to K_{Leg} at 1.5 Hz for both motion capture and accelerometry, and that changes in both K_{Leg} values were positively correlated. Also, we expected greater K_{Leg} assessed via accelerometer at self-selected frequency was associated with greater peak vertical force and greater loading rate. Additionally, we hypothesized that each of loading rate and peak force derived from accelerometry were positively correlated with each of loading rate and peak force derived from motion capture, respectively, during single leg hopping at self-selected frequency.

Aim 3: Lower extremity stiffness is most commonly assessed during bilateral hopping, but this may not reflect the stresses experienced during running. K_{Leg} and K_{Vert} are synonymous during vertical hopping; however, they are different during running given the leg compresses in a non-vertical direction during a running stride. Therefore, we completed a correlation analysis between laboratory based measurement of K_{Leg} during single-leg hopping at self-selected frequency via motion capture and K_{Leg} during running. We also evaluated the relationship between K_{Leg} during single-leg hopping at self-selected frequency via motion capture and K_{Vert} during running. These two analyses will provide a more complete assessment of leg stiffness during running and hopping. We hypothesized that K_{Leg} during hopping was positively correlated between K_{Vert} during running. Additionally, we hypothesize that K_{Leg} during hopping was positively correlated with K_{Leg} during running.

Power Analysis

A priori power analysis was completed using G*Power 3.1 to determine the sample size necessary to achieve power of 0.80 for $\alpha = 0.05$ for all analyses.

Aim 1: In our previous study evaluating the relationships between 3 clinical indicators of passive mobility and K_{Leg} during running in 92 subjects we achieved a significant correlation of 0.49 with an estimated Cohen's *d* effect size of 0.23. We included the 3 previous "passive" predictor variables along with our 4 "active" predictor variables from this study for a total of 7 predictor variables for our multiple linear regression equations to predict the variance in lower extremity stiffness during single leg hopping across all frequencies and self-selected running. A small effect size of 0.23 for a multiple linear regression will require a sample of 65 subjects to achieve 0.80 power.

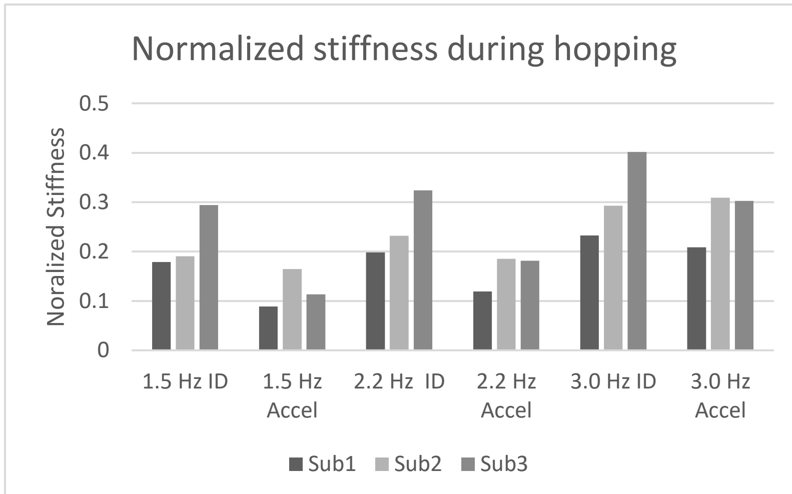


Figure 3.11: K_{Leg} assessed during hopping via motion capture (ID) and accelerometer data (Accel). Subjects completed hopping at 1.5, 2.2 and 3.0 Hz.

Aim 2: Our pilot data from 3 subjects indicated that normalized K_{Leg} was greater during hopping calculated via motion capture (0.25 ± 0.06) than via waist-mounted accelerometry (0.16 ± 0.03) (Figure 3.11). K_{Leg} calculated via motion capture and K_{Leg} calculated via accelerometer yielded a correlation of 0.53. We estimate a moderate expected correlation of 0.5 between K_{Vert} assessed via motion capture and K_{Vert} assessed via waist mounted accelerometer during single leg hopping at self-selected frequency. We estimate a required sample size of 29 subjects to achieve 0.80 power when testing bivariate correlational coefficient versus 0 (two-tailed).

Our additional analysis examined if the waist mounted accelerometer is sensitive to changes in lower extremity stiffness induced by hopping frequency. Our pilot data from 3 subjects indicated that the change in K_{Leg} during hopping assessed via waist-mounted accelerometry (0.15 ± 0.03) was greater than that calculated with motion capture (0.08± 0.02) when increasing frequency from 1.5 Hz to 3.0 Hz (Figure 3.12). The relationship between changes scores between hopping at 1.5 Hz and 3.0 Hz yielded a positive correlation of 0.82 indicating the accelerometer is sensitive to changes in K_{Leg} during hopping. To be conservative we estimated a moderate correlation of 0.7 for a two-tailed bivariate correlational coefficient which would require a sample size of 13 to achieve 080 power for a test versus 0 (two-tailed).

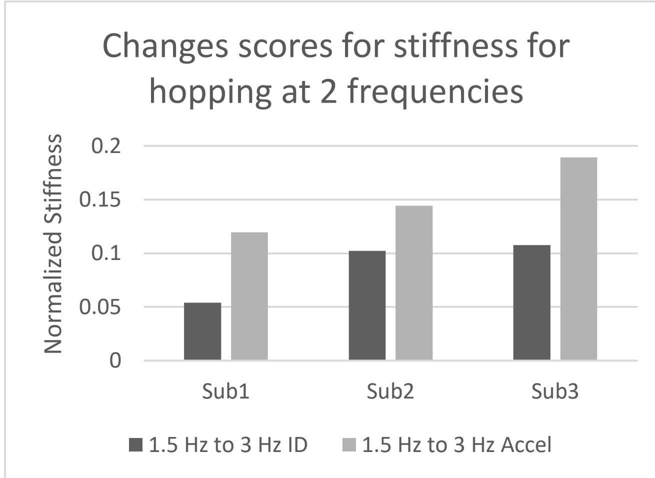


Figure 3.12: Vertical stiffness changes scores assessed during bilateral hopping via motion capture (ID) and accelerometer data (Accel) comparing hopping at 1.5 Hz and hopping at 3 Hz.

Aim 3: Our pilot data from 3 subjects indicate that normalized K_{Leg} during single leg hopping at 2.2 Hz (0.25 ± 0.06) was highly similar to normalized K_{Vert} during running (0.24 ± 0.10), and both of these

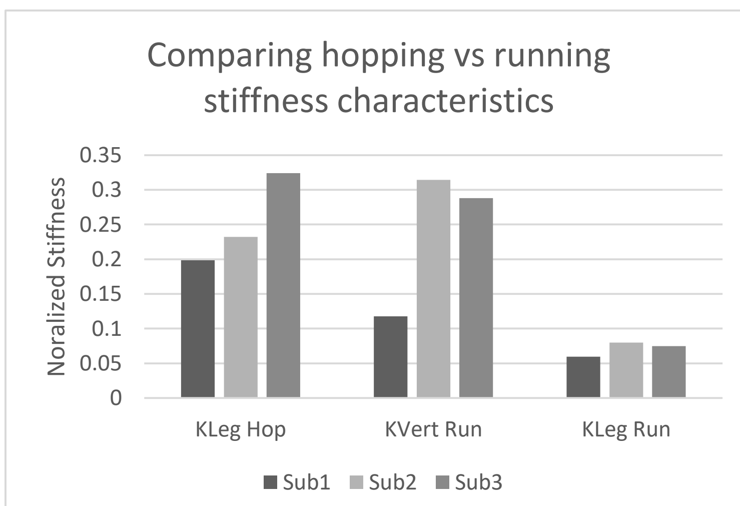


Figure 3.13: Stiffnesses for K_{Leg} during hopping, K_{Vert} during running and K_{Leg} during running.

values were substantially higher than normalized K_{Leg} during running (0.07± 0.01) (Figure 3.13). Our evaluation of the relationship between K_{Vert} during hopping and running yielded a correlation of 0.61. To be conservative, we estimated a moderate expected correlation of 0.5 between K_{Leg} during single hopping at self-selected frequency and K_{Vert} during running. We estimate a required sample size of 29 subjects utilizing a two-tailed bivariate correlation coefficient to achieve 0.80 power (two-tailed) for a test versus 0. Evaluation of the relationship between K_{Leg} during hopping and K_{Leg} during running yielded a correlation of 0.51. We estimate a moderate expected correlation of 0.5 between K_{Leg} during single limb hopping at self-selected frequency and K_{Leg} during running.

Given the results of these power analyses, we recruited 70 total subjects to ensure all of our aims are properly powered. This will account for potential subject attrition given the data collection occurs over two testing sessions.

Chapter 4: Summary Results

Aim 1: To identify neuromuscular contributions to K_{Leg} during hopping and running.

We utilized forward stepwise selection multiple regression to ascertain the neuromechanical variables with the greatest influence on lower extremity stiffness (K_{Leg}) during running and hopping (Table 4.1).

Condition	Variables	R^2	p
K_{Leg} Hop 1.5	None	*	*
K_{Leg} Hop Self	EMG_{Knee}, EMG_{Ankle}, Hop Frequency	0.41	<0.0001
K_{Leg} Hop 3.0	MTS_{Knee}, MTS_{Ankle}, Hip IR ROM, EMG_{Ankle}	0.24	0.001
K_{Leg} Run	Hip IR ROM, MTS_{Knee}, Run velocity	0.23	0.0005
K_{Vert} Run	Hip IR ROM, MTS_{Knee}, Run velocity	0.30	<0.0001

Table 4.1: Significant variables selected for model inclusion via forward stepwise selection.

K_{Leg} during hopping at a self-selected frequency demonstrated the strongest regression model ($R^2 = 0.41$) with greater K_{Leg} associated with less EMG activity of the knee extensors, greater EMG activity of the ankle plantarflexors, and greater hopping frequency. This supports previous research demonstrating that active muscle contributions are predominantly associated with modulating lower extremity kinematic response during landing.[23] Greater ankle plantarflexor EMG activity was also associated with greater K_{Leg} during hopping at 3.0 Hz in addition to greater ankle plantarflexor MTS, lesser knee extensor MTS, and greater Hip IR ROM. Greater K_{Leg} and K_{Vert} during running were associated with less Hip IR ROM, greater knee extensor MTS, and greater running velocity. These relationships indicate that our neuromechanical variables are significantly associated with K_{Leg} however only display a moderate relationship. These analyses are addressed in detail in Chapter 5: Manuscript 1.

