

The Pennsylvania State University  
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**INFLUENCE OF FOOT AND ANKLE STRUCTURE ON OPTIMAL  
PERFORMANCE IN DIFFERENT MOTOR TASKS**

A Dissertation in  
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by  
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## **ABSTRACT**

The plantarflexor muscles play a critical role in the successful performance of many motor tasks performed by humans. Foot and ankle structure determines the leverage of the plantarflexors and thus affects how the plantarflexor muscles function. Variation in foot and ankle structure across individuals suggests that certain individuals are better adapted to perform certain tasks, since tasks vary with respect to the requirements placed on the muscles. Previous studies have shown that subjects who specialize in performance of certain activities have foot and ankle structure and muscle-tendon force-generating properties that are different in many respects from those of controls. Results have, however, occasionally not been consistent across studies and the mechanisms that explain how variation in structure affects performance are not always clear. The purpose of this dissertation was to investigate how variation in several ankle and foot structural properties allow for optimal performance in various motor tasks.

In our first study a computational model was created to study maximal-height single-joint ankle jumping. Simulations suggested that bouncing was the optimal strategy for this task, and the best human subject jumpers also jumped highest when bouncing. The model showed that a bouncing strategy allowed for increased elastic energy storage in the plantarflexor tendon. The frequencies of bouncing employed by the computational model and by the best human jumpers further suggest that subjects made use of mechanical resonance to improve performance in this maximal-height explosive task. The exploitation of muscle resonance has been identified previously as a successful strategy in endurance activities requiring a large amount of metabolic energy. The use of resonance for maximizing performance in an explosive type of activity, however, has not been previously reported.

The second study was designed to evaluate the effect of variation of foot and ankle structure on performance in maximal-height single-joint jumping. In this study subjects employed a single upward thrust with no countermovement or bouncing allowed. Significant correlations were found between jump height and heel length, and jump height

and toe length. Subjects with longer heels and longer toes jumped highest, although these subjects were not necessarily the largest subjects.

The third study aimed to explain the mechanism by which shorter heels could potentially reduce the metabolic cost in running. It has been suggested that distance runners with shorter heels experience increased plantarflexor muscle force, thereby increasing tendon energy storage and return and thus reducing metabolic cost. An inverse relationship was found between heel length and peak force, but neither heel length nor peak force were significantly correlated with rate of oxygen consumption during running. Increased tendon force could be associated with increased muscle activation which would cause an increase in metabolic cost, negating the benefit of increased energy storage and return.

In the fourth study we employed a modified version of the computational model from the first study to investigate how variation in foot and ankle structure influences performance in three different motor tasks: maximal vertical-energy pushoff, maximal horizontal acceleration, and maximal static load support. In both explosive movements, short heels and small normalized tendon lengths allowed for larger force production. High stiffness had a positive effect on performance in the maximal vertical-energy pushoff task, whereas a low stiffness improved performance in the maximal horizontal acceleration task. In the isometric task a long heel, small normalized tendon length, and high stiffness caused larger force production. Variation in toe length did not substantially affect performance. To our knowledge there had been no previous work investigating variation in multiple foot and ankle structural parameters and its influence on performance in multiple tasks.

In conclusion, this dissertation aimed to increase our understanding of the effect of foot and ankle structural variation on performance in different tasks. While our findings have expanded our knowledge of these effects, questions remain as to the exact nature of how variation in structure influences performance of different activities in different individuals.

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He worked hard every day with a smile on his face. I hope some of him will  
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# **Chapter 1**

## **Introduction**

### **1.1 Background**

The human foot and ankle constitute a lever mechanism that facilitates plantarflexor muscle function during locomotion and in the performance of other vital motor tasks. Muscles generate force according to their activation level, length, and shortening velocity. Muscle length and shortening velocity are both affected by joint structure, which determine muscle moment arm, and tendon dynamics. There is good evidence that muscle and joint architecture vary considerably across individuals. This variation allows for some individuals to be optimal performers in certain tasks, but could also potentially explain the cause of functional impairments in others. While performing various tasks our muscles are also subjected to different stresses as the requirements for muscle force, moment, and power vary considerably across different tasks.

The capability of our muscles to generate movement through the production of muscle force affects our ability to perform activities of daily living. Knowledge of how muscle architectural properties and joint structure influence our capacity to perform motor tasks is essential if we are to understand the effects of interindividual variation on locomotor function. For example, it is important to recognize the muscle and joint architectural factors that influence mobility for older adults. Understanding the mechanisms by which muscle and joint properties contribute to maintaining motor function during later life could lead to interventions that preserve mobility and quality of life among older adults. In addition, investigations into muscle and joint properties of athletes with differing abilities could better explain why certain athletes perform better in certain events.

Previous investigations have illustrated the critical role played by the plantarflexor muscles in many locomotor activities. The plantarflexor group is the largest contributor to lower extremity positive joint power during walking and running (DeVita et al., 2007;

Farris and Sawicki, 2011; Winter, 1983) and it has been shown that elderly show a significant decline in plantarflexor muscle function (*e.g.* Winter et al., 1990; Morse et al., 2005). Conclusions regarding plantarflexor architectural properties and ankle joint structure have mainly focused on differences between specific populations, such as athletes versus non-athletes (*e.g.* Lee and Piazza, 2009) or sprinters versus distance runners (*e.g.* Abe et al., 2000). Some studies have produced conflicting results regarding the optimal muscle and joint properties for a given task as well as the underlying mechanisms by which certain properties affect performance. For example, evidence is not conclusive on whether stiffer tendons are beneficial to distance runners (Arampatzis et al., 2006; Kubo et al., 2010), or whether two different mechanisms explain the benefit of shorter plantarflexor moments arms to both sprinters and distance runners (Baxter et al., 2012; Scholz et al. 2008). Overall, our understanding of the connections between plantarflexor function and foot and ankle structure remains incomplete and in need of clarification.

## **1.2 Purpose of the Dissertation**

The purpose of the dissertation is to evaluate how different ankle plantarflexor muscle architectural properties and ankle joint structural properties affect performance in different motor tasks. Specifically, I will use human subjects experiments and computer simulations to investigate which properties allow for optimal performance in a variety of motor tasks and the mechanisms by which these properties permit optimal performance. A computational modeling approach will be used to simulate movements involving the ankle joint and its musculature. The aim will be to create a model that is detailed enough to study the underlying properties of the ankle joint and muscles, yet simple enough that extraneous effects can be ignored. Using the model, ankle joint and muscle properties will be optimized separately for different motor tasks and then compared across motor tasks. Comparison across different motor tasks using a single computational model will allow a critical analysis of previously proposed mechanisms that try to explain why certain ankle joint and muscles properties appear beneficial for certain motor tasks. Experimental

measurements will be made that explore the effects of selected ankle joint properties on jumping and distance running performance.

### *Aims and Hypotheses*

**Aim 1: To create a computational musculoskeletal simulation of a maximal-height jumping task (to be used in subsequent aims to investigate structure-function relationships) and test its predictions using experimental data from human subjects experiments.**

*H1.1. The computational model of maximal-height single-joint ankle jumping will use a countermovement jump as optimal strategy.*

*H1.2. Human subjects performing maximal-height single-joint ankle jumping will also use a countermovement jump as optimal strategy.*

*Rationale:* Creating a simple single-joint ankle model with a lumped plantarflexor and dorsiflexor muscle will allow for a focused investigation into the optimal muscle and joint properties and help elucidate the mechanism affording performance enhancement. However, a model needs to be evaluated in order to be confident that it can effectively simulate a given movement. Maximal-height jumping is an unambiguous task with a clear objective function that can be used to assess the validity of the model. Previous studies of multijoint and single-joint jumping have shown that a countermovement jump increases jump height. Once the computational model has been evaluated against previous results and subject data, it can be implemented with greater confidence to study other motor tasks.

**Aim 2: To examine how variation in ankle and foot structure affect human performance in a single-joint jumping task**

*H2.1. Maximal jump height will be negatively correlated with heel length, which is representative of plantarflexor moment arm.*

*H2.2. Maximal jump height will be positively correlated with toe length.*

*Rationale:* Differences in foot and ankle structure have been found when comparing athletes with non-athletes in sprinting (Baxter et al., 2012; Lee and Piazza, 2009) and correlations have been found between ankle structure and running economy in distance



runners (Scholz et al. 2008). Little attention has been paid to how foot and ankle structural variation could also explain performance differences in other optimal-performance tasks. Maximal-height jumping is one such activity. Higher joint moments at the ankle would be expected to produce higher jumps, but which factors influence the ability to produce joint moment? One might assume that for a given muscle force, individuals with longer heels, or larger plantarflexor moment arms, should be able to produce higher torque and hence jump higher. However, recent simulation studies have shown that longer heels might reduce the force-generating capacity of the ankle muscles at fast ankle rotation speeds (Baxter et al., 2012, Nagano et al., 2003). It is therefore not clear whether a short or long heel length would be beneficial in a vertical jump task. Interaction between the ground and foot influences the dynamics at the ankle joint, and it is of interest to see how toe length could potentially affect performance during a jumping task. Toe length changes the lever arm of the ground reaction force and could negatively affect the mechanical advantage of the plantarflexor muscles. On the other hand, computational modeling studies of sprinting have shown that longer toes may allow the foot to stay on the ground longer and thus enhance impulse for forward propulsion (Lee and Piazza, 2009). It is not clear whether jumpers would benefit from having shorter or longer toes.

**Aim 3 – To perform experiments to examine mechanisms for enhanced distance running performance suggested by previous studies.**

*H3.1. The rate of energy consumption of endurance runners will be positively correlated to heel length, which is representative of plantarflexor moment arm*

*H3.2. The rate of energy consumption of endurance runners will be negatively correlated with the gear ratio at the initiation of the pushoff phase of the running stride*

*H3.3. The rate of energy consumption of endurance runners will be positively correlated with toe length*

*Rationale:* The results of two recent studies have suggested that distance runners benefit from having a shorter plantarflexor moment arm, with a correlation between rate of energy consumption and moment arm measures (Scholz et al., 2008; Raichlen et al., 2011).

Scholz et al. (2008) proposed that a shorter moment arm allows for increased tendon energy storage and an ultimate reduction in the metabolic cost of running. Others have shown that smaller moment arms also benefit sprinters (Lee & Piazza, 2009; Baxter et al., 2012). Baxter et al. (2012) further explored the dynamics of the foot structure of sprinters and found that not just the muscle moment arm, but also the ratio of forefoot length to rearfoot length are important factors in determining the action of the plantarflexor muscles. Further, dynamic changes under control of the runner can affect the forces transmitted to the plantarflexor muscles. Runners can vary the degree of foot rotation or the area of the foot with which they strike the ground, and both of these will change the ground reaction force lever arm and ultimately the ratio of the output to input lever arms about the ankle, the ankle joint “gear ratio” (Bojsen-Moller, 1978; Erdemir and Piazza, 2002; Ardigo et al., 1995). The physical properties of the foot structure, as well as dynamics of foot-ground contact, can be modulated in order to affect the forces transmitted to the muscles and the energy stored in the tendon. A smaller moment arm in itself might benefit energy storage. However, athletes who demonstrate larger gear ratios during the pushoff of running, either by having larger forefeet or through dynamic changes in the ground reaction force lever arm, could also see increases in elastic energy storage and reduction in metabolic cost. The gear ratio will be affected by the length of the foot and the nature of the foot-ground interaction. Increased toe length could potentially increase the gear ratio, increase elastic energy storage, and reduce metabolic cost. Conversely, recent work has shown that shorter toes reduce positive work at the metatarsophalangeal joint, which could potentially reduce metabolic cost during running (Rolian et al, 2009).

**Aim 4 – To use a modified version of the computational model from Aim 1 in optimizations that will determine optimal ankle and foot structural properties for different motor tasks**

*H4. Optimal ankle muscle and joint properties will differ across different motor tasks when simulated and optimized using a single computational model.*

Task specific hypotheses:

*H4.1. The properties most essential for maximal-height jumping will be:  
Compliant tendon; short plantarflexor moment arm; long toes*

*H4.2. The properties most essential for a sprint pushoff will be: Long fascicles;  
short plantarflexor moment arm; long toes*

*H4.3. The properties most essential for a maximal load support will be: Stiff  
tendon; long plantarflexor moment arm*

*Rationale:* Findings relating specific ankle and foot properties to optimal performance in a given motor tasks, and mechanisms by which this occurs, have not been conclusive. Using a single computational model to simulate movement permits one to investigate optimal muscle and joint parameters for different motor tasks, while the effects of extraneous subject variability are eliminated. Using a modeling approach, muscle and joint properties, as well as factors external to the ankle joint, can be optimized simultaneously and the interplay between various properties can be evaluated. Having one model to simulate three different motor tasks (maximal vertical-energy pushoff, maximal average horizontal acceleration, and maximal static load support) gives more credibility to comparisons between different ankle muscle and joint properties as well as foot structural properties that are optimal for different tasks.

### **1.3 Organization of the Dissertation**

In the following chapters a literature review will be presented followed by four research studies and a final discussion. Chapter 2 contains a survey of previous work that relates to the topic of this dissertation. Chapters 3 describes the creation and validation of a computational model using a maximal-height single-joint jumping protocol. Chapter 4 contains an investigation into foot and ankle structure and its influence on performance in a single-joint jumping tasks. In chapter 5 we examine possible mechanisms for performance enhancement in running due to foot and ankle structural variation. A modeling approach is used in chapter 6 to examine optimal foot and ankle structure for different motor tasks. The final chapter, chapter 7, contains a final discussion and conclusion of the dissertation.

## **Chapter 2**

### **Literature Review**

#### **2.1 Introduction**

The aim of this literature review is to provide background on the structure and function of the ankle joint and its musculature in order to shed light on how structural variations affect function. Firstly, the special role of the ankle joint during various motor tasks will be discussed. The review will then consider anatomical difference between different groups, and groups with different abilities performing different motor tasks to investigate the possibility that ankle joint or muscle properties allows for enhanced performance in given tasks. Subsequently, the dynamic action of the musculature of the ankle joint during various motor tasks will be reviewed. How the muscles function during various motor tasks, for example whether fibers shorten or contract isometrically, can highlight the importance of certain properties of the muscles during specific movements. Finally, other considerations regarding the modulation of the dynamics of the ankle joint will be considered. This will include investigations into other anatomical factors not specific to the ankle joint as well as forces that significantly affect the action of the musculature at the ankle joint. Throughout the review, the focus will be on the plantarflexor muscles and their joint action, as they are a major contributor to propulsion during locomotion.

#### **2.2 The Contribution of the Ankle Joint During Movement**

Despite the fact that all joints work together to perform various movements, numerous studies have highlighted the specific importance of the ankle joint (*e.g.* DeVita et al., 2007; Winter et al., 1990). Ankle joint function or dysfunction affects how we perform tasks in everyday life as well as more specific tasks performed by special populations such as elite athletes.

During the stance phase of walking the ankle produces a significant portion of the positive mechanical power (DeVita et al., 2007), and during running the ankle joint contributes more to the total average positive power than the hip and knee (Farris and Sawicki, 2011; Winter, 1983). The transition from walking to running speeds has also been linked to the role of the plantarflexor muscles. Forward dynamics simulation of walking and running at the transition speed have shown that plantarflexor muscles are able to produce double the force at similar activation levels when running compared with walking at the transition speed (near 2 m/s) (Neptune and Sasaki, 2005). These modeling results were later confirmed in an experimental study when Farris and Sawicki (2011) used ultrasound to identify muscle behavior during walking and running and found that during running at the transition speed, the muscle fascicle velocity was reduced at the time of peak force development, which related to an increase in peak and average force development in the medial gastrocnemius.

Furthermore, the elderly show a reduction in ankle joint function which need to be compensated for at the other joints. There is a significant decline in the pushoff work generation at the ankle joint during walking for healthy elderly compared with young individuals walking at the same speed (Winter et al., 1990), as well as increases in muscle activation of ankle muscles at midstance (Schmitz et al., 2009). Changes in performance have also been shown to be related to the physical abilities of the elderly. Elderly with low-physical performance increase ankle power output with increased walking speed, and were unable to match peak power developed by the healthier and fitter elderly (Graf et al., 2005). Lee and Piazza (2012) have found some structural differences between elderly with different physical abilities, in that a higher correlation existed between plantarflexor muscle moment arm in slower walking elderly compared with faster walking elderly.

Sports performance studies have also considered the influence of ankle action on performance. The plantarflexors contribute significantly to large support forces during acceleration phase of sprint running (Dorn et al., 2012) and these large forces are related to maximal running speeds in humans (Weyand et al., 2010). Scholz et al. (2008) have shown that the rate of energy consumption in distance runners is correlated with the length

of the plantarflexor muscle moment arm. In general, various ankle muscle and joint properties differ between athletes specializing in different events (see below).

## **2.3 Architecture of the Plantarflexor Muscles**

### **2.3.1 Muscle fiber type**

Muscle fiber type differs between individuals and it seems evident that variable muscle fiber type composition could favor individuals depending on the task they perform. A general classification of fiber types divides muscle fiber types into fast fibers that have low endurance compared with slow fibers with a higher endurance (Lieber, 2009). The fast twitch fibers have potential of fast force production, but cannot maintain the force producing levels, whereas slow twitch fibers cannot produce force quickly, although maintenance of the force production is possible. It has been shown that endurance athletes have a higher percentage of slow twitch fibers compared with untrained individuals and sprinters, and conversely sprinters have a higher percentage of fast twitch fibers compared with the untrained and endurance athletes in the gastrocnemius muscle (Bergh, 1978; Costill et al., 1976). Zajac et al. (1984) suggested that faster plantarflexor muscle fibers should increase jump height for a single joint ankle jump task. More recently, Baguet et al. (2011) compared the gastrocnemius muscles of different athletes and found a greater percentage of the cross-sectional area occupied by fast twitch fibers for power and sprint athletes (jumps, throws, 100-800m) compared with control subjects, with endurance athletes (from 1500 m to marathon) having the lowest values. Comparisons between athletes of different abilities within the same discipline have also found differences in fiber type composition. A significant negative correlation ( $r = -0.62$ ) exists between the percentage of slow twitch fibers in the lateral gastrocnemius and distance runner's best six-mile time (Fink, 1977), and Clarkson et al. (1980) found a significant negative correlation ( $r = -0.94$ ) between percentage of slow twitch fibers in the medial gastrocnemius and the maximal isometric ankle strength of weight lifters.

### 2.3.2 Muscle fiber length

Muscle fiber length is an important, if not the most important architectural characteristic of the muscle (Lieber and Ward, 2011). The excursion of the muscle, as well as how the fibers are arranged within the muscle, is prescribed by the muscle fiber length (Lieber and Ward, 2011). An increase in fiber length permits a larger excursion and allows the muscle to operate at higher contraction velocities. Shorter fibers cannot maintain their maximal force production over changes in length, and a small change in length causes a drop off the optimal value on the force-length relation curve. Studies investigating fiber or fascicle length and interindividual variability have found gastrocnemius fascicle lengths are longer in sprint athletes (Abe et al., 2000; Kumagai et al., 2000; Lee and Piazza, 2009). Even when comparing sprint athletes of varying ability (100 m time of below and above 11 s), Kumagai et al. (2000) found that better sprinters have longer fascicles in the gastrocnemius medialis. However, Karamanidis et al. (2011)) did not find differences in gastrocnemius fascicle length for two more homogenous groups (100 m time of below and above 10.7 s). Modeling results have also shown that longer fascicles are beneficial for activities that are of a fast concentric nature (Nagano et al., 2007). These results seem to suggest that sprinters have longer muscle fascicles than non-sprinters and distance runners. The advantage that long fibers afford sprinters stems from the fact that longer fibers can shorten at higher velocities while maintaining relatively high force output. On the other hand, for a given muscle volume short fibers increase the physiological cross-sectional area (PSCA), which allows for greater force production during isometric tasks (Lieber and Ward, 2011). This is shown by the following relationships:

$$PSCA = \frac{V_{muscle} \cdot \cos\theta}{l_f} \quad (1)$$

$$F_{isomax} = \sigma_{muscle} \cdot PSCA \quad (2)$$

With  $V_{muscle}$  being the muscle volume,  $l_f$  the muscle fiber length and  $\theta$  the pennation angle of the muscle fibers.  $F_{isomax}$  is the maximum isometric force the muscle can produce, whereas  $\sigma_{muscle}$  is the muscle specific tension, which is assumed constant for mammalian muscles (Lieber and Ward, 2011).

### ***2.3.3 Pennation angle***

The maximum isometric force that a muscle can produce is proportional to the PCSA (Equation 2), which in turn is proportional to the cosine of the pennation angle of the fibers (Equation 1). A larger pennation angle reduces to force in line with the muscle pull. However, for a given volume a larger pennation angle packs more shorter length fibers into the muscle. These muscles fibers arranged in parallel increases the PCSA and therefore the maximum isometric force. The effect of pennation angle has on the PCSA and maximum isometric force is minimal (Zajac, 1989). When considering the pennation angle of the soleus, which has the largest pennation angle of muscles crossing the ankle joint, approximately 25° (Biewener et al., 2004; Friederich & Brand, 1990; Ward et al., 2008; Wickiewicz et al., 1983), the decrease in PCSA due to this angle would be less than 10%. However, reports of pennation angles in various populations have found significant results. Distance runners appear to have larger pennation angles than both sprinters and control subjects (Abe et al., 2000; Karamanidis and Arampatzis, 2006), with no difference being shown between sprint athletes and controls (Abe et al., 2000; Lee and Piazza, 2009).

### ***2.3.4 Tendon properties: stiffness and resting length***

Tendon contributes significantly to the overall properties and dynamics of the muscle-tendon unit. Tendon stiffness affects the amount of energy stored in a tendon as well as force safety margins inside the muscle-tendon unit (*e.g.* Kongsgaard et al., 2005; McNeill Alexander, 2002). Numerous studies (*e.g.* Arampatzis et al., 2006; Kubo et al., 2007; Rosager et al., 2002) have considered the effect of tendon stiffness on performance and results have not always been consistent. Kubo et al. (2010) found lower tendon stiffness to be correlated with better performance in long distance running (5000 m), which compares favorably to previous modeling study results (Voigt et al., 1995). Voigt and colleagues varied Young's modulus of tendons and found that a decrease in stiffness increased gross mechanical efficiency. Others, however, have found higher tendon stiffness in the triceps surae for the most economical runners (Arampatzis et al., 2006) or no difference in tendon stiffness between runners and non-runners (Rosager et al., 2002). Results show that jumping performance seems to be enhanced by reduced stiffness, or more



compliant tendons (Bobbert, 2001; Zajac et al., 1984). However, Kubo et al. (2007) found a positive correlation between a single-joint ankle jump height and Achilles tendon stiffness. Sprint runners have been found to have higher tendon stiffness compared with endurance runners and controls (Arampatzis et al., 2007). Muraoka et al. (2005) have also shown that the Achilles tendon was stiffer in subjects with greater muscle strength, which may play a role in reducing the probability of tendon strain injuries.

Tendon slack length ( $l_{TSLACK}$ ) is the maximal length of the tendon before it transmits force (Zajac, 1989), and differences in tendon slack length seem to affect efficiency during jumping. A decrease in tendon length reduces efficiency whereas a lengthening increases efficiency (Voigt et al., 1995). Assuming tendon stiffness stays constant for tendons of different lengths, Zajac (1989) has pointed out that the longer the  $l_{TSLACK}$  the more compliant the tendon is and that actuator compliance depends on the  $l_{TSLACK}$ -to-optimum fiber length ratio, and not necessarily the  $l_{TSLACK}$  value alone. Using a computer simulation, Nagano et al. (2004) studied the effect of varying the  $l_{TSLACK}$ -to-optimal fiber length ratio on propelling a single mass upward and it was found that higher heights are obtained with a longer series elastic element, or longer  $l_{TSLACK}$  when the mass is small, while with a shorter series elastic element, or shorter  $l_{TSLACK}$  when the mass is large.

### **2.3.5 Muscle moment arm**

Muscle function cannot be deduced by considering muscle architecture alone, since muscles act through a moment arm to produce a moment at a given joint (Lieber and Ward, 2011). Therefore moment arm variations as well as muscle architectural properties play a role in determining joint moments and ultimately movement. Although there are many muscles contributing to the plantarflexor joint moment at the ankle, the triceps surae group, consisting of the soleus and medial and lateral gastrocnemius, are the prime movers, responsible for more than 90% of joint moment produced at neutral angle (from OpenSim: Lower Limb Model 2010 - Arnold et al. (2010)). The triceps surae group attaches to the Achilles tendon which insert on the calcaneus on the foot. The triceps surae muscle (or

Achilles tendon) moment arm affects the force producing capability of the muscle itself, as well as affecting the joint moment through the following equation:

$$\text{Moment} = \text{Moment arm} \times \text{Force} \quad (3)$$

The moment produced at the joint is the product of the muscle moment arm and the force produce by the muscle. The moment arm also affects the range of movement of the muscle tendon unit. For a given rotation at the ankle joint, a large plantarflexor muscle moment arm will mean that the muscle itself will lengthen more than if the moment arm was small. This has implications for the velocity of shortening of the muscle. Shorter moment arms will allow the muscle to have a slower shortening velocity which is favorable from a force-velocity standpoint. However, during isometric contractions large moment arms multiplied with a given force will increase the torque produced at the joint. Recently a number of studies have considered the ankle joint moment arm of the plantar flexors and difference between different populations. Sprinters appear to have smaller plantarflexor moment arms than controls (Baxter et al., 2012; Lee and Piazza, 2009). It has been proposed that the reduced moment arm allows the plantarflexor muscles to operate at a reduced velocity for a given ankle rotation velocity, which is beneficial for force production based on the muscle force-velocity relationship (Baxter et al., 2012). Reduced plantarflexor moment arms have also been found in better endurance athletes (Raichlen et al., 2011; Scholz et al., 2008), and Scholz et al. (2008) suggested that shorter moment arms increase the necessary force in the plantarflexors for a given joint moment. This increase in force would cause an increase in the amount of energy stored in the tendon, thereby allowing the tendon to make better use of elastic energy storage and return to reduce the metabolic cost of running. Voigt et al. (1995) used a modeling approach to show that decreasing the moment arm by 10% will lead to a 13% increase in the gross efficiency for a hopping task. In another modeling study, Nagano and Komura (2003) explained how smaller moment arms increase joint moment, joint power, and joint work compared with large moment arms for fast movements (greater than 120°/s). Given that both sprinting and distance running require the ankle joint to rotate at velocities higher than 120°/s (Nagano et al., 2007), it seems likely that both these activities would benefit from having smaller moment arms at the ankle joint. This small moment values for seemingly different

activities are mirrored by a study by an investigation showing smaller moment arm for volleyball players and long distance runners compared with control subjects (Watanabe, 2008). Results are not necessarily conclusive though, with Rosager and colleagues (2002) finding no difference between moment arm of runners and non-runners.

It is important to note that the different muscle architectural properties are interrelated, and although many of the studies mentioned above have measured certain specific muscle or joint properties separately, interplay between properties could affect joint function in ways that might not be expected when only one factor is considered. For example, it has been found that a large range of muscle fiber lengths of the medial gastrocnemius, (45 to 70 mm), combined with a large range of tendon stiffness values (150–500 N/mm) can allow movement close to optimal efficiency for both walking and running (Lichtwark and Wilson, 2008). Also, although muscle fiber length in itself in part determines the range over which the fibers operate, the actual rotation that the joint undergoes significantly affects the muscle's force producing capabilities. Small moment arms reduce fiber excursion for a given joint rotation, whereas longer moment arms will have the opposite effect. For this reason the ratio of fiber length to moment arm, and not fiber length alone, seem to be an important characteristic. A high fiber length to moment arm ratio contributes to the production of a relatively constant joint torque, whereas a low fiber length to moment arm ratio indicate a motor producing large force changes in the muscle as the fiber length changes and therefore large torque output variation during joint rotation. Lee and Piazza (2009) found the ratio of fascicle length to moment arm was 50% larger in sprinters compared with controls, implying that sprinters can maintain the torque at the joint over a large portion of the ankle motion.

## **2.4 Dynamic Action During Movement**

Movement of the muscle-tendon unit during activities affects how force is developed based on the force-length and force-velocity properties of the muscle fibers as well as the force-length properties of the tendons. It is important to distinguish between concentric, eccentric and isometric actions of the muscle-tendon unit as a whole, compared

with muscle contractions in which the muscle fibers shorten, lengthen or stay at a constant length, since the velocity of contraction of the whole muscle tendon unit is not equivalent to the velocity of the muscle fibers (Fukunaga, 2002). Therefore, although the muscle-tendon unit can contract isometrically, the muscle fibers might be shortening (as in the start of any isometric force production). Also the muscle-tendon unit could be lengthening with the muscle fibers contracting isometrically. Muscle fiber and tendon interaction within the muscle-tendon unit is also important as it affects how much energy is stored in the tendon, which could reduce the metabolic cost of a given activity.

Kinetic and kinematic patterns differ for different motor tasks and these differences put different demands on the muscle tendon unit. Studies looking at walking and the function of the plantar flexors have used ultrasound to show that the medial gastrocnemius fascicles act near isometric during the stance phase of walking and that the tendon stretches and then recoils to allow energy storage and return (Fukunaga et al., 2002; Ishikawa et al., 2005; Lichtwark and Wilson, 2007). In running however, it seems that the fascicle of the medial gastrocnemius shortens during the contact phase (Ishikawa and Komi, 2007; Lichtwark and Wilson, 2007) and it has been suggested that the shortened fascicles might allow more tendon stretch which could possible enhance elastic energy storage in the tendons.

During squat jumping the medial gastrocnemius fascicles shorten during the first part of the push off as the tendon lengthens, keeping the muscle-tendon unit at constant length after which the fascicles contract again near isometrically during the final part of the pushoff phase, with the muscle-tendon unit and tendon rapidly shortening (Kurokawa et al., 2003). Kawakami et al. (2002) have shown that during a countermovement jump that the change in fascicle length is lower during the final pushoff phase compared with a squat jump, whereas others (Fukashiro et al., 2006) suggest that the medial gastrocnemius fascicles act nearly isometrically during squat jumps, countermovement jumps and drop jumps, with shortening taken up mostly by the tendon. In investigating the muscle fascicle behavior during drop jumps, Sousa et al. (2007) indicates that an optimal drop height exists. When the drop height exceeded a specific height, the medial gastrocnemius fascicles are lengthening during the braking phase, whereas this is not the case for lower heights.

The authors argued that the stretch may reduce the ability of the tendon to return energy during the final pushoff phase.

During an ankle bending activity it has been shown that during a rapid plantarflexion phase following dorsiflexion the plantarflexor muscle fascicle stayed near isometric during first half of plantarflexion phase while the tendon shortened, after which both shortened (Fukashiro et al., 2006). The authors suggested that it shows a near isometric fiber action at the transition between the dorsiflexion and plantarflexion.

It seems evident that isometric contraction of muscle fibers is beneficial for many different activities. Hof et al. (1983) have described this behavior as ‘concerted contraction’, where the muscle is activated in such a manner as to ensure that muscle fibers contract isometrically.

