The role of the ankle plantar flexors, extrinsic foot muscles, and intrinsic foot muscles in performing propulsive tasks

by

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Abstract

The ability of the extrinsic foot muscles, intrinsic foot muscles, and foot ligaments to modulate foot arch rigidity may influence the transmission of energy generated through ankle plantar flexor work to the ground. However, the extent to which the function of these structures may affect the ankle plantar flexor work performed during vertical jumping is not well understood. The purpose of this thesis was threefold: First, to examine the effects of modifying centre of pressure (COP) location during jump propulsion on vertical jump height, and distribution of ankle, knee, and hip work during the subsequent landing. Second, to estimate the contributions from the extrinsic foot muscles, intrinsic foot muscles, and foot ligaments to foot arch support. Third, to examine the effects of 11-weeks of heel-raise exercise performed on an incline plane compared to a flat box on vertical jump performance, ankle plantar flexor strength, and toe flexor strength. Three studies were undertaken in pursuit of these purposes. In study 1, female volleyball (n = 17, age = 18 ± 2 yrs., height = 1.76 ± 0.07 m, mass = 67.1 ± 7.3 kg) and ice hockey players (n = 19, age = 17 ± 2 yrs., height = 1.64 ± 0.06 m, mass = 62.9 ± 7.8 kg) performed vertical countermovement jumps while recorded using motion analysis. When jumping with the COP close to the forefoot, participants performed more ankle plantar flexor and knee extensor work, compared to jumps with the COP close to the heel ($p < 0.05$). Jumping with the COP close to the forefoot also resulted in greater jump height and more ankle plantar flexor work ($p < 0.05$), but less hip extensor work being performed during landings ($p < 0.05$). In study 2, a musculoskeletal model was developed and used to estimate the contribution of the extrinsic foot muscles, intrinsic foot muscles, and foot ligaments to the midfoot moment in four different loading conditions, with and without maximal voluntary toe flexor activation. Input data was obtained from 6 female (age: 23 ± 3 yrs., stature: 1.62 ± 0.05 m, mass: 57.3 ± 5.6 kg) and 6 male (age: 27 ± 6 yrs., stature: 1.78 ± 0.08 m, mass: 78.7 ± 8.8 kg) participants who volunteered for the study. Midfoot net joint moment increased with increasing loads, while midfoot angles decreased with increasing load ($p < 0.05$). The static optimization algorithm used to estimate muscle and ligament contribution to midfoot moment could only find viable solutions to the force sharing problem when a specific tension of 60 N/cm^2 was used. The extrinsic foot muscles were the largest contributors to the midfoot net joint moment, followed by the intrinsic foot muscles and foot ligaments, for all external loads and with and without maximal voluntary toe flexor contraction. In study 3, female volleyball players completed 11 weeks of heel-raise exercise performed on an incline plane (n = 14, age = 16 ± 1 yrs., stature = 1.77 ± 0.08 m, mass = 67.1 ± 11.1 kg) or a flat box (n = 11, age = 17 ± 2 yrs., stature = 1.80 ± 0.07 m, mass = 70.3 ± 7.2 kg). Vertical countermovement and 3-step approach jump performance before, after 7 weeks, and after 11 weeks of training was assessed using motion analysis. Ankle plantar flexor and toe flexor strength was assessed using dynamometry. Both exercise groups improved vertical countermovement and 3-step approach jump height and hallux flexion strength over 11 weeks of training ($p < 0.05$). However, no changes were observed in ankle plantar flexor strength or work performed during vertical or approach jumps ($p > 0.05$). This work provides evidence that the extrinsic and intrinsic foot muscles may contribute to foot arch support, and that the hallux flexors can be strengthened using heel-raise exercise. Further, this thesis suggests that ankle plantar flexor work is an important determinant of vertical jump height. However, the mechanism through which an improvement in ankle plantar flexor strength affects jump height remains unclear.