Aim 2: To determine the relationship between K_{Leg} during hopping via laboratory (motion capture) and clinical (waist-mounted accelerometer) measurement techniques.

Our analysis revealed that the accelerometer significantly and substantially overestimated K_{Leg} at each frequency compared to motion capture. However, the accelerometer was sensitive to changes in K_{Leg} induced by changes in hopping frequency, as the accelerometer change score from 1.5 Hz to 3 Hz was significantly correlated with the motion capture change score ($r = 0.36$, $p = 0.001$).

ANOVA revealed a significant interaction with between analysis method and hopping frequency ($p<0.0001$) (Figure 4.1). Post-hoc analyses revealed for motion capture that K_{Leg} at 3.0 Hz hopping was significantly greater than K_{Leg} at self-selected hopping ($p<0.0001$) and K_{Leg} hopping at 1.5 Hz ($p<0.0001$). K_{Leg} at hopping at self-selected frequency was also significantly greater than K_{Leg} hopping at 1.5 Hz ($p<0.0001$) for motion capture as well.

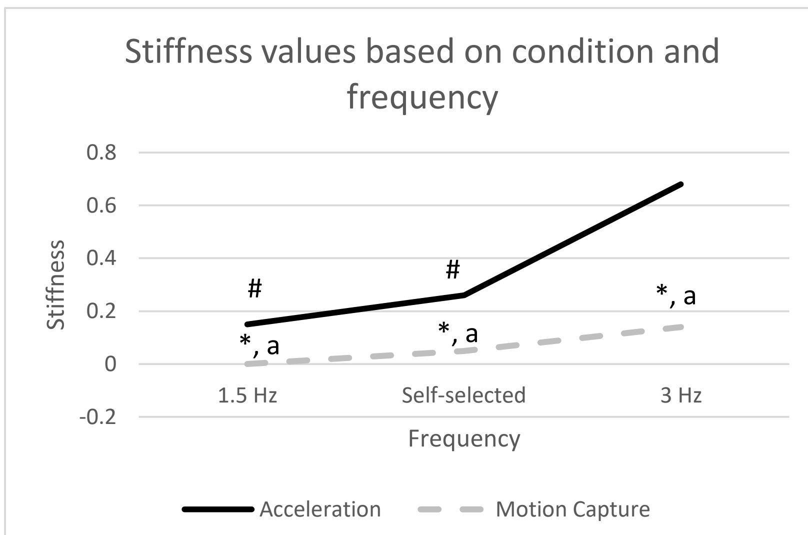
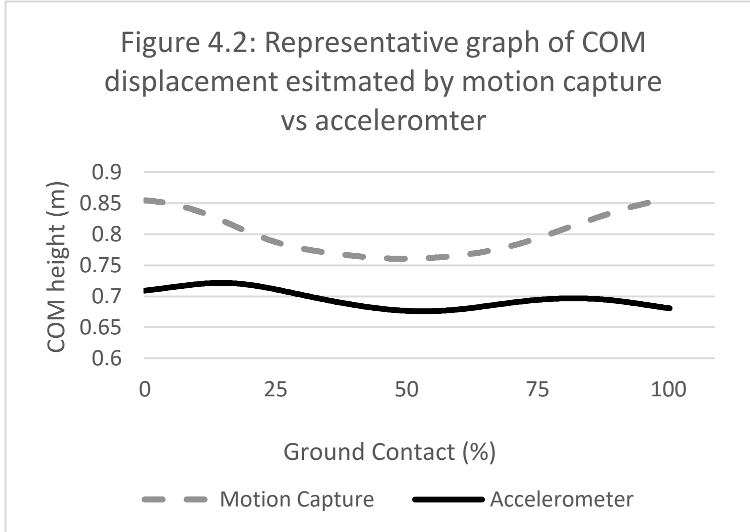


Figure 4.1: (*) = significant difference between all frequencies for Motion Capture; (#) = significant difference compared to 3Hz for Acceleration; (a) = significant difference between Motion Capture and Accelerometer.

Similarly for accelerometer derived values, K_{Leg} at 3.0 Hz hopping was significantly greater than K_{Leg} hopping at self-selected frequency ($p<0.0001$) and K_{Leg} at 3.0 Hz hopping was significantly greater than K_{Leg} hopping at 1.5 Hz ($p<0.0001$). However, K_{Leg} hopping at self-selected frequency was not significantly different than K_{Leg} hopping at 1.5 Hz ($p=0.24$).



Finally, K_{Leg} values for the accelerometer were significantly greater compared to K_{Leg} values assessed via motion capture for hopping at 1.5 Hz ($p<0.05$), hopping at self-selected frequency ($p<0.005$) and hopping at 3.0 Hz ($p<0.0001$). These relationships demonstrate that while our accelerometer was sensitive to changes in hopping frequency, they are likely influenced by error of the accelerometer underestimating COM displacement during the double integration of the acceleration curve (Figure 4.2). These analyses are addressed in detail in Chapter 6: Manuscript 2.

Aim 3: To determine the relationship between K_{Leg} during running and hopping.

We found that while K_{Leg} values derived from running and hopping were related within our subjects, the relationship was small and likely not clinically relevant. K_{Leg} during hopping at a self-selected frequency was significantly correlated with K_{Leg} at self-selected running ($r = 0.24$, $p = 0.04$) and K_{Vert} self-selected running ($r = 0.26$, $p = 0.02$). K_{Vert} self-selected running was significantly correlated with K_{Leg} self-selected running ($r = 0.88$, $p < 0.0001$). This demonstrates that hopping should not be used in the clinical sense as a proxy for running. These analyses are addressed in detail in Chapter 6: Manuscript 2.

Summary:

Our project revealed that the selected neuromechanical variables explained a significant amount of variance associated with K_{Leg} during running and hopping. However >59% of the variance was still left unexplained and further research is necessary to identify factors that explain the remaining variance. Additionally, we demonstrated that waist-mounted accelerometers are capable of detecting a change K_{Leg} caused by an increase in hopping frequency however our accelerometer overestimated K_{Leg} values by underestimating COM displacement. Finally, we demonstrated that subjects who demonstrate greater K_{Leg} during hopping display greater K_{Leg} during running; however, this association was small and likely not clinically relevant. Single-leg hopping should not be used as a clinical proxy for running biomechanical performance.

Chapter 5: Manuscript 1

Introduction

Running is a popular mode of exercise in the U.S. with over 48 million race participants in 2015.[1] Running has positive health benefits including reduced mortality and disability but it also incurs notable risk for lower extremity injury.[2, 3] Lower extremity injuries occur in up to 79% of runners[4] and subsequently lead to reduced activity, prolonged recovery, healthcare utilization, and time away from work.[5, 6] A conservative estimate of a 35% lower extremity injury rate in U.S. runners would incur an annual financial burden in excess of $1.6 billion.[49]

Runners must attenuate forces resulting from repetitive single leg landings via eccentric muscle action and passive resistance from skeletal tissues.[23] Lower extremity stiffness (K_{Leg}) quantifies resistance to deformation of the lower extremity following ground contact, and reflects the kinematic and kinetic contributions to force attenuation. Stiffness is typically assessed by modeling the lower extremity as a massless spring where K_{Leg} is calculated as the ratio of peak vertical ground reaction to the compression of the leg.[27] K_{Leg} is also commonly assessed during hopping in the literature to estimate the loading experienced during dynamic activities and may serve as a proxy for K_{Leg} experienced during running.[27]

Greater K_{Leg} is associated with a shorter ground contact time[16] and less joint excursion, resulting in greater loading rate and magnitude to passive tissues such as bone and cartilage.[13, 14, 17] Lesser K_{Leg} is associated with greater joint excursion and greater reliance on active muscle contributions to attenuate landing forces.[81, 184, 185] Greater loading rates have been linked to greater risk for bony injury,[9-12] while lesser loading rates have been linked to greater soft tissue injury risk.[13, 14] These data suggest that excessive or insufficient K_{Leg} may influence lower extremity injury risk. As such, K_{Leg} could be a clinical target for identifying individuals at heightened risk of lower extremity injury and a modifiable variable for reducing injury risk.

K_{Leg} is multifactorial in nature, reflecting contributions from a variety of active and passive characteristics of the musculoskeletal system.[21, 22, 82] We previously demonstrated that less passive foot, ankle, and hip mobility identified during a clinical exam explained 49% of the variance in K_{Leg} during running. However, factors explaining the remaining variance have yet to be elucidated. K_{Leg} is regulated in large part by lower extremity muscle activity and passive muscle characteristics. Musculotendinous stiffness (MTS) quantifies a muscle's resistance to lengthening and subsequent joint motion,[25] and likely plays a role in determining K_{Leg}. MTS potentially influences musculoskeletal injury risk by providing resistance to joint loading.[141-143] However, excessive MTS is associated with greater soft-tissue injury incidence, as previously injured subjects demonstrate greater MTS[144] and individuals with greater MTS prospectively experience higher soft-tissue injury rates.[116]

Greater MTS contributes to greater K_{Leg} during dynamic activities.[131, 139, 140] The level of activity (EMG) of the lower extremity musculature also contributes to K_{Leg}, with greater EMG associated with greater K_{Leg}.[76, 126] However, these variables have not, to our knowledge, been analyzed together to determine their combined influence on K_{Leg}. These findings suggest that, independently, greater MTS and greater EMG activation are associated with greater K_{Leg}.