## **2.5 Other Considerations**

Anatomical variations distal to the ankle joint have the ability to affect what happens at the ankle joint. Baxter et al. (2012) have shown that ratio of the length of the forefoot to the length of the rearfoot influences ankle muscle functioning irrespective of the absolute length of the plantarflexor moment arm. Model results showed that a larger forefoot-to-rearfoot ratio increase work done by the plantarflexor muscles during faster motions similar to a sprint start. This increased ratio allows the plantarflexor muscle fibers to shorten at a lower velocity for a given rotational velocity. Lee and Piazza (2009) found similar results employing a slightly different model, but also found that increased toe lengths allows the sprinter to spend more time on the ground and produce more work due to this. Both Baxter et al. (2012) and Lee and Piazza (2009) found significant differences between the toe lengths of sprinters compared with non-sprinters. Longer toes also permits the out lever, or the moment arm of the ground reaction force (GRF) to be larger, which similarly allows for a reduction in velocity for the plantarflexor muscle-tendon unit, and probably the plantarflexor muscle fibers. This finding relates to the terms gear ratio and effective mechanical advantage. Gear ratio is the ratio for the GRF moment arm to the ratio for the plantarflexor muscle moment arm (Carrier et al., 1994). Other authors (*e.g.*

Biewener et al., 2004) have used other terminology - effective moment arm (EMA), which is the inverse of gear ratio to quantify differences between the moment arms. Findings for sprinters indicate that a high gear ratio, or low EMA, is beneficial for force-velocity properties of the plantarflexor muscles. Others (Stefanyshyn and Fusco, 2004; Toon et al., 2009) have also argued that increasing the sole stiffness of shoes in sprinters could effectively increase the output lever or GRF moment arm, which should increase the gear ratio, possibly increase the moment produced at the ankle joint (Stefanyshyn and Fusco, 2004), but also reduce shortening velocity in the plantarflexor muscles. Another factor that seems to be beneficial with respect to increased stiffness is the reduction in negative work performed at the metatarsophalangeal (MTP) joint. A stiffer foot/shoe allows less MTP joint extension, which in turn reduces the negative work performed at that joint. Stefanyshyn and Nigg (2000) also showed increase performance (and reduction in negative MTP work) during a jumping task. Results on this have however not been consistent (*e.g.* Krell and Stefanyshyn, 2006). Higher stiffness does not always mean better sprinting performance and the amount of stiffness is probably tuned to individual athletes. Reduction of MTP joint extension might however negatively affect the ability for energy to be stored in the windlass mechanism, although an increase in rigidity that increases the lever arm might compensate for this (Toon et al., 2009). For distance running it seems that shorter toes might be more favorable, due to the fact that shorter toes will reduce the amount of negative work at the MTP joint (Rolian et al., 2009). However, another way to reduce the negative work done at the MTP joint is by using stiffer soles. Roy and Stefanyshyn (2006) tested this and found a small decrease in metabolic energy consumption for stiffer soles, but no difference in MTP joint work. Braunstein et al. (2010) suggested this when their results indicated that while running with shoes the ankle joint showed a lower gear ratio in early stance and a higher ratio in the late stance and that the higher gear ratio in late stance could benefit force producing capabilities of the plantarflexor muscles. This could imply that the increase in gear ratio rather than reduction in negative work at the MTP joint could be an important factor.

It should be noted that changes in gear ratio and effective mechanical advantage is not something that is static. Although anatomical variations certainly affect gear ratio in a

more fixed way, the gear ratio changes naturally throughout normal stance phase of locomotion. Results from many studies show how the gear ratio starts small during heel strike for running which then increases during the toe-off (Carrier et al., 1994). The larger gear ratio at toe off should have the beneficial effect of reducing muscle-tendon unit velocity. Carrier et al. (1994) suggests that small gear ratio during heel strike should increase the amount of stretch of the tendon, which enhances elastic energy storage. This is however different from suggestions that a smaller moment arm (and in effect a small gear ratio) should increase the force stored in the tendon and through this means allow for a larger amount of energy storage (Scholz et al., 2008). The amount of energy storage in the tendon is a factor of force, stiffness and stretch and how these factors interrelate seems to still require further investigation. Gear ratio or EMA also changes when transitioning from walking to running, with the EMA at the ankle joint increasing 23% for the plantarflexor muscles (Biewener et al., 2004).

The ankle joint is not only affected by anatomical variations distal to the joint, but also due to active modulation of foot-ground contact kinetics and kinematics by a subject. For example, rotation of the foot in the transverse plane changes the GRF lever. During toe-out the foot rotates about a transverse MTP joint axis which increases the gear ratio, whereas during toe-in the foot rotates about an oblique-axis which is closer to the ankle joint and therefore leads to a reduced gear ratio (Bojsen-Moller, 1978; Erdemir and Piazza, 2002). Bojsen-Moller (1978) suggested that subjects can vary this depending on the activity, whereas a high gear ratio could be used for sprinting and a low gear ratio for walking up hill and the sprint start. Sagittal plane foot placement during running also has the ability to affect the gear ratio and ankle muscle functioning. Using a forefoot striking pattern compared with a rearfoot striking pattern in running seems to produce higher mechanical work without differences in metabolic expenditure (Ardigo et al., 1995). Ardigo and colleagues speculated that when a forefoot strike pattern is used, storage of elastic energy (and subsequent return) is better utilized. This would be similar to suggestions made by Scholz et al. (2008). Karamanidis and Arampatzis (2007) have shown that elderly walking at similar speed to the young modulate the gear ratios by shortening

the GRF moment arm. They argue that this will increase the EMA, which should allow plantarflexor muscles to improve force production during walking.

## **2.6 Summary**

It is evident that there are many factors that contribute to how well the ankle musculature can function during various activities. These factors include mechanisms that can be thought of as internal to the joint, as well as external factors. Internal factors include the ankle musculature and the muscle moments arms and many studies have considered how variability in these internal factors might affect execution of various motor tasks. Results seem to suggest that differences do exist between various populations and that certain properties appear to be beneficial for certain activities. Many studies focus on a single variable and often neglect the effect of interplay between different factors.

Factors external to the joint have the ability to modulate the dynamic action of the internal joint structures. Not only fixed external factors, for example foot geometry, are important, but dynamic factors, specifically the kinetics and kinematics of foot-ground contact influences what happens at the ankle joint and need to be considered.

To gain better understanding of the mechanisms responsible for optimal performance in various motor tasks, it seems that internal or external factors cannot be investigated in isolation. Current results on the effects of certain internal joint properties, as well as external factors on motor task performance are not always conclusive, and further investigation seems justified.



## Chapter 3

### **Computational Model of Maximal-Height Single-Joint Jumping Predicts Bouncing as an Optimal Strategy**

*This study was published in the Journal of Biomechanics*

#### **3.1 Introduction**

Maximal-height jumping has been studied extensively using both experimental paradigms (Domire and Challis, 2007; Harman et al., 1990; Komi and Bosco, 1978) and computational modeling (Bobbert and Casius, 2005; Domire and Challis, 2007; Pandy et al., 1990). This task is well suited for experimental study of neuromuscular control and musculoskeletal function because it is a well-defined task with a clear instruction to the subject. Similarly, maximal-height jumping may be numerically simulated with the aid of an unambiguous performance criterion.

Several previous investigations involving both experiment and computer simulation have considered possible control strategies for achieving maximal height during a jump (Anderson and Pandy, 1993; Bobbert et al., 1996; Harman et al., 1990). Most of these studies have considered jumping movements that involve multiple joints, and results from both experimental and modeling studies of varying complexity have indicated that a countermovement (CM), or downward movement of the center of mass, prior to takeoff from the ground enhances the jump height (Bobbert et al., 1996; Harman et al., 1990).

Consideration of single-joint maximal-height jumping offers certain advantages over studying multijoint jumping. Limiting the task to a single joint reduces the complexity of the task (for example, the effects of biarticular muscles and interjoint coordination are eliminated) and thus allows more focused study of the mechanisms underlying optimal performance at the joint. Previous investigators (Levine et al., 1983; Zajac et al., 1984) used a combined simulation and experimental approach to study jumping in which the ankle was the only joint modeled. They used optimal control techniques together with a torque-actuated model (Levine et al., 1983) and a muscle-actuated model (Zajac et al.,

1984) to study maximal-height jumping. Because the foot was constrained to start on the ground in these studies, the effect of CM on jumping performance was not investigated. Kawakami et al. (2002) showed that Achilles tendon work was larger during a CM “jump” compared with a non-CM jump during a single-joint ankle propulsion task where subjects lay on a horizontal sled and pushed off from a platform with their ankles. The horizontal distance moved by the body (analogous to jump height) was not recorded in that study. In another experimental study using a horizontal sled, Kubo et al. (2007) found a mean difference of 3 cm in movement distance between CM jumps and non-CM jumps. The results of these studies suggest that a CM strategy would also be optimal for maximal-height vertical jumping driven by ankle joint muscles alone.

The purposes of this study were: (1) to develop a simple model of the body suitable for computational simulation of maximal-height jumping in which ankle muscles alone are used for propulsion; (2) to use the simulation to investigate optimal strategies for completing this task; and (3) to test the predictions of the simulation against experimental data collected in human subjects. We hypothesized that parameter optimization would reveal CM to be the optimal strategy found for the simulation, and that subjects would jump higher when employing a CM. In addition to identification of the optimal strategy from the simulation and from experimental results, we used the computational model to identify specific mechanisms related to muscle and tendon properties and timing of muscle activation input that enable optimal performance.

## **3.2 Methods**

### ***3.2.1 Computer Simulation***

A simplified mathematical model of one-half of the human body was created that had two degrees of freedom and two massless segments, a foot and a leg, with a point mass representing half of the body mass ( $m = 37.5$  kg) positioned atop the 1.0 m leg segment (Figure 3-1). The ankle was modeled as a revolute joint, as was the pivot between the foot and the ground, which was located at the approximate position of the metatarsophalangeal (MTP) joint, 14 cm distal to the ankle. Ground contact beneath the heel was not modeled,

allowing the heel to move downward initially and thus permitting a countermovement. Each simulation began in a static position with the foot horizontal and the body mass directly above the MTP joint, with the ankle joint in 8° dorsiflexion. This configuration was achieved by applying the minimum muscle activation necessary to maintain the system with zero acceleration in this position.

Two musculotendon actuators, a lumped plantarflexor and a lumped dorsiflexor, acted across the ankle joint. These were modeled as Hill-type actuators with first-order activation dynamics and the muscle model included force–length relations for muscle and tendon and a force–velocity dependence for muscle (Schutte, 1992). Both actuators originated 40 cm proximal to the ankle. Muscle architecture properties were derived by a parameter optimization that minimized the differences between experimental isometric torque–angle curves (both passive and active) from the literature and model-generated ankle torque curves. Parameters determined in this way included: optimal fiber length ( $L_{Fopt}$ ); tendon slack length ( $L_{TS}$ ); maximum isometric force ( $F_{MAX}$ ); tendon elastic modulus ( $E_T$ ); and tendon insertion point on the foot ( $M_{INSERT}$ ), which was the horizontal location with respect to the ankle joint. Zero pennation angle was assumed for both muscles. The experimental maximum isometric joint torque–angle results for plantarflexion and dorsiflexion used in this optimization were those reported by Sale et al. (1982) and Marsh et al. (1981), respectively. The experimental passive joint torque–angle relationship results used in the optimization for both plantarflexion and dorsiflexion were the curves published by Riener and Edrich (1999). Optimizations were performed using the *fmincon()* routine in Matlab (The Mathworks, Inc.; Natick, MA) and converged with RMS errors between simulated and experimental torques of 4.77 Nm (2.8% of maximum) for the effective plantarflexor and 0.65 Nm (1.3% of maximum) for the effective dorsiflexor. The optimization routine was run multiple times with varied initial guesses and repeatedly converged on the same set of parameter values, suggesting that a global minimum was reached. The bounds on the parameters used in the optimization and their values following from the optimization are given in Table 3-1.

Maximal-height jumping was simulated using a separate parameter optimization based on forward dynamic integration of the model equations of motion. The following

objective function, modified from Levine et al. (1983), was maximized during the optimization:

$$J(t_f, u) = y_c(t_{end}) + \frac{1}{2g} \dot{y}_c^2(t_{end}) - K \frac{G_{end}}{mg} - K \frac{(t_f - t_{end})}{t_f} \quad (1)$$

The 43 optimization parameters were the muscle excitations  $u$  ( $0 \leq u \leq 1$ ) with 21 nodes for each muscle and final time  $t_f$ . Pandy et al. (1992) previously used 21 nodes in a simulation of maximum height jumping because this number reduced sensitivity to small changes in the controls without unnecessarily increasing computation time. The first two terms in Eq. 1 represent the jump height and the last two were penalty terms.  $G_{end}$  is the vertical ground reaction force at the end of the simulation. If the model failed to jump,  $G_{end}$  remained positive throughout integration from time zero to time =  $t_f$  and a penalty was incurred. If the model jumped before  $t = t_f$  was reached, the simulation halted at  $t_{end}$  and a penalty was incurred.  $K$  is a weighting factor set equal to 100, which was used to ensure that  $G_{end}$  and the difference  $t_f - t_{end}$  were driven very close to zero upon convergence. The model was developed using Simulink SimMechanics (MathWorks, Inc.; Natick, MA, USA) and the optimization problem was solved using a combination of non-gradient based and gradient-based algorithms functions in MATLAB (MathWorks, Inc.; Natick, MA, USA). Several particle swarm optimization (PSO) (Kennedy and Eberhart, 1995) runs were performed first, with particles initially positioned randomly within the search space with re-seeding based on the results of previous runs. The best result from PSO was used as an initial guess in a constrained gradient-based optimization solved using the *fmincon()* function in Matlab. Following convergence of the gradient-based optimizer, the model successfully left the ground at  $t_{end}$ :  $G_{end}$  was  $2.2 \times 10^{-11}$  N and  $(t_f - t_{end})$  was  $4.1 \times 10^{-11}$  s.

The computer simulation results indicated that a bouncing strategy, and not a single countermovement, might be optimal for the jumping task (see below), and the experimental protocol was designed with this result in mind. To create a computer simulation that corresponded to trials in which subjects did not bounce, we performed an additional optimization in which bouncing was prevented by adding to Eq. 1 another term that penalized the number of sign changes in ankle angular velocity.

### **3.2.2 Experiments**

Eight healthy male subjects (age =  $24.1 \pm 3.2$  y, height =  $175.9 \pm 5.7$  cm; body mass =  $78.7 \pm 8.1$  kg) performed maximal-height jumps using only their ankles. The experimental protocol was approved by the Institutional Review Board of The Pennsylvania State University and informed consent was obtained prior to data collection.

Knee motions were restricted by universal knee immobilizers (Bledsoe Brace Systems, Grand Prairie, TX, USA) that were held in place by Velcro™ straps. Subjects were instructed to refrain from moving the rest of the body (hips, trunk, and head) with arms folded across the chest. Subjects wore platform shoes (JumpSoles; Metapro, Mountain View, CA, USA) (Figure 3-1), which permitted subjects to begin each jump with the heel above the floor, thus allowing an initial CM. Because simulation results suggested that a bouncing strategy might be optimal for the jumping task (see below), subjects were encouraged to explore different strategies for achieving the highest possible jump. Subjects were always instructed to jump as high as possible and first performed five trials with no further instruction followed by five trials with the additional instruction to move down initially, and finally five trials with the instruction to try “bouncing” on the toes before jumping up. Trials were excluded from further analysis whenever subjects left the floor briefly while bouncing before the final takeoff, or when subjects’ feet did not leave the floor at all. Only the highest bouncing and the highest non-bouncing trial for each subject were further analyzed.

Kinematic data were recorded using a six-camera Eagle System (Motion Analysis Corporation, Santa Rosa, CA, USA). Clusters of four reflective markers each were placed over the sacrum and on the shank and foot of the right leg (Figure 3-1). Marker cluster data along with the coordinates of markers placed over anatomical landmarks (collected during preliminary standing trials; these were markers on the tibial condyles, malleoli, heel, and head of 2<sup>nd</sup> metatarsal) were used to generate anatomically oriented segment-fixed coordinate systems for both the shank and foot. These two coordinate systems were aligned for quiet standing trials, which defined neutral ankle position (zero degrees, with the plantar surface of the foot perpendicular to the long axis of the shank), and used as reference for determination of plantarflexion and dorsiflexion angles in the sagittal plane. Jump height

was taken to be the difference in height for the sacrum cluster centroid between its peak and in the reference standing position.

### 3.3 Results

The optimal jumping strategy that resulted from the computer simulation consisted of an initial movement from 8° dorsiflexion to 22° plantarflexion followed by a bouncing motion featuring two additional plantarflexion peaks before takeoff (Figure 3-2, *top*). Optimal muscle activation patterns were characterized by alternating maximal or near-maximal plantarflexor and dorsiflexor activity (Figure 2, *bottom*). The jump height attained by the model was 12.8 cm, but was only 9.7 cm when bouncing behavior was penalized and bouncing did not occur (Table 3-2). The model jump height was calculated as the difference between the peak point mass height achieved during a jump and the height of the point mass at a nominal configuration of 0° ankle angle (1 m).

Four of the eight subjects jumped highest when they used a bouncing strategy similar to the strategy employed by the model (Figure 3-3). For these subjects (whom we called BOUNCERS), the final bounce frequency ( $f_{\text{BOUNCE}}$ ) was  $2.53 \pm 0.47$  Hz (Figure 3-3 and Table 3-2).  $f_{\text{BOUNCE}}$  was defined as the inverse of the time interval between the last two dorsiflexion peaks. “Bouncing” was defined to have occurred when there was a plantarflexion motion of at least a 10° between the last two dorsiflexion peaks. Those subjects whose highest jumps did not occur when bouncing (NON-BOUNCERS) had a  $f_{\text{BOUNCE}}$  of  $1.46 \pm 0.45$  Hz (Figure 3-3 and Table 3-2) when they tried bouncing.  $f_{\text{BOUNCE}}$  for the optimized computer simulation was found to be 2.78 Hz.

The average maximal jump height for all subjects was  $16.3 \pm 4.6$  cm (Table 3-2). For BOUNCERS the mean maximal jump height was  $18.7 \pm 4.5$  cm but for NON-BOUNCERS the mean was lower:  $13.9 \pm 3.3$  cm. This difference was not statistically significant, although it did trend towards significance ( $p = 0.10$ ). There were no significant differences in anthropometric measures (mass, height, lower leg length, foot length, calf circumference) between the two groups (all  $p > 0.47$ ).

Results for the computational model revealed that the energy stored in the plantarflexor tendon increased with successive bounces (Figure 3-4). Plantarflexor fiber length patterns showed an initial decrease followed by periods of constant length when tendon energy storage peaked (Figure 3-4).

The optimization used to determine muscle architectural properties produced tendon elastic modulus values ( $E_T$ ) for both muscles at the lower limit of the range suggested by Zajac (1989), 0.6 GPa. The sensitivity of simulation results to  $E_T$  was further investigated by increasing  $E_T$  by 50% (0.9 GPa) and rerunning the optimization. With this increased tendon  $E_T$  value, the model still produced the previously observed bouncing behavior. Maximal jump height with increased tendon stiffness was lower (12.0 cm) and  $f_{\text{BOUNCE}}$  was increased to 3.23 Hz.

### 3.4 Discussion

The goal of this study was to develop a computational model with which to investigate maximal-height single-joint jumping in which only the ankle muscles are used for propulsion. We hypothesized that the simulation would employ a countermovement to achieve maximal performance. We were surprised, however, to find that the optimized simulation performed multiple countermovements, or bounces, before leaving the ground. Although the subjects achieved their highest jumps using different strategies, the best jumpers recorded their maximal jump height when they employed a bouncing strategy similar to that employed in the simulation. The lower-jumping NON-BOUNCER subjects bounced with a lower frequency than did either BOUNCER subjects or the computer model. Model jump height was similar to those recorded experimentally, differing from the mean for the human subjects by less than one standard deviation. The bouncing frequency of the model compared well to that of the BOUNCER subjects and it was found that both the model and the BOUNCERS reduced their height of jump by a similar amount when a bouncing strategy was not allowed.

Our findings were novel in that (1) the simulation clearly identified a bouncing strategy as being optimal; and (2) the best performing subjects in our study achieved their

highest jumps when bouncing. Previous investigations of maximal-height multijoint jumping (Bobbert et al., 1996; Harman et al., 1990) and single-joint maximal pushoff tasks (Kawakami et al., 2002; Kubo et al., 2007) have found that a single CM enhances performance. While rhythmic bouncing on the toes has been considered in studies of movement dynamics (Takeshita et al., 2006), we have not found previous descriptions of bouncing as a strategy for optimizing performance in a discrete movement such as a maximal-height jump.

Simulation results were examined to investigate the mechanism by which bouncing enhances jumping performance. Previous authors have suggested that a single CM contributes to performance through facilitation of active state development (Bobbert and Casius, 2005; Kubo et al., 2007; van Ingen Schenau, 1984) and storage of elastic energy (Komi and Bosco, 1978; Kubo et al., 2007). We considered each of these possibilities to determine their influence on simulated jump height in our study.

Benefits due to active state development are related to muscle excitation-activation dynamics and the fact that it takes time for a muscle to develop force (Bobbert & Casius, 2005). A countermovement allows the muscle to be maximally activated as it starts to shorten during the final pushoff phase. The time necessary for active force development, however, does not appear to be the reason for bouncing in the present simulation. Full muscle activation, although not instantaneous, can be achieved after a single countermovement; the system does not require multiple bounces to reach maximal muscle activation during the final pushoff phase. Our results indicate that the muscles were maximally activated during each of the successive pushoff phases of bouncing (Figure 3-2). Active state development in our simulation was therefore unaffected by the use of a bouncing strategy.

Increased tendon energy storage is the more likely mechanism for increased performance due to bouncing in our simulation. During countermovements, energy is stored in the plantarflexor tendon and such energy stored in the tendon has the potential to contribute to raising the body during the final pushoff phase of the movement (Kawakami et al., 2002). Arakawa et al. (2010) have shown that for muscles with relatively long series elastic elements, like the plantarflexors, work output is enhanced to a much greater degree



by additional elastic energy utilization than by increasing the time available for active state development. Analysis of the energy stored in the tendon of the present model demonstrated that with each successive bounce the energy stored in the tendon increased. The model muscle fibers acted essentially isometrically during the final two bounces, likely enhancing muscle force production by reducing fiber shortening velocity to near zero. In a recent experimental study comparing single-CM movements with movements without a CM, Kawakami et al. (2002) showed that muscle fascicles contracting isometrically allow for increased power generation by the muscle-tendon complex, enabling the tendon to recoil during the shortening phase and enhancing the total work produced. Muscle fascicles acting isometrically have also been observed in activities such as walking (Fukunaga et al., 2001). It seems plausible that if a single CM increases energy storage and subsequent work production, multiple bounces should also be able to increase the final amount of muscle work produced during the final pushoff phase, and therefore increase jump height.

We performed additional analyses to test (1) whether summed gravitational and elastic potential energy was indeed maximal at the time of the last dorsiflexion peak in the bouncing simulation; and (2) whether differences in tendon energy storage could account for the nearly 3 cm difference in jump height between the simulation in which multiple bounces were permitted and the simulation in which only one bounce was permitted. It was found that the ankle angle, plantarflexor activation state, and plantarflexor fiber length that maximized the sum of the gravitational potential energy and the elastic potential energy stored in the tendon had values identical to those exhibited by the bouncing simulation at the initiation of the final plantarflexor phase before takeoff.

Comparisons of muscle and tendon work between the bouncing and non-bouncing simulations showed that the difference in jump heights between these two simulations could not be attributed to elastic energy storage directly. We found that the non-bouncing simulation did produce less tendon energy storage (9.8 J versus 38.9 J) but the plantarflexor muscle fibers did more work in raising the mass prior to takeoff (41.6 J versus 7.7 J). More negative work was done by the dorsiflexors (-12.0 J versus -1.4 J) in the non-bouncing case, however, and the resulting total difference in work explains the 3 cm difference in jump height almost exactly. The negative dorsiflexor work in the non-bouncing simulation

enhances plantarflexor force by actively braking plantarflexion to reduce plantarflexor fibers shortening velocity. We confirmed this conclusion with a non-bounce simulation in which the dorsiflexor were made passive during the early part of the final plantarflexion and the result was a much more rapid plantarflexion and a reduction in the work done by the plantarflexor actuator.

Another factor related to elastic energy storage that is likely to contribute to jump height among bouncers is mechanical resonance. The viscoelastic properties of the muscle-tendon unit make its behavior dependent on the frequency of movement (Bach et al., 1983; Dean and Kuo, 2011; Takeshita et al., 2006). Studies of rhythmic bouncing that have focused on the metabolic cost of movement (Bach et al., 1983; Dean and Kuo, 2011) have shown that when subjects bounce close to the resonant frequency most of the work to maintain the bouncing is performed passively. Takeshita and colleagues (2006) found that at the resonant frequency muscle fiber excursions are minimal compared with the length changes of the entire muscle tendon unit, placing the muscle fiber in a more favorable position to generate muscle force that tensions the tendon in a manner similar to that observed in the present model. Previous experimental results show the mean resonant frequencies for similar ankle joint movements to be 2.67 Hz (Takeshita et al., 2006), 3.07 Hz (Dean and Kuo, 2011), and 3.33 Hz (Bach et al., 1983). In our study, the mean frequencies of the final bounce ( $f_{\text{BOUNCE}}$ ) for the model and BOUNCERS were 2.78 Hz and 2.53 Hz respectively. These frequencies are similar to the resonant frequencies from the cited studies and suggest that the model and some of the subjects used resonance to increase performance. Making the plantarflexor tendon stiffer in the model should have increased the resonant frequency, and the optimization appeared to take advantage of this; when we increased tendon elastic modulus, bounce frequency increased to 3.2 Hz following optimization. The NON-BOUNCERS, who were unsuccessful at jumping highest by bouncing, bounced at only 1.46 Hz, perhaps indicating a failure to make effective use of resonance.

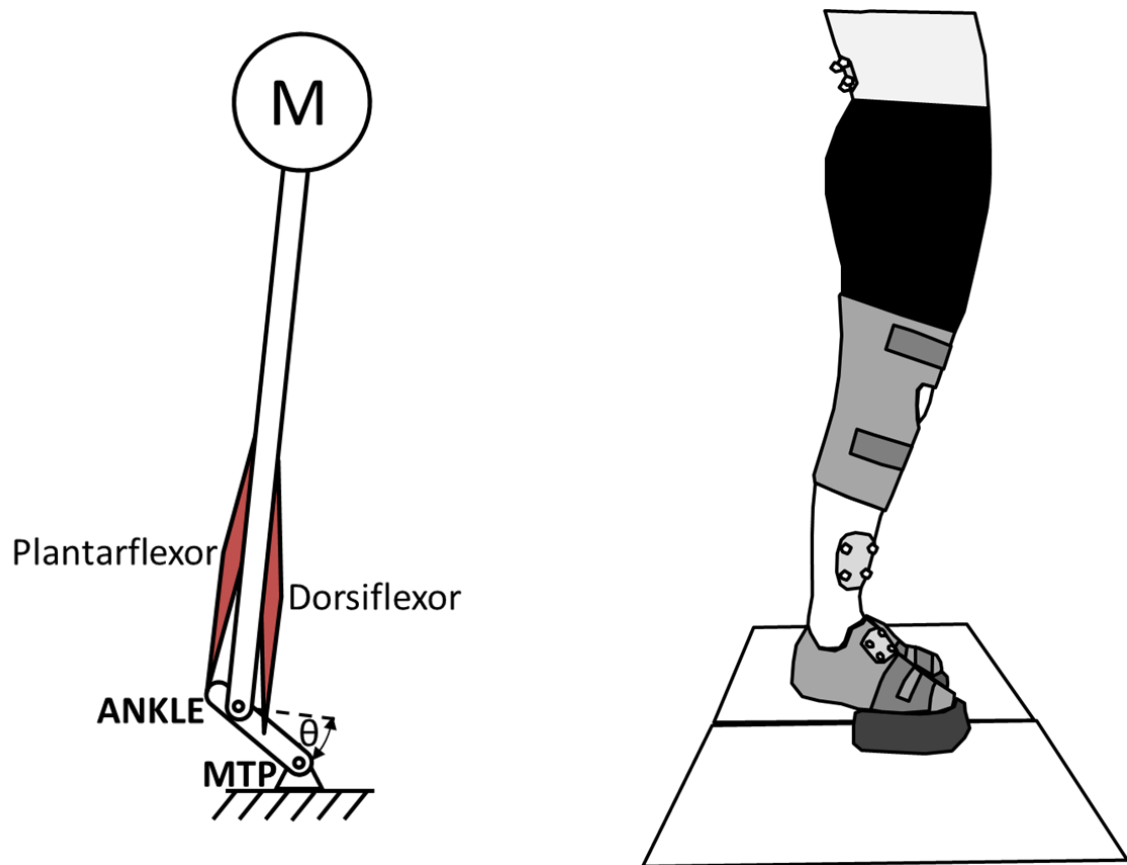
Certain limitations affected our computer simulation. Employing only one effective plantarflexor and one dorsiflexor, each of which crossed a single joint, represents a substantial simplification of a complex system. Further, in order to focus on ankle joint

function in only the sagittal plane, the model was made planar. During experimental data collection it was assumed that subjects' feet and bodies above the ankle were rigid bodies. Subjects wore knee-immobilizing braces and were asked not to bend at the hips or move their heads or arms, but such movements are likely to have occurred despite these efforts. These additional movements may explain why the mean jump height for subjects was higher than the jump height found for the model. Neural factors that may influence performance during a countermovement like task, such as the stretch reflex or force potentiation may explain some of the disparities we observed between model and experiment, but were not considered in our study. Kurokawa et al. (2003), however, did not find medial gastrocnemius stretch during a countermovement jump, suggesting that the stretch reflex and force potentiation did not contribute to the work done by the muscle-tendon unit. Unfortunately, the difference in frequency of jumping between the two groups was only discovered after the jumping data were analyzed and we were unable to ask the NON-BOUNCER subjects to jump at a higher frequency to see whether they would have jumped as high as the BOUNCERS.