Preface

This thesis is an original work by Torstein E. Dæhlin. The research projects, of which this thesis is a part, received ethics approval from the University of Alberta Research Ethics Board as follows:

- Comparison of Foot and Ankle Mechanics During Vertical Jumping in Athletes from Sports with Free and Restricted Foot Plantar Flexion, Pro00067479, approved August 29, 2016.
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The author of this thesis was responsible for conceptualization of the study, recruitment of research participants, data collection and analyses, and manuscript composition. L.Z.F. Chiu was the supervising author and was involved with conceptualization of the study and manuscript composition.

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Chapter 1

Literature review

1.1. Introduction

The feet are the only points of contact with the ground during the propulsion phase of many tasks, including walking, running, and jumping (Harcourt-Smith & Aiello, 2004). Consequently, energy generated by the lower extremity muscles must be transmitted to the ground via the foot to propel the body. However, the foot does not constitute a rigid lever, but deforms under the influence of external and internal forces, such as the ground reaction force and muscle forces (Blackman et al., 2009; Kelly et al., 2014; Thordarson et al., 1995). This foot deformation absorbs energy, reducing the energy available to perform propulsive work (Riddick et al., 2019; Takahashi et al., 2017). The aim of this thesis was to further our understanding of how the foot muscle function may affect the ability to utilize ankle plantar flexor work. Specifically, the aim was to investigate how foot muscle and ankle plantar flexor function may affect performance in vertical jumping as a propulsive task. Particular attention was given to understanding the extent that the ankle plantar flexors influence vertical jump performance and the extent to which the foot muscles can contribute to the moments generated at the midfoot. Moreover, whether the foot and ankle plantar flexor muscles may act together to improve performance in vertical jumping as a propulsive task was experimentally tested.

1.2 The arched foot

The bipedal mode of locomotion acquired by early hominids between two and three million years ago, is likely one of the main factors that influenced the evolution of the human foot (Latimer & Lovejoy, 1989, 1990a, 1990b). The human foot evolved from one similar to that of early African apes, which is characterized by an opposable hallux (big toe) and an efficient prehensile (grasping) function (Crompton et al., 2008; Harcourt-Smith & Aiello, 2004; Latimer & Lovejoy, 1990a, 1990b; Raichlen et al., 2011; Wang et al., 2014; Wang & Crompton, 2004). These traits are considered essential for efficient arboreal ambulation, such as tree climbing (Latimer & Lovejoy, 1990a, 1990b). In contrast, early hominids, such as *Australopithecus afarensis*, have a foot morphology that resembles that of modern *Homo sapiens* (Latimer & Lovejoy, 1989, 1990a, 1990b). This suggests a gradual evolution from a foot specialized for arboreal habitats, to a foot specialized for life in terrestrial habitats.

Among the foot's most essential evolutionary adaptations for terrestrial propulsion is an adducted and more robust hallux, relatively shorter phalanges and longer metatarsals, more compact and reoriented tarsal bones, and the development of longitudinal foot arches (Crompton et al., 2008; Latimer & Lovejoy, 1990a, 1990b; Rolian et al., 2009; Wang & Crompton, 2004). Some of these evolutionary traits have been demonstrated to provide an advantage when propelling the body mass in a terrestrial environment. For example, compaction and reorientation of the tarsal bones prevent the "mid-tarsal break" observed in great apes with prehensile feet (Crompton et al., 2008). A "mid-tarsal break" would make for a less rigid lever for the lower extremity muscles to act on. The arched foot of *H. sapiens* does not display this mid-tarsal break (Crompton et al., 2008). Although this suggests that the human foot is more rigid than that of other primates, it appears that the rigidity of the foot arch differs between different tasks, and between different phases within a task.

The human foot consists of 26 bones, excluding the sesamoid bones (Gray, 1918). Due to their configuration, these bones form three arches, the medial and lateral longitudinal arches, and the transverse arch (Gray, 1918). The medial longitudinal arch is formed from posterior to anterior by the calcaneus, talus, navicular, medial, intermediate, and lateral cuneiform, and 1st through 3rd metatarsal (Gray, 1918). The lateral longitudinal arch on its hand is formed by the calcaneus, cuboid, and 4th and 5th metatarsal. The transverse arch is formed by the cuboid and three cuneiforms (Gray, 1918). As the medial and lateral longitudinal arches are more prominent than the transverse arch, the foot is often conceptualized as a single longitudinally oriented arch (foot arch), supported by viscoelastic elements connecting the anterior and posterior aspects of the arch (Simkin & Leichter, 1990; Figure 1.1).