However, increasing muscle activity increases MTS,[186, 187] suggesting that EMG activity influences MTS during running and hopping. This relationship allows active muscle tissue to effectively "tune" the muscle to the appropriate level of MTS to generate the appropriate amount of overall braking and propulsive forces during dynamic tasks.[112, 146]

The purpose of this study was to identify neuromechanical contributions to lower extremity stiffness during hopping and running. We hypothesized that greater K_{Leg} would be associated with greater ankle plantarflexor and knee extensor MTS. Greater K_{Leg} would also theoretically be associated with greater ankle plantarflexor and knee extensor EMG. However, we hypothesized that MTS would mediate the influence of EMG on K_{Leg} such that individuals with greater MTS would require lesser EMG to achieve a given level of K_{Leg}. This notion is supported by Spurrs et al.[131] who found that increasing MTS in runners corresponded with improved running performance (3k race time), but in the absence of a change in metabolic demand. As metabolic demand is an indirect assessment of energy expenditure, these data suggest that greater MTS affords the ability to attenuate force effectively with less active effort (i.e. EMG activity). This study will allow for improved understanding of clinical targets of MTS and EMG for rehabilitation to alter K_{Leg}.

Methods

Seventy healthy runners (42f/28m; 22.8 ± 4.71 y.o, range: 18-40 y.o.; 63.8 ± 11.8 kg; 1.7 ± 0.09 m) volunteered for participation in this study. Subjects were required to be between 18-40 y.o., run a minimum of 15 miles/week over the previous 3 months (mean = 27±11 miles/week for the previous 3 months), be free from lower extremity orthopedic injury over the previous 6 months, and have no history of chronic neurological disorder or lower extremity reconstructive surgery. Subjects read and signed a biomedical IRB approved informed consent prior to participation. Data collection occurred over two sessions separated by one week

Procedures:

Session 1:

Upon reporting to the laboratory subject height and mass were recorded and subjects completed a dynamic 5-minute shod running warm-up on a treadmill at a self-selected pace they could "comfortably sustain for 20-30 minutes" (mean running velocity = 3.1 ± 0.4 m/s). This pace was used to determine running speed for the running protocol in Session 2.

Passive lower extremity range of motion was assessed in the dominant leg via tests we previously demonstrated were associated with K_{Leg} during running.[183] Leg dominance was defined as leg preference for kicking a ball for maximum distance. A licensed orthopedic physical therapist completed all clinical exams (Figure 5.1). We obtained 1st ray mobility with the subject seated on an exam table and the feet hanging freely off the edge. The examiner moved the base of the first metatarsal superiorly and inferiorly while the four lateral metatarsal heads were stabilized. Mobility was graded on an ordinal scale of hypomobile, normal mobile, or hypermobile. Hip internal rotation (IR) range of motion (ROM) was also assessed in

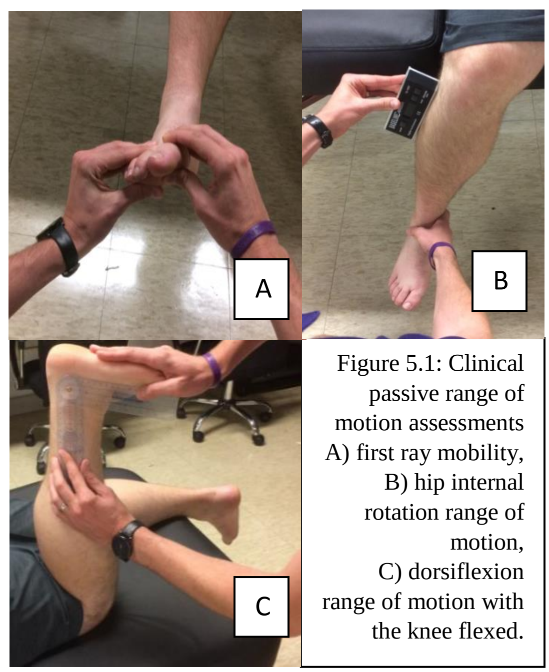


Figure 5.1: Clinical passive range of motion assessments A) first ray mobility, B) hip internal rotation range of motion, C) dorsiflexion range of motion with the knee flexed.

the seated position on an exam table with the foot and shank allowed to move freely. The examiner moved the foot laterally to induce hip rotation in the transverse plane and visually and physically tracked the subject's ipsilateral iliac crest. Once a firm joint end feel and superior movement of the iliac crest were noted, the examiner measured the angle of the shank relative to the vertical using a digital inclinometer. Finally, with the subject prone on the exam table and the feet off the end of the table the subject's knee was flexed 90° so that the shank was directed towards the ceiling. With the knee held in this position, the subject's ankle was moved into maximum dorsiflexion and the ankle angle was measured between the foot and long axis of the shank via goniometer (Ankle DF_{Flex} ROM).

Figure 5.2: Experimental set-up for ankle plantarflexion MVIC with ankle joint in neutral position

Following the clinical exam, subjects were fit with wireless EMG sensors (Trigno, Delsys, Inc, MA, USA) over the rectus femoris, vastus lateralis, vastus medialis, biceps femoris, semitendinosus, medial gastrocnemius, lateral gastrocnemius, soleus, and tibialis anterior of the dominant leg. Sensors were placed on the area of largest muscle bulk determined via manual muscle test. Prior to sensor placement, the skin was shaved if necessary, debrided, and cleaned with an alcohol prep pad to reduce skin impedance. EMG data were sampled at 2000 Hz via the Motion Monitor motion capture software (Innovative Sports Training, Inc, IL, USA).

Plantarflexor and Knee Extensor Strength Assessments

Following the passive range of motion assessments subjects completed knee extensor and ankle plantarflexion strength and MTS assessments in a counterbalanced order. For plantarflexion strength, subjects were seated in a custom chair which placed the hip, knee, and ankle in 90° of flexion (Figure 5.2). The metatarsal heads of the test leg were placed on a custom wood block anchored on top of a force plate (Bertec, Columbus, OH, USA). A second wooden block of equal height was placed under the heel to ensure the ankle was maintained in 90°. A custom loading device was placed on the superior aspect of the distal thigh and secured to the floor to fix the leg in the testing position and prevent ankle joint motion. Subjects were then instructed to plantarflex maximally while the vertical ground reaction force was sampled at 1000 Hz and lowpass filtered at 75 Hz (4th order Butterworth).[171] Subjects completed 1 practice trial to ensure proper adherence with testing protocol followed by 3 collection trials separated by 1 minute of rest to reduce the likelihood of fatigue. The largest peak vertical ground reaction force of the three trials was utilized as the peak plantarflexion maximal isometric voluntary contraction (MVIC) force. Additionally, subjects completed 1 MVIC trial of isometric ankle dorsiflexion to obtain MVIC EMG amplitude for the tibialis anterior.

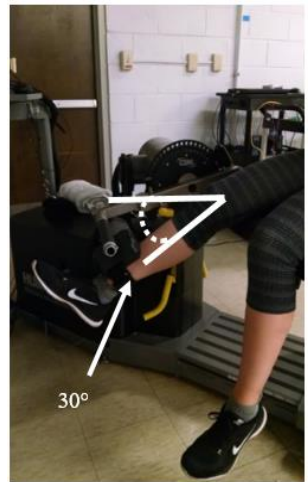


Figure 5.3: Experimental set-up for knee extensor MVIC

For knee extension strength, subjects were seated in the HUMAC dynamometer (CSMI USA, MA, USA) with the knee in 30° of flexion and performed maximal isometric knee extension against a bolster fixed at the distal shank (Figure 5.3). Torque data was sampled at 600 Hz and low pass filtered at 50 Hz (4th order Butterworth). Subjects were instructed to extend maximally for 5 seconds from which the peak torque was defined as the knee extension MVIC. Subjects were given 1 practice trial and subsequently completed 3 trials on the dominant leg.

Additionally, subjects completed 1 MVIC trial of knee flexion to obtain MVIC EMG amplitude for the hamstrings.

Plantarflexor and Knee Extensor Musculotendinous Assessments

After obtaining plantarflexion strength, subjects then completed the ankle plantarflexor musculotendinous stiffness (MTS_{Ankle}) assessment. The strap was removed from the loading device to permit sagittal plane ankle motion (Figure 5.4). A load representing 30% of the subject's plantarflexion MVIC force was placed on the loading device in the form of weighted plates. Subjects donned headphones playing white noise and a blindfold to occlude visual and auditory feedback. The wooden block was removed from under the subject's heel, and the subject was required to maintain isometric ankle plantarflexion to support the applied load with the ankle at 90°. A downward manual perturbation was then applied randomly to the loading device within 10 seconds following the isometric contraction. The subject was instructed to attempt to maintain this position without volitionally resisting the perturbation. The perturbation induced a series of plantarflexion/dorsiflexion oscillations that was characterized in the vertical ground reaction force and used to assess MTS.

Figure 5.4: Experimental set-up for ankle plantarflexion MTS. Downward perturbation noted by arrow.

Following the knee extension MVIC assessment, subjects completed knee extensor musculotendinous stiffness (MTS_{Knee}) testing. Subjects were supine on a table with the thigh supported on a wooden wedge in 30° of hip flexion (Figure 5.5).The knee and shank were not supported to allow for free sagittal plane movement of the knee. A rigid Orthoplast splint was attached to the subject’s foot and shank to maintain the ankle in a neutral position. A load equal to 30% MVIC was placed on the distal ankle. The subjects then activated the quadriceps isometrically to support the applied load with the knee in 30° of flexion with the shank parallel to the floor. Subjects again were blindfolded and wearing headphones playing white noise to reduce anticipation of the perturbation. A downward manual perturbation was applied to the distal shank to induce oscillatory knee flexion/extension within 10 seconds after initial knee contraction. Subjects were instructed to attempt to maintain the initial position without volitionally resisting the perturbation. The perturbation induced oscillations of the shank about the knee into flexion and extension that was captured in the tangential acceleration derived from an accelerometer (352C65, PCB Piezotronics, Depew, NY, USA) attached to the splint sampled at 1000 Hz and low passed filtered at 50 Hz (4^{th} order Butterworth).

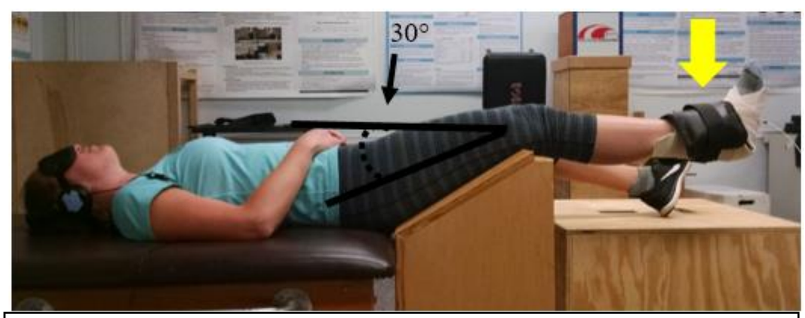


Figure 5.5: Experimental set-up for knee extensor MTS. Downward perturbation noted by arrow.