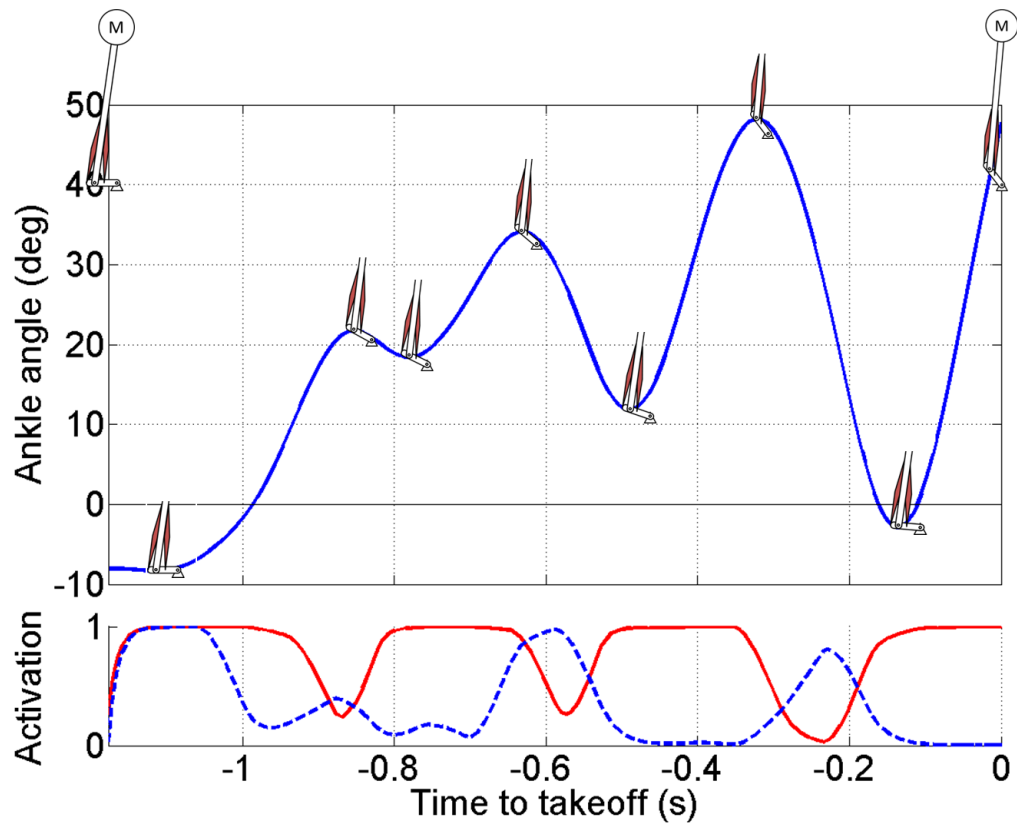
Maximal-height, single-joint jumping is a task not performed in everyday life. This unusual paradigm was chosen to gain insight into single joint mechanics and control without accounting for factors such as interjoint coordination and biarticular muscle function that would be relevant to a more complex movement. Whether conclusions drawn about human movement from the study of a simple task like single-joint jumping are transferrable to other tasks should be the subject of further investigation.

In conclusion, results from a computational simulation and from human subjects experiments suggest that a bouncing strategy is optimal for maximal-height single-joint jumping in which only the ankle joint muscles are used for propulsion. The increase in elastic energy storage that occurred during successive bounces and the selection of a bounce frequency that takes advantage of mechanical resonance are the most likely mechanisms for this performance enhancement. Although this study and its results were limited to a single-joint system, it is conceivable that the human body may also take advantage of the resonant properties of muscles while performing motor tasks employing multiple muscles and spanning multiple joints. We intend to use the current computational

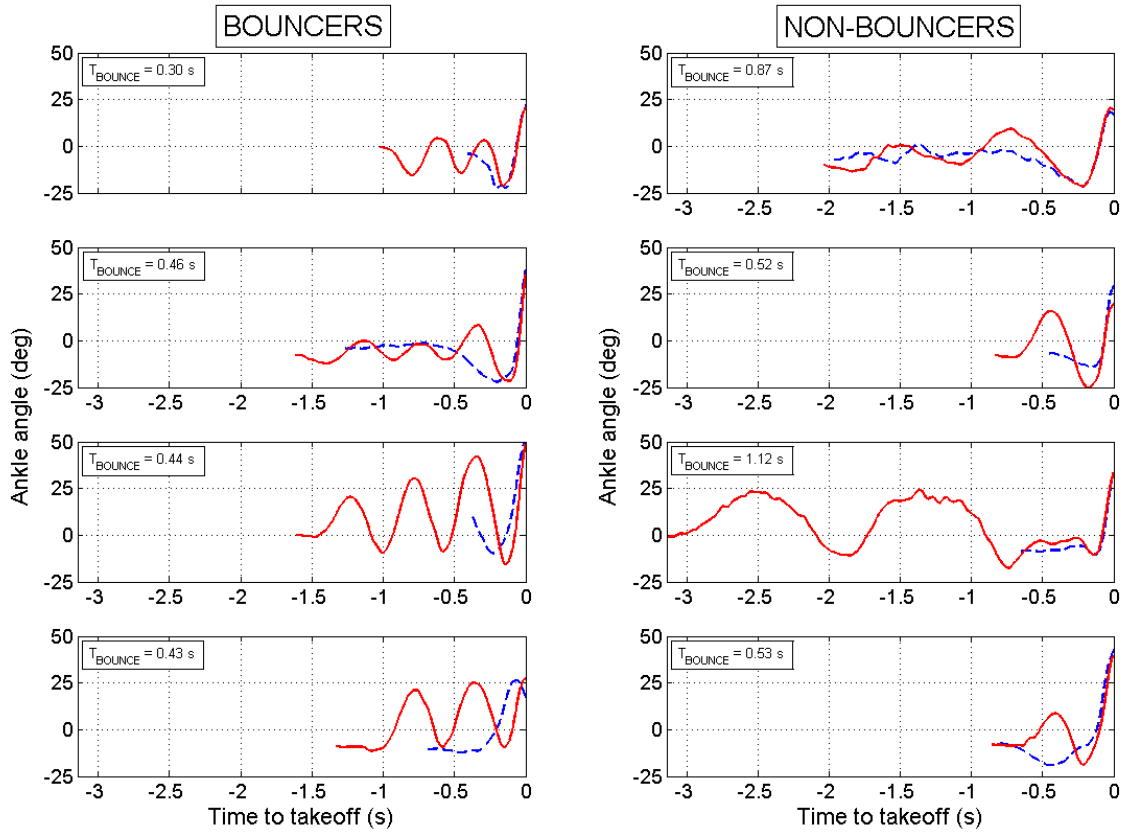
model in future work to investigate the effects of varied muscle and joint properties on maximal-height jumping and other motor tasks.



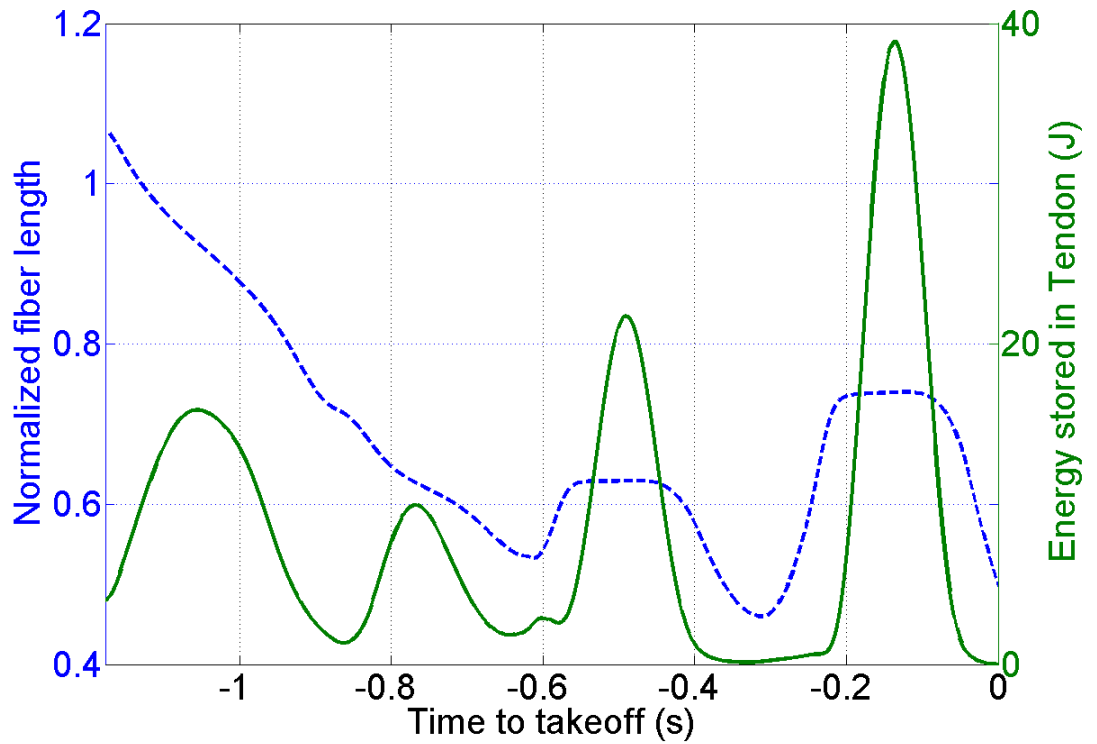
**Figure 3-1:** The two-degree-of-freedom planar model used in the computer simulation (*left*). Two muscles, a lumped plantarflexor and dorsiflexor, cross the ankle joint. Contact between the heel and floor was not modeled, allowing a countermovement to occur. The ankle angle  $\theta$  was defined as shown. During experiments (*right*) subjects wore platform shoes that allowed countermovement and braces to limit knee motion. Clusters of reflective markers were affixed to the foot, lower leg, and sacrum.



**Figure 3-2:** *Top:* Ankle angle plotted versus time to takeoff for the simulation (positive angles correspond to plantarflexion). The model starts in 8° dorsiflexion and then executes successive bounces until lift off occurs 1.18 s later. Insets show the model configuration at each ankle angle peak. *Bottom:* Muscle activations following optimization for the plantarflexor (*solid line*) and dorsiflexor (*dashed line*) musculotendon actuators.



**Figure 3-3:** Ankle angle plotted versus time to takeoff for all eight subjects (positive angles correspond to plantarflexion). Only the best bouncing trial (*solid line*) and the best non-bouncing trial (*dashed line*) are shown for each subject. In the left column are ankle angles for the BOUNCERS, subjects whose maximal-height jump occurred when they bounced, and angles for NON-BOUNCERS are found in the right column. The bounce period,  $T_{\text{BOUNCE}}$ , for each subject's best bouncing trial is given in each subplot.



**Figure 3-4:** Plantarflexor fiber length normalized by optimal fiber length (*dashed line*) and plantarflexor tendon elastic energy (*solid line*) plotted versus time to takeoff for the simulation. Tendon energy increased with each successive bounce and muscle fibers operated nearly isometrically when tendon energy peaked.



**Table 3-1:** Computational model muscle architecture properties derived through parameter optimization, with bounds used in the optimization given in parentheses.

	Units	Lumped Plantarflexor	Lumped Dorsiflexor
Max isometric force ( $F_{MAX}$ )	N	5451 (3157 – 6577)	1254 (664 – 1254)
Tendon elastic modulus ( $E_T$ )	GPa	0.6 (0.6 – 1.7)	0.6 (0.6 – 1.7)
Muscle insertion point ( $M_{INSERT}$ )	cm	3.5 (3.5 – 5.3)	4.6 (3.2 – 4.8)
Optimal fiber length ( $L_{Fopt}$ )	cm	4.4 (1.0 – 45.0)	12.6 (1.0 – 45.0)
Tendon slack length ( $L_{TS}$ )	cm	35.2 (1.0 – 45.0)	30.3 (1.0 – 45.0)

Notes:

$F_{MAX}$  bounds were between the maximum isometric force of all ankle joint muscles combined and the maximum of the most powerful single-joint muscle as prescribed by (Arnold et al., 2010).

E bounds as suggested by Zajac (1989).

For  $M_{INSERT}$  the range was set by using 80% and 120% of the moment arm values of the strongest plantarflexor and dorsiflexor muscles from Arnold et al. (2010)

Bounds on  $L_{Fopt}$  and  $L_{TS}$  were set to extremes to allow these results to be actively constrained by the ratio  $L_{TS} : L_{Fopt}$  (Zajac, 1989) with  $L_{TS} : L_{Fopt}$  of  $\pm 20\%$  of gastrocnemius and soleus average of 10 (8.0 to 12.0) and  $L_{TS} : L_{Fopt}$  of  $\pm 20\%$  of dorsiflexor value of 3 (2.4 to 3.6).

**Table 3-2:** Comparison of model results with maximal jumps of subjects. Means are given with standard deviations in parentheses.

	Units	Model	Subjects		
			All	BOUNCERS	NON-BOUNCERS
Overall maximum height	cm	12.8	16.3 (4.6)	18.7 (4.5)	13.9 (3.3)
Bounce maximum height	cm	12.8	15.7 (4.8)	18.7 (4.5)	12.7 (2.9)
Non-bounce max. height	cm	9.7	14.8 (3.8)	15.7 (4.0)	13.9 (3.3)
T <sub>BOUNCE</sub>	s	0.36	0.58 (0.25)	0.41 (0.06)	0.76 (0.25)

## **Chapter 4**

### **Foot and Ankle Structural Measures are Correlated with Performance in a Maximal-Height Single-Joint Jumping Task**

#### **4.1 Introduction**

Evidence from studies of animals as well as of humans suggests that variability in foot and ankle structure both across species and within species has the potential to influence function. For example, McGowan et al. (2008) found calcaneus and metatarsal lengths to differ between similar wallaby species that inhabit different environments, suggesting that these structural differences reflect the varying demands of the environment. Similar structural differences have been found in primates living in different habitats (Goto and Kumakura, 2013). Several recent studies have pointed to the potential functional significance of the foot and ankle structure in humans. It has been demonstrated that sprinters have smaller Achilles tendon moment arms as well as longer toes than non-sprinters (Baxter et al., 2012; Lee and Piazza, 2009) and that volleyball players jump higher and have smaller plantarflexor moment arms than controls, presumably non-athletes (Watanabe et al., 2008). Studies of distance runners have revealed a correlation between heel length and running economy (Raichlen et al., 2011; Scholz et al., 2008). Toe length has been found to correlate with toe flexor work in running humans, perhaps signifying that shorter toes reduce the metabolic cost of force generation (Rolian et al., 2009). These studies suggest that interindividual variation in human foot and ankle structure allows for function that is adapted to various motor tasks.

Understanding the relationship between foot and ankle structure and motor function has important implications. Variability in structure and associated function occurring across humans has the potential to influence the ability of certain individuals to perform at an elite level, while leading others to struggle with basic mobility and activities of daily living. A better understanding of how structural variability influences functional capacity will improve our ability to assess individual capacity to perform a given task and potentially will allow us to better understand what is required to help improve performance.

In addition, differences in foot and ankle structure between modern humans and human ancestors can tell us about the evolution of functional capacity if we understand better these structure-function relationships. It has been suggested that the shorter toes of modern humans relative to those of other hominids reduces mechanical work during running (Rolian et al., 2009) and that our shorter heels compared with those of Neanderthals allows for better tendon energy storage and reduced metabolic cost during running (Raichlen et al., 2011). Both of these findings suggest human evolution for endurance running.

It is not clear whether correlations similar to those found between foot and ankle structure and running performance would also be present for other motor tasks. One such task of interest is maximal-height jumping, an activity with a clear objective making it well suited for an experimental research paradigm. Watanabe et al. (2008) have found differences in ankle structure between volleyball players and controls, but to our knowledge the effect of variation in foot and ankle structure on jump performance has not been studied using correlational measures. Because muscular joint moment is the product of muscle force and muscle moment arm, it might be expected that individuals with longer heels (*i.e.* larger plantarflexor moment arms) would be better able to produce ankle joint moments that raise the body's center of mass during a maximal-height jump. However, experimental results from Watanabe et al. (2008), as well as results from recent musculoskeletal computer simulations suggest a different hypothesis: Large moment arms may reduce plantarflexor force generating capacity during rapid joint movements (Baxter et al., 2012; Nagano and Komura, 2003). Nagano and Komura (2003) computed plantarflexor kinetics subject to varied joint angular velocity using models with different plantarflexor muscle moment arms. A larger moment arm caused the muscle to shorten more in a given time, resulting in a higher shortening velocity that reduced muscle force generating capacity due to the force-velocity relation. A smaller muscle moment arm had the opposite effect, reducing muscle shortening velocity and increasing muscle force and power output, overcoming the simultaneous deficiency in leverage that was created.

It is also not clear what effect toe length will have on jumping performance. Longer toes, in increasing the length of the foot, may compromise the mechanical advantage of the plantarflexors in lifting the body. Conversely, it has been suggested that the longer toes of

sprinters may confer an advantage by increasing ground contact time and thus increasing propulsive impulse when a sprinter accelerates at the start of a race (Lee and Piazza, 2009). Longer toes may offer a similar benefit during maximal-height jumping: Longer toes may prolong contact with the ground, enhancing vertical impulse and leading to higher jumps.

The purpose of this study was to test for correlations between foot and ankle structural measures and performance measured during a single-joint maximal-height jumping task. Using a single-joint movement framework allows for a focused investigation into the mechanisms of optimal performance and has been used previously to explain movement behaviors of interest at the ankle joint (*e.g.* Kawakami et al., 2002; van Werkhoven and Piazza, 2013). We hypothesized that single-joint jump height would correlate negatively with heel length and that single-joint jump height would correlate positively with toe length.

## **4.2 Methods**

Ten healthy male subjects (age:  $23.8 \pm 3.1$  y; height:  $176.5 \pm 5.5$  cm; body mass:  $77.9 \pm 7.8$  kg) performed maximal-height static jumps using only their ankles for propulsion. The experimental protocol was approved by the Institutional Review Board of The Pennsylvania State University and informed consent was obtained prior to data collection.

To limit motion at joints other than the ankle, subjects wore universal knee immobilizers (Bledsoe Brace Systems, Grand Prairie, TX, USA) that were held in place by Velcro™ straps to restrict motion. Further, subjects were asked not to move the rest of the body (hips, trunk, and head) and also hold their arms folded across the chest. When atypical motion at joints other than the ankle was noticed by the experimenter, the subject was asked to repeat the trial. Subjects wore platform shoes (JumpSoles; Metapro, Mountain View, CA, USA) with a wooden block below the heel to prevent a countermovement (Figure 4-1). Subjects were instructed to jump as high as possible using a single upward movement (*i.e.* no countermovement or bouncing was allowed). Each subject performed five trials and the average jump height for each subject over these trials was used for further analysis. The

jump heights measured for trials where subjects were unable to lift off the ground (one subject, three trials) were included in the average jump height calculations.

The coordinates of reflective markers were recorded at 100 Hz using a six-camera Eagle System (Motion Analysis Corporation, Santa Rosa, CA, USA) to track the locations of markers, from which jump height and ankle joint angles were subsequently computed. Marker position data were low-pass filtered at 15 Hz using a 4<sup>th</sup> order Butterworth filter. Marker clusters consisting of four reflective markers each were placed over the sacrum and on the shank and foot of the right leg (Figure 4-1). Additional markers were placed on the tibial condyles, medial and lateral malleoli, heel, and head of 2nd metatarsal in order to establish anatomically oriented segment-fixed coordinate systems for the right shank and right foot. These markers were removed after an initial static calibration trial, leaving only the cluster markers during dynamic trials. The anatomical coordinate systems attached to the shank and foot were assumed to be aligned for quiet standing trials, meaning that neutral ankle position was defined by the plantar surface of the foot being perpendicular to the long axis of the shank. The ankle plantarflexion-dorsiflexion angle was computed by Euler angle decomposition of the rotation of the foot coordinate system with respect to that of the shank. In this decomposition, the first rotation (the sagittal-plane ankle angle of interest) occurred about the shared flexion axis, followed by rotations about the anteroposterior and superior-inferior axes attached to the foot. Jump height was calculated as the difference in the vertical position of the sacrum cluster centroid between its peak value and its value in the reference standing position.

Several anthropometric measures were recorded for each subject, including body stature and body mass as well as measures specific to the lower leg. These leg measures included: lower leg length (from the lateral tibial plateau to the floor surface in a standing posture); maximum lower leg circumference; foot length; hallux length (distance from first metatarsal head to distal end of the hallux); lateral heel length (horizontal distance from lateral malleolus to back of the heel); and medial heel length (horizontal distance from medial malleolus to back of the heel).

Lateral and medial heel lengths were measured from digital photographs taken of each subject's right foot while the subject was seated and the foot placed on a wooden

reference block to which a millimeter-scale measuring tape was attached. Before taking photographs, the subject's anterior tibia was aligned vertically and an x-mark was drawn on the tips of the lateral and medial malleoli with a black marker. Subsequently, points were digitized on the reference measuring tape and markings on the tips of the malleoli and the posterior aspect of the ankle at the same height as the malleoli. Custom written Matlab routines (The Mathworks, Inc., Natick, MA) were written to apply calibration, correct for parallax, and calculate actual lateral and medial heel lengths. As the foot "gear ratio" has often been cited as a potential determinant of the ability of humans and animals to perform certain tasks (*e.g.* Hildebrand, 1960), we also computed this quantity. Gear ratio was defined as the ratio of the distance from ankle to toe to the distance from ankle to heel. The lateral heel length was used as the ankle-to-heel length measurement and ankle-to-toe length was estimated by subtracting the ankle-to-heel length from the total foot length. Bivariate correlation analyses were performed between all anthropometric variables and the average jump height for each subject.

To test for the possibility that a longer toe or a longer heel is simply an indication of a larger foot or larger stature, a factor analysis was performed to investigate whether some underlying feature related to body stature could explain jumping performance. A factor analysis with principal component extraction (Kim and Mueller, 1978) was carried out that included all measured anthropometric values except mean heel length, which was excluded due to its clear relationship with both medial and lateral heel length. After the factor analysis was performed, only factors with eigenvalues larger than unity were retained. An orthogonal rotation (varimax) was implemented in order to have variables maximally weighted onto single factors.

### **4.3 Results**

Significant positive correlations were found between lateral heel length and jump height, between hallux length and jump height, as well as between mean heel length and jump height (Table 4-1; Figures 4-2 and 4-3). There was also a significant negative correlation between gear ratio of the foot and jump height (Table 4-1; Figure 4-4).

None of the other anthropometric variables measured were found to correlate significantly with jump height. Taller subjects did not necessarily jump higher and there was no significant correlation between stature and jump height (Table 4-1, Figure 4-5).

The number of retained factors (eigenvalues larger than unity) for the factor analysis was three. Results of the factor analysis after varimax rotation are shown in Table 4-2. Using 0.5 as a cutoff for weighting on a given factor it is evident that height, lower leg length and foot length and mass to a lesser extent loaded onto factor one. Mass, calf circumference and medial heel length loaded on factor two; and hallux length and heel length loaded onto factor three.

#### **4.4 Discussion**

The aim of this study was to test for correlations between performance in a maximal height single-joint jumping task and foot and ankle anthropometric characteristics. Our first hypothesis stating that maximal single-joint jump height would be negatively correlated with heel length was not supported. Experimental results showed the opposite to be the case: The best jumpers had longer lateral heel lengths. However, single-joint jump height was found to positively correlate with hallux length, supporting our second hypothesis. We also considered gear ratio as a possible anthropometric based indicator of jumping ability and, contrary to previous findings that animal jumpers possess higher gear ratios, we found a negative correlation between jump height and gear ratio in our human subjects. None of the other anthropometric variables were correlated with jump height. Finally, the factor analysis indicated that the anthropometric variables considered could potentially be distinguished into three underlying factors.

Our finding that humans with longer moment arms jumped higher seems at odds with previous investigations showing that sprinters have smaller plantarflexor moment arms (Baxter et al., 2012; Lee & Piazza, 2009). Due to the fact that both jumping and sprinting require rapid energy generation by the plantarflexors, we expected that the jumpers in our study would also benefit from having shorter plantarflexor moment arms or shorter heels. Watanabe et al. (2008) also found that volleyball players jumped higher and



have smaller plantar flexor moment arms than controls and experimental studies comparing trained sprinters with non-sprinter controls have also shown that sprinters have shorter plantarflexor moment arms. Simulation studies have proposed a potential benefit for smaller moment arms: During fast movements a smaller heel length reduces the speed of shortening of the muscles and allows the muscles to produce forces higher than those that would be generated with a longer heel, due to force-velocity effects (Baxter et al., 2012; Nagano and Komura, 2003). Examination of gear ratios in our subjects also revealed the opposite of what was expected: Gear ratio was negatively correlated with jump height. Studies of comparative anatomy have shown that animals specialized for sprinting and jumping have larger gear ratios (ratio of the distance from ankle to toe to the distance from ankle to heel) (*e.g.* Goto and Kumakura, 2013; Hildebrand, 1960). A larger gear ratio should allow faster velocities at the foot endpoint (toe) for a similar angular rotation of the ankle, which might be seen as desirable for both sprinters and jumpers.

The unexpected positive correlation between jump height and plantarflexor moment arm and the unexpected negative correlation between jump height and gear ratio may be attributable to the fact that our subjects were not trained athletes. Studying a group of non-elite athletes, Blazeovich et al. (2009) found that larger knee extension moments were produced by subjects with larger knee extensor moment arms regardless of the rate of knee rotation. It has been suggested that the best animal jumpers have larger gear ratios that allow for larger range of motion over which to accelerate, requiring smaller peak forces to cover a certain distance. This is compared with so-called generalists with smaller gear ratios, who need to produce higher forces over a smaller joint range of motion to cover the same distance (Demes et al., 1999; Goto and Kumakura, 2013). It is possible that elite human athletes undergo specialized adaptations that differentiate them from the average individual. Athletes' training or their genetic characteristics may give them smaller plantarflexor moment arms that enhance force production during rapid plantarflexion. Individuals who are not trained athletes, such as the subjects in the present study, might be seen as generalists who make use of their musculoskeletal structure in a different manner.

Subjects with longer toes (hallux length) jumped higher in the current study. In a similar explosive movement type of activity, it has been shown that sprinters have longer

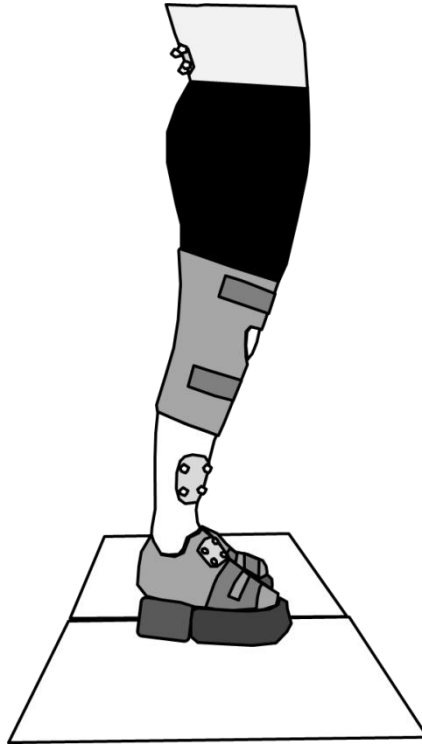
toes compared with control subjects (Baxter et al., 2012). It was suggested that longer toes potentially allow the foot to stay on the ground longer and thereby increase the impulse generated. Since the subjects in this study wore platform shoes with blocks underneath the heels, it was not considered to measure movement contact time during this protocol. The use of the special shoes and blocks could potentially have affected the relative toe movement of our subjects.

Factor analysis indicated that hallux length and heel length are separate anthropometric variables, neither of which is strongly associated with body stature. The variables aligning with the first retained factor of our factor analysis were body height, lower leg length and foot length. These can be classified as general stature measures. The second factor contained the following variables: body mass, calf circumference and horizontal distance from the medial malleolus to the back of the heel. Body mass and calf circumference are both measures of the size of a person. It was interesting, however, that horizontal distance from the medial malleolus to the back of the heel weighted with the body size factor more than with the associated horizontal distance from the lateral malleolus to the back of the heel, which was our proxy for heel length and associated plantarflexor moment arm. It is not clear why this is the case. The third factor contained the measures of hallux length and heel length, indicating that these variables could be considered as distinct from body size measures and that it cannot be generally said that subjects with larger toes and larger heels are simply those individuals with greater body size.

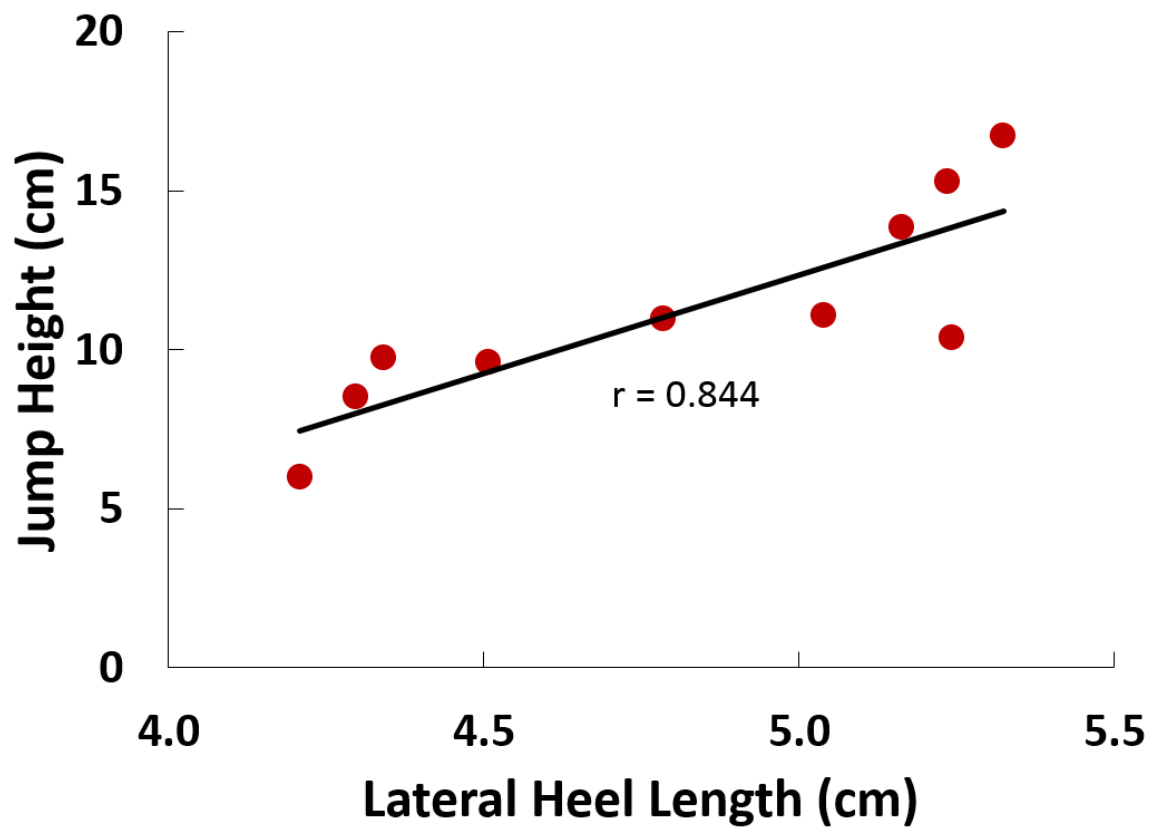
Our study had certain limitations. Although the subjects' knees were immobilized using knee braces, small motions may have occurred at the knee and other joints in the body. The subjects wore special platform shoes and blocks below the heels as this study formed part of a larger study design. These modifications could have potentially affected the subjects' movement patterns and caused them to move differently from what they would while performing the movement barefoot or wearing regular shoes. Jump height was measured as the vertical displacement of markers placed on the sacrum. Calculating jump height using other means (*e.g.* flight time method, vertical impulse method, displacement of estimated center of mass) could potentially lead to different results. An external measure of the lateral heel length is only a proxy for the plantarflexor moment arm. Several previous

investigators have used similar measures to identify correlations between moment arm and performance in a given task (*e.g.* Raichlen et al., 2011; Scholz et al., 2008). In those studies, a mean heel length measure (average of the medial and lateral heel lengths) was used. For this study both medial and lateral heel length was measured and average heel length was calculated. A stronger correlation was found between lateral heel length and jump height than either average heel length or medial heel length. Statistical analysis using PCA for factor identification has historically been used with large sample sizes, although several authors have recently suggested that this technique could potentially be used with much smaller samples as is encountered in this study (Mundfrom et al., 2005; de Winter et al., 2009)

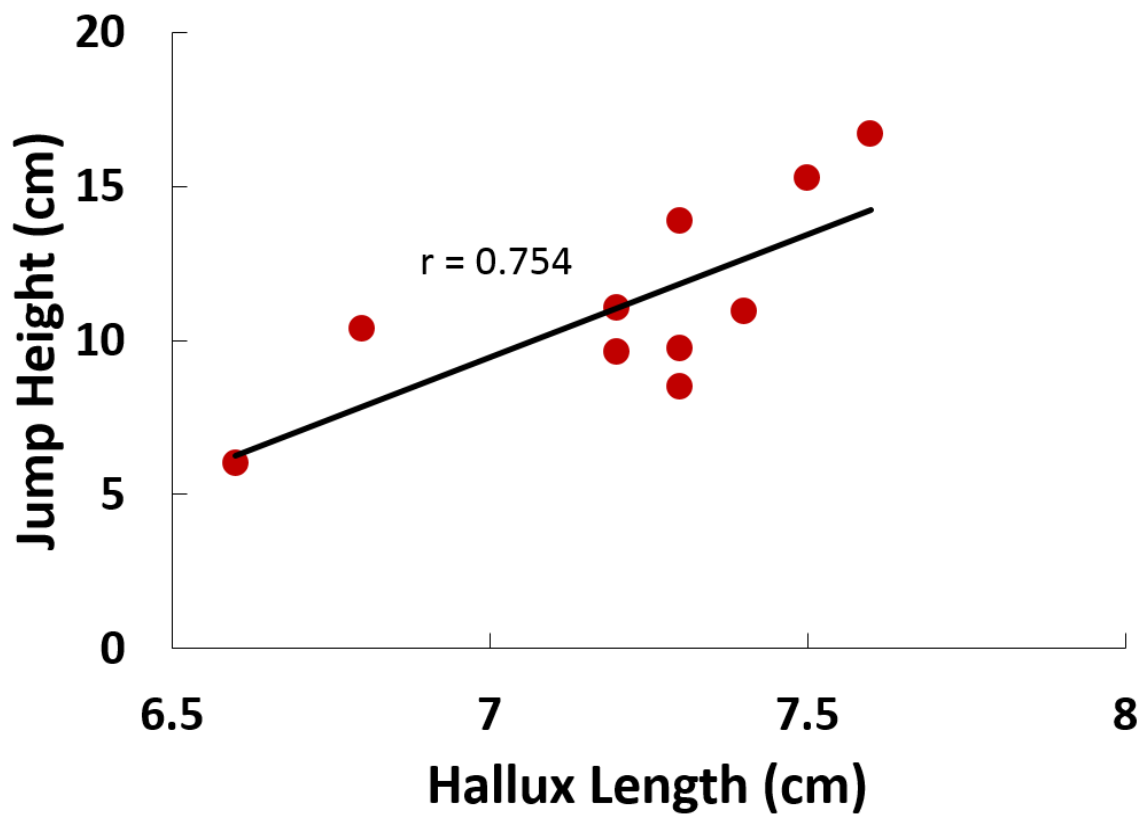
In conclusion, results from this study show that subjects with longer toes and longer heels perform better in a maximal height single-joint jumping task. To our knowledge, this is the first study to find significant correlations between foot and ankle anthropometric measures and jump height. The suggestion that different populations (*e.g.* elite athletes vs. non-athletes) might have different optimal foot and ankle structures allowing for maximal performance in the same task should be further explored in future research. This concept could have significant implications on how optimal musculoskeletal structure is defined. The use of musculoskeletal modeling might be helpful in understanding the mechanisms responsible for maximal performance given a certain anthropometric structure of the foot and ankle.