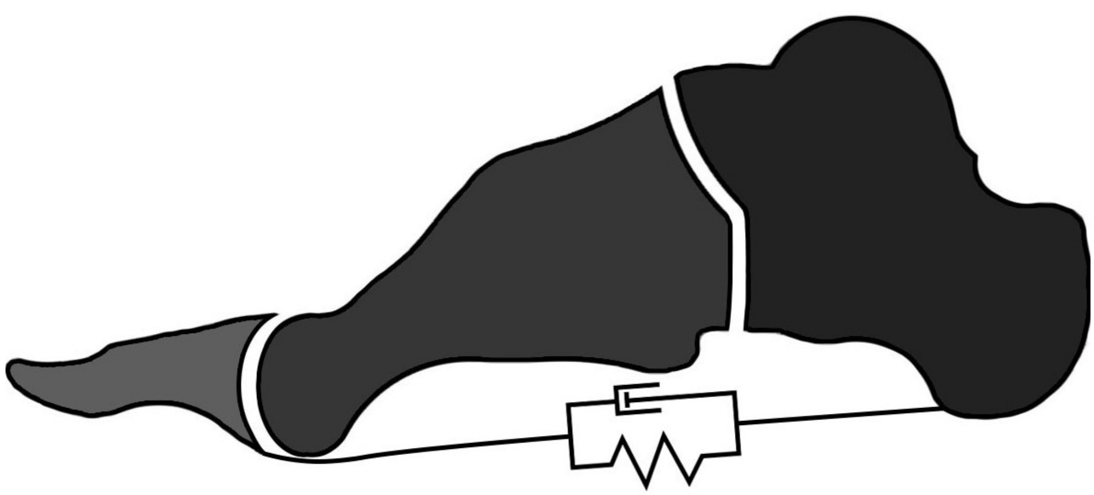

Figure 1.1. Common conceptualization of the foot arch supported by viscoelastic elements. The rearfoot, forefoot, and toes are illustrated in dark, medium, and light gray.

The foot arch is purported to serve multiple functions. One of these is to store, generate, and dissipate energy during foot ground contact (Kelly et al., 2018; Ker et al., 1987; Riddick et al., 2019; Simkin & Leichter, 1990; Takahashi et al., 2017; Wager & Challis, 2016). Energy storage and release is observed during level gait, where the foot arch deforms during the load acceptance phase and recoils during terminal stance (Chang et al., 2008; Hunt et al., 2001; Ker et al., 1987;

Leardini et al., 2007; Riddick et al., 2019; Stamm & Chiu, 2016; Wager & Challis, 2016). However, when walking on uneven terrain, the foot arch may also contribute to energy generation and absorption (Riddick et al., 2019). The energy storage function of the foot arch is also affected by the structural alignments of the foot. Specifically, the foot's capacity for energy storage is compromised with *pes planus* and *pes cavus*, in which the medial longitudinal arch is lower or higher than normal, respectively (Kim & Voloshin, 1995; Simkin & Leichter, 1990). A second function of the foot arch is to provide a rigid lever for transmission of energy generated more proximal in the lower extremity to the ground (Arakawa et al., 2013; Aronow et al., 2006; Takahashi et al., 2016). This function is purported to be essential for efficient propulsion (Kelly et al., 2015; Takahashi et al., 2017).

To serve these functions, the foot arch must meet the two seemingly paradoxical requirements of being both compliant and rigid (Crompton et al., 2008; Latimer & Lovejoy, 1989). To accomplish this, one must be able to modulate the foot's rigidity according to needs of a specific task. The modulation of foot arch rigidity is achieved by a combination of active and passive structures. However, the extent to which modulation is desired depends on the task and the presence of forces tending to deform the foot arch.