Linear stiffness (k) of the ankle plantarflexors (MTS_{Ankle}) and knee extensors (MTS_{Knee}) was estimated utilizing the damped oscillatory method in which the damped frequency of oscillation (f) and total mass of the system (m) are utilized in the equation $k = 4\pi^2 mf^2$.[144, 171, 173] The total mass of the system is equal to the summed masses of the foot and shank (6.1% body mass) and the applied load.[153, 172, 174] The damped frequency of oscillation was calculated from the inverse of time between the first two peaks of the associated oscillatory signals [$f = 1/(t_2-t_1)$] (Figure 5.6). To screen for excessive muscular coactivation during the MTS assessments, mean EMG values for the ankle plantarflexors and knee extensors during the MTS_{Ankle} and MTS_{Knee} trials respectively were required to be ≤ 30% MVIC for a minimum of 1 second prior to delivery of the manual perturbation.[175] Trials with excessive antagonist activation (≥10% mean MVIC EMG values), tibialis anterior and medial and lateral hamstring respectively for MTS_{Ankle} and MTS_{Knee} were excluded from our analysis. We normalized MTS_{Knee} and MTS_{Ankle} to body mass to reduce the influence of subject anthropometrics.

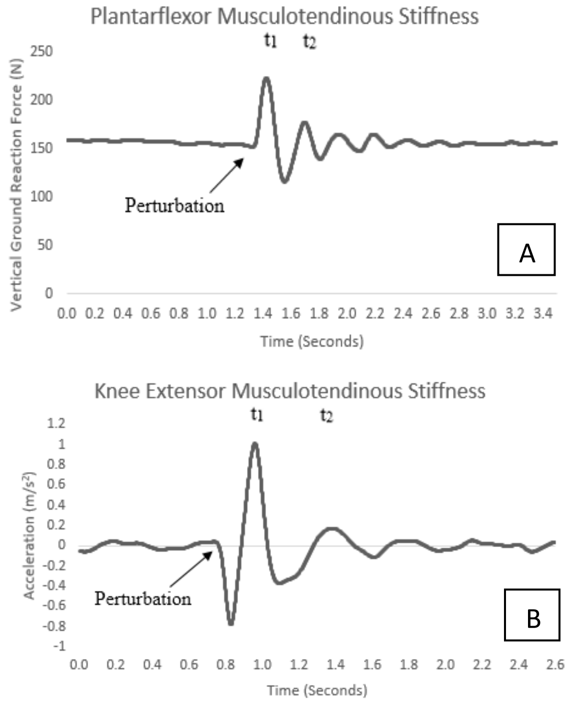


Figure 5.6: Representative foot and shank oscillations following downward manual perturbation for the ankle plantarflexors (A) and knee extensors (B).

Session 2:

Subjects were required to wait 1 week between testing sessions (8.3 ± 2.2 days between sessions). Upon arrival to the lab subjects completed a 5-minute dynamic warm-up on a cycle ergometer. Subjects were then fit with retroreflective makers placed on the C7 spinous process, sacrum, and bilaterally on the anterior superior iliac spines, posterior superior iliac spines, greater trochanters, medial and lateral femoral epicondyles, medial and lateral malleoli, calcaneus, and 1st and 5th metatarsal heads. Additionally, plates with retroreflective marker clusters were placed bilaterally on the thighs and shanks. This marker set was utilized to create a link segment model of the torso, pelvis, thighs, shanks and feet. Subjects were also fit with wireless EMG sensors (Trigno, Delsys, Inc, MA, USA) bilaterally on the vastus lateralis, vastus medialis, medial gastrocnemius, lateral gastrocnemius, and soleus (Figure 5.7).

Figure 5.7: Experimental kinematic marker and electromyography sensors marker set-up for running and hopping protocol. Waist mounted accelerometer noted in yellow circle.

A standing calibration trial was recorded to establish joint centers and segment coordinate systems, and following the calibration trial, the medial malleolus and medial epicondyle markers were removed for ease of mobility. Subjects then completed counterbalanced running and hopping protocols. Three-dimensional kinematic and kinetic data were sampled via a 14-camera Cortex motion capture system (Motion Analysis, Santa Rosa, CA, USA) and a dual belt instrumented treadmill (Bertec, Columbus, OH, USA). For the running protocol, subjects completed a 2-minute run on the treadmill at their preferred comfortable running velocity determined in Session 1. For the hopping protocol, subjects completed 30s of single-leg hopping on the dominant leg at 3 counterbalanced frequencies: 1.5 Hz, self-selected, and 3.0 Hz.[16, 178, 179] During the 1.5 Hz and 3.0 Hz hopping conditions, subjects were instructed to match their hopping frequency to a metronome that provided auditory and visual cues. During both running and hopping ground reaction forces were sampled at 1000 Hz and whole body kinematic data was sampled at 100 Hz. EMG sensors were sampled at 1000 Hz. Subjects were given a 5-minute rest between the hopping and running protocols and 2 minutes of rest between hopping frequencies to minimize the likelihood of fatigue.

Data processing

All data were reduced using a custom LabVIEW program (National Instruments, Austin, TX, USA). EMG signals were corrected for DC bias, rectified, and bandpass (10-350 Hz) and notch (59.5-60.5 Hz) filtered (4th order Butterworth). Mean knee extensor EMG (EMG_{Knee}) was averaged across the vastus lateralis and vastus medialis, and mean ankle plantarflexor EMG (EMG_{Ankle}) was averaged across the soleus, medial gastrocnemius, and lateral gastrocnemius. EMG_{Knee} and EMG_{Ankle} amplitudes were averaged over the 150 ms interval centered about ground contact (vertical ground reaction force > 20N) and normalized as a percentage of the mean EMG activity over 3 seconds of during the quiet standing calibration trial.

Kinematic and kinetic data were lowpass filtered at 10 Hz and 75 Hz, respectively. K_{Leg} was calculated during running (K_{Leg} Run) as the ratio of the peak vertical ground reaction force (F_{max}) to the downward displacement of the lower extremity (ΔL). ΔL is equal to the change in vertical leg length ($\Delta L = \Delta y + L_0 (1\text{-}\cos \Theta)$ where $\Theta = \sin^{-1}(ut_c/2L_0)$, Δy = maximum vertical displacement of center of mass; L_0 = standing leg length; Θ = half angle of the arc swept by the leg; u = horizontal velocity; and t_c = contact time) (Figure 5.8). K_{Vert} was also assessed during running as the ratio of F_{max} to the downward displacement of the center of mass ($F_{max}/\Delta y$). Center of mass location was estimated from the sacrum marker and standing leg length was estimated from the height of the greater trochanter relative to the floor during the standing calibration. Maximal vertical displacement was measured as the maximum inferior displacement of the sacrum marker following initial ground contact. Contact time defined as the interval from initial ground contact (vertical ground reaction force >20N) to toe off (vertical ground reaction force <20 N). Horizontal running velocity was determined from the treadmill velocity.[27, 38, 71] Stiffness was assessed during single-leg hopping (K_{Leg} Hop 1.5, K_{Leg} Hop Self, K_{Leg} Hop 3.0) as the ratio of F_{max} to the downward displacement of the center of mass ($F_{max}/\Delta y$).

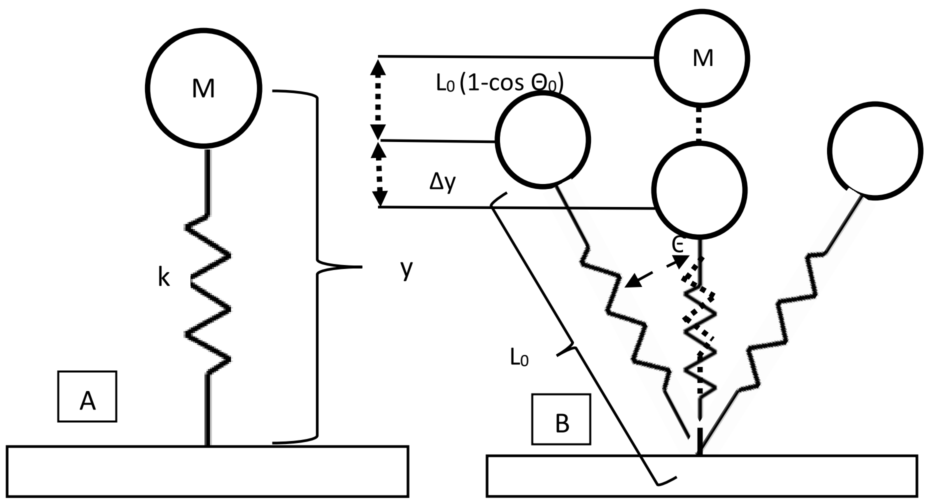


Figure 5.8: (A) Ideal spring mass model for K_{Vert} with the leg orientated vertically. (B) Ideal spring mass model for K_{Leg} with the leg orientated in non-vertical position.

We normalized all K_{Leg} and K_{Vert} values to reduce the influence of subject anthropometrics.[181] Specifically, F_{max} was normalized to body weight (BW) and Δy was normalized to standing leg length (L_0) [e.g. Normalized $K_{Vert} = (Fmax/BW)/(\Delta y/L_0)$]. [181]

Statistical Analysis

All analyses were performed using JMP Pro v13.0 statistical software (SAS Institute Inc., Cary, NC), and significance was established *a priori* as $\alpha = 0.05$. We completed stepwise forward selection multiple regression to evaluate the relationship between the linear combination of three clinical passive variables (Hip IR ROM, Ankle DF_{Flex} ROM, 1st ray mobility), two musculotendinous stiffness variables (MTS_{Knee}, MTS_{Ankle}), and two EMG variables (EMG_{Knee}, EMG_{Ankle}) and the criterion variable (K_{Leg}). Variable inflation factors were calculated to assess the multicollinearity of our variables. This analysis was separately repeated using K_{Leg} during running (K_{Leg} Run and K_{Vert} Run) and hopping (K_{Leg} Hop 1.5, K_{Leg} Hop Self, and K_{Leg} Hop 3.0) as the criterion variable. Given K_{Leg} is greatly influenced by hopping frequency and running velocity, these values were included in the regression models for self-selected hopping (2.05±0.23 Hz) and running (3.1±0.4 m/s). Frequency was not included in the analysis of hopping at prescribed frequencies given the limited variability among subjects (K_{Leg} Hop 1.5 = 1.51±0.03 Hz; K_{Leg} Hop 3.0 = 2.95±0.06 Hz) due to the use of the metronome.