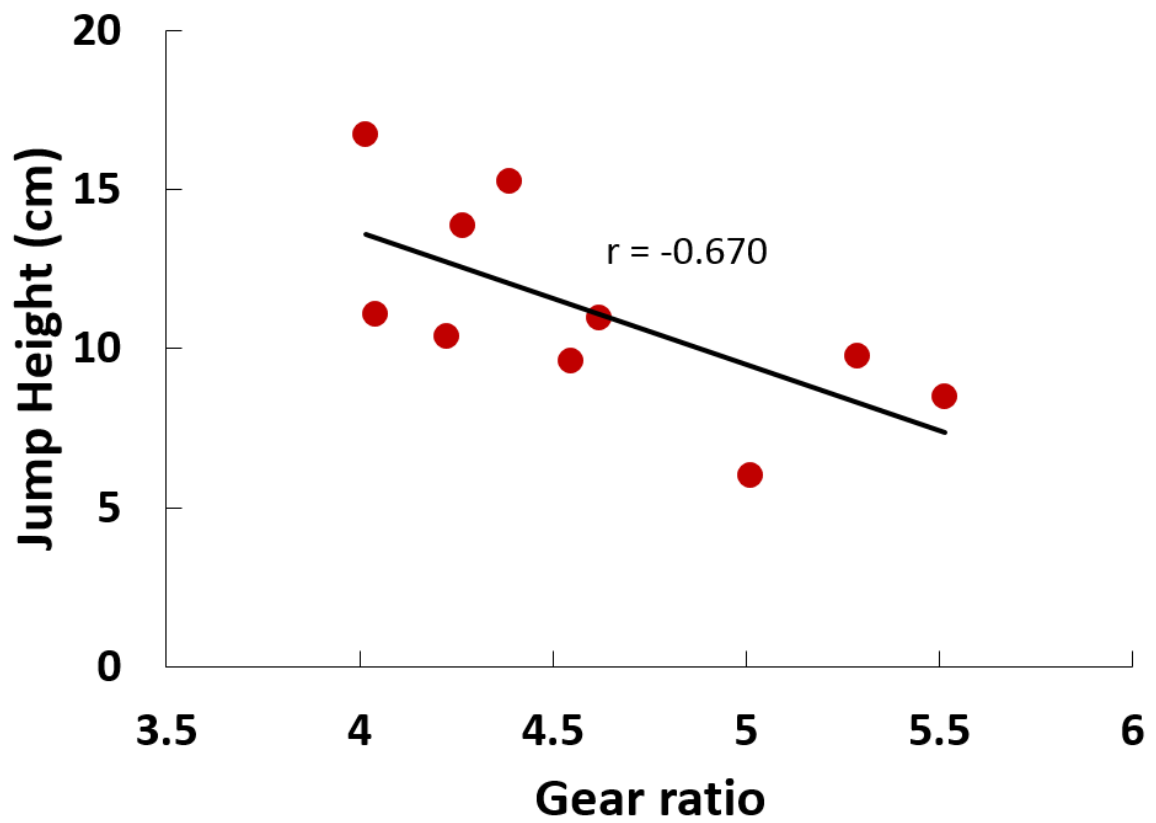
**Figure 4-1:** Experimental setup showing a subject wearing knee braces and platform shoes with a wooden block affixed below the heel of the shoe. Clusters of reflective markers affixed to the foot, shank, and sacrum.



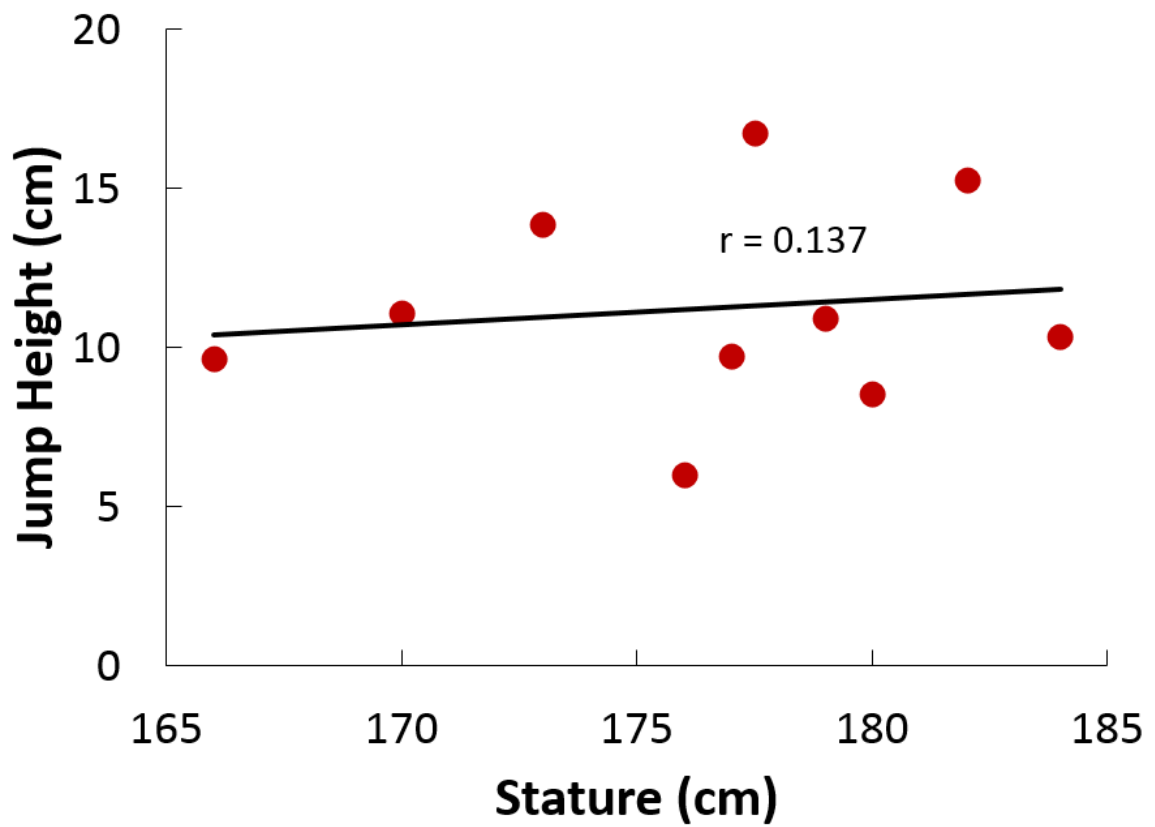
**Figure 4-2:** Jump height plotted against lateral heel length for all participants. Jump height was found to correlate significantly with lateral heel length ( $p = 0.002$ ).



**Figure 4-3:** Jump height plotted against hallux length for all participants. Jump height was found to correlate significantly with hallux length ( $p = 0.012$ )



**Figure 4-4:** Jump height plotted against gear ratio for all participants. Jump height was found to correlate significantly with gear ratio ( $p = 0.034$ ).



**Figure 4-5:** Jump height plotted against stature for all participants. Jump height was found not to correlate significantly with stature ( $p = 0.706$ ).



**Table 4-1:** Correlations between jump height and anthropometric variables; jump height and gear ratio.

Anthropometric variables	r	p-value
Stature	0.137	0.706
Mass	0.018	0.961
Lower Leg Length	0.383	0.274
Calf Circumference	0.294	0.409
Foot length	0.382	0.276
Medial Heel Length	0.475	0.165
Lateral Heel Length	<b>0.844</b>	<b>0.002</b>
Mean Heel Length*	<b>0.717</b>	<b>0.020</b>
Hallux Length	<b>0.754</b>	<b>0.012</b>
Gear ratio	<b>-0.670</b>	<b>0.034</b>

\* Mean heel length is the average of lateral and medial heel lengths

**Table 4-2:** Factor analysis results showing the three retained factors after varimax rotation. Variable weightings of larger than 0.5 are shown in bold.

	Component		
	<b>1</b>	<b>2</b>	<b>3</b>
Stature	<b>0.981</b>	0.041	-0.093
Mass	<b>0.572</b>	<b>0.679</b>	-0.345
Lower Leg Length	<b>0.935</b>	-0.001	0.259
Calf Circumference	0.028	<b>0.933</b>	0.062
Foot length	<b>0.868</b>	0.169	0.283
Medial Heel Length	0.010	<b>0.967</b>	0.172
Lateral Heel Length	0.215	0.464	<b>0.630</b>
Hallux Length	0.096	-0.042	<b>0.909</b>

## **Chapter 5**

### **Shorter Heels are Accompanied by Higher Achilles Tendon Forces, but Not by Better Running Economy**

#### **5.1 Introduction**

The ankle joint and associated foot structure plays an important role in many tasks of daily activity. The ankle joint generates a greater share of the muscle power during running than do either the hip or the knee joint (Farris & Sawicki, 2011; Winter, 1983). It has also been shown that ankle joint kinetics are affected more severely during old age, with joint power reduced more at the ankle than at any other joint in the lower extremity (Winter et al., 1990). It is however not only the structure of the ankle joint that influences the dynamics at the ankle joint. The foot interacts with the ground during over ground activities and this foot-ground interaction plays a large part in determining ankle joint dynamics. Perhaps it is because the foot and ankle play such an important role in locomotor function that these structures and their relation to locomotion have been the topic of many recent investigations (*e.g.* Baxter et al., 2012; Perl et al., 2012; Scholz et al., 2008). Several of these studies have found correlations between foot and ankle structure and performance, while others have demonstrated differences between elite performers and controls for explosive types of activities. For example, it has been shown that sprinters have smaller plantarflexor moment arms and longer toes compared with non-sprinters (Baxter et al., 2012; Lee and Piazza, 2009) and volleyball players have smaller plantarflexor moment arms compared with controls (Watanabe et al., 2008). Significant correlations have also been found between maximal jump height and heel length and as well as maximal jump height and toe length in a single-joint jumping task (see Chapter 4).

In distance runners, similar functional differences between runners of varying ability have been found or suggested. Watanabe et al. (2008) have found distance runners to have smaller plantarflexor moments arms compared with controls and two separate groups have reported negative correlations between running economy and heel length (Raichlen et al., 2011; Scholz et al., 2008). The investigators (Raichlen et al., 2011; Scholz

et al., 2008) measured metabolic cost in highly trained runners while they ran on a treadmill at 16 km/h and found that the mass-specific metabolic cost of running correlated significantly with heel length. Scholz and colleagues (2008) suggested a theoretical explanation for their findings: To produce a given joint moment, a smaller moment arm requires larger muscle force. Larger muscle force should increase tendon energy storage and the subsequent return of energy, which may save metabolic energy. While Scholz et al. (2008) and Raichlen et al. (2011) have reported correlations between heel length and running economy, the dynamics of this proposed mechanism have not been tested experimentally. It is not known whether smaller moment arms do in fact lead to larger muscle forces and whether larger muscle forces in turn lead to higher tendon energy storage. Additionally, if larger muscle forces do increase energy storage in the tendon, it might also be possible for runners to dynamically change their foot strike patterns in order to increase tendon energy storage. By changing foot strike patterns in this way, runners might increase muscle forces for a given moment arm and therefore influence energy storage in the tendon and metabolic energy expenditure. The recent findings of Perl et al. (2012) may be evidence of such a relationship: They found that forefoot strikers have larger Achilles tendon forces than rear foot strikers. This increased force, however, did not correspond to lower energy consumption for forefoot strikers.

The length of the toes has also been suggested as a structural characteristic that may influence function in distance running (Rolian et al., 2009). It has been proposed that shorter toes may reduce the metabolic cost of running by decreasing the mechanical work required at the metatarsophalangeal joint. Rolian et al. (2009) documented reduced metatarsophalangeal joint work for individuals with smaller toes, but these work differences were not linked to running economy.

Despite this recent attention in the literature, it remains unclear whether the reported correlations between heel length and running economy are associated with higher forces in the Achilles tendon, and whether the length of the toes does in fact affect running economy. The purpose of this study was to investigate further how ankle and foot structure affect running economy and to elucidate the underlying mechanisms for such effects. We first aimed to replicate the findings of Scholz et al. (2008) and Raichlen et al. (2011) that shorter

heels are accompanied by better running economy. We sought to investigate the dynamics of the foot and ankle to determine if runners with smaller heels would be found to have similar plantarflexor moments and thus also have larger Achilles tendon forces. We also wanted to investigate whether more economical runners have different foot strike patterns that might affect tendon forces and therefore energy storage. Finally we wanted to study the effect of toe length on metabolic cost. Based on previous findings, we hypothesized that there would be a positive correlation between moment arm (as estimated by heel length) and energy consumption. Following the theory proposed by Scholz et al. (2008), we hypothesized that shorter-heeled runners would have higher estimated peak Achilles tendon forces. We also hypothesized that runners with shorter toes would be more economical as they would have shorter levers to work against as they push off the ground in terminal stance.

## **5.2 Methods**

Fifteen healthy male distance runners (age:  $22.5 \pm 3.1$  y, height:  $176.0 \pm 7.3$  cm; body mass:  $67.6 \pm 7.7$  kg) were recruited for this study. The experimental protocol was approved by the Institutional Review Board of The Pennsylvania State University and informed consent was obtained prior to data collection. To qualify for this study, each subject had to be a competitive distance runner with the ability to run a 10 km in less than 36 minutes or the ability to comfortably run on a treadmill for 15 minutes at a 6 min/mile pace.

### ***5.2.1 Anthropometric measures***

The following anthropometric measurements were recorded for each subject: body height; body mass; leg length (from the tip of the greater trochanter to the floor surface in a standing posture); lower leg length (from the lateral tibial plateau to the floor surface in a standing posture); foot length; hallux length (distance from the head of the first metatarsal to hallux distal end); lateral heel length (horizontal distance from the most lateral tip of the fibular (lateral) malleolus to back of the heel); and medial heel length (horizontal distance

from the most medial tip of the tibial (medial) malleolus to back of the heel). The lateral heel length served as a proxy for the plantarflexor moment arm. Heel lengths were measured with the technique used by Scholz et al. (2008). Digital photographs of each subject's right foot were taken with the subject seated and their foot placed on a wooden reference block with an attached millimeter-scale measuring tape to provide scaling. Before taking the photographs the subject's anterior tibia was aligned vertically and the tips of the lateral and medial malleoli of each subject's right foot was marked with a black marker. Images were used to digitize points on the reference measuring tape and markings on the tips of the malleoli and the posterior aspect of the ankle at the same height as the malleoli and custom Matlab routines (The Mathworks, Inc., Natick, MA) were written to calculate actual lateral and medial heel lengths.

### ***5.2.2 Rate of oxygen consumption***

Prior to each running trial the resting rates of oxygen consumption ( $\dot{V}O_2$ ) was measured while the subject stood quietly on the treadmill for 10 minutes. This was measured using an open-circuit respirometry system (TrueOne 2400; ParvoMedics, Sandy, UT). The collected air was analyzed and oxygen and carbon dioxide concentrations collected during the last two minutes of the resting trial were used to calculate the resting metabolic cost. Subjects then warmed up on a motorized treadmill for 10 minutes at a self-selected speed. Following the warm-up, the running trial started with the treadmill speed immediately increased to 16 km/hr. During the 10 minute running trial, the rate of oxygen consumption ( $\dot{V}O_2$ ) was again measured and the last two minutes of the running trial were used to calculate the metabolic cost during running. The net rate of oxygen consumption during running was calculated by subtracting the resting rate of oxygen consumption from the rate of oxygen consumption during running. The rate of oxygen consumption was normalized by body mass to obtain a mass-specific rate of energy consumption, referred to as the relative rate of oxygen consumption ( $\dot{V}O_{2-REL}$ ). During the running trial of one subject a power failure occurred after 7½ of the 10 minute trial and the running trial was not completed. Since post-hoc analysis revealed a very strong correlation ( $r^2 = 0.949$ ,  $p < 0.001$ ) between  $\dot{V}O_2$  from minutes 5-7 and  $\dot{V}O_2$  from minutes 8-10 for all other subjects,

it was deemed appropriate to estimate this specific subject's 8-10 minute  $\dot{V}O_2$  using a linear regression between 5-7 minute  $\dot{V}O_2$  and 8-10 minute  $\dot{V}O_2$ .

### ***5.2.3 Motion Analysis***

After completion of the treadmill running trial to measure metabolic cost, subjects performed further running trials in a motion laboratory. A six-camera Eagle System (Motion Analysis Corporation, Santa Rosa, CA, USA) was used to collect kinematic data that were synchronized with force-platform data. Marker clusters (4 markers each) were placed on the right thigh, shank, and foot, as along with single markers placed on the heel and the second metatarsal head. All of these markers remained in place during the subsequent running trials. For an initial static standing trial, markers were also placed on the following anatomical landmarks of the right leg, but were removed prior to running trials: greater trochanter, medial and lateral femoral epicondyles, and medial and lateral malleoli. Coordinates from the marker clusters as well as anatomical landmark coordinates were used to create anatomically referenced coordinate systems for both the shank and foot. These segment-fixed coordinate systems were defined such that they were aligned during standing trials, making the ankle position neutral ( $0^\circ$ , with the plantar surface of the foot perpendicular to the long axis of the shank) in standing position. This definition was used as a reference for later determination of plantarflexion and dorsiflexion angles in the sagittal plane. The ankle angles were computed through Euler angle decomposition of the rotation of the foot coordinate system with respect to that of the shank.

Following the initial standing trials, subjects were asked to familiarize themselves with the 30 m runway in the laboratory before attempting running trials at the specified 16 km/h. Running speed was monitored using infrared timing gates to ensure subjects stayed within  $\pm 5\%$  of the required speed. Subjects' starting positions were adjusted in order to ensure that the right foot contacted the floor-mounted force platform (Kistler, Winterthur, Switzerland). Five acceptable trials were recorded. For one subject, only two acceptable trials were recorded (error due to force platform saturation) and the subsequent analyses for this subject were based on these two trials.

#### 5.2.4 Data processing

Both kinematic marker data and kinetic ground reaction force (GRF) data were low-pass filtered at the same frequency (20 Hz), as suggested by Kristianslund et al. (2012), using a 4<sup>th</sup> order Butterworth filter. Ground contact time was calculated as beginning when vertical GRF increased above 5% of bodyweight for each subjects and ending when vertical GRF went below 5% of bodyweight.

Ground reaction force and center of pressure data were used to calculate the GRF lever arm ( $\mathbf{R}$ ) based on the following equation, adapted from Pandy (1999):

$$\mathbf{R} = \frac{\mathbf{M}_{GRF-ANKLE}}{F} = (\mathbf{U}_{ANKLE} \cdot \mathbf{r}_{COP} \times \hat{\mathbf{F}}) \mathbf{U}_{ANKLE} \quad (1)$$

Where  $\mathbf{M}_{GRF-ANKLE}$  is the component of the ankle joint moment associated with the GRF,  $\mathbf{U}_{ANKLE}$  is the unit vector along the ankle mediolateral axis,  $\mathbf{r}_{COP}$  is the vector from the ankle joint center to the center of pressure under the foot, and  $\hat{\mathbf{F}}$  is the unit vector parallel to the resultant GRF.

Subsequently the ratio of ground reaction force lever arm (magnitude of  $\mathbf{R}$ ) to the plantarflexor muscle moment arm ( $r$ ) was computed over the duration of the stance phase. This relationship is known as the gear ratio (Carrier et al., 1994):

$$\text{Gear Ratio} = \frac{R}{r} \quad (2)$$

Gear ratio at maximal dorsiflexion angle was chosen as the variable of interest as this is the position where the series elastic element in the muscle is maximally stretched during running (Lichtwark & Wilson, 2007).

The plantarflexor moment arm value ( $r$ ) is dependent on ankle joint angle and changes throughout the stance phase as the ankle joint plantarflex and dorsiflex. In order to account for the changing moment arm, a dynamic plantarflexor moment arm was estimated. Previous experimentally measured plantarflexor moment arm versus ankle angle data were used to define this subject specific dynamic moment arm (Maganaris et al., 2000). The data from the previous study were interpolated over the ankle angle range of motion experienced by each individual subject in this study and multiplied by a scaling factor. The scaling factor was the ratio of subject's lateral heel length to the interpolated moment arm at zero degrees (ankle in neutral position) from Maganaris et al. (2000).



Ankle joint moments and power were calculated using an inverse dynamic approach and the Newton-Euler equations. Estimated Achilles tendon (AT) force was calculated by dividing the ankle plantarflexor joint moment by subject specific dynamic moment arm.

Several other kinematic, kinetic and spatiotemporal measures were made, including: ankle joint angle at heel strike, toe-off and maximal dorsiflexion angle during stance; stance phase GRF impulse and Achilles tendon impulse; and ground contact time.

#### *Statistical analysis*

Correlation analyses were performed between mass-specific rate of oxygen consumption and the various anthropometric variables and between mass-specific rate of oxygen consumption and gear ratio. Further correlation analyses were performed between heel length (normalized to subject height) and maximum ankle plantarflexor moment (normalized to body mass) and between heel length (normalized to subject height) and maximum AT force (normalized to body mass)

### **5.3 Results**

The net relative rate of oxygen consumption ( $\dot{V}O_{2-REL}$ ) averaged across subjects was 43.46 mL/kg/min ( $\pm 2.68$ ). Neither heel length ( $r = -0.273$ ,  $p = 0.325$ ) nor hallux length ( $r = -0.490$ ,  $p = 0.064$ ) was significantly correlated with  $\dot{V}O_{2-REL}$ , although hallux length was close to significant and showed a negative correlation with  $\dot{V}O_{2-REL}$  (Table 5-1, Figure 5-1 and Figure 5-2). Neither were any significant correlations found between  $\dot{V}O_{2-REL}$  and any other of the anthropometric variables measured (Table 5-1).

There were no significant correlations found between  $\dot{V}O_{2-REL}$  and time of contact or between  $\dot{V}O_{2-REL}$  and any of the ankle kinematic variables measured (Table 5-2). Neither was there a correlation between  $\dot{V}O_{2-REL}$  and any of the kinetic variables of interest (Table 5-3).

Maximal ankle plantarflexor moment normalized to body mass was not found to correlate significantly with normalized heel length ( $r = -0.034$ ,  $p = 0.905$ ) (Figure 5-3). However, maximal AT force normalized to bodyweight was found to be significantly and

negatively correlated with heel length normalized to body height ( $r = -0.654$ ,  $p = 0.008$ ) (Figure 5-4).

## 5.4 Discussion

The purpose of this study was to investigate how foot and ankle structure and the associated biomechanics could potentially contribute to interindividual variation in running economy. Our results showed no significant relationship between any of the anthropometric measurements and the net relative rate of oxygen consumption ( $\dot{V}O_{2-REL}$ ) at 16 km/hr. With respect to our specific variables of interest,  $\dot{V}O_{2-REL}$  was not correlated with heel length but normalized maximal Achilles tendon force was negatively correlated with normalized heel length. There was a trend toward a significant negative correlation between  $\dot{V}O_{2-REL}$  and hallux length.

The ankle joint is of special interest during locomotion as the long lever of the foot permits the ground reaction force to act at a substantial distance from the joint during locomotion. For this reason, internal ankle joint torques tend to be large and typically vary considerably as the magnitude, direction, and point of application of the ground reaction force vector change throughout the stance phase. The foot's role as a lever suggests that the geometry of the foot and ankle joint are likely to be a prominent factor in determining the energetic cost of transport during running. The present study was meant to replicate previous findings showing metabolic cost to be correlated with heel length (Raichlen et al., 2011; Scholz et al., 2008). The authors of those studies explained the relationships they found by citing increased potential for elastic energy storage in the Achilles tendon that would follow from shorter heels. For a given muscular plantarflexion moment, a smaller moment arm would require a larger Achilles tendon force, which would in turn result in increased elastic energy storage and reduced metabolic cost. We did find a correlation between normalized values of force in the Achilles tendon and heel length, lending support to the central premise of this argument: smaller Achilles tendon moment arms do seem to be associated with larger Achilles tendon forces. However no correlation was found between Achilles tendon forces and  $\dot{V}O_{2-REL}$  during running.

The mean moment arm or heel length measured values and coefficient of variation (CV) were similar between our results (mean = 4.93 cm, CV = 7.1%) and those of Scholz et al. (2008) (mean = 4.85 cm, CV = 7.4%). The average  $\dot{V}O_{2-REL}$  across our subjects also compared well with the average value of the subjects in the Scholz study (net  $\dot{V}O_{2-REL}$  of 43.46 ml/kg/min vs  $\dot{V}O_{2-REL}$  of 48.45 ml/kg/min). The average value from the Scholz study is roughly 5 ml/kg/min higher probably due to the fact that resting rate of oxygen consumption (5.35 ml/kg/min for subjects in our study) was not subtracted. However, differences in the CV of  $\dot{V}O_{2-REL}$  between our study and that of Scholz et al. (2008) were considerable. Measurements from Scholz et al. (2008) showed a CV in  $\dot{V}O_{2-REL}$  of 11.7%, whereas our subjects were less variable, with a CV of 6.2 %. Previous investigators performing similar measurements of running economy have reported variability across subjects that was similar to that of our subjects: Heise and Martin (2001) reported a CV of 7.4% (n = 16); Williams & Cavanagh (1987) found a CV across 31 subjects of 5.3%.

There are several possible explanations for why larger Achilles tendon muscle forces might not lead to reduced oxygen consumption during running. Large tendon forces would be expected to correspond to increased tendon energy storage if it is assumed that tendon properties, such as tendon slack length and tendon stiffness, do not vary across individual runners. Variations in tendon stiffness and tendon slack length have the potential, however, to affect the amount of tendon energy storage for a given tendon force. Previous research on the homogeneity of tendon properties has been inconclusive with some investigators finding no difference in tendon stiffness in the plantarflexors between runners and non-runners (Arampatzis et al., 2007; Rosager et al., 2002) and others finding highly economical runners to have higher tendon stiffness (Albracht and Arampatzis, 2013; Arampatzis et al., 2006). In comparing forefoot striking runners with rearfoot striking runners, Perl et al. (2012) found that runners using a forefoot striking pattern also seem to have larger maximal Achilles tendon forces than their rearfoot striking counterparts. Forefoot strikers, however, were not found to be more economical than rearfoot strikers. The authors suggested that, while it is true that larger Achilles tendon forces should increase tendon stretch and energy storage in the tendon, larger forces in the plantarflexor muscle tendon unit also require larger forces to be carried by the active muscle fibers. For

muscles with similar force producing capabilities, larger forces would be associated with higher muscle activation, which would have the effect of increasing metabolic cost. It might be that increased energy storage in the tendon is negated by the increased energy required to sustain higher forces in the muscle tendon unit.

It is unclear why we did not find the same negative correlations between oxygen consumption and heel length that were reported in previous studies (Raichlen et al., 2011; Scholz et al., 2008). Recently, in a study comparing forefoot and rearfoot strikers, Gruber (2012) measured heel length similar to this study and the Scholz et al. (2008) study and failed to find a significant correlation between  $\dot{V}O_{2-REL}$  and heel length for both groups of runners, confirming our result. We used lateral heel length instead of mean heel length (the mean of lateral and medial heel length) as our proxy for plantarflexor moment arm. Others (e.g. Gruber, 2012; Scholz et al., 2008) used mean heel length, but we did not find significant correlations in the present study even when the analyses were performed again with mean heel length. Our choice of lateral heel length was guided by recent unpublished results from our laboratory showing better correlation between plantarflexor moment arms measured from magnetic resonance images (MRI) and lateral heel length than mean heel length.