1.3 Foot arch deformation

The foot arch may deform under the influence of external forces such as the ground reaction force, and internal forces such as muscle forces (Blackman et al., 2009; Cheung et al., 2006; Kelly et al., 2014; Thordarson et al., 1995). Since the foot is the only point of contact with the ground during many propulsive tasks, the ground reaction force is a significant contributor to foot arch deformation. Specifically, the ground reaction force acting under the metatarsals and that acting on the heel have an arch deforming effect as they tend to plantar flex the rearfoot and dorsiflex the forefoot, respectively. This is exemplified during gait, where peak foot arch deformation occurs at the same instances as the ground reaction force peaks associated with heel strike and heel-off (Caravaggi et al., 2010; Hunt et al., 2001). Foot arch deformation due to plantar flexion of the rearfoot and dorsiflexion of the forefoot is also observed when external loads are applied such that they act axially through the leg segment (Arangio et al., 1998; Blackman et al., 2009; Kelly et al., 2014).

In addition to the external forces acting on the foot, muscle forces may also cause foot arch deformation. Most notably, the triceps surae muscles can exert a large arch deforming force on the calcaneus (Chiu et al., 2020; Jones, 1941). The triceps surae consists of two muscles, the soleus and the gastrocnemius, the latter of which has medial and lateral heads. The two heads of the gastrocnemius originate above the medial and lateral femoral condyles, respectively (Fukunaga et al., 1992). In contrast, the soleus originates on the posterior aspect of the tibia near the soleal line, and from the head and proximal shaft of the fibula (Fukunaga et al., 1992). Distally, the aponeuroses of these muscles combine to form the strong calcaneal tendon. The triceps surae can exert a large calcaneal plantar flexion moment, due to its long moment arm and large physiological cross-sectional area (Baxter et al., 2012; Chiu et al., 2020; Klein Horsman et al., 2007; Klein et al., 1996; Ward et al., 2009; Figure 1.2). Plantar flexion of the calcaneus deforms the foot arch, which has been demonstrated both *in vivo* and *in vitro*. For example, exerting a pull on the calcaneal tendon in cadaveric foot specimens decreases foot arch height and increases arch length (Blackman et al., 2009; Cheung et al., 2006; Thordarson et al., 1995). Similarly, the triceps surae muscle has been found to shorten more than the amount predicted by rotation of the entire foot due to rearfoot rotation occurring during isometric contractions (Iwanuma et al., 2011). Calcaneal plantar flexion has also been observed during the mid-stance and early push-off in gait, where calcaneal tendon forces are known to be large (Finni et al., 1998; Hunt et al., 2001; Leardini et al., 2007; Stamm & Chiu, 2016). Collectively, these findings corroborate the suggestion that the triceps surae may cause considerable arch deformation (Jones, 1941).