Results

Normalized MTS_{Ankle} and normalized MTS_{Knee} were 82.7 ± 25.69 N/m/kg and 25.79 ± 4.92 N/m/kg, respectively. From our clinical exam, subjects demonstrated an average Hip IR ROM of 36 ± 7° and an average DF_{Flex} ROM of 20 ± 5°. For 1st ray mobility, 17 subjects were hypomobile, 50 subjects were normal mobile, and 3 subjects were hypermobile. K_{Leg} values obtained during Session 2 are detailed in Table 5.1.

Condition	Stiffness (KN/m)	Normalized Stiffness	EMG_{Knee} (%)	EMG_{Ankle} (%)
K_{Leg} Hop 1.5	20.8±10.72	0.02±0.02	2531.64±2500.60	1143.46±696.56
K_{Leg} Hop Self	30.88±18.66	0.05±0.03	1536.17±1020.73	1125.58±828.56
K_{Leg} Hop 3.0	86.68±41.08	0.14±0.06	1649.42±1400.28	1020.73±518.80
K_{Leg} Run	23.95±10.81	0.03±0.01	1243.31±1241.24	730.56±899.30
K_{Vert} Run	52.69±30.34	0.08±0.05	1243.31±1241.24	730.56±899.30

Table 5.1: K_{Leg} and EMG values from Session 2 categorized via hopping and running

Significant variables selected via forward selection are included in Table 2. The selected predictor variables did not explain a significant amount of variance in K_{Leg} during hopping at 1.5 Hz. Greater hopping frequency and greater EMG of the ankle plantarflexors and lesser EMG of the knee extensors explained a significant amount of variance in K_{Leg} during hopping at a self-selected frequency. K_{Leg} during hopping at 3.0 Hz was associated with greater ankle plantarflexor EMG, greater hip IR ROM, greater ankle plantarflexor MTS, and lesser knee extensor MTS. Finally, greater K_{Leg} and K_{Vert} during running were associated with lesser hip IR ROM, greater running velocity, and greater MTS_{Knee}. Final models for each multiple regression model are included in Tables 5.3-5.6.

Condition	Variables	R^2	p
K_{Leg} Hop 1.5	None	*	*
K_{Leg} Hop Self	EMG_{Knee}, $EMG_{Ankle,}$ Hop Frequency	0.41	<0.0001
K_{Leg} Hop 3.0	MTS_{Knee}, $MTS_{Ankle,}$ Hip IR ROM, EMG_{Ankle}	0.24	0.001
K_{Leg} Run	Hip IR ROM, $MTS_{Knee,}$ Run velocity	0.23	0.0005
K_{Vert} Run	Hip IR ROM, $MTS_{Knee,}$ Run velocity	0.30	<0.0001

Table 5.2: Significant variables selected for model inclusion via forward stepwise selection.

	Estimated Unstandardized Coefficients	**Standard Error**	**t-statistic**	**p-value**	**Variance Inflation Factors**
Intercept	-0.09	0.02	-3.48	<0.001*	
EMG_{Knee}	-9.55e-6	0.31	-2.81	0.006*	1.28
EMG_{Ankle}	9.31e-6	0.52	2.13	0.030*	1.39
Hop Frequency	0.07	0.04	5.38	<0.001*	1.09

TABLE 5.3. Multiple regression model results for the K_{Leg} Hop Self model containing the 3 predictor variables. * p<0.05

	Estimated Unstandardized Coefficients	**Standard Error**	**t-statistic**	**p-value**	**Variance Inflation Factors**
Intercept	0.03	0.05	0.78	0.43	
MTS_{Knee}	-0.2	0.06	-3.13	0.002*	1.35
MTS_{Ankle}	0.03	0.01	2.15	0.030*	1.33
EMG_{Ankle}	2.67e-6	0.00001	1.93	0.050	1.03
Hip IR ROM	0.002	0.001	2.64	0.010*	1.01

TABLE 5.4. Multiple regression model results for the K_{Leg} Hop 3.0 model containing the 4 predictor variables. * p<0.05

	Estimated Unstandardized Coefficients	**Standard Error**	**t-statistic**	**p-value**	**Variance Inflation Factors**
Intercept	-0.07	0.05	-0.13	0.890	
Hip IR ROM	-0.002	0.0008	-2.53	0.010*	1.17
MTS_{Knee}	0.11	0.04	2.59	0.010*	1.08

Run velocity	0.03	0.01	3.09	0.002*	1.10

TABLE 5.5. Multiple regression model results for the K_{Vert} Run model containing the 3 predictor variables. * p<0.05

	Estimated Unstandardized Coefficients	Standard Error	t-statistic	p-value	Variance Inflation Factors
Intercept	0.04	0.01	2.12	0.030*	
Hip IR ROM	-0.0009	0.0002	-3.39	0.001*	1.17
MTS_{Knee}	0.03	0.01	2.19	0.030*	1.08
Run velocity	0.006	0.004	1.35	0.180	1.10

TABLE 5.6. Multiple regression model results for the K_{Leg} Run model containing the 3 predictor variables. * p<0.05

Discussion

Our study found significant moderate associations between the selected neuromechanical variables and K_{Leg} during running and hopping. However, our hypothesis of an inverse relationship between MTS and EMG activity was not supported. K_{Leg} during hopping at a self-selected frequency demonstrated the strongest regression model ($R^2 = 0.41$) with greater K_{Leg} associated with less EMG activity of the knee extensors, greater EMG activity of the ankle plantarflexors, and greater hopping frequency. This agrees with previous research demonstrating that muscle activity is a predominant contributor to the lower extremity kinematic response during landing.[23] Greater ankle plantarflexor EMG activity was also associated with greater K_{Leg} during hopping at 3.0 Hz in addition to greater ankle plantarflexor MTS, lesser knee extensor

MTS, and greater Hip IR ROM. Greater K_{Leg} and K_{Vert} during running were associated with lesser Hip IR ROM, greater knee extensor MTS, and greater running velocity.

Our previous research found that Hip IR ROM, DF_{Flex} ROM, 1^{st} ray mobility, and body mass explained 48% of the variance in K_{Leg} during running. In the current study, we normalized K_{Leg} to body mass and standing leg length to minimize the influence of anthropometrics. To compare studies, we reanalyzed our data from the previous study while normalizing to body anthropometrics and only including Hip IR ROM, DF_{Flex} ROM, and 1^{st} ray mobility in our regression equation. These 3 variables explained 19% of the variance associated with K_{Leg} during running at 3.35 m/s ± 5%. This value is similar to the regression analysis for K_{Leg} during running in the current study ($R^2 = 0.23$). Additionally, running speed differed between the current (3.1 ± 0.4 m/s) and previous (3.35 ± 5% m/s), likely partially explaining the difference in explained variance across studies. Additionally, our cohort was younger (23 ± 5 y.o) compared to the previous study (50 ± 11 y.o, range: 25-81 y.o.). This discrepancy in age might explain the difference in explained variance in K_{Leg} between studies given aging causes reductions in muscle force, muscle mass, and overall muscle capacity.[188]

Our data indicate that K_{Leg} during single-leg hopping at self-selected frequencies and above is predominantly modulated by EMG activity and MTS of the ankle. This finding is supported by previous research indicating that stiffness during double-leg hopping at or above self-selected frequencies is predominately modulated by ankle joint stiffness.[75, 76, 82] Our study differed from the majority of previous studies in that we had subjects complete single-leg rather than double-leg hopping, which demonstrates K_{Leg} values 24% less than double-leg hopping.[169]

When increasing double-leg hopping frequency from 1.5 Hz to 3.0 Hz, Hobara et al.[16] found that subjects increased K_{Leg} by landing in a more extended position (i.e. less knee flexion). Given the limited amount of knee flexion displacement, single-leg hopping may limit the influence of knee extensor musculature which highlights the greater influence of the ankle musculature.

During hopping at 3.0 Hz and self-selected frequency, greater K_{Leg} was associated with lesser EMG activity and MTS of the knee extensors. This finding does not agree with previous literature reporting that K_{Leg} during double leg hopping is significantly influenced by knee joint stiffness.[16, 36, 75, 82, 92] Runners demonstrate muscular changes at the quadriceps and hamstrings that reduce flexibility which may contribute to greater passive contributions at the knee to increase K_{Leg}.[189, 190] Lesser potential joint mobility (i.e. stiffer musculature and less passive joint range of motion) may result in less availability for the leg to compress during ground contact and subsequently result in greater K_{Leg}.[141, 149, 191]

MTS increases with training, and all our subjects were habitually trained runners.[192-195] This training background may explain the influence of MTS during hopping at 3.0 Hz and running. Spurrs et al.[131] found that through plyometric training, distance runners increased lower extremity MTS which corresponded with an improved 3k time trial running performance. This improved performance occurred without increased metabolic demand. As metabolic demand is an indirect assessment of energy expenditure, these data suggest that greater MTS affords the ability to attenuate force effectively with less active effort (i.e. EMG activity).

Therefore, runners may condition their knee and ankle musculature to utilize more passive contributions (e.g. stiffer aponeurosis) to generate resistance to muscle lengthening and subsequently reduce demand on metabolic tissue (i.e. active muscle contributions). Also, the muscle-tendon unit is viscoelastic in nature such that MTS increases with greater loading rates. By increasing hopping frequency, there is a reduction ground contact time. This reduced time requires the subject to more rapidly generate force to meet the task demands. Thus, the muscle tendon unit is more rapidly loaded to generate the appropriate force to meet the higher frequency. This viscoelastic behavior may explain the greater influence of ankle plantarflexor MTS on K_{Leg} when shifting from hopping at a self-selected frequency to a higher frequency of 3.0 Hz.