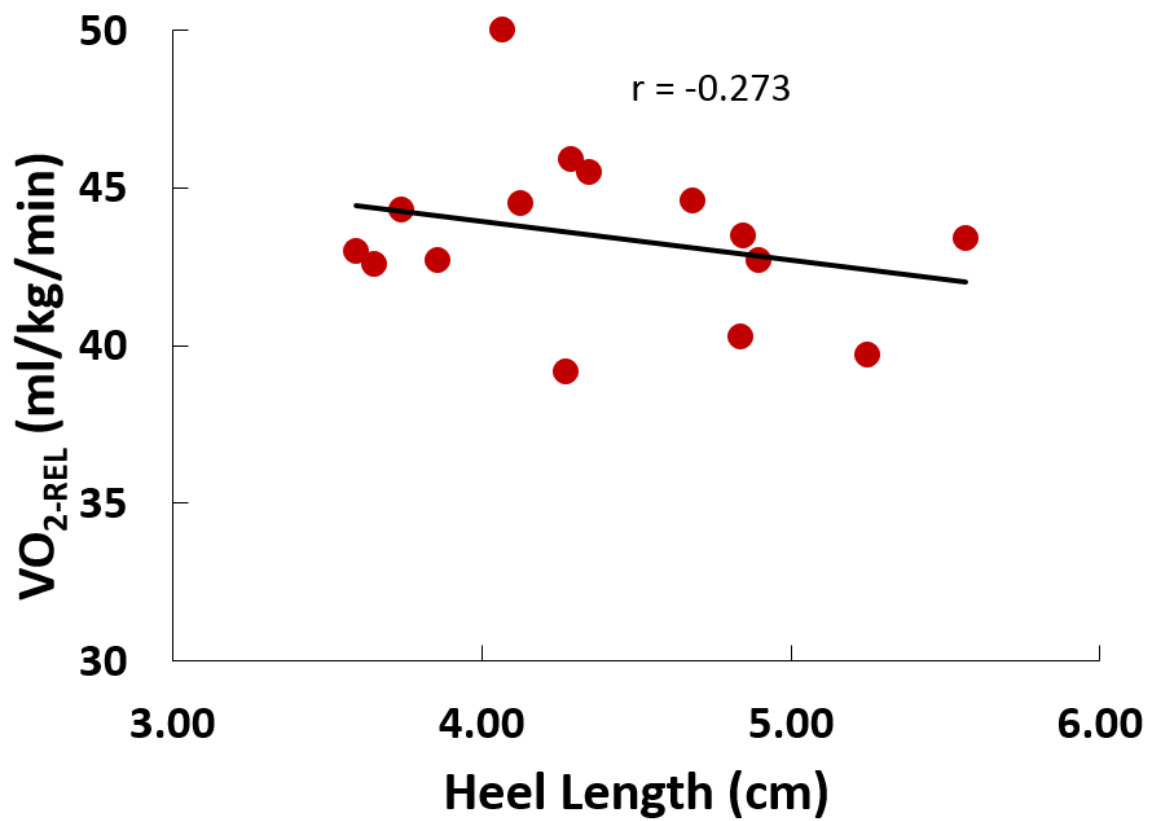
Some differences between the populations studied in our experiments and in the previous two studies should be considered. Subjects from the Scholz et al. (2008) study ( $n = 15$ ) all had personal best 10 km times less than 34 minutes, whereas subjects from the Raichlen et al. (2011) study ( $n=8$ ) had personal best times less than 36 minutes. The athletes in our group were not all specialists at the 10 km distance and instead specialized in different distances. Of our subjects who did report 10 km times, five of six had personal best times less than 34 minutes. All of our subjects had to be able to run a 10 km in under 36 minutes or comfortably run at a 6 min/mile pace on a treadmill for 15 minutes. Post hoc analysis revealed that two of the 15 subjects had a respiratory exchange ratio of 1.0 or greater during the last 2 minutes of the running trial, which could suggest that the activity was exhaustive for these two subjects (McArdle et al., 2006). Even when these two subjects' data were removed, however, the correlations did not change substantially, and we decided against excluding their results.

The variability in  $\dot{V}O_{2-REL}$  measured in our subjects, although similar to values from previous studies (*e.g.* Heise and Martin, 2001; Williams and Cavanagh, 1987) was lower than the variability reported by Scholz et al. (2008). Larger variability in  $\dot{V}O_{2-REL}$  would increase the probability of finding correlations between cost and other variables (*i.e.*, it is not possible to find any correlations if there is no variability in cost). While we cannot identify the source of the diversity, it may be that a more diverse subject pool was used by Scholz et al. (2008) and could have led to differences between their results and ours.

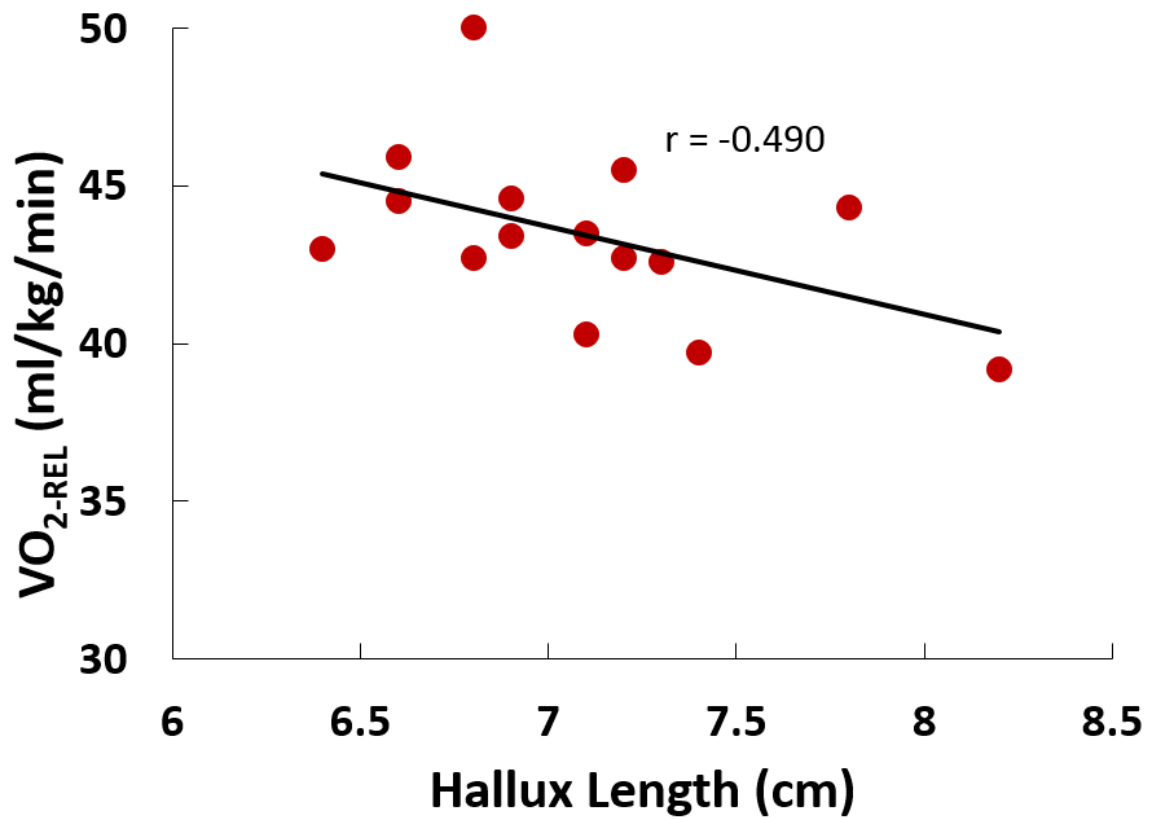
In considering the effect of hallux length on running economy, it was found that a longer toe length seems to suggest a lower rate of oxygen consumption, although this result only trended towards significance ( $p = 0.064$ ; Figure 5-2). This finding contrasts to the suggestion that shorter toes may reduce metabolic cost (Rolian et al., 2009). Rolian et al. (2009) found that shorter toes were accompanied by reduced work at the metatarsophalangeal (MTP) joint and therefore potentially reduce metabolic cost. Unfortunately, joint work at the MTP joint was not measured in the present study.

The lack of correlation we found between rate of oxygen consumption during running and the other measured variables have been reported before in some cases, but in other cases previous investigators have found significant correlations. As in our study, Williams and Cavanagh (1987) found no significant difference in peak GRF between groups with different running ability, and no differences in anthropometric and inertial measures (segment lengths, masses and moments of inertia) of various body segments between groups with different running ability. However, contrary to our results, Williams and Cavanagh (1987), did find plantarflexion angle at toe-off was significantly smaller in runners with better running economy. In our study, contact times were not correlated with rate of oxygen consumption. Results from previous studies have been inconclusive. Kyröläinen et al. (2001) found no significant relationship between contact time and running economy, as in the current study. Williams and Cavanagh (1987) found a trend towards runners with a lower rate of oxygen consumption showing longer contact times while Paavolainen et al. (1999) showed that runners with improved 5 km times exhibited a decrease in contact time during a constant velocity running trial.

In conclusion, the results of the present study do little to clarify the potential influence of ankle and foot anthropometry on running economy. From the contrasting findings evident from our work and the studies published by Scholz et al. (2008) and Raichlen et al. (2011), it is clear that more research is necessary before we can understand the nature of the relationship between plantarflexor moment arm and running economy. It may not be possible to find foot and ankle anthropometric measures that effectively predict running economy. Previous authors have discussed the large number of factors that may influence running economy and their general consensus is that it is probable that only a combination of several of these factors would allow prediction of running economy rather than a single factor (*e.g.* Anderson, 1996; Heise and Martin, 2001; Williams and Cavanagh, 1987).

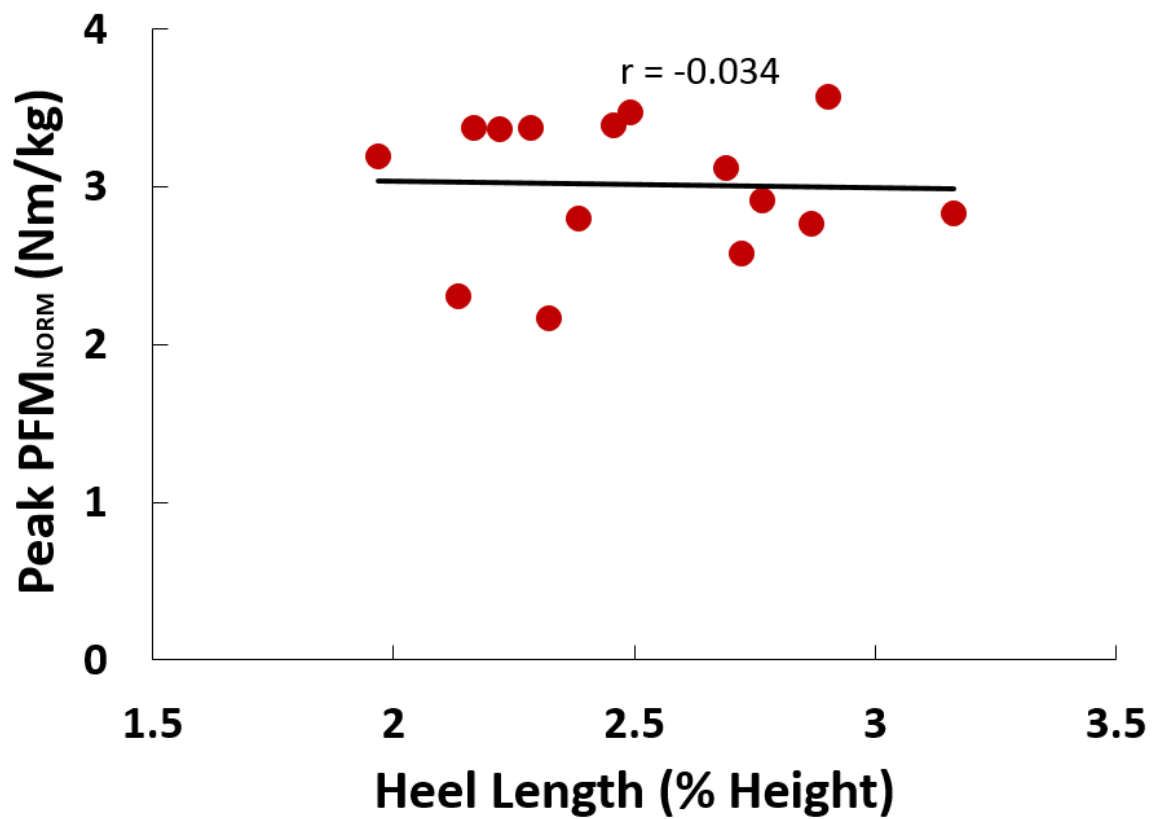


**Figure 5-1:**  $\dot{V}O_{2-REL}$  of running plotted versus the heel length for all subjects. Regression analysis revealed a non-significant negative correlation ( $p = 0.325$ ).

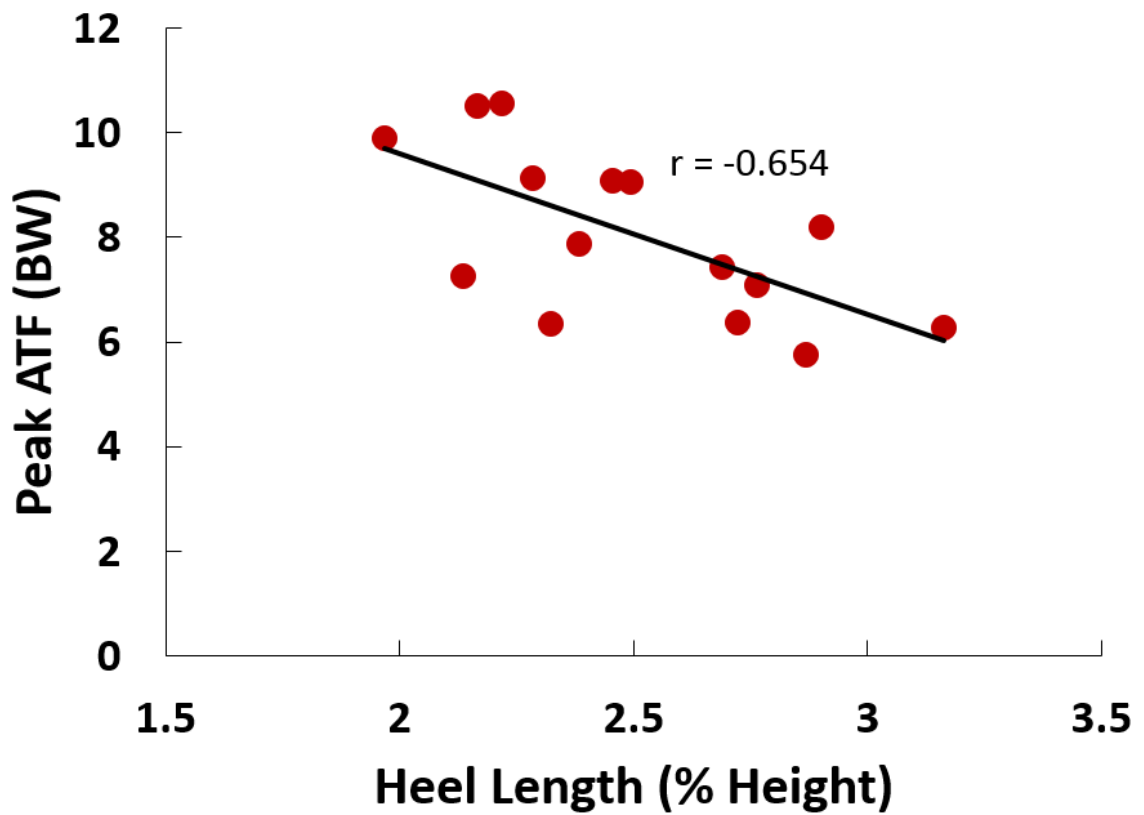


**Figure 5-2:**  $\dot{V}O_{2-REL}$  of running plotted versus the length of the hallux for all subjects. Regression analysis revealed a moderate negative correlation that trended toward significance ( $p = 0.064$ ).





**Figure 5-3:** Peak plantarflexor moment normalized to body mass versus heel length normalized to body height. Regression analysis revealed a non-significant negative correlation ( $p = 0.905$ ).



**Figure 5-4:** Estimated peak Achilles tendon force normalized to bodyweight versus heel length normalized to body height. Regression analysis revealed a strong negative correlation that was significant ( $p = 0.008$ ).

**Table 5-1:** Correlations between  $\dot{V}O_{2-REL}$  and anthropometric variables.

Anthropometric variables	Mean (SD)	r	p-value
Height (cm)	176.0 (7.3)	-0.276	0.319
Mass (kg)	67.6 (7.7)	-0.166	0.554
Total Leg Length (cm)	90.7 (5.5)	-0.416	0.123
Lower Leg Length (cm)	49.0 (2.5)	-0.317	0.250
Foot Length (cm)	26.4 (1.4)	-0.438	0.102
Lateral Heel Length (cm)	4.4 (0.6)	-0.273	0.325
Mean Heel Length (cm)	4.9 (0.4)	-0.347	0.205
Hallux Length (cm)	7.1 (0.5)	-0.490	0.064

**Table 5-2:** Correlations between  $\dot{V}O_{2-REL}$  and kinematic and spatiotemporal variables.

Variables	Mean ( $\pm$ SD)	r	p-value
Time of contact (s)	0.20 (0.02)	0.365	0.181
Ankle angle at HS (°)	-1.18 (13.14)	0.297	0.283
Ankle angle at TO (°)	-23.55 (6.89)	-0.033	0.906
Max. dorsiflexion (°)	19.19 (5.04)	0.334	0.223

**Table 5-3:** Correlations between  $\dot{V}O_{2-REL}$  and kinetic variables. All force and impulse variables are normalized to bodyweight (BW).

Variables	Mean ( $\pm$ SD)	r	p-value
Max AT force / BW	8.05 (1.58)	-0.030	0.916
Max GRF / BW	2.82 (0.37)	-0.291	0.293
AT impulse / BW (1/s)	0.75 (0.15)	0.025	0.929
GRF impulse / BW (1/s)	0.34 (0.02)	0.080	0.778
Gear ratio at max. dorsiflexion	3.15 (0.52)	0.287	0.299

## **Chapter 6**

### **Computational Model Predictions of Optimal Foot and Ankle Structure for Different Activities**

#### **6.1 Introduction**

Certain musculoskeletal features of the foot and ankle seem to be adapted for optimal performance in various motor tasks. It has been shown, for example, that sprinters have longer muscle fascicles, shorter plantarflexor moment arms and longer toes than non-sprinters (Baxter et al., 2012; Lee and Piazza, 2009). Plantarflexor moment arms correlate with maximal isokinetic torque production in a dynamometer task (Baxter and Piazza, 2014), and correlations between jump height and toe length as well as heel length have been found for a single-joint maximum-height jumping task (see Chapter 4). Scholz and colleagues (2008) as well as Raichlen et al. (2011) have found heel dimensions to be positively correlated with rate of oxygen consumption during treadmill running.

Computational modeling techniques are particularly well suited for investigating the influence of variation in musculoskeletal structure on human performance. A modeler can use a model with a specific set of musculoskeletal architectural properties as the basis for a simulation of a given movement using a forward dynamic approach. The effect of variation of the architectural properties on the outcome performance is then investigated by specifying a different set of architectural properties. This modeling approach is analogous to having subjects with different musculoskeletal properties perform a given task, but computational modeling allows the modeler to change only the parameters of interest without simultaneous changes to other variables, which is unavoidable in human subjects testing. Further, computational modeling allows consideration of internal mechanisms not easily studied in vivo. One can, for example, assess muscle fiber length and internal muscle force throughout the movement and through this analysis better understand the mechanisms that potentially allow for specific architectural variations to improve (or adversely affect) performance.

Several attempts have been made to use computational modeling to elucidate the mechanisms that underlie the connections between structural properties of the ankle and foot and human performance (*e.g.* Baxter et al., 2012; Nagano and Komura, 2003; Rolian et al., 2009). Using a simple single-degree of freedom model of the ankle joint, Baxter et al. (2012) showed that sprinters may benefit from shorter heels due to a reduction in plantarflexion contraction velocity. A similar investigation by Nagano and Komura (2003) showed that smaller plantarflexor moment arms allows for larger joint power production during fast ( $>120$  °/s) isokinetic plantarflexion movements. Lee and Piazza (2009) employed a model with an ankle and metatarsophalangeal (MTP) joint and found that longer toes allowed a sprinting model to stay on the ground longer, increasing its ability to generate horizontal impulse during a simulated sprint pushoff.

Simulation studies are often performed with the purpose of answering a specific question related to specific movement such as the sprint start (*e.g.* Baxter et al., 2012) or maximal-height jumping (*e.g.* Bobbert, 2001). To our knowledge, there are no studies employing a single computational model to study the effect of structural variability on multiple tasks of human movement. Using the same model to study various tasks and analyze the effects of varying structural properties on task performance facilitates comparisons across tasks and eliminates the variability that would accompany using different models to study different tasks.

Another limitation of previous modeling work is the limited number of foot and ankle structural parameters investigated. Simulation studies have generally focused on one or two variables to try to understand their influence on performance. Considering a larger number of parameters simultaneously would enhance our understanding of how several parameters combine to produce optimal performance. While it is not possible to design a computational model that incorporates all properties of the system, it is feasible to include several properties with the potential to affect performance in order to understand the effects of this combined set of parameters on performance in different motor tasks.

The purpose of this study was to create a simplified computational model incorporating the foot with ankle and MTP joints to study how variation in structural properties of a combined plantarflexor muscle, ankle and toe affect performance across

different motor tasks. For this study the tasks were: i) maximizing vertical energy during pushoff, similar to a vertical jump; ii) maximizing average horizontal acceleration from an upright stance position, similar to sprinting from a standing start; and iii) maximizing static load support with the heels lifted off the ground. Properties of the computational model were analyzed to elucidate the mechanisms of optimal performance and their relation to variation in musculoskeletal architectural properties.

## 6.2 Methods

### 6.2.1 Model characteristics

The musculoskeletal model created was based on the three dimensional lower limb musculoskeletal model described by Arnold et al. (2010). Only one half of the body was modeled and the movements of the model were restricted to the sagittal plane. The segments of the simplified model consisted of: a toe segment; a foot segment; a massless leg segment; and a point mass representing the rest of the body that was rigidly attached to the leg (Figure 6-1). Inertial properties for all segments are given in Table 6-1. The model had five degrees of freedom and its position was defined by five generalized coordinates: x-position of the toe; y-position of the toe; toe-angle between the toe and the ground; metatarsophalangeal (MTP) angle between the foot segment and toe segment; and ankle angle between the leg segment and the foot segment. A complete description of model and muscle attachment coordinates is given in Appendix A.

Contact between the foot and the ground was assumed to occur at three possible locations: below the distal end of the toe segment; below the MTP joint; and at back of heel below the insertion of the plantarflexor muscle (Figure 6-1). The contact model used is a modified version of the model employed by Celik and Piazza (2013). The vertical component of foot-contact force was defined according to:

If  $\delta > 0$

$$GRF_y = a\delta^3(1 + b\dot{\delta})$$

else



$$GRF_y = 0$$

where:  $\delta$  is penetration of the foot into the floor;

$GRF_y$  is vertical ground reaction force;

$a = 1 \times 10^9 \text{ N/m}^3$  (stiffness coefficient); and

$b = 1.0 \text{ s/m}$  (damping coefficient)

The spring stiffness was set such that a reasonable amount of foot penetration occurred during simulated quiet standing. The value of  $1 \times 10^9 \text{ N/m}^3$  resulted in penetration values of 5-6 mm at each contact node. The damping value was chosen to minimize oscillations when penetration and pushoff occurred. Horizontal contact forces between the foot and floor were defined as follows:

$$GRF_x = \mu GRF_y \tanh \frac{\dot{x}}{s}$$

where:  $GRF_x$  is horizontal ground reaction force;

$\dot{x}$  is the horizontal velocity of the contact point;

$\mu = 0.8$  (Coulomb friction coefficient); and

$s = 0.1$  (Velocity scaling parameter)

Joint range of motion (ROM) limits were included to resist hyperflexion and hyperextension. Excessive joint rotations were counteracted by means of a torsional spring-damper that produced a resistive joint torque when joint angle  $\theta$  exceeded its limit  $\theta_{\text{limit}}$  as follows:

If  $\varepsilon > 0$

$$T_{\text{resist}} = \alpha \varepsilon^3 (1 + \beta \dot{\varepsilon})$$

else

$$T_{\text{resist}} = 0$$

where:  $\varepsilon = \theta - \theta_{\text{limit}+}$  or  $\varepsilon = \theta_{\text{limit}-} - \theta$  is angle of overextension;

$T_{\text{resist}}$  is the resistive joint torque;

$\alpha = 0.5 \text{ N.m/deg}^3$  (stiffness coefficient); and

$\beta = 0.001 \text{ s/deg}$  (damping coefficient)

The coefficients  $\alpha$  and  $\beta$  were set to produce values for  $T_{resist}$  that were comparable to the previously reported passive ankle joint torques (Riener & Edrich, 1999), and that just minimized oscillations in joint angle when a range of motion limit was encountered.

The following limits on joint rotation were assumed:

MTP joint:  $\theta_{limit+} = 30^\circ$  (extension);  $\theta_{limit-} = 45^\circ$  (flexion); and

Ankle joint:  $\theta_{limit+} = 20^\circ$  (dorsiflexion);  $\theta_{limit-} = 50^\circ$  (plantarflexion)

The model was actuated by four lumped muscle-tendon units, each of which represented a muscle group: plantarflexor, dorsiflexor, toe flexor and toe extensor (Figure 6-1). The parameters used to define the lumped actuators (maximum isometric force, optimal fiber length, tendon slack length, pennation angle, stiffness scaling factor) were adapted from measurements made by Ward et al. (2008). Lumped muscle parameters were calculated using a weighted average of the volume of muscles as a basis for weighting of all factors (see Appendix B for details). The final muscle parameters used for the current model are shown in Table 6-2. The muscle paths of the single lumped muscles were created by averaging coordinates for the origin, insertion and intermediate routing points (“via points”) of the original Arnold et al. (2010) model as described below.

- i) Plantarflexor: The single plantarflexor was a lumped model of the two heads of the gastrocnemius and the soleus. Although other plantarflexors (tibialis posterior, peronius brevis, peroneus longus) contribute to plantarflexor joint torque, their contribution is significantly lower and these other muscles do not insert on the Achilles tendon, whose plantarflexor moment arm is one of the musculoskeletal parameters under investigation in this study. The lumped plantarflexor muscle path was created by first replacing the original wrapping surface that defined the path of the gastrocnemius posterior to the knee joint with two points just distal to this wrapping surface. These points represented the origins for two single-joint plantarflexors, the lateral and medial gastrocnemius. The average of these two points was then used as an effective origin for a single combined gastrocnemius muscle. Finally, the lumped plantarflexor origin point was taken as midpoint between the combined gastrocnemius muscle origin and the original soleus muscle

origin. The insertion points for these muscles were the same in the original Arnold et al. model and their coordinates were used for the lumped plantarflexor insertion.

ii) Dorsiflexor: The original origin, insertion, and via points of the tibialis anterior in the Arnold et al. (2010) model were used to define the path of the lumped dorsiflexor muscle. Although other muscles contribute to dorsiflexion, their contribution is small in comparison to the tibialis anterior they were excluded from the model.

iii) Toe flexor: This actuator was a combination of the flexor digitorum longus and flexor hallucis longus. The midpoints of the two muscles' origins and insertions were used for the lumped actuator, along with via points representative of both muscles. During initial testing of the model, it was found that the model was unable to successfully pushoff from the toes during the final stage of movement in preparation for a simulated vertical jump. Further investigation revealed that the toe flexors in the original Arnold et al. model were unable to produce MTP flexion moments similar to those measured experimentally. Goldmann and Brüggemann (2012) reported a maximum isometric flexion moment of 12 Nm at 25° MTP extension, whereas the original model could only produce 3.6 Nm at the same angle. It was decided to increase the maximal isometric muscle force,  $f_{MAX}$ , of the toe flexor from its original value of 685 N by a factor of 4, resulting in a new  $f_{MAX} = 2740$  N. This increased the model's maximum MTP flexion moment from 3.6 N m to 14.5 N m. Although this  $f_{MAX}$  value for the lumped toe flexor muscle is much higher than the original model value, recent modeling work showed toe flexor forces in excess of 2000 N during running trials when estimated from ground reaction force data (Rolian et al., 2009).

iv) Toe extensor: The midpoints between the origins and insertions of the extensor digitorum longus and extensor hallucis longus were used along with via points representative of each muscle.

The model was made planar by projecting all muscles path coordinates onto the sagittal plane and joints axes were rotated such that joint rotations occurred in the sagittal

plane. Muscle-tendon unit lengths and moment arms of the muscles of the planar model were compared with those of the original three-dimensional model to ensure that they were similar. Where muscle moment arms or muscle lengths differed substantially (i.e., by more than  $\pm 5\%$  throughout the joint range of motion) muscle-tendon paths of the planar model were adjusted to ensure good agreement. The new planar model was created and edited in OpenSim before being adapted for use with SimMechanics toolbox in Matlab/Simulink (MathWorks, Inc.; Natick, MA, USA). All four lumped muscles were modeled as Hill-type muscle-tendon actuators driven by excitation inputs. The muscle model included first-order activation dynamics and force-length relations for muscle and tendon and a force-velocity dependence for muscle (Schutte, 1992).

### 6.2.2 Musculoskeletal optimization parameters

The following 4 parameters were selected as musculoskeletal parameters to be adjusted in order to achieve optimal performance in the simulated motor tasks:

- Plantarflexor insertion point (*lHeel*): The horizontal distance from the ankle axis to the plantarflexor muscle insertion on the posterior of the foot segment (the heel), which was representative of plantarflexor moment arm.
- Plantarflexor stiffness scaling factor ( $k^*$ ). This scaling factor was used to scale the nominal normalized tendon force versus strain curve (Zajac, 1989) in the muscle model employed. This had the effect of changing the slope of the normalized tendon force versus strain curve, similar to a change in stiffness.
- Plantarflexor normalized tendon length ( $\rho$ ): The ratio of tendon slack length ( $lT_{SLACK}$ ) to optimal fiber length ( $lF_{OPT}$ ) for a given muscle. A change in  $\rho$  affects both  $lT_{SLACK}$  and  $lF_{OPT}$ . The following constraint was placed on these values to define  $lT_{SLACK}$  and  $lF_{OPT}$  for different values of  $\rho$ :

$$(lT_{SLACK} + lF_{OPT} \cos \theta) / lMTU = \text{CONSTANT} @ 0^\circ$$

This constraint was necessary to ensure that the relative combined length of  $lT_{SLACK}$  and  $lF_{OPT}$  (corrected for pennation angle  $\theta$ ) with respect to the total muscle tendon unit

- length ( $l_{MTU}$ ) remained constant as  $\rho$  was varied. Variations in  $l_{Heel}$ , however, also affected  $l_{MTU}$ . This was accounted for by noting that each different  $l_{Heel}$  produced a different  $l_{MTU}$ , from which  $\theta$  and  $\rho$  were calculated.
- Toe length ( $l_{Toe}$ ). Variation in  $l_{Toe}$  changed the dimension of the toe segment. The location of contact points remained at the distal end of the toe segment and below the MTP joint. The location of the insertion points of the toe flexor and toe extensor stayed in the same position relative to the MTP joint and changes in  $l_{Toe}$  did not therefore affect muscle function.