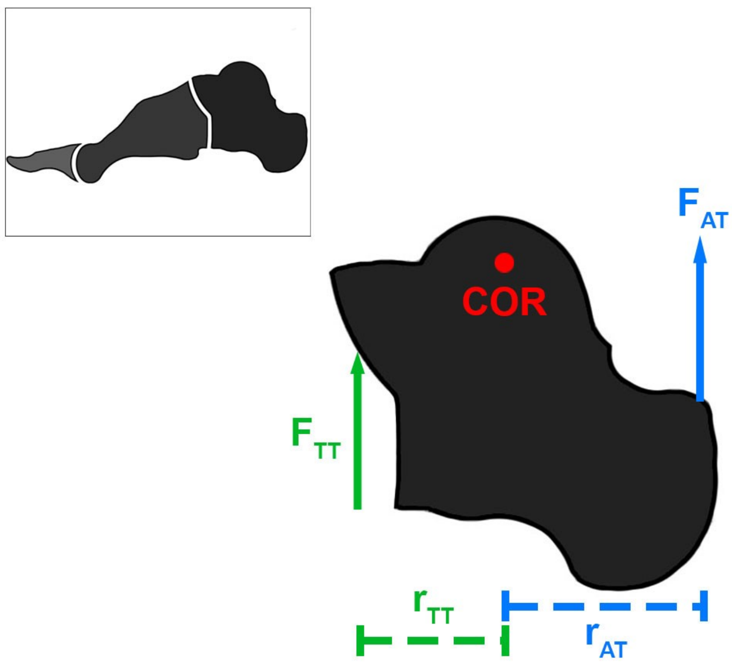

Figure 1.2. Illustration depicting the division between the forefoot and rearfoot (inset), and the moment arm (r_{AT}) of the Achilles tendon force (F_{AT}) and moment arm (r_{TT}) of the reaction force acting at the transverse tarsal joint (F_{TT}). The moments are taken to act about the centre of rotation (COR) or the ankle joint and only vertical force components are illustrated for simplicity.

1.4 Foot arch support

Multiple mechanisms may be responsible for preventing the foot arch from collapsing under the influence of external and internal forces acting on it. The support mechanisms of the foot arch may be divided into two categories: passive and active support mechanisms.

1.4.1 Passive mechanisms

The passive structures that support the foot arch include the bones that form the arch, the plantar ligaments, and the joint capsules of the intertarsal and tarsometatarsal joints. The bony configuration of the foot has been demonstrated to affect deformation of the longitudinal arches (Arangio et al., 1998; Kim & Voloshin, 1995; Simkin & Leichter, 1990). For example, musculoskeletal computer models have been used to establish that the unloaded configuration of the foot arch affects the degree to which it deforms under a given load (Arangio et al., 1998; Kim & Voloshin, 1995; Simkin & Leichter, 1990). Specifically, the foot arch is predicted to deform more with foot configurations consistent with *pes planus* and *pes cavus* deformity, compared to a normal foot configuration (Arangio et al., 1998; Kim & Voloshin, 1995). These findings indicate that bone-on-bone reaction forces contribute to the passive support of the foot arch.

In addition to bone-on-bone contact forces, the plantar ligaments, intertarsal joint capsule, and tarsometatarsal joint capsule play a significant role in foot arch support. Among the plantar ligaments and joint capsules, the plantar aponeurosis, long and short plantar ligaments, and spring ligament are considered most important for maintaining the structural integrity of the foot arch (Cheung et al., 2004; Chu et al., 2001; Crary et al., 2003; Huang et al., 1993; Iaquinto & Wayne, 2010; Ker et al., 1987; Kitaoka et al., 1997). The long and short plantar ligaments span the plantar aspect of the calcaneocuboid joint. Specifically, the long plantar ligament originates on the plantar aspect of the calcaneus, just anterior to the calcaneal tuberosity, and inserts on the plantar surface of the cuboid and the base of the 2nd through 4th metatarsals (Ward & Soames, 1997). The short plantar ligament is deep to the long plantar ligament, originates from the calcaneal tubercle and inserts on the plantar surface of the cuboid (Ward & Soames, 1997). The long and short plantar ligaments reinforce the plantar aspect of the calcaneocuboid joint and are hypothesized to play an essential role in supporting the lateral longitudinal arch. The spring ligament is medial to the long and short plantar ligaments. It consists of two distinct portions, commonly referred to as the superomedial and inferior calcaneonavicular ligaments (Davis et al., 1996). The superomedial calcaneonavicular ligament originates from the medial aspect of sustentaculum talus of the calcaneus and inserts to the entire medial margin of the navicular facet for articulation with the talar head (Davis et al., 1996; Taniguchi et al., 2003). The inferior calcaneonavicular ligament is narrower and originates in the notch between the middle and anterior articular facets of the calcaneus and inserts to the cortical bone on the plantar aspect of the navicular (Davis et al., 1996; Taniguchi et al., 2003). Together, these ligaments form a sling-like structure that supports and reinforces the plantar and medial aspects of the talonavicular joint (Davis et al., 1996). The importance of the long and short plantar ligaments, and spring ligament for the structural integrity of the foot arch has been demonstrated in both computer and cadaveric models. Specifically, division of either of these ligaments results in a marked decrease in arch height and elastic energy stored during arch deformation (Cheung et al., 2004; Crary et al., 2003; Huang et al., 1993; Ker et al., 1987; Kitaoka et al., 1997). However, the decrease in arch height resulting from removing these ligaments is small compared to that resulting from removal of the plantar aponeurosis (Crary et al., 2003; Huang et al., 1993; Kitaoka et al., 1997). For this reason, the plantar aponeurosis is considered to play a more important role in supporting the foot arch.