Recent data has suggested that greater knee stiffness is associated with greater incidence of lower extremity injury.[196] Our data supports this with greater knee extensor musculotendinous stiffness serving as the primary neuromechanical contribution to K_{Leg} during self-selected running. Further studies should analyze the specific contributions to knee stiffness to overall K_{Leg}. Additionally, altering knee musculature stiffness through exercises may affect lower extremity injury risk. A quadriceps stretching protocol can improve knee extensor extensibility and conversely, a knee extensor plyometric protocol can increase knee extensor tissue stiffness. These mechanisms can serve to alter K_{Leg} and potentially reduce lower extremity injury occurrence.

Limitations

Our strongest regression equation explained 41% of the variance in K_{Leg} during hopping during self-selected frequency. Our other equations describing K_{Leg} and K_{Vert} during running and hopping explained less than 30% of the variance in K_{Leg} (R^2 = 0.24-0.30). These findings may be explained by the fact that our selected variables consisted of a parsimonious set of 3 passive clinical measurements, 2 laboratory measurements of MTS, and a gross EMG profile centered on ground contact ±75ms. Passive ROM outcomes may not reflect joint ROM during hopping and running. Additionally, ankle plantarflexor and knee extensor MTS were assessed with the musculature active to 30% MVIC and may not reflect MTS during running and hopping, as these tasks likely require differing activation levels. Further analysis of muscle activity prior to ground contact or activation following ground contact may yield more insight compared to our profile that combined muscle activation prior to and after ground contact. This would highlight the potential influence of the short stretch reflex cycle (i.e. 30-60 ms) or long latency reflex activity (i.e. 60-90ms) following ground contact. Alterations in preactivation and reflex activity have been associated with changes in K_{Leg} in double-leg hopping between 1.5 Hz and 3.0 Hz. [36] Additionally, we did not measure neuromechanical characteristics of any musculature surrounding the hip which has been to influence K_{Leg} during hopping.[16] Further clinical tests (i.e. hip abduction strength, single hop test for distance) may yield stronger associations with K_{Leg} during hopping and running.

Conclusion

Our study demonstrated that K_{Leg} during single-leg hopping and running is significantly associated with lesser passive hip mobility, MTS, and EMG activity of the knee extensors and ankle plantarflexors, as well as hopping frequency and running velocity, respectively. However, these variables are limited in this study in potentially predicting K_{Leg} given that >59% of the variance associated with K_{Leg} during dynamic activities is still unexplained. We also demonstrated that hopping and running display different strategies for modulating K_{Leg} (i.e. ankle- vs. knee-dominant). The remaining variance in K_{Leg} is likely attributable to additional anatomical characteristics (e.g. transverse ROM at the knee and ankle) and additional muscle contributions (e.g. hip extensor and abductor EMG). Given that runners utilize passive contributions of the muscle tendon unit to generate appropriate MTS during dynamic activities, future research should evaluate the influences of additional passive neuromuscular contributions (e.g. tendon stiffness) to K_{Leg} to determine the influence of non-metabolic tissues during dynamic activities. This will allow for potentially improved rehabilitation techniques and greater sport participation while minimizing overall metabolic demand to limit the effects of fatigue.

Chapter 6: Manuscript 2

Introduction

Running is an increasingly popular mode of exercise with over 48 million participants in the U.S. in 2015.[1] Running is associated with reduced mortality and disability but also incurs notable risk for lower extremity soft tissue and bony injury,[2, 3] which occurs in up to 79% of individuals.[4] Furthermore, these injuries lead to reduced activity, prolonged recovery, healthcare utilization, and time away from work.[5, 6] A conservative estimate of a 35% lower extremity injury rate in U.S. runners would incur an annual financial burden of over $1.6 billion.[49]

Runners have to mitigate forces resulting from repetitive single-leg landings through eccentric muscle action and passive resistance from skeletal tissues.[23] The inability to do so alters loading rates and magnitudes, which are associated with injury risk.[9-14, 23, 185] During running, the lower extremity can be modeled to have the unique stiffness characteristics of a spring.[70, 71] Lower extremity stiffness (K_{Leg}) describes the resistance to deformation of the lower extremity following ground contact, and can be calculated as the ratio of the peak vertical ground reaction force to the compression of the leg.[27] K_{Leg} may serve as a global surrogate for force attenuation by the lower extremity during running. Lesser K_{Leg} is associated with greater joint excursion and lesser loading rate[81, 184, 185] while greater K_{Leg} is associated with less joint excursion and greater loading magnitudes and rates.[13, 14] Greater loading rates have been linked to greater risk for bony injury,[9-12] while lesser loading rates have been linked to greater soft tissue injury risk.[13, 14] As such, inadequate or excessive K_{Leg} may be associated with greater running-related injury risk.

K_{Leg} is typically assessed during running in a laboratory setting utilizing motion capture. However, this is not clinically feasible due to the burdens of training, time, and cost.[27] K_{Leg} can be more easily assessed during hopping, which may be a more clinically-feasible approach.[18, 87, 154] However, despite similar biomechanical profiles during ground contact (i.e. force attenuation via flexion of the lower extremities), it is unclear if K_{Leg} derived during hopping is indicative of K_{Leg} during running. Additionally, vertical stiffness (K_{Vert}) is often measured during running given its relative ease to calculate compared to K_{Leg}.[27, 36, 75] While both measures are calculated using peak vertical ground reaction force (F_{max}) and vertical displacement (Δy) of the center of mass, K_{Leg} also accounts for contact time, the arc swept by the leg during ground contact, and forward velocity. K_{Vert} and K_{Leg} during hopping are synonymous given there is no forward velocity. K_{Leg} is a more robust indicator of lower extremity stiffness during running compared to K_{Vert}, thus it is important to determine if lower extremity stiffness estimated during hopping is indicative of this characteristic.

Portable body-worn accelerometers have emerged as a novel method for tracking activity level, and may permit an estimate of K_{Leg}.[155, 156] Accelerometers have been used to track overall physical activity,[158] jump take-off and peak vertical jump height,[28, 159-162] as well as loading magnitude and rate during running and hopping.[28, 29, 163] This demonstrates the potential for commercially available accelerometers (e.g. cell phones[165]) to assess characteristics associated with injury risk[11, 12, 164] and estimate K_{Leg} in the clinical setting.

The primary purpose of this study was to assess the relationship between K_{Leg} during hopping and running to determine whether leg stiffness during hopping could be used as a clinical proxy for leg stiffness during running. Our secondary purpose was to evaluate the feasibility of using waist-mounted accelerometers to approximate K_{Leg} during hopping compared to the gold standard of motion capture. These aims seek to advance clinical evaluation of K_{Leg} and improve understanding of observed clinical metrics potentially associated with running related injuries.

Methods

Seventy healthy runners (42f/28m; 23±5 y.o; 63.8±11.8 kg; 1.7±.09 m) volunteered for participation in this study. Subjects were required to be between 18-40 y.o., average running a minimum of 15 miles/week over the previous 3 months, be free from lower extremity orthopedic injury over the previous 6 months, and have no history of chronic neurological disorder or lower extremity surgery. Subjects averaged 27±11 miles/week over the 3 months prior to participation. Subjects read and signed an IRB approved informed consent prior to participation.

Procedures

These data were collected as part of a larger two-session study evaluating the neuromechanical contributions to $K_{Leg.}$ The data reported here were obtained during the 2nd session. Preferred running velocity was determined during the 1st session as subjects completed a 5-minute running warm-up on a treadmill at a velocity they could “comfortably sustain for 20-30 minutes” (average running velocity = 3.1±0.4 m/s).

Upon arrival to the laboratory for the 2nd session subjects completed a 5-minute dynamic warm-up on a cycle ergometer. Subjects were then fit with retroreflective markers placed on the C7 spinous process and sacrum, and bilaterally on the anterior superior iliac spines, posterior superior iliac spines, greater trochanters, medial and lateral femoral epicondyles, medial and lateral malleoli, calcaneus, and 1st and 5th metatarsal heads. Additionally, plates with retroreflective marker clusters were placed bilaterally on the thighs and shanks. A wireless triaxial accelerometer (Trigno, Delsys, Inc, MA, USA) was secured medial to the anterior superior iliac spine of the dominant leg (defined as leg preference for kicking a ball for maximum distance). A standing calibration trial was recorded to establish joint centers and segment coordinate systems, after which the medial malleolus and medial epicondyle markers were removed for ease of mobility.

Subjects then completed counterbalanced running and hopping protocols. Three-dimensional kinematic and kinetic data were sampled via a 14-camera motion capture system (Cortex, Motion Analysis, Santa Rosa, CA, USA) interfaced with a dual belt instrumented treadmill (Bertec, Columbus, OH, USA). For the running protocol, subjects completed a 2-minute run on the treadmill at their preferred speed identified in the 1st session. For the hopping protocol, subjects completed single-leg hopping on the dominant limb for 30 seconds at 3 counterbalanced frequencies: 1.5 Hz, self-selected, and 3 Hz.[16, 178, 179] During the 1.5 Hz and 3 Hz hopping trials, subjects were instructed to match their hopping frequency to a metronome that provided auditory and visual cues. During both running and hopping ground reaction forces were sampled at 1000 Hz and whole body kinematic data were sampled at 100 Hz. The Trigno accelerometer sensors have a sensitivity of ± 6g with an intrinsic sampling rate of 148 Hz. The accelerometer digital signal was reconstructed as an analog signal and resampled at 1000 Hz in order to time sync accelerometer derived variables with 3D kinematics and kinetics. Subjects were given a 5-minute rest between the hopping and running protocols and 2 minutes of rest between hopping frequencies to minimize the likelihood of fatigue.