### **6.2.3 Optimization procedure**

To reduce the complexity of the optimization, the values of the musculoskeletal optimization parameters were varied in binary fashion. Each musculoskeletal optimization parameter was assigned either a high value or a low value. Where possible, these high and low values were estimated from previously reported experimental values. High values were one standard deviation (SD) above the nominal parameter value, and low values were one SD below the nominal parameter value. For example, Lee and Piazza (2009) measured toe lengths of  $7.3 \pm 0.9$  cm SD for twelve non-sprinter subjects, yielding a coefficient of variation (CV) of 12%. The high and low  $l_{Toe}$  values used in the present study were thus set to the nominal parameter value  $\pm 12\%$ . This method produced the high and low values given in Table 6-3. Running separate optimization routines for each of these high/low combination for each of the 4 parameters resulted in  $2^4 = 16$  different musculoskeletal configurations (in addition to the nominal model) for which optimal performance in each of the motor tasks was simulated.

### **6.2.4 Task-specific simulation characteristics**

#### **6.2.4.1 Maximal vertical-energy pushoff**

‘Vertical-energy’ was defined as the sum of the gravitational potential energy and the kinetic energy associated with vertical movement.

*Movement initialization.* The model started in a static position with the heel on the ground and zero muscle activation across all muscles. This initial position was found using optimization that minimized the sum of four penalty functions:

$$J_{INIT} = P_{FORCE} + P_{STRAIN} + P_{ACCEL} + P_{HEEL}$$

where  $P_{FORCE}$  ensured static equilibrium in the muscles, by penalizing differences between muscle and tendon forces;

$P_{STRAIN}$  penalized initial tendon lengths that produced negative (compressive) strain;

$P_{ACCEL}$  ensured a static initial position by penalizing joint and model center of mass accelerations; and

$P_{HEEL}$  ensured initial heel contact by penalizing heel height above floor level

For further details, including the formulation of the above penalties, the reader is referred to Appendix C.

For this movement initialization optimization there was a total of 8 parameters to optimize: initial fiber lengths for all four muscle ( $lPF_{INIT}$ ,  $lDF_{INIT}$ ,  $lTF_{INIT}$ ,  $lTE_{INIT}$ ) and four generalized coordinates: y-position of the toe; toe angle with respect to ground; metatarsophalangeal (MTP) angle; and ankle angle. It was not necessary to include the x-coordinate of the toe as it did not affect the objective function value. For each of the nominal model and the 16 different musculoskeletal architectures of the model (17 in total) an optimized initial configuration was found using the Covariance Matrix Adaptation algorithm (CMA), a non-gradient based optimization algorithm (Hansen et al., 2003). Acceptable starting positions were found for each of the 17 combinations of musculoskeletal architecture. A starting position was deemed acceptable when the objective function value ( $J_{INIT}$ ) approached zero (see Appendix C for details)

*Movement.* Initial simulations of maximal height jumping showed that optimization produced a bouncing strategy, similar to the strategy employed by a simulation and human subjects studied by van Werkhoven and Piazza (2013). For the present study, we wanted to study a simpler movement that consisted of a single ankle plantarflexion without any bouncing. To prevent the model from bouncing, a time limit was placed on the movement

such that the simulation could not last more than 0.5 s. This value represented the highest values observed during experimentally measured single-joint ankle jumps (van Werkhoven and Piazza, 2013). With the 0.5 s final time constraint in place and a penalty for not leaving the ground at the final time, it was found that the model ‘jumped’ by dorsiflexion of the ankle after it raised on the toes in order to create a larger negative (downward) center of mass vertical velocity. We were interested in the capability of the model to produce an upward thrust, even if the model was unable to jump, and therefore removed the constraint that forced the model to leave the ground.

The objective function was therefore to maximal vertical energy at 0.5 s, given by the equation below:

$$J_{PUSHOFF}(t_f, u) = J_{VERT} - J_{HORIZ}$$

where  $J_{VERT}(t_f, u) = y_c(t_f) + \frac{1}{2g} \dot{y}_c^2(t_f)$

was the task specific objective, the mechanical energy associated with vertical movement at  $t_f$  normalized to bodyweight (Levine, 1983);

and  $J_{HORIZ}$  was a term that penalized the energy associated with horizontal velocity, ensuring that the model moves straight upwards at  $t_f$ .

For further details, including the exact formulation of this penalty function, the reader is referred to Appendix C.

Optimal pushoff was simulated using a parameter optimization based on forward dynamic integration of the model equations of motion. The objective was function of final time  $t_f$  and muscle excitations  $u$  (with 21 nodes evenly spaced in time between  $t = 0.0$  and  $t = t_f$  for each of the four muscles), giving a total of 85 movement based optimization parameters. Excitation values could range between 0 and 1, and the range of values for  $t_f$  was set between 0.1 and 0.5 s.

The jump simulation of the model with nominal musculoskeletal architecture was optimized using a dual CMA and pattern search (PS) approach (Hooke & Jeeves, 1961). Random initial guesses were used as input into both CMA and PS algorithms for the nominal musculoskeletal architectural configuration. Results from both optimizers were then used as initial guesses for the other optimizer, until an optimal result was found. An optimal solution was assumed to have been found when at least two different initial guesses resulted in the same maximal objective function result, or if no better solutions were found after starting the optimizer with at least four different initial guesses. After the nominal optimal result was found, this result was used as initial guess for the 16 different musculoskeletal configurations. Different initial guesses were also used during this stage of the optimization in an attempt to ensure that globally optimal results were found.

#### 6.2.4.2 Maximal average horizontal acceleration

*Movement initialization.* The model started in the same static position with the heel on the ground as described for the *maximal vertical-energy pushoff* simulation.

*Movement.* The objective function was to maximize the average horizontal acceleration, given by the equation below:

$$J_{SPRINT} = J_{ACCEL} - J_{PENSPRINT}$$

$$\text{where } J_{ACCEL}(t_f, u) = \frac{\dot{x}_C(t_f) - \dot{x}_C(0)}{t_f}$$

which is the average horizontal acceleration given that the model starts with a zero initial horizontal velocity;

$$\text{and } J_{PENSPRINT}(t_f, u) = P_{GRF} + P_{GROUND} + P_{BOUNCE} + P_{VCOM}$$

where  $P_{GRF}$  guaranteed that the model left the ground at  $t_f$ ;

$P_{GROUND}$  ensured that the point mass did not move through the floor during jumping;

$P_{BOUNCE}$  penalized the model for any bouncing movement by allowing only one change in ankle rotation direction throughout the movement; and



$P_{VCOM}$  penalized negative vertical end velocity - to ensure that the model did not dive at  $t_f$

For further details regarding the equations to implement the above penalties, refer to Appendix C.

The maximal average horizontal acceleration simulation was similar using the same technique as the vertical-energy approach, using parameter optimization based on forward dynamic integration of the model equations of motion. The objective was function of final time  $t_f$  and muscle excitations  $u$  (with 21 nodes for each of the four muscles) giving a total of 85 movement based optimization parameters. Excitations values could range 0 and 1 and  $t_f$  was allowed to range between 0.1 and 2.0 s.

A similar optimization approach to the vertical-energy optimization was employed, using both PS and CMA algorithms.

#### 6.2.4.3 Maximal static load support

This problem was not solved through dynamic optimization but rather through static optimization similar to the initialization problem described above. The objective was to maintain a static position with the body center of mass lifted 5 cm from the original standing posture with the greatest possible mass added to the point mass in the model. The following objective function was used:

$$J_{SUPPORT} = J_{MASS} - J_{PEN_{SUPPORT}}$$

where  $J_{MASS}(a, l_M, q) = m_{LIFT}$

is the extra mass carried by the body;

$a$  = initial muscle activation levels;

$l_M$  = initial muscle fiber lengths;

$q$  = four generalized coordinates;

and  $J_{PEN_{SUPPORT}}(a, l_M, q) = P_{FORCE} + P_{STRAIN} + P_{ACCEL} + P_{POS}$

where  $P_{FORCE}$  ensured static equilibrium in the muscles, by penalizing differences between muscle and tendon forces;  
 $P_{STRAIN}$  penalized initial tendon lengths that produced negative (compressive) strain;  
 $P_{ACCEL}$  ensured a static initial position by penalizing joint and model center of mass accelerations; and  
 $P_{POS}$  ensured that the CoM is lifted 5 cm above CoM at model resting position

For further details regarding the equations to implement the above penalties, refer to Appendix C.

There were a total of 13 parameters to optimize: the mass lifted, as well as initial activations for all four muscles, initial fiber lengths for all four muscle and four generalized coordinates: y-position of the toe; toe-angle with respect to ground; metatarsophalangeal (MTP) angle with respect to toe segment; and ankle angle with respect to foot segment. As with the previous optimization, it was not necessary to include the x-coordinate of the toe as it did not affect the objective function value. For each of the 17 different musculoskeletal architecture combinations of the model, an optimized initial configuration was found using the Covariance Matrix Adaptation algorithm (CMA). An optimal result was assumed to have been found when different initial guesses resulted in the same maximal objective function result.

## 6.3 Results

### 6.3.1 Maximal vertical-energy pushoff

All optimal solutions for the maximal vertical-energy pushoff simulations produced a similar movement pattern in which the model initially leaned forward before plantarflexing to propel the body upwards. On average across all configurations simulated,

43% of the movement time of 0.5 s was spent moving the body over the toes, and 57% of the movement time was spent moving the body upwards.

The nominal architectural properties produced an objective function value, which is maximal vertical energy normalized to body weight of 1.0910 m. The combination of factors that gave the best performance was a short heel, short toe, small normalized tendon length and high tendon stiffness, which produced a vertical energy value of 1.1011 m. The performance ranged between a minimum of 1.0798 m and a maximum of 1.1011 m. Variation in  $l_{Heel}$  had the biggest effect on performance, followed by variation in  $\rho$ ,  $k^*$ , and  $l_{Toe}$  (Figures 6-2 and 6-3).

In order to understand the possible mechanisms that allow better performance due to structural variation, the following internal variables were analyzed: plantarflexor fiber length, plantarflexor fiber velocity, and plantarflexor force throughout the final push off phase of the movement (final 57% of movement time). When analyzing the effect of the different parameters, it is useful to examine how parameter variation affects the internal variables of interest (fiber length, fiber velocity, muscle force). The plantarflexor fiber length averaged over this pushoff phase for all configurations with short  $l_{Heel}$  stayed much closer to optimal fiber length throughout the movement compared with the simulation with a long  $l_{Heel}$ . Small  $\rho$  showed a marginally better operating range compared with large  $\rho$ . Variation in  $k^*$  and  $l_{Toe}$  had minimal effect on fiber length throughout movement (Figure 6-4). When considering fiber velocity, variation in  $l_{Heel}$  and  $\rho$  had the largest effects. In this case however, short  $l_{Heel}$  on average reduced fiber velocity, whereas small  $\rho$  showed an increase in fiber velocity. Variation in  $k^*$  and  $l_{Toe}$  both had minimal effect on fiber velocity characteristics (Figure 6-5). Plantarflexor muscle force was most sensitive to  $l_{Heel}$  variation (shorter better) followed by  $\rho$  (smaller better). Variation in  $k^*$  and  $l_{Toe}$  again showed minimal effect (Figure 6-6).

### **6.3.2 Maximal average horizontal acceleration**

All solutions for the maximal average horizontal acceleration simulations showed an initial dorsiflexion movement to get center of mass in a position to move horizontally, followed by a plantarflexion pushoff. The average total movement time was 0.9277 s, with

86% spent in a dorsiflexion motion, and the final 14% of time spent performing a plantarflexion motion for pushoff which propelled the body forward horizontally.

The nominal architectural properties produced an average horizontal acceleration of  $1.3905 \text{ m/s}^2$ . The best performance was  $1.4839 \text{ m/s}^2$ , achieved with a short heel, short toe, small normalized tendon length and low stiffness. The worst performance was an acceleration of  $1.2183 \text{ m/s}^2$ . Similar to the vertical maximal energy task results, variation in  $l_{Heel}$  and  $\rho$  contributed more to performance differences compare to  $k^*$  and  $l_{Toe}$  (Figures 6-7 and 6-8). Configurations with a short  $l_{Heel}$  stayed closer to optimal fiber length through the movement time, as did configurations with a small  $\rho$  and low  $k^*$  (Figure 6-9). Variation in  $l_{Toe}$  had no effect on fiber operating length (Figure 6-9). Fiber velocity showed the largest variation due to variation in  $l_{Heel}$ , followed by variation in  $\rho$  and  $k^*$  and no noticeable effect from variation in  $l_{Toe}$  (Figure 6-10). Plantarflexor muscle force had a higher peak value with short  $l_{Heel}$ , and marginally higher with small  $\rho$  (Figure 6-11).

### **6.3.3 Maximal static load support**

For maximal static load support the combination of factors that gave the best performance was long  $l_{Heel}$ , long  $l_{Toe}$ , small  $\rho$  and high  $k^*$ , in which the model was able to support 189.9 kg. The nominal configuration produced a value of 161.3 kg, and the lowest mass supported was 133.1 kg. Variation in  $l_{Heel}$  had a significantly larger effect on performance than did the other variables (Figures 6-12 and 6-13). Variation in  $l_{Toe}$  had almost no effect on the fiber length at which the mass was held or the plantarflexor force. The optimal values for all other parameters ( $l_{Heel}$ ,  $k^*$  and  $\rho$ ) resulted in fiber lengths closer to optimal length and higher plantarflexor muscle forces (Table 6-4).

## **6.4 Discussion**

The goal of this study was to use a computational model to investigate the effect of simultaneous variation in selected foot and ankle musculoskeletal architecture properties on performance in several motor tasks. To our knowledge this was the first attempt to use

a single computational model to simulate different motor tasks in order to investigate the effect of musculoskeletal variation on performance. We simulated three different activities: a maximal vertical-energy task similar to a maximal height jump or pushoff; a sprint start from an upright standing position with maximum average horizontal acceleration; and an isometric task in which the model maintained a static position while supporting the greatest possible additional mass while the heel was lifted off the ground.

Performance in both explosive tasks, the vertical-energy task as well as the horizontal acceleration task, was significantly affected by variation in heel length. Short heels resulted in superior performance for both activities. Examination of internal plantarflexor muscle dynamics during simulated movements showed that short heels allowed the muscle fibers to operate closer to the optimal fiber length than did long heels. Short heels also allowed for a reduced shortening velocity of the muscle fibers throughout the movement and increased force production. This is consistent with results from previous simulation studies in which the effects of heel lengths or plantarflexor moment arm during movements with fast ankle joint rotations were considered (Baxter et al., 2012; Nagano and Komura, 2003). Results from experimental studies of explosive ankle tasks have generally found similar results when comparing good performers to controls. It has been shown that sprinters have shorter plantar flexor moment arms than non-sprinters (Baxter et al., 2012; Lee and Piazza, 2009) and volleyball players have been found to have smaller plantarflexor moment arms compared with controls (Watanabe et al., 2008). However, results from studies in which performance is correlated to plantarflexor moment arm do not necessarily support these findings. In Chapter 4 we showed that a significant correlation exists between jump height and heel length for a group of healthy young men and others have found correlations between isokinetic strength measured in dynamometer and plantarflexor moment arm (Baxter and Piazza, 2014). These correlational studies of young healthy male subjects indicate that longer, not shorter, heels seem to afford a performance benefit. The findings in these studies may relate to the different subject populations; individuals who are not trained athletes might be adapted differently than are more athletic individuals. Musculoskeletal properties of sprinters and volleyball players might be adapted to allow for optimal performance with shorter heels, whereas normal non-athletic

population benefit from longer heels in combination with their other musculoskeletal properties. It was predicted that by varying the combinations of musculoskeletal properties in this study we would be able to explore interactions between properties that could allow for better performance with long heels (compared with elite athletes with short heels) based on a more favorable combination of  $\rho$ ,  $k^*$ , and  $l_{Toe}$  properties. However, the effect of variation in  $l_{Heel}$  was so extreme that no combination of other factors could override its effect on performance.

Variation in normalized plantarflexor tendon length ( $\rho$ ) also substantially affected performance in both explosive tasks, with smaller values corresponding to a shorter tendon and longer muscle fibers yielding better performance. Further analysis showed that a small  $\rho$  allowed fibers to operate closer to optimal fiber length during the movements. Interestingly, the muscle fiber contraction velocities were not lower for small  $\rho$ , suggesting that for this particular amount of variation, remaining closer to optimal fiber length might be more important than having a reduced contraction velocity. Configurations with small  $\rho$  were capable of producing slightly larger maximal force on average. Variation in  $\rho$  causes the isometric force length relation of a muscle to vary. Small  $\rho$  allows the muscle to produce maximal force at shorter muscle-tendon unit lengths, whereas large  $\rho$  requires muscle to be at a longer length the produce maximal force. During the final pushoff phase of the movement, as the muscles are shortening considerably, a muscle with a small  $\rho$  would be capable of producing larger forces (Zajac, 1989). Not many previous studies have considered the effect of the  $\rho$  as a differentiator in performance. Zajac (1989) analyzed the effect of small and large  $\rho$  on the force-length relation of muscles and found that larger  $\rho$  tend to widen the force-length curve, with maximal isometric force being produced at longer total muscle tendon unit length. It has also been suggested that small  $\rho$  are better for lifting large masses, whereas large  $\rho$  are better for lifting small masses (Nagano et al., 2004), although the computational model used by Nagano and colleagues (2004) only simulated linear motion of an isolated muscle lifting a single mass. When considering muscle fascicle or fiber lengths separately, results have consistently showed that sprinters have longer muscle fascicles compared with controls (Abe et al., 2000; Lee

and Piazza, 2009). This has also been investigated with use of a computational modelling study, which showed that longer fascicles are beneficial for fast concentric activities by allowing sarcomeres to experience lower relative shortening and therefore higher forces due to the force velocity relation (Nagano et al., 2007).

Variation in stiffness ( $k^*$ ) did not have a considerable effect on performance and results were not consistent for the two movements. For the vertical movement a stiffer tendon produced marginally better performance. Experimental results from previous studies have been inconclusive. Using a single-joint ankle jumping protocol similar to what our model employed, Kubo et al. (2007) found a significant correlation between jump height and Achilles tendon stiffness for 24 subjects. This is contrary to other experimental results (Bobbert, 2001) and computational modeling results (Zajac et al., 1984) that have shown more compliant tendons to be beneficial for performance in jumping movements. In the task where the model maximized horizontal acceleration from a standing start, lower stiffness improved performance. This is contrary to experimental results showing that sprinters have higher tendon stiffness compared with endurance runners and controls (Arampatzis et al., 2007). Our task was a unique single-joint pushoff movement that occurred from an initial upright stationary position, whereas during sprint start athletes are crouched during the initial phase of the movement and then continue moving upright after the initial acceleration. The fact that variation in stiffness had such a small effect on performance was surprising. The one standard deviation variation above or below nominal values that were used in this study might be too small to detect large differences in performance. Arampatzis et al. (2007) have found that sprinters have mean plantarflexor muscle normalized stiffness that far exceed a one standard deviation difference from non-active subjects' muscle stiffness (sprinters:  $37.2 \pm 12.1$  kN/strain; non-active:  $21.9 \pm 4.5$  kN/strain)

Toe length had almost no effect on performance in either of the explosive tasks. Variation in toe length however, has been shown to affect performance during experimental protocols. Sprinters have longer toes compared with controls (Baxter et al., 2012; Lee & Piazza, 2009), and a correlational study of single-joint jumping has shown that better jumpers had longer toes (see Chapter 4). Although toe length has the potential to keep the

foot on the ground longer, it also may affect the mechanical advantage of the plantarflexor muscles. Longer toes increase the output lever arm and potentially put the muscles at a disadvantage. The benefit of staying on the ground longer might be negated by this potential change in mechanical advantage. It may be that our current model is not detailed enough to fully capture the foot-ground interaction that occurs during movement and prevents examination of the possible role a longer toe plays in performance enhancement. Lee and Piazza (2009) employed a similar model with toe, foot, and leg segments and did find that longer toes improved contact time and horizontal impulse during a sprint like task. However, the model that was employed in that study did not have foot-contact at the heel and toe and the movement started from a non-zero initial velocity.

For the isometric maximal force production task, in which the model lifted the heel and held a maximal weight while maintaining static equilibrium, not all parameters affected the performance in the same manner as found for the explosive tasks. Longer heels produced better performance, allowing the plantarflexor muscle to operate at a length that was closer to optimal and consequently produce more force. Variation in heel length produced the largest effect on performance even though the difference in force production was higher for variations in  $\rho$  and  $k^*$ . This is because heel length also has a direct effect on torque production, with longer heels allowing for larger ankle torque. Baxter and Piazza (2014) have experimentally shown that larger plantarflexor moment arms produce larger plantarflexor joint torque in a group of healthy young men. Variation in  $\rho$  affected performance with small  $\rho$  increasing the weight supported. Again this is because fiber lengths were able to operate closer to optimal value for smaller  $\rho$  values and muscles were able to produce more force. A stiffer tendon enabled the model to support more weight by allowing the fiber to be closer to optimal fiber length and therefore enable larger force. This is consistent with experimental findings indicating that subjects with greater strength have stiffer Achilles tendons (Arampatzis et al., 2007; Muraoka et al., 2005). As with the other two tasks, variation in toe length had no effect on performance. This static task did not require transfer of force while the subject moved over the toes or onto the toes as in the dynamic tasks, and as such it was not expected to play a role in performance.

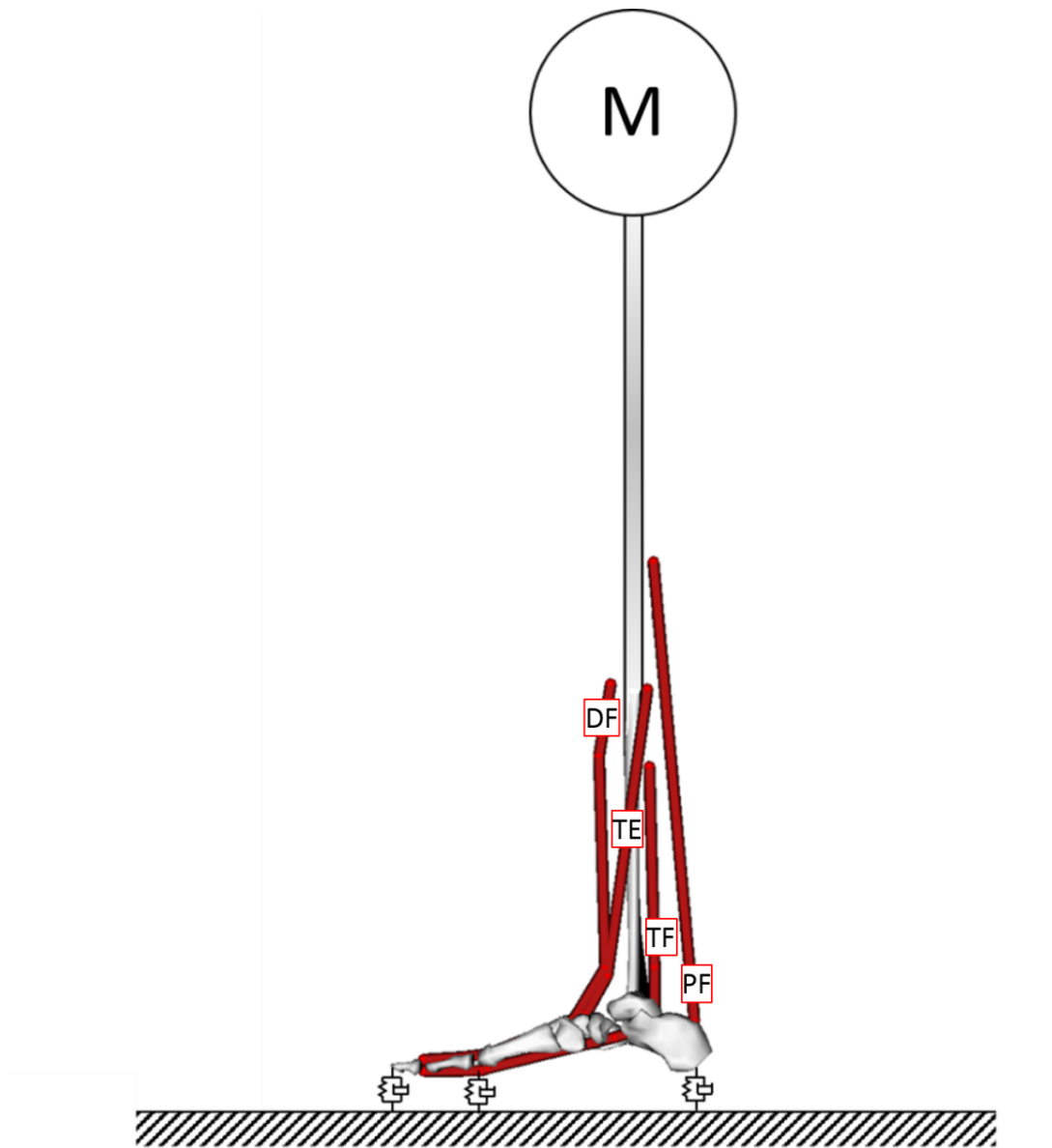


It is clear from the results that variation in certain architectural properties affected performance more than others for each simulated task. The variation in each parameter was set to one standard deviation above or below the nominal value, except for  $\rho$ , which was set at 10%. This similar amount of variation (10-13%) produced significantly different results. It was evident that variation in *lHeel* had the largest effect on performance across all tasks, followed by  $\rho$ , then  $k^*$  and finally *lToe*.

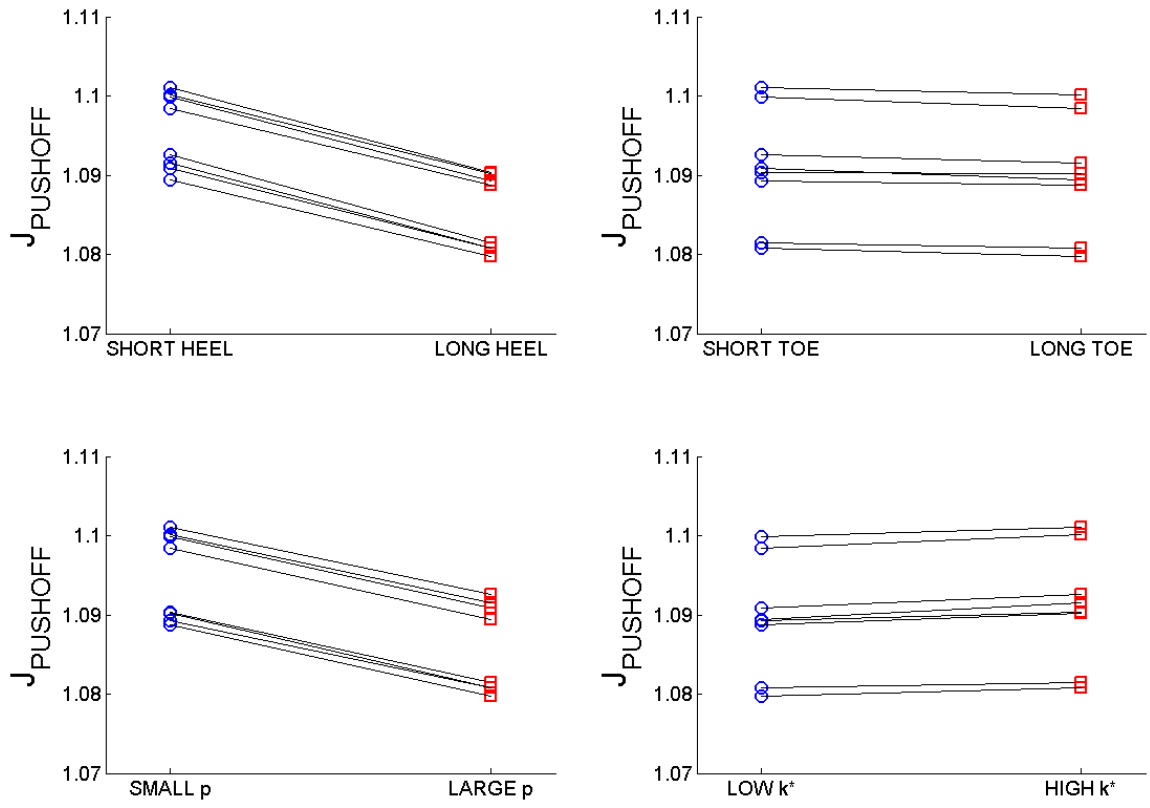
There were certain limitations associated with this study. This computational model is a simplified model of the human body with many fewer degrees of freedom. This approach was chosen to sharpen the focus on specific variables of interest. To the extent that the results can be extrapolated to whole-body movement, this should be done with caution. Not all foot and ankle architectural properties were included in this model. Maximal fiber shortening velocity, for example, was kept constant, and the effect of the plantar fascia was not considered. In this study muscle volume varied as muscle fiber length changes occurred. Changes in  $\rho$  caused variation in muscle fiber length, which meant that long fibered muscles had an increased muscle volume compared with short fibered muscles. Similarly, fiber pennation angle was dependent on muscle fiber length: longer fibers were less pennated, whereas shorter fibers exhibited an increased pennation angle. Keeping the muscle volume constant (by increasing isometric maximal force for shorter muscle fibers) and/or varying pennation angle independently from fiber length could potentially lead to different results. Rather than finding optimal solutions from a continuous range of parameter values, it was decided instead to reduce the complexity of the optimization by using only two values for each parameter, a high or a low value based on experimentally measured variation. It is possible that parameter values between these extremes could give better performance results. Use of a lumped muscle model prevents consideration of differences between muscles like the gastrocnemius and the soleus. Both of these muscles are plantarflexors, but there are important differences between these muscles in their architecture and fiber composition.

In conclusion, the computational model used in this study showed how variation in certain musculoskeletal properties of the foot and ankle affect performance in three different motor tasks. Heel length, which predicts plantar flexor moment arm, and

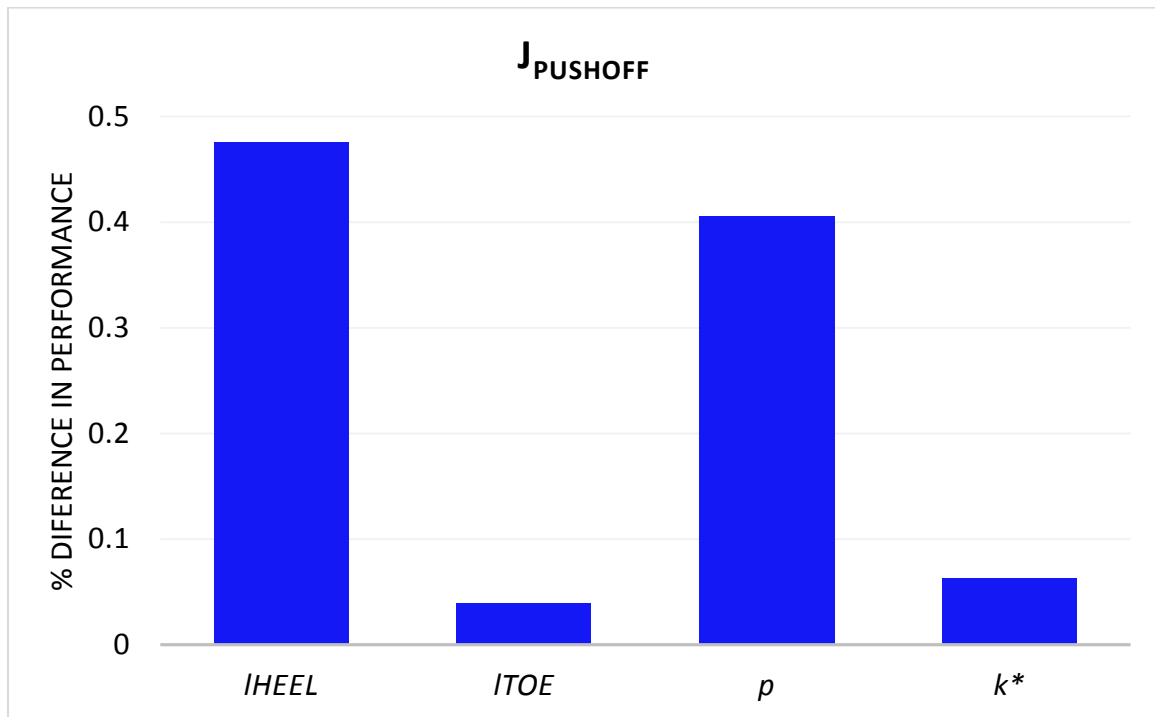
normalized tendon length variation significantly influenced performance in all tasks, but variation in tendon stiffness and toe length did not influence performance as strongly. Future studies should include a more complex foot and toe model that could predict how toe length could change performance, a finding suggested by experimental studies. The current model could potentially be used to investigate other tasks, for example an aerobic endurance activity, to determine which structural properties are most relevant to such activities.



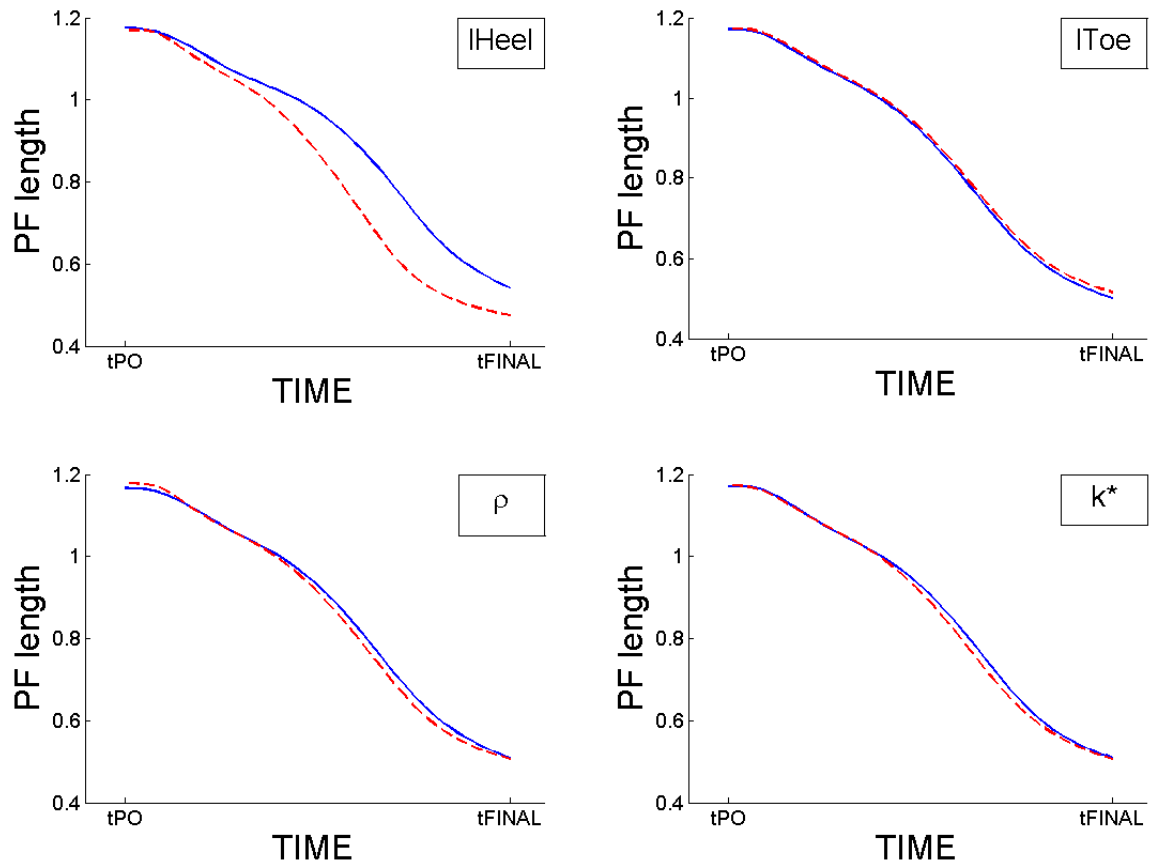
**Figure 6-1:** Planar computational model showing the bodies and four lumped muscle-tendon actuators (plantarflexor – PF; dorsiflexor – DF; toe flexor – TF; toe extensor – TE). The three ground-contact points was modeled with a spring-damper systems underneath the tip of the toe segment, below the MTP joint as well as below the insertion point of the plantarflexor muscle. Model figure is a modified version of the model by Arnold et al (2010) from OpenSim.



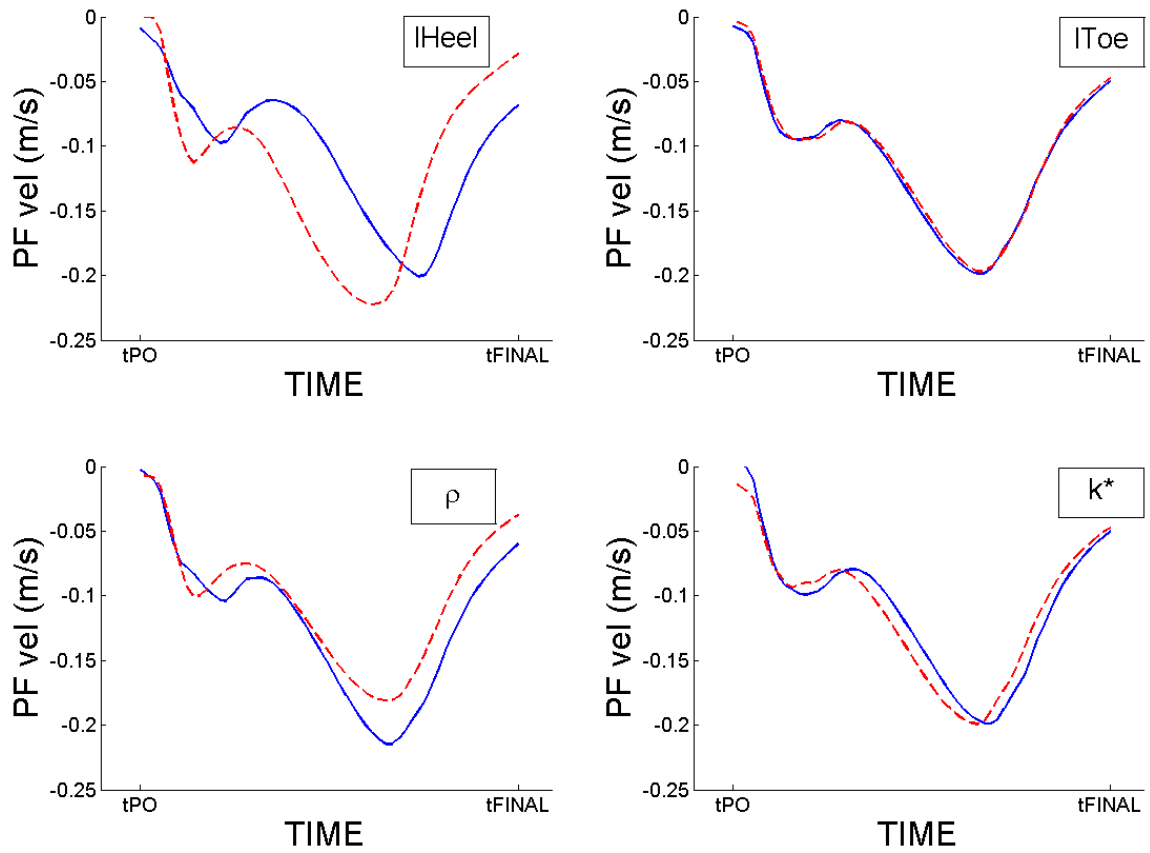
**Figure 6-2:** Maximal vertical-energy performance ( $J_{\text{PUSHOFF}}$ ) grouped according to LOW (blue circles) and HIGH (red squares) parameter values for all 4 parameters. Black lines are drawn between matching pairs of parameter combinations. Matching pairs are results where only the one parameter of interest is different –  $l_{\text{Heel}}$  in this case of the top left figure. For example, the result for the short  $l_{\text{Heel}}$ , short  $l_{\text{Toe}}$ , small  $\rho$ , low  $k^*$  is matched with the result for the long  $l_{\text{Heel}}$ , short  $l_{\text{Toe}}$ , small  $\rho$ , low  $k^*$ . Variation in  $l_{\text{Heel}}$  and  $\rho$  had the biggest effect on performance, evident from the larger slope of the lines.



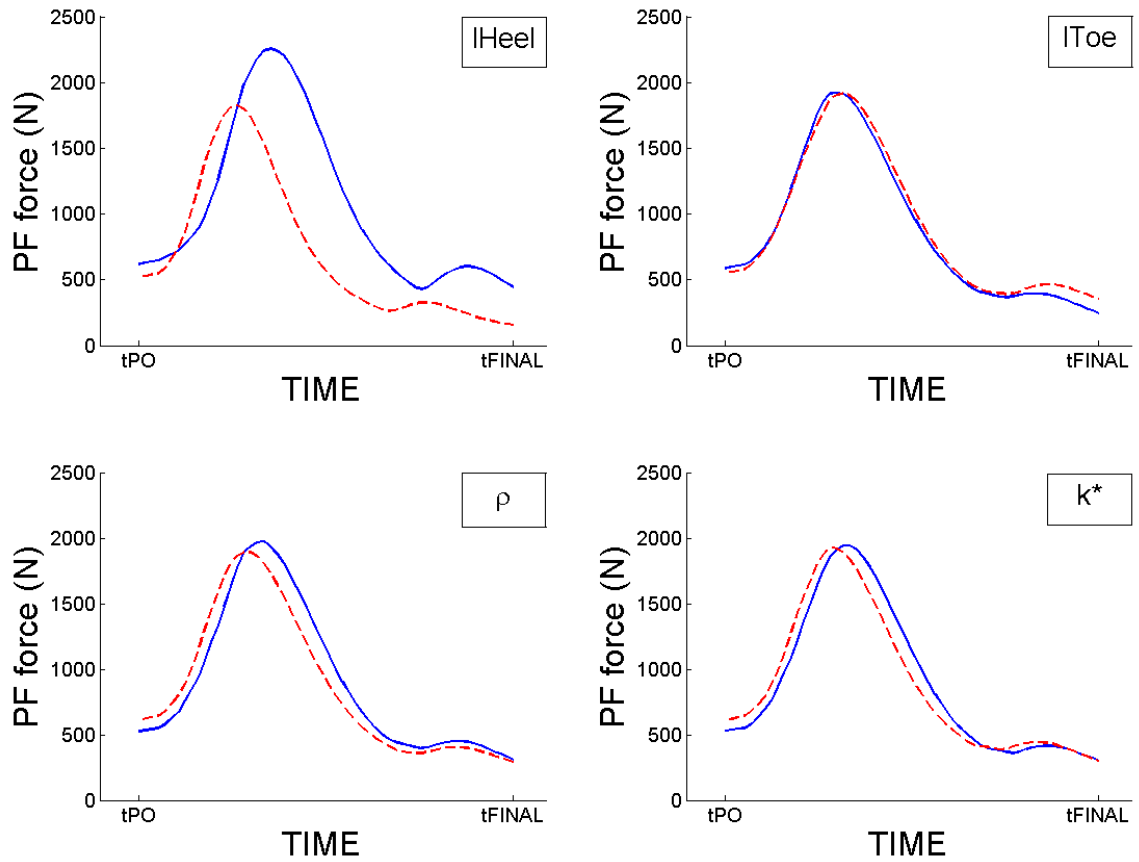
**Figure 6-3:** Percent difference in  $J_{PUSHOFF}$  performance relative to nominal performance due to one SD variation in the different parameters. Mean performance difference in simulations with short heel lengths compared with long heel lengths show the highest difference (~0.5%), followed by variation in  $\rho$ . Much smaller differences in performance occurred with variation  $k^*$  and  $I_{Toe}$ .



**Figure 6-4:** Plantarflexor fiber operating length normalized to optimal fiber length throughout the pushoff phase (tPO to tFinal) of the movement. The results shown are for the optimal  $J_{PUSHOFF}$  simulations for architectures with LOW (solid blue line) and HIGH (dashed red line) parameter values. In the top left figure the solid blue line, indicating configurations with short  $l_{Heel}$  values, stayed closer to the optimal fiber length of 1.0 compared with the dashed red line which shows configurations with a long  $l_{Heel}$ . Differences in fiber operating length due to variation in the other parameters are less evident.

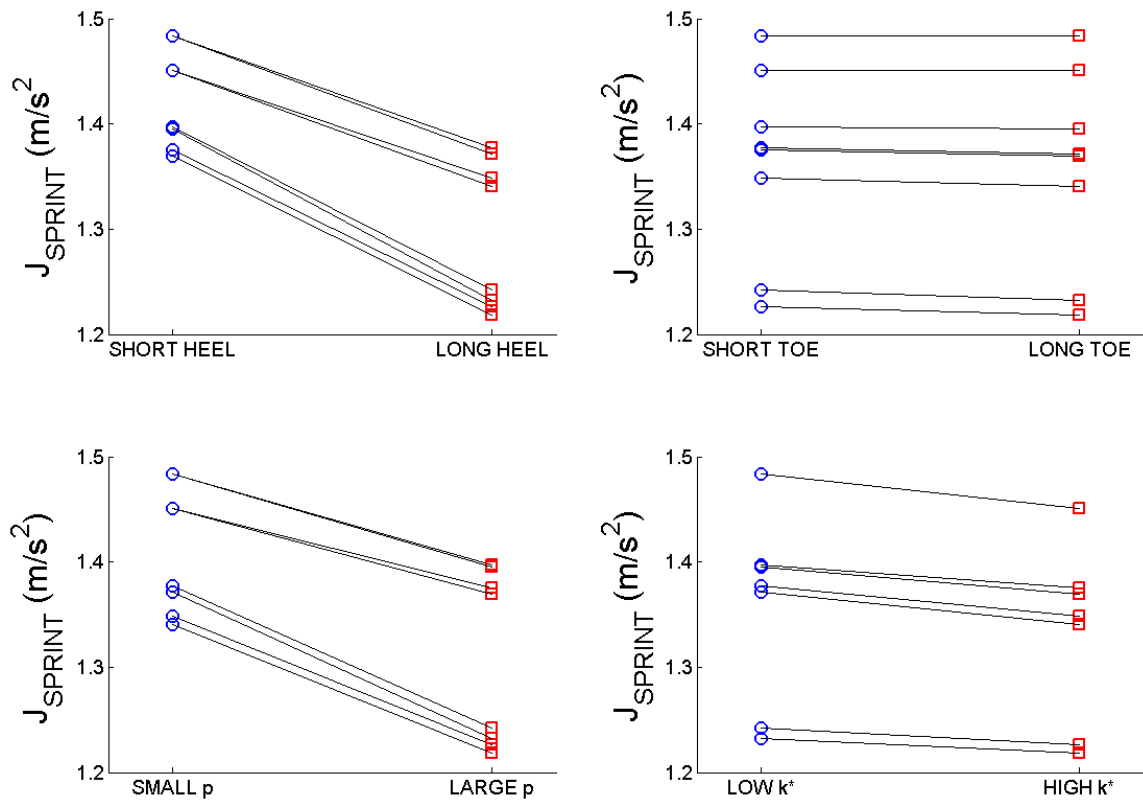


**Figure 6-5:** Plantarflexor fiber velocity throughout the pushoff phase (tPO to tFinal) of the movement for optimal  $J_{\text{PUSHOFF}}$  simulations for architectures with LOW (solid blue line) and HIGH (dashed red line) parameter values. A short  $l_{\text{Heel}}$  configuration resulted in a lower maximal contraction velocity whereas a smaller  $\rho$  had a higher maximal contraction velocity. Differences in contraction velocities due to variation in  $k^*$  and  $l_{\text{Toe}}$  were minimal.

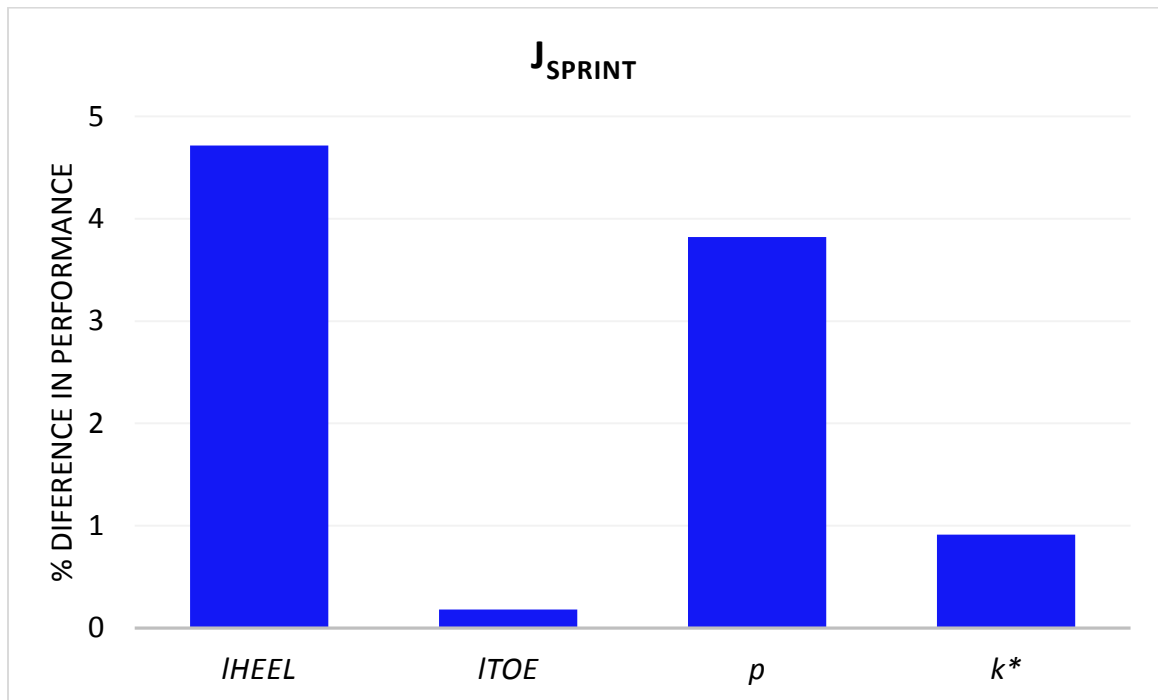


**Figure 6-6:** Plantarflexor muscle force throughout the pushoff phase ( $t_{PO}$  to  $t_{FINAL}$ ) of the movement for optimal  $J_{PUSHOFF}$  simulations for architectures with LOW (solid blue line) and HIGH (dashed red line) parameter values. Shorter heels were capable of producing a much larger peak force. Differences in force production due to variation in  $k^*$  and  $l_{Toe}$  were minimal.

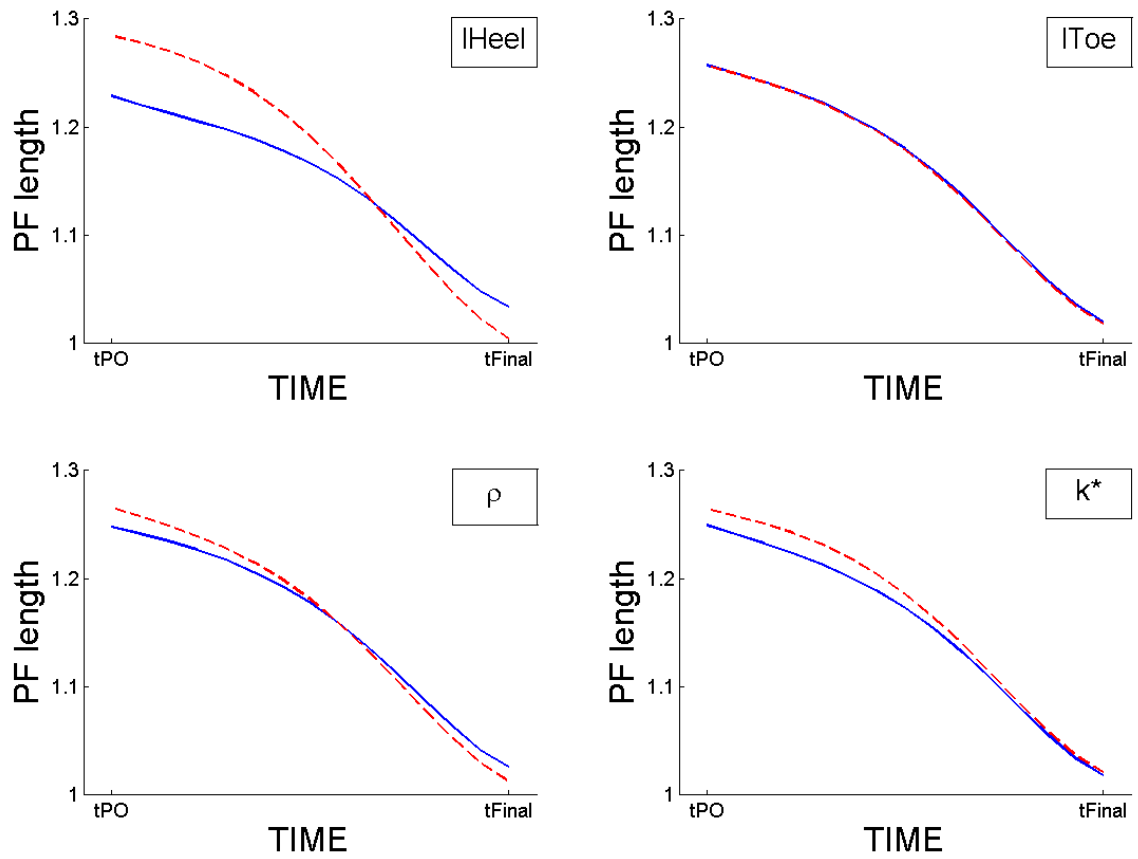




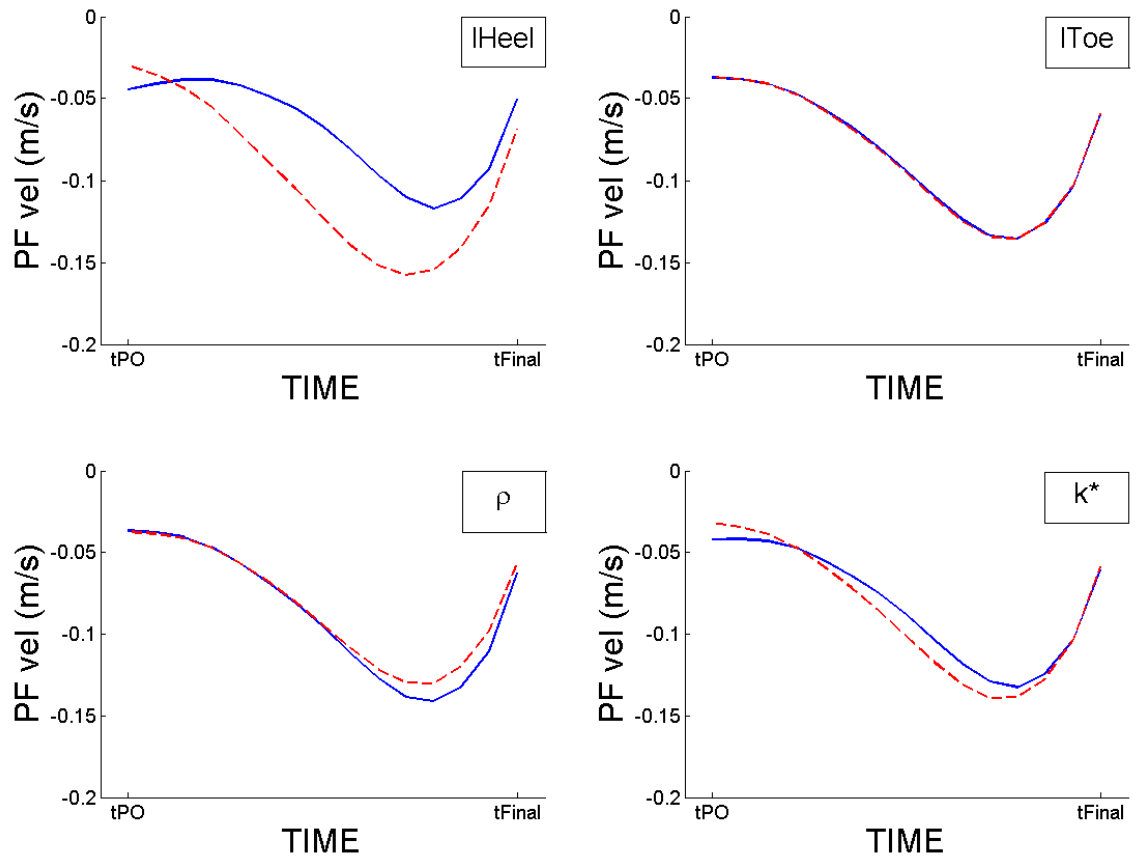
**Figure 6-7:** Maximal average acceleration performance ( $J_{\text{SPRINT}}$ ) grouped according to LOW (blue circles) and HIGH (red squares) parameter values for all 4 parameters. Black lines are drawn between matching pairs of parameter combinations. Variation in  $l_{\text{Heel}}$  and  $\rho$  had the biggest effect on performance, evident from the larger slope of the lines.



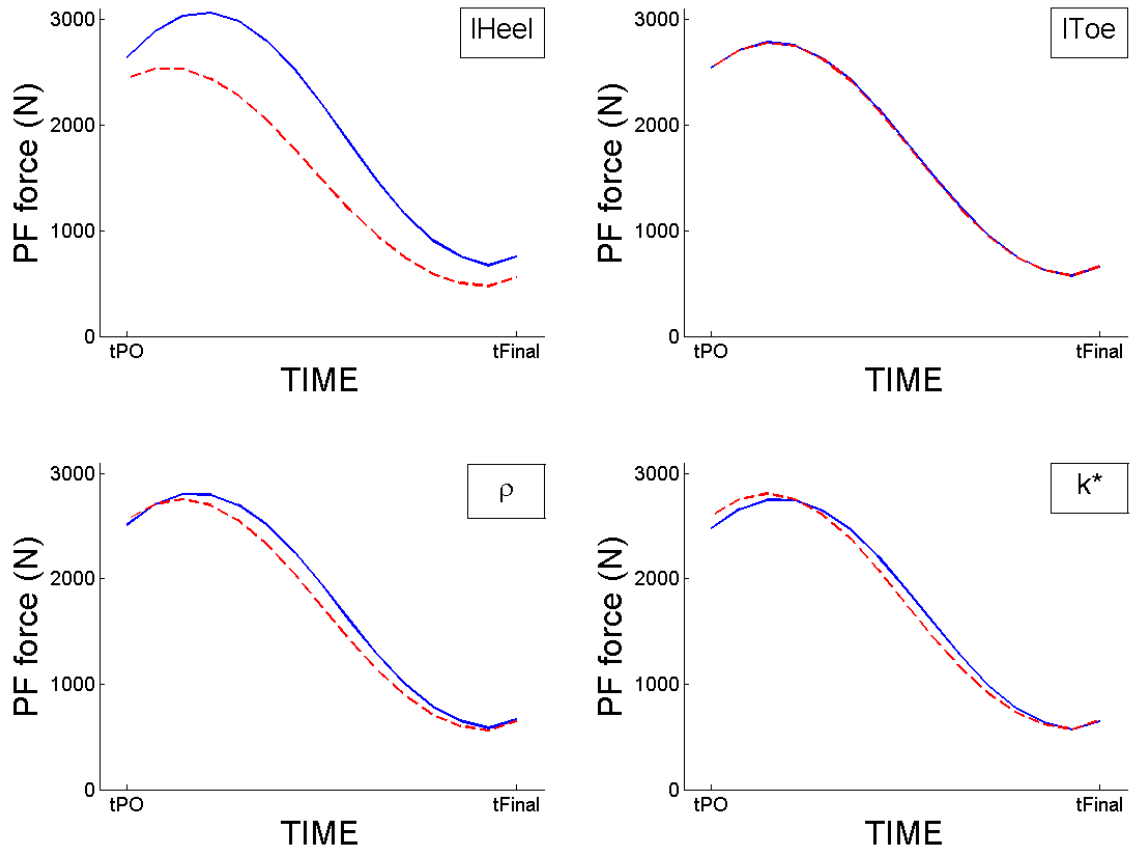
**Figure 6-8:** Percent difference in  $J_{SPRINT}$  performance due to variation in the different parameters. Mean performance difference in simulations with short heel lengths compared with long heel lengths show the highest difference (~4.5%), followed by variation in  $\rho$ ,  $k^*$ , and  $l_{Toe}$ .



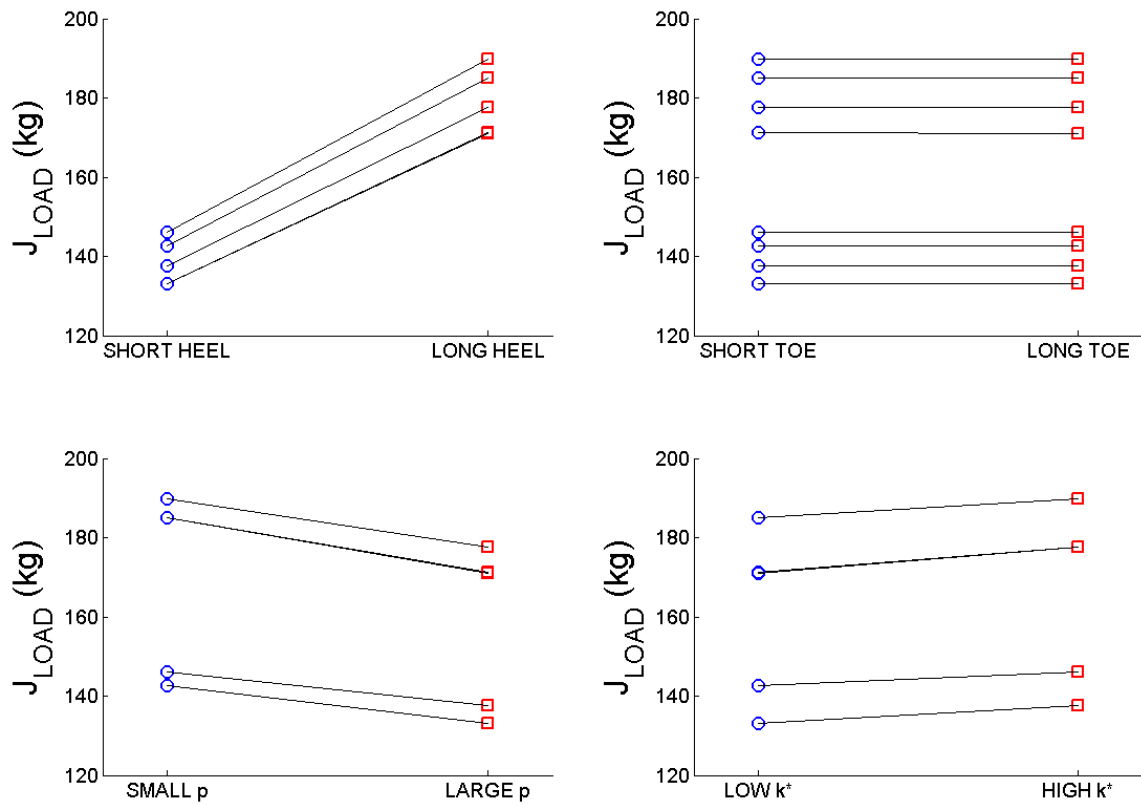
**Figure 6-9:** Plantarflexor fiber operating length throughout the pushoff phase ( $t_{PO}$  to  $t_{Final}$ ) of the movement for optimal  $J_{SPRINT}$  simulations for architectures with LOW (solid blue line) and HIGH (dashed red line) parameter values. A short heel configuration stayed closer to optimal fiber operating range, as did simulations with small  $\rho$  and low  $k^*$  to a lesser extent.