Data processing

All data were reduced using a custom LabVIEW program (National Instruments, Austin, TX, USA). Kinematic and kinetic data were low-pass filtered at 10 Hz and 75 Hz, respectively, and the accelerometer data were low-pass filtered at 50 Hz. A standard motion capture procedure was utilized to calculate K_{Leg} during running as the ratio of the peak vertical ground reaction force (F_{max}) to the change in leg length (ΔL) (K_{Leg} Mocap Run = $F_{max}/\Delta L$). Change in leg length was calculated as $\Delta L = \Delta y + L_0 (1 - \cos \Theta)$ where $\Theta = \sin^{-1}(ut_c/2L_0)$, Δy = maximum vertical displacement of center of mass; L_0 = standing leg length; Θ = half angle of the arc swept by the leg; u = horizontal velocity; and t_c = contact time (Figure 6.1). Center of mass location was estimated from the sacrum marker and standing leg length was estimated from the height of the greater trochanter relative to the floor during the standing calibration. Maximal vertical

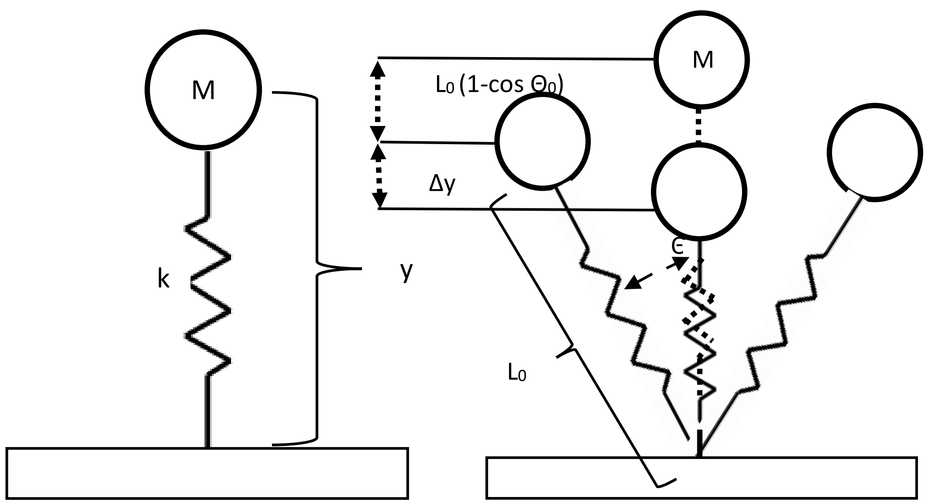


Figure 6.1: (A) Ideal spring mass model for K_{Vert} with the leg orientated vertically. (B) Ideal spring mass model for K_{Leg} with the leg orientated in non-vertical position.

displacement was measured as the maximum inferior displacement of the sacrum marker following initial ground contact. Contact time was defined via the force plate as the interval from initial ground contact (vertical ground reaction force >20N) to toe off (vertical ground reaction force <20 N). Horizontal running velocity was determined from the treadmill velocity.[27, 38, 71] Stiffness was also assessed during single-leg hopping (K_{Leg} Mocap Hop 1.5, K_{Leg} Mocap Hop Self, K_{Leg} Mocap Hop 3.0) and running (K_{Vert} Mocap Run) as the ratio of the peak vertical ground reaction force (F_{max}) to the downward displacement of the center of mass (Δy).

For accelerometer-derived parameters during hopping, vertical ground reaction force was estimated by multiplying acceleration by the subject's mass. The acceleration curve was then double integrated to estimate center of mass (COM) position. The acceleration and position curves were time synced with the vertical ground reaction force from the force plate to establish ground contact and stance phase duration. K_{Leg} was then estimated as the ratio of the peak vertical ground reaction force to center of mass displacement derived from the accelerometer data for hopping at 1.5 Hz (K_{Leg} Accel Hop 1.5), self-selected frequency (K_{Leg} Accel Hop Self), and 3.0 Hz (K_{Leg} Accel Hop 3.0). Additionally, we utilized the vertical force curve derived from both accelerometer data and force plate data to assess peak vertical force and loading rate. Loading rate was calculated as the slope of the vertical ground reaction force curve from initial ground contact to peak vertical ground reaction force.[182] The vertical force curve from the accelerometer was time synced with the force curve from the force plate to identify ground contact.

We normalized K_{Leg} and K_{Vert} to reduce the influence of subject anthropometrics.[181] Specifically, F_{max} was normalized to body weight (BW) and Δy was normalized to standing leg length (L_0) [i.e. Normalized $K_{Vert} = (Fmax/BW)/(\Delta y/L_0)$].[181] Additionally, peak forces and loading rates were normalized to body weight (BW).

Analysis

All statistical analyses were performed using JMP Pro v13.0 statistical software (SAS Institute Inc., Cary, NC), and significance was established *a priori* as $\alpha = 0.05$. Pearson product-moment correlations were calculated to evaluate relationships between 1) K_{Leg} during hopping at the self-selected frequency (K_{Leg} Mocap Hop Self) and running (K_{Leg} Mocap Run), 2) K_{Leg} during hopping at the self-selected frequency and K_{Vert} during running (K_{Vert} Mocap Run), and 3) K_{Leg} Mocap Run and K_{Vert} Mocap Run. We also assessed the relationship between leg stiffness during hopping at the self-selected frequency derived from the accelerometer (K_{Leg} Accel Hop Self) and from motion capture (K_{Leg} Mocap Hop Self) via Pearson product-moment correlations.

A 2(method) x 3(hopping frequency) mixed-model repeated measures analysis of variance (ANOVA) was used to compare K_{Leg} between hopping frequencies using both the accelerometer data and motion capture data to determine if the waist-mounted accelerometer was sensitive to changes in lower extremity stiffness induced by changes in hopping frequency. Bonferroni adjusted p-values were generated to evaluate significant post-hoc ANOVA models. We also computed change scores between 1.5 and 3 Hz for K_{Leg} assessed via motion capture and waist-mounted accelerometry, and evaluated the relationship between these change scores using Pearson product-moment correlation.

Pearson product-moment correlations were calculated to determine the relationship between loading rate and peak vertical ground reaction force determined via accelerometry and motion capture during hopping at the self-selected frequency. Additionally, we utilized separate paired t-tests to compare loading rate and peak vertical force determined via accelerometry and via motion capture, respectively.

Results

Means and standard deviations for all stiffness variables are reported in Table 6.1.

Condition	Normalized Stiffness
K_{Leg} Mocap Run	0.03±0.01
K_{Vert} Mocap Run	0.08±0.05
K_{Leg} Mocap Hop 1.5	0.02±0.02
K_{Leg} Mocap Hop Self	0.05±0.03
K_{Leg} Mocap Hop 3.0	0.14±0.06
K_{Leg} Accel Hop 1.5	0.15±0.26
K_{Leg} Accel Hop Self	0.26±0.33
K_{Leg} Accel Hop 3.0	0.68±0.82

Table 6.1: Normalized stiffness values based on hopping and running condition

K_{Leg} Mocap Hop Self was significantly correlated with K_{Leg} Mocap Run ($r = 0.24$, $p = 0.04$) and K_{Vert} Mocap Run ($r = 0.26$, $p = 0.02$). K_{Vert} Mocap Run was significantly correlated with K_{Leg} Mocap Run ($r = 0.88$, $p < 0.0001$).K_{Leg} Mocap Hop Self was not significantly correlated with K_{Leg} Accel Hop Self ($r = 0.09$, $p = 0.41$). However, the accelerometer was sensitive to changes in K_{Leg} induced by changes in hopping frequency, as the accelerometer change score from 1.5 Hz to 3 Hz was significantly correlated with the motion capture change score ($r = 0.36$, $p = 0.001$).

In general, the accelerometer significantly and substantially overestimated K_{Leg} at each frequency. ANOVA revealed a significant interaction with between analysis method and hopping frequency ($p < 0.0001$; Figure 6.2). Post-hoc analyses revealed that K_{Leg} Mocap Hop 3.0 was significantly greater than K_{Leg} Mocap Hop Self ($p < 0.0001$) and K_{Leg} Mocap Hop 1.5 ($p < 0.0001$). K_{Leg} Mocap Hop Self was also significantly greater than K_{Leg} Mocap Hop 1.5 ($p < 0.0001$). Similarly, K_{Leg} Accel Hop 3.0 was significantly greater than K_{Leg} Accel Hop Self ($p < 0.0001$) and K_{Leg} Accel Hop 1.5 ($p < 0.0001$). However, K_{Leg} Accel Hop Self was not significantly different than K_{Leg} Accel Hop 1.5 ($p = 0.24$). Finally, K_{Leg} values for the accelerometer was significantly greater compared to K_{Leg} values assessed via motion capture for hopping at 1.5 Hz ($p<0.05$), self-selected frequency ($p<0.005$) and 3.0 Hz ($p<0.0001$).

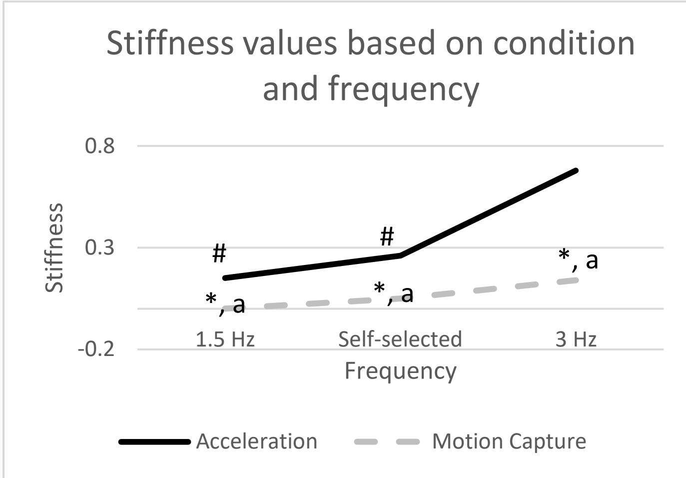


Figure 6.2: (*) = significant difference between all frequencies for Motion Capture; (#) = significant difference compared to 3Hz for Acceleration; (a) = significant difference between Motion Capture and Accelerometer.

Peak force during self-selected hopping determined via the force plate (3.52±1.47 BW) was significantly correlated with peak force determined via accelerometer (3.70±0.84 BW) ($r = 0.44$, $p < 0.0001$), and these magnitudes were not statistically different ($p = 0.19$). However, loading rate during self-selected hopping assessed via accelerometer (316.06±411.95 BW/s) was not significantly correlated with loading rate assessed via force plate (36.30±19.28 BW/s) ($r = -0.15$, $p = 0.18$), and the difference between these values was statistically significant ($p < 0.0001$).