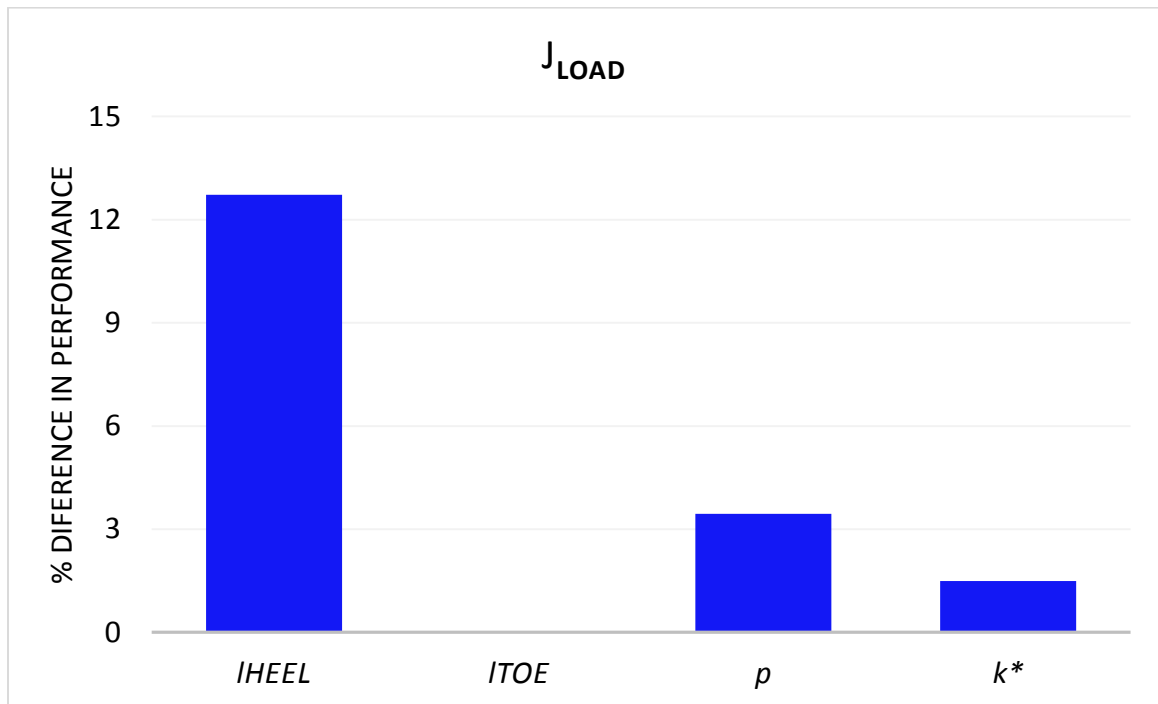
**Figure 6-10:** Plantarflexor fiber velocity throughout the pushoff phase ( $t_{PO}$  to  $t_{Final}$ ) of the movement for optimal  $J_{SPRINT}$  simulations for architectures with LOW (solid blue line) and HIGH (dashed red line) parameter values. A short  $l_{Heel}$  meant lower fiber contraction velocity, as did low  $k^*$ . Small  $\rho$  produced higher contraction velocities.



**Figure 6-11:** Plantarflexor muscle force throughout the pushoff phase (tPO to tFinal) of the movement for optimal J<sub>SPRINT</sub> simulations for architectures with LOW (solid blue line) and HIGH (dashed red line) parameter values. Short heels and small  $\rho$  produced higher peak plantarflexor force. Differences in force production due to variation in  $k^*$  and  $l_{Toe}$  were minimal.



**Figure 6-12:** Maximal static load support performance ( $J_{LOAD}$ ) grouped according to LOW (blue) and HIGH (red) parameter values for all 4 parameters. Black lines are drawn between matching pairs of parameter combinations. Variation in  $l_{Heel}$  had the largest effect, with variation in  $\rho$  and  $k^*$  also showing some differences in slope.



**Figure 6-13:** Percent difference in  $J_{LOAD}$  performance relative to nominal performance due to one SD variation in the different parameters. Mean performance difference in simulations with short heel lengths compared with long heel lengths show the highest difference (~13%), followed by variation in  $\rho$  and  $k^*$ .  $I_{Toe}$  showed minimal effect (< 0.002%)

**Table 6-1:** Inertial properties for all the segments of the model

Body	m (kg)	$I_{zz}$ (kg.m <sup>2</sup> ) about CoM
Toes	0.2166	0.0010
Foot	1.2600	0.0041
Leg	0.0000 <sup>a</sup>	0.0000 <sup>a</sup>
Rest of Body <sup>1</sup>	36.0234	0.0000 <sup>b</sup>

<sup>1</sup> – Rest of body mass calculated by subtracting toe, foot and leg mass from body mass, which was half of 75 kg

<sup>a</sup> – Leg segment was massless and therefore had no MoI

<sup>b</sup> – Rest of body was a point mass with no MoI



**Table 6-2:** Lumped muscle architectural parameters used in the model.

<i>Muscle architectural properties</i>					
	$lF_{OPT} (m)$	$lT_{SLACK} (m)$	$f_{MAX} (N)$	$PA (^\circ)$	$k^*$
Muscle					
Plantarflexor	0.0478 <sup>x</sup>	0.3300 <sup>x</sup>	5456.4	21.4	1.0 <sup>x</sup>
Dorsiflexor	0.0683	0.2390	677.5	9.6	1.0
Toe Flexor	0.0499	0.3635	2740.0*	15.8	1.0
Toe Extensor	0.0712	0.3552	502.5	10.3	1.0

$lF_{OPT}$  – Optimal fiber length;  $lT_{SLACK}$  – Tendon slack length;  $f_{MAX}$  – Maximal isometric force;  $PA$  – Pennation angle;  $k^*$  – Plantarflexor stiffness scaling factor

<sup>x</sup> Values shown for the plantarflexor are the nominal values. During the optimization process these values were varied to investigate the effect of variation on performance.

**Table 6-3:** Nominal, low and high values used to simulate variation in ankle and foot architectural parameters

	NOMINAL	LOW	HIGH
$l_{Heel} (m)^1$	0.0438	0.0381	0.0495
$k^{*2}$	1.00	0.87	1.13
$\rho^3$	6.904	6.210	7.587
$l_{Toe} (m)^4$	0.0700	0.0616	0.0784

<sup>1</sup> – CV of 13% (Lee & Piazza, 2009)

<sup>2</sup> – CV of 13% (Maganaris & Paul, 2002). Variation in  $k^*$  was based on measures of Young's modulus for the plantarflexor tendon.

<sup>3</sup> – Range of 10% (Zajac, 1989). These values were not based of CV values, which were not available from the literature. Zajac (1989) found  $\rho$  values of 9 for the gastrocnemius and 11 for the soleus. This range of  $\pm 10\%$  was used to define the low and high values.

<sup>4</sup> – CV of 12% (Lee & Piazza, 2009)

**Table 6-4:** The effect of variation on the four different architectural parameters on plantarflexor fiber length normalized to optimal fiber length ( $lPF$ ) and plantarflexor force ( $fPF$ ) during the maximal static load support ( $J_{LOAD}$ ).

	Short $l_{Heel}$	Long $l_{Heel}$	Short $l_{Toe}$	Long $l_{Toe}$	Small $\rho$	Large $\rho$	Low $k^*$	High $k^*$
$lPF$	0.8270	0.8019	0.8148	0.8141	0.8253	0.8036	0.7987	0.8302
$fPF$ (N)	4580	4497	4540	4538	4676	4402	4485	4593

## **Chapter 7**

### **Discussion**

#### **7.1 Summary**

The purpose of this dissertation was to investigate ankle and foot structural properties that allow for optimal performance in various motor tasks. To that end both computational modeling and experimental frameworks were employed.

The goal of our first study was to create a computational model to study maximal-height single-joint jumping in which only the ankle muscles were used for propulsion and to compare performance results from the computational model to results from experimental jumping of human subjects. Results from our simulation suggested that bouncing was the optimal strategy for this single-joint jumping task, and this was confirmed experimentally when the best jumpers also jumped higher when they employed a bouncing strategy (Figure 3-2 and Figure 3-3). Further analysis revealed that when non-successful jumpers employed a bouncing strategy for jumping, they used a bouncing frequency lower than the successful jumpers employed and the computational model suggested (Table 3-2). An optimal bouncing strategy potentially allowed for increased energy storage in the tendon (Figure 3-4). Comparison with results from metabolic bouncing studies suggest that resonance of the plantarflexor muscles might have been used to improve performance in this maximal-height explosive task. The use of resonance for maximizing performance in an explosive type of activity has not been previously reported. Whether the human body can take advantage of resonant properties of muscles in multijoint tasks should be explored.

The second study was experimental in nature and focused on variation of the foot and ankle structure and its effect on performance in a maximal-height single-joint jumping task. Previous work using experimental approaches has consistently shown that sprinters benefit from having shorter heels and longer toes (Baxter et al., 2012; Lee and Piazza, 2009). Whether foot and ankle structural variation also affect performance in maximal-height jumping was of interest. In this study no countermovement or bouncing was allowed and the subjects performed a single upward thrust. Various anthropometric measures were

taken and significant correlations were found between jump height and lateral heel length and as well as between jump height and hallux length (Figure 4-2 and Figure 4-3). This implied that the subjects with longer heels and longer toes jump highest. Our data suggested that these subjects (with longer heels and toes) were not necessarily the larger subjects, or the subjects with bigger feet (Table 4-2). Results from this study did not support the idea that short heels allow for better performance in explosive types of activity, and idea that is suggested by previous studies comparing sprinter and non-sprinter foot morphology. More research is needed to explain how variation in heel length affects explosive movements.

The third study aimed to explain the mechanism by which shorter heels could potentially reduce metabolic cost in running. Correlational studies have found a relationship between heel length and rate of oxygen consumption in distance runners (Raichlen et al., 2011; Scholz et al., 2008). Scholz et al. (2008) proposed that for a given joint moment shorter heels will increase muscle force, which should increase tendon elastic energy storage and return and thus lead to a reduction in metabolic cost. Although an inverse relationship was found between heel length and estimated muscle force (Figure 5-4), the increase in muscle force was not accompanied by a reduction in rate of oxygen consumption. Neither heel length nor peak Achilles tendon force, or any other of the kinematic or kinetic variable measured, were significantly correlated with rate of oxygen consumption during running (Table 5-1, Table 5-2 and Table 5-3). Even if short heels increased tendon force, it is possible that higher tendon force requires higher active muscle force, which would cause an increase in metabolic cost, potentially negating the positive effects of increased energy storage and return.

In the fourth study we employed a computer model to investigate how variation in foot and ankle structure influences performance in three different motor tasks: maximal vertical-energy pushoff, maximal horizontal acceleration, and maximal static load support. To our knowledge there has been no previous work investigating how variation in multiple foot and ankle structural parameters in combination influence performance in multiple tasks. Of all the structural parameter variations investigated (heel length, toe length, normalized tendon length, tendon stiffness), it was heel length variation that most influenced performance across all tasks, followed by variation in normalized tendon length

(Figure 6-3, Figure 6-8, and Figure 6-13). Variation in stiffness had a very small influence on performance and toe length did not appear to influence performance. For both explosive tasks (vertical-energy and horizontal acceleration) shorter heels were beneficial, allowing for fiber operating lengths closer to optimal and lower contraction velocities. For the isometric task longer heels allowed for larger muscle force due to fiber operating lengths closer to optimal as well as increased torque production. For both explosive and isometric tasks, small normalized tendon length appeared to be advantageous, potentially due to the fact that it allowed the fibers to operate closer to the optimal fiber length. High stiffness values improved performance in the vertical-energy and isometric heel lift task, whereas low stiffness improved the horizontal acceleration performance, although the improvements were very small. Our findings related to heel length and normalized tendon length compare well with experimental studies that contrast differences between athletes and non-athletes in similar types of activities. Athletes have shorter heel lengths and longer muscle fascicles, due to small normalized tendon lengths, than non-athletes (Abe et al., 2000; Lee & Piazza, 2009). Experimental study results are less clear on the effect of stiffness on performance in explosive tasks, but in static maximal load tasks which requires maximal isometric force production it is evident that larger stiffness is beneficial (Arampatzis et al., 2007), similar to what was found in the current study.

Results from our second and fourth studies seem to contradict one another: Our maximal height single-joint jumping task results suggest that longer heels are advantageous, whereas our computational model results show the opposite: Shorter heels are beneficial for optimal performance in a task in which vertical-energy is maximized. Other experimental work has shown that volleyball players have shorter moment arms than controls and jump higher than controls (Watanabe et al., 2008). For other explosive tasks, like sprinting, differences in heel length have been found between sprinters and non-sprinters, with sprinters having smaller plantarflexor moment arms (Baxter et al., 2012; Lee and Piazza, 2009). It seems that only studies focusing on differences between athletes and non-athletes have shown differences in the size of the plantarflexor moment arm and related this difference to the activities of the populations being compared. To our knowledge, no correlations have yet been reported indicating that smaller plantarflexor

moment arms allow for increased performance in an explosive type of activity. Our jumping results clearly show that subjects with longer heels (larger plantarflexor moment arms) jumped higher (Figure 4-2). Similarly, Baxter and Piazza (2014) used a protocol involving isokinetic torque production and found a significant correlation between torque production and plantarflexor moment arm at various speeds of movement: Larger, not smaller plantarflexor moment arms appear to produce more torque in subjects who are not trained athletes.

Difference in subject populations might explain the apparent conflicting results presented in Chapters 4 and 6. Subjects in both our jumping study and in the Baxter and Piazza study (2014) were non-athletes. Could it be that different musculoskeletal properties allow for optimal performance in different populations? Could the within-human variability, whether due to genetic variation or adaptation due to training (or lack thereof), allow different groups to use the musculoskeletal structure differently to perform the same function? The animal kingdom clearly shows how variation in structure allows for differences in performance due to interspecies variability. In the animal kingdom there are different animals that are well adapted for sprinting, running, and digging, but within the single human species we could potentially have good sprinters, runners, and diggers who show variation in structure optimized for their specific activity. This idea should be further investigated.

## **7.2 Future Work**

With this dissertation we tried to better understand the influence of variation in foot and ankle architecture on performance in a variety of tasks. Although this body of work enhanced our understanding, some questions remain and some interesting findings from this research warrant further investigation.

The fact that muscle resonance could be beneficial in an explosive maximal performance task has not been previously shown. Although the single-joint jumping framework is unique and not employed during everyday life, it would be interesting to investigate whether multiple countermovements or bouncing, which makes use of

resonance factors, could potentially improve performance in tasks other than maximal height single-joint jumping.

Analysis of muscle activation during the single-joint jumping tasks revealed that dorsiflexor muscle activity occurs at the end of the plantarflexion movement while plantarflexor muscles were active. Our results suggest that this antagonist (dorsiflexor) activation during a maximal plantarflexor activity might allow for increased muscle force production by slowing down agonist muscle contraction velocity. Muscle cocontraction has been historically been thought of as a means of stabilizing joints and protecting joints against injury. The idea that cocontraction can enhance the function of the agonist muscle by modulation of muscle contraction dynamics needs further attention.

Results from our simulation study failed to find any connection between toe length variation and performance. Experimental results have shown that different populations (sprinters vs. non-sprinters) show differences in toe length. It might be that our model of the toe was too simplistic to correctly capture what happens during foot-contact. Creating a better foot model to investigate how variation in toe length might influence performance should be explored.

The different tasks that were simulated in our investigation of variation on performance were two explosive maximal tasks and an isometric maximal task. A more metabolic or aerobically taxing task was not simulated. Experimental results have shown that foot and ankle structure variation might influence performance in metabolically demanding tasks. Defining a metabolic task for the current model and simulating the task could shed light on the influence of heel length on muscle force production and tendon energy storage.

### **7.3 Conclusion**

As a result of this dissertation we know that a unique bouncing technique in which muscles possibly make use of resonance and increased elastic energy storage increases jump height in a single-joint jumping task. We have shed some doubt on the concept that shorter heels allows for better running economy. Finally, results from our single-joint



jumping study and simulation study of a similar maximal vertical-energy pushoff task contradict each other. We cannot definitively say whether long or short heels are better for explosive types of activities. Variable human adaptation for different activities might allow different people to utilize their musculoskeletal structure differently. It is clear that the mechanisms by which variation in foot and ankle structure affect performance across different tasks and for different subject groups are not yet fully understood.

## Appendix A

### Model Coordinate Systems and Coordinates (XYZ)

All values are given for the nominal model.

Location of the TOE relative to the world:

- TOE origin: (0.0000 0.0000 0.0000)

Toe coordinate system (ToeCS) located at TOE origin

- Locations defined in ToeCS:
  - TOE contact: (0.0000 0.0000 0.0000)

Location of MTP joint center relative to TOE origin

- MTP: (0.0700 0.0000 0.0000)

Foot coordinate system (FCS) located at MTP joint

- Locations defined in FCS:
  - MTP contact: (0.0000 0.0000 0.0000)
  - HEEL contact: (0.1738 0.0000 0.0000)

Location of the ANKLE joint center (AJC) relative to FCS

- ANKLE: (0.1300 0.0440 0.0000)

Locations defined relative to the AJC:

- HIP: (0.0000 1.0000 0.0000)

Location of the tibial coordinate system (TibCS) relative to AJC

- TIBIA: (-0.0100 0.400 0.0000)

**Table A-1:** Muscle path definitions including origins (O), insertions (I), and via points.

MUSCLE	COORDINATES			REFERENCE FRAME	
Plantarflexor					
PF_P1 (O)	(	0.0200	-0.0300	0)	TibCS
PF_P2 (I)	(0.1688	0.0320	0)		FCS
Dorsiflexor					
DF_P1 (O)	(-0.0154	-0.1312	0)		TibCS
DF_P2	(-0.0251	-0.1906	0)		TibCS
DF_P3	(-0.0200	-0.3660	0)		TibCS
DF_P4 (I)	(0.0598	0.0180	0)		FCS
Toe Flexor					
TF_P1 (O)	(0.0166	-0.1997	0)		TibCS
TF_P2	(0.0209	-0.3658	0)		TibCS
TF_P3	(0.1358	0.0280	0)		FCS
TF_P4	(0.1030	0.0180	0)		FCS
TF_P5	(-0.002	-0.0076	0)		FCS
TF_P6	(0.0667	-0.0076	0)		ToeCS
TF_P7 (I)	(0.0198	-0.0080	0)		ToeCS
Toe Extensor					
TE_P1 (O)	(0.0150	-0.1354	0)		TibCS
TE_P2	(-0.0182	-0.3727	0)		TibCS
TE_P3	(0.0793	0.0374	0)		FCS
TE_P4	(-0.0005	0.0045	0)		FCS
TE_P5	(0.0545	0.0038	0)		ToeCS
TE_P6 (I)	(0.0197	0.0015	0)		ToeCS

## Appendix B

### Lumped Muscle Parameter Identification

Muscle parameters for the lumped muscles used in this study was computed using a weighted average of the muscle volume of the original muscles from which the lumped muscles were built up of. Fukunaga et al. (2001) have found a strong correlation between muscle volume and joint torque production. The following steps were employed:

- STEP 1:
  - Calculate contribution (CONT) of each muscle to overall muscle volume of combined muscle. See Table B-1.
- STEP 2:
  - Calculate lumped muscle optimal fiber length ( $l_{FOPT}$ ) using the same contribution of each muscle as defined by volume contribution.
- STEP 3
  - Repeat step 2 for pennation angle (PA)
- STEP 4
  - Repeat step 2 for tendon slack length ( $l_{TSLACK}$ ). Further analysis of the model characteristics in OpenSim showed that calculating  $l_{TSLACK}$  using this method changed the passive force-length relationship significantly from the relationship due to the summed contribution of the constituent muscles.  $l_{TSLACK}$  was modified to match summed passive tendon force-length curve from Arnold model more closely.
- STEP 5
  - Calculate physiological cross sectional area (PCSA) for newly defined lumped muscle:
    - $PCSA = Volume / l_{FOPT}$
- STEP 6
  - Calculate  $f_{MAX}$ 
    - $f_{MAX} = PCSA * 61 \text{ N/cm}^2$  - using the value of specific tension as proposed by Delp (1990)

**Table B-1:** Lumped plantarflexor muscle parameter estimation

	Volume	CONT (%)		lF <sub>OPT</sub> (cm)	PA (°)	lT <sub>SLACK</sub> (cm)		PCSA (cm <sup>2</sup> )	f <sub>MAX</sub> (N)
Soleus	261.2	61		4.40	28.3	28.2			
Medial gastrocnemius	107.5	25		5.10	9.9	40.1			
Lateral gastrocnemius	58.9	14		5.88	12.0	38.2			
<b>TOTAL or WEIGHTED AVE</b>	<b>427.6*</b>	<b>100*</b>		<b>4.78~</b>	<b>21.4~</b>	32.6~ <b>33.0<sup>#</sup></b>		<b>89.4</b>	<b>5456</b>

\* Summed value

~ Weighted average based on each muscle's volume contribution (see text for details)

<sup>#</sup> lT<sub>SLACK</sub> value used in model. This value was modified from original weighted average to better match the summed passive tendon force-length curve from the Arnold model (see text for details)

**Table B-2:** Lumped dorsiflexor muscle parameter estimation

	Volume	CONT (%)		lF <sub>OPT</sub> (cm)	PA (°)	lT <sub>SLACK</sub> (cm)		PCSA (cm <sup>2</sup> )	f <sub>MAX</sub> (N)
Tibialis anterior	80.1	100		6.83	9.6	24.1			
<b>TOTAL or WEIGHTED AVE</b>	<b>80.1</b>	<b>100</b>		<b>6.83</b>	<b>9.6</b>	24.1~ <b>23.9<sup>#</sup></b>		<b>11.1</b>	<b>678</b>

<sup>#</sup> lT<sub>SLACK</sub> value used in model. This value was modified from original weighted average to better match the summed passive tendon force-length curve from the Arnold model (see text for details)

**Table B-3:** Lumped toe flexor muscle parameter estimation

	Volume	CONT (%)		IF <sub>OPT</sub> (cm)	PA (°)	IT <sub>SLACK</sub> (cm)		PCSA (cm <sup>2</sup> )	f <sub>MAX</sub> (N)
Flex. hallucis longus	36.8	66		5.27	16.9	35.6			
Flex. digitorum longus	19.2	34		4.46	13.6	37.8			
<b>TOTAL or WEIGHTED AVE</b>	<b>56.0*</b>	<b>100*</b>		<b>4.99~</b>	<b>15.8~</b>	<b>36.4~</b>		<b>11.1</b>	<b>685 2740<sup>#</sup></b>

\* Summed value

~ Weighted average based on each muscle's volume contribution (see text for details)

<sup>#</sup> f<sub>MAX</sub> value used in model due to inability of original toe flexor muscle properties to create experimentally comparable joint torques (Goldmann and Brüggemann, 2012)

**Table B-4:** Lumped toe extensor muscle parameter estimation

	Volume	CONT (%)		IF <sub>OPT</sub> (cm)	PA (°)	IT <sub>SLACK</sub> (cm)		PCSA (cm <sup>2</sup> )	f <sub>MAX</sub> (N)
Ext. hallucis longus	19.8	34		7.48	9.4	33.2			
Ext. digitorum longus	38.8	66		6.93	10.8	36.7			
<b>TOTAL or WEIGHTED AVE</b>	<b>58.6*</b>	<b>100*</b>		<b>7.12~</b>	<b>10.3~</b>	<b>35.5~</b>		<b>8.2</b>	<b>502.5</b>

\* Summed value

~ Weighted average based on each muscle's volume contribution (see text for details)

## Appendix C

### Penalty Function Descriptions

#### 1. Initialization objective function: $J_{INIT}$

The initial position for both the maximal vertical-energy pushoff task as well as the maximal average horizontal acceleration task was found by minimizing:

$$J_{INIT} = P_{FORCE} + P_{STRAIN} + P_{ACCEL} + P_{HEEL}$$

The penalty terms were defined and equated as follows:

- $P_{FORCE}$  ensured static equilibrium in the muscles:

$$P_{FORCE} = 1e6 * (fPF_{EQ}^2 + fDF_{EQ}^2 + fTF_{EQ}^2 + fTE_{EQ}^2)$$

$$\text{with } fPF_{EQ} = fPF_{TENDON} - (aPF_{INIT} * fPF_{ACTIVE} + fPF_{PASSIVE})$$

Definitions:

$fPF_{EQ}$	test for equal force in plantarflexor (PF) muscle and tendon
$fPF_{TENDON}$	PF tendon force
$aPF_{INIT}$	initial PF muscle activation
$fPF_{ACTIVE}$	active PF muscle force
$fPF_{PASSIVE}$	passive PF muscle force

. . . similar equation for dorsiflexor, toe flexor and toe extensor

- $P_{STRAIN}$  penalized initial tendon lengths that produced negative (compressive) strain:

$$P_{STRAIN} = pPF-STRAIN + pDF-STRAIN + pTF-STRAIN + pPE-STRAIN$$

if  $PFstrain < 0$

$$pPF-STRAIN = -1e6*PFstrain$$

else

$$pPF-STRAIN = 0$$

Definitions:

$pPF-STRAIN$       penalty due to negative tendon strain

$PFstrain$           PF tendon strain

. . . similar equation for dorsiflexor, toe flexor and toe extensor

- $P_{ACCEL}$  ensured a static initial position by penalizing joint and model center of mass accelerations:

$$P_{ACCEL} = aCOM^2 + 1e6*aCOMy^2 + \alpha TOE^2 + \alpha MTP^2 + \alpha ANKLE^2$$

Definitions:

$aCOMx$           horizontal acceleration of body center of mass

$aCOMy$           vertical acceleration of body center of mass

$\alpha TOE$           angular acceleration of the TOE with respect to ground

$\alpha MTP$           angular acceleration of the MTP joint

$\alpha ANKLE$         angular acceleration of the ANKLE joint

Initially it was difficult to initialize the model in a static position. The heel did not settle into a position where the spring-damper system balanced the weight of the body. This caused the acceleration to remain close to  $-9.81 \text{ m/s}^2$ . It was decided to heavily weight ( $1e6$ ) the vertical acceleration in order to find an initial position in which the heel penetrated to a distance where ground reaction force was equal to model weight.



- $P_{HEEL}$  ensured initial heel contact by penalizing heel height above floor level:

*if*  $h_{HEEL} \geq 0$

$$P_{HEEL} = 1e6 * h_{HEEL}$$

*else*

$$P_{HEEL} = 0$$

Definitions:

$h_{HEEL}$             height of heel contact point above ground

After optimization  $J_{INIT}$  was less than  $2 \times 10^{-9}$  for all 17 combinations of musculoskeletal architecture

## 2. Maximal vertical-energy pushoff penalty function: $J_{HORIZ}$

$$J_{HORIZ}(t_f, u) = K \dot{x}_c^2(t_f)$$

Definitions:

$\dot{x}_c(t_f)$             horizontal velocity at final time

$K$                     penalty weighting

Horizontal velocity at final time reached  $\dot{x}_c(t_f)$  was minimized in order to ensure that the model would be able to return to original position after the vertically directed jump occurred, by jumping straight up in the air.  $K$  is a weighting factor set equal to 10, which was used to ensure that the penalty term do not significantly affect the jump height objective and final results showed that the horizontal velocity at the end of the simulation was never higher than 0.005 m/s for all simulations.

3. *Maximal average horizontal acceleration penalty function:*  $J_{PEN\_SPRINT}$

$$J_{PEN\_SPRINT}(t_f, u) = P_{GRF} + P_{GROUND} + P_{BOUNCE} + P_{V_{COM}}$$

The separate penalty terms were defined and equated as follows:

- $P_{GRF}$  guaranteed that the model left the ground at  $t_f$ :

$$P_{GRF} = GRF_y(t_f)$$

Definitions:

$GRF_y$  ground reaction force at the end of simulation time ( $t_f$ )

- $P_{GROUND}$  ensured that the point mass did not move through the floor during jumping:

$$\text{if any}(COM_y) \leq 0$$

$$P_{GROUND} = 1e3$$

else

$$P_{GROUND} = 0$$

Definitions:

$COM_y$  center of mass y position

Initial simulations showed that the model used a pendulum movement to gain maximal horizontal velocity. It started in the upright position and then swing downward, through the floor and up again. To limit this moment, a penalty was instituted when the center of mass moved below the floor level at any instant during the simulation.

- $P_{BOUNCE}$  penalized the model for any bouncing movement:

$$\text{if } PEAKS > 1$$

$$P_{BOUNCE} = 1e3 * (PEAKS - 1)$$

*else*

$$P_{BOUNCE} = 0$$

Definitions:

*PEAKS*            *number of transition PEAKS (or directional changes) from dorsiflexion to plantarflexion*

Similar to results from our initial jumping simulation (van Werkhoven and Piazza, 2013), the simulation showed a bouncing behavior. In order to minimize this the model movement was restricted to a single dorsiflexion followed by a single plantarflexion phase.

- $P_{VCOM}$  penalized any negative vertical velocity:

*if*  $vCOMy(t_f) < 0$

$$P_{VCOM} = 1e3 * vCOMy(t_f)$$

*else*

$$P_{VCOM} = 0$$

This was to ensure that the model did not dive downward at  $t_f$

After optimization all terms of  $J_{PEN_{SPRINT}}$  was 0, except for  $P_{GRF}$  which had a value of less than 0.006 N for all 17 combinations of musculoskeletal architecture.

#### 4. Maximal static load support penalty function: $J_{PEN_{SUPPORT}}$

$$J_{PEN_{SUPPORT}}(a, l_M, q) = P_{FORCE} + P_{STRAIN} + P_{ACCEL} + P_{POS}$$

The first three terms of this penalty function were exactly the same as  $P_{FORCE}$ ,  $P_{STRAIN}$ , and  $P_{ACCEL}$  for the initialization objective function (see 1 above). The other penalty term was defined as follows:

- $P_{POS}$  ensured that the CoM is lifted 5 cm above CoM at model resting position:

$$P_{POS} = -1e6*(COMy - (COMy_{REST} + 0.05))^2$$

Definitions:

$COMy$  center of mass vertical position

$COMy_{REST}$  center of mass when model is standing stationary yon ground  
with all joint angles set to 0° and zero penetration into floor

After optimization  $J_{PEN_{SUPPORT}}$  was less than 0.1981 for all 17 combinations of musculoskeletal architecture.

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# CURRICULUM VITAE

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