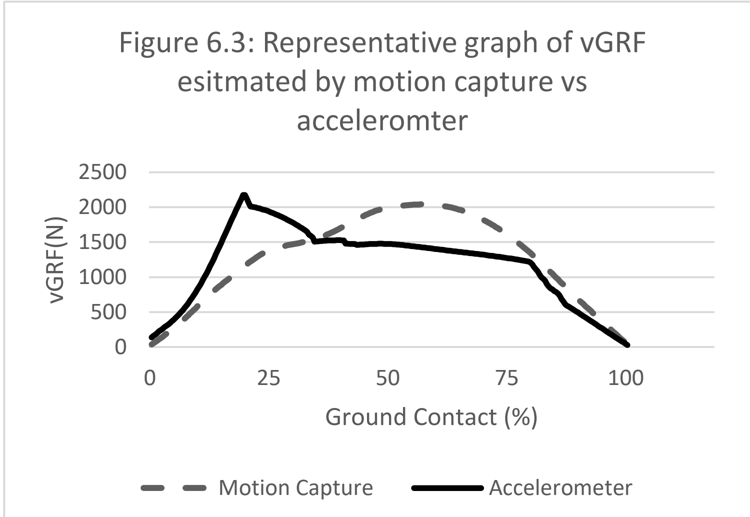
Discussion

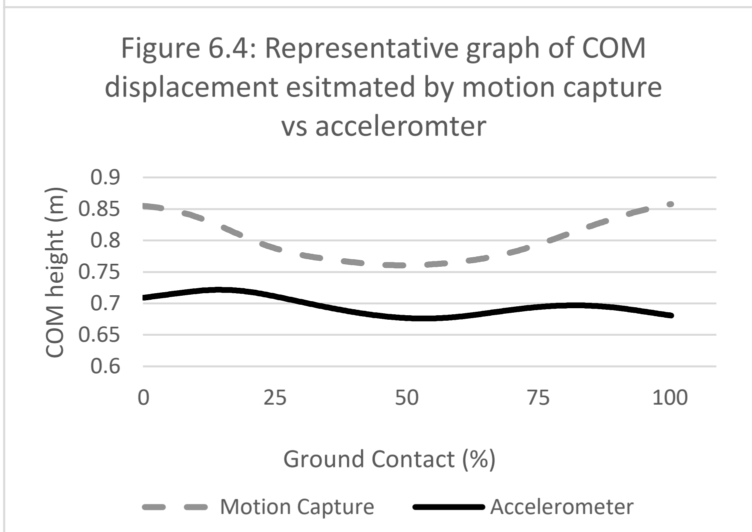
Our first hypothesis was supported in that K_{Leg} during hopping was significantly and positively correlated with K_{Leg} and K_{Vert} during running. However, the limited variance shared between K_{Leg} during hopping and running ($r^2 = 0.06 – 0.07$) suggests that leg stiffness during hopping cannot be used as a clinical proxy for leg stiffness during running. This is likely influenced by lower extremity position during landing, the effects of frequency, and forward velocity of running compared to the strictly inferior/superior direction during hopping. During single-leg hopping, all subjects landed in a plantarflexed position in order to meet the demands of the hopping task. Anecdotally, the majority of our subjects demonstrated a rear-foot or mid-foot strike pattern during running. This varied movement pattern suggests single-leg hopping is predominantly influence by ankle-dominant variables whereas running involves a gait pattern that is more reliant on knee flexion displacement to successfully complete the task. This variation likely contributes to the discrepancies in K_{Leg} during hopping and running.

K_{Leg} derived via motion capture increased as hopping frequency increased from 1.5 Hz to self-selected frequency (2.1 Hz) to 3 Hz (0.02±0.02 vs. 0.05±0.03 vs. 0.14±0.06). This pattern was replicated when K_{Leg} was assessed via accelerometer. The change scores across frequencies for each measurement technique were moderately correlated ($r = 0.36$, $p = 0.001$), but the values derived from the accelerometer were substantially larger (0.15±0.26 vs. 0.26±0.33 vs. 0.68±0.82). As such, our second hypothesis was partially supported in that the accelerometer was sensitive to changes in K_{Leg}.

K_{Leg} determined via motion capture differed significantly between each hopping frequency, but the accelerometer was not sensitive to differences in K_{Leg} between 1.5 Hz and hopping at self-selected frequency (2.1 Hz). Previous research suggests that waist-mounted accelerometers can detect differences in K_{Leg} during single-leg hopping from 2.2 Hz to 2.6 Hz to 3.0 Hz.[197] Hopping at 1.5 Hz is on the lower end of the spectrum of human hopping frequency and is typically completed during double-leg hopping.[16, 80] Our subjects reported and demonstrated difficulty maintaining the 1.5 Hz hopping frequency and typically required 3-5 seconds to achieve steady-state frequency. This difficulty likely influenced the lack of predictor variables explaining significant variance at 1.5 Hz hopping given our subjects were unfamiliar with the task and did not display coordinated movements patterns during the task. Our study is the first to our knowledge to analyze single-leg hopping at such a low frequency with a body worn accelerometer, and the accelerometer may not be sensitive enough at this low of frequency. Furthermore, the accelerometer approach does not accurately reflect K_{Leg} assessed via motion capture, as the accelerometer values consistently overestimated K_{Leg} at all hopping frequencies. Collectively, these data suggest that K_{Leg} displayed during hopping should not be used as a clinical estimate for K_{Leg} during running.

For our third hypothesis, we compared peak force and loading rate derived from the force plate and accelerometer data. While peak force obtained from the accelerometer was correlated with and similar in magnitude to that derived from the force plate, loading rates differed substantially between measurement techniques. As such, our third hypothesis was partially supported. The discrepancy in loading rates is likely due to the temporal shift of the peak force determined via the accelerometer occurring earlier following ground contact, thus resulting in a higher loading rate (Figure 6.3). This is likely due to the errors associated with acceleration drift and magnified with the double integration of the force curve to determine COM position (Figure 6.4).[198] Given COM displacement (Δy) is the denominator of the K_{Leg} equation ($F_{max}/\Delta y$), any reduction in Δy would artificially inflate the K_{Leg} values. This is supported by previous data suggesting that accelerometer derived data may be influenced by systemic error in determining take-off and landing phases during vertical jump by overestimating flight time.[159] Our study is in agreement with Rowlands et al.[199] who reported a strong correlation between peak impact force and average





resultant force determined via a waist-mounted accelerometer. However, this study also reported a strong correlation with waist-mounted accelerometer raw acceleration output and peak loading rate determined via force plate which our study did not agree. Rowlands et al. tested 2 types of commercial accelerometers which both differed from our selected accelerometer. Interestingly, there were discrepancies in acceleration values between the similarly located accelerometers based on the specific activity. (i.e. one accelerometer displayed significant higher output during "low jumps" compared to the other accelerometer). This demonstrates the activity sensitivity that certain accelerometers display and caution that should be taken when comparing accelerometers.

Accelerometers have been utilized previously to quantify overall physical activity[155] and spatio-temporal characteristics (e.g. cadence and velocity) during slower paced movements such as walking.[161, 200] Recent studies have also shown that body-worn accelerometers are sensitive to temporal and loading parameters during running and jumping tasks,[159, 160, 197] but are not sensitive to velocity, power, and contact time.[159] Previous studies have demonstrated that accelerometers overestimate vertical oscillation during running compared to motion capture.[201-203] Given our K_{Leg} values derived via accelerometer were greater than K_{Leg} determined via motion capture, our accelerometer likely underestimated COM displacement thereby driving up the K_{Leg} values. This is likely due to errors introduced and compounded during the double integration of the accelerometer curve. To further analyze the discrepancies between motion capture and accelerometer. We analyzed the ratio of the peak force detected via accelerometer vs. motion capture and determined that discrepancy ratio was 1.37 ± 1.01, indicating that the accelerometer slightly overestimated the peak force compared to the vGRF obtained via the force plate. We computed the same ratio for COM displacement determined via accelerometer (4.90 ±

5.05 cm) and motion capture (9.76 ± 7.34 cm) which generated a ratio of 0.57 ± 0.5, indicating the accelerometer substantially underestimated COM displacement. These data collectively suggest that errors introduced through double integration of the acceleration curve to derive COM displacement is the primary contributor to the discrepancy in K_{Leg} between measurement techniques by reducing the denominator of the K_{Leg} equation ($F_{max}/\Delta y$).[204]

Previous research has utilized a variety of positions for body worn accelerometers including the malleoli,[29] distal tibia,[29, 160, 205] proximal tibia,[30] superior knee,[200] waist,[158, 199, 206] lumbar spine,[161, 165] sacrum,[159] torso,[201] wrist,[158, 199] and head.[30, 160] We placed the accelerometer medial to the anterior superior iliac spine of the dominant limb in an attempt to assess overall acceleration experienced by the center of mass during dynamic activity. This location likely limited the sensitivity of the accelerometer given the damping effect of the ankle, knee, and hip joints on impact loading experienced by the pelvis. Caution should be used when comparing acceleration parameters from different studies given accelerations detected near the ground-body interface are more likely to display greater values compared to more proximally position accelerometers.[30] Future research should utilize a variety of accelerometers in different locations to determine the most appropriate location to approximate K_{Leg}.

Conclusion

Our study demonstrates there is a significant relationship between K_{Leg} during self-selected hopping and K_{Leg} during self-selected running. However, given the weak magnitude of this relationship, stiffness obtained during hopping should not be used as a proxy for stiffness during running in the clinical setting. This study demonstrates that it is important to observe running biomechanics in the clinical setting to determine potential injury risk and make determinations about rehabilitative targets. Additionally, accelerometers are capable of estimating peak forces during hopping, but loading rate is not accurately reflected. Given these data, utilizing waist-mounted accelerometers to estimate K_{Leg} is not clinically appropriate. Future studies should incorporate a multitude of accelerometers along with a variety of accelerometer locations to improve the efficacy of body worn accelerometers to estimate K_{Leg}.

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