The plantar aponeurosis is a fibrous tissue sheath that covers the plantar aspect of the foot, and consists of a distinct medial, central, and lateral portion (Hedrick, 1996; Moraes do Carmo et al., 2008). The medial and lateral portion of the plantar aponeurosis originates from the medial calcaneal tuberosity and fan out to cover the abductor hallucis and abductor digiti minimi, respectively (Hedrick, 1996; Moraes do Carmo et al., 2008). The role these portions of the plantar aponeurosis play in maintaining normal arch function is poorly understood (Hedrick, 1996). However, the common consensus is that they are of little significance (Moraes do Carmo et al., 2008). In contrast, the central portion of the plantar aponeurosis is purported to play an important role for normal foot function. At its origin on the medial calcaneal tuberosity, the central portion of the plantar aponeurosis is approximately 2.0-5.0 mm thick and 15 mm wide (Bolton et al., 2005; Chen et al., 2014; Hedrick, 1996; Kamel & Sakla, 1961; Lobo et al., 2016; Moraes do Carmo et al., 2008; Ozdemir et al., 2005; Pascual Huerta & Alarcon Garcia, 2007; Uzel et al., 2006). Around the middle of the metatarsal bones' longitudinal axis, the plantar aponeurosis splits into multiple tracts (Bojsen-Moller & Flagstad, 1976; Hicks, 1954). These tracts form sagittal septa that insert to the fascia covering the interossei and the transverse head of the adductor hallucis proximally, as well as to the plantar ligaments and the base of the proximal phalanges of each digit (Bojsen-Moller & Flagstad, 1976).

The plantar aponeurosis assists in maintaining the structural integrity of the foot arch, stores and releases elastic energy during movement, and modulates the rigidity of the foot. The role of the plantar aponeurosis in maintaining the structural integrity of the foot arch is readily observed when it is completely removed, a procedure that clinically is referred to as complete plantar fasciotomy. For example, computer and cadaveric models have demonstrated that arch height decreases, and arch length increases by approximately 20% following complete fasciotomy (Arangio et al., 1998; Cheung et al., 2004; Huang et al., 1993; Kim & Voloshin, 1995). Further, plantar fasciotomy decreases the stiffness of the foot arch by approximately 25% (Arangio et al., 1998). In contrast, partial fasciotomy, in which the plantar aponeurosis is only partially divided, appears to have little effect on its role in maintaining the structural integrity of the foot arch. For example, Chen et al. (2014) found no changes in foot arch height or distribution of plantar pressures after transecting the first ray of the plantar aponeurosis at its distal end in cadaveric specimens.

Experiments have shown that the plantar aponeurosis behaves approximately as a viscoelastic spring-dampener system when it is stretched, and results from cadaveric studies suggest that its failure strength is between 850 N and 1750 N (Erdemir et al., 2004; Kim & Voloshin, 1995; Kitaoka et al., 1994; Wright & Rennels, 1964). One mechanism through which the plantar aponeurosis may lengthen is due to foot deformation. As the plantar aponeurosis has viscoelastic properties, it is not surprising that tension in the aponeurosis increases with increasing deformation of the foot arch (Arangio et al., 1998; Cheung et al., 2006). Foot deformation in the form of calcaneal plantar flexion also occurs when the triceps surae exerts large moments on the calcaneus through the calcaneal tendon (Blackman et al., 2009; Cheung et al., 2006). If the toes remain in a fixed position, the plantar aponeurosis will lengthen and plantar aponeurosis tension increases (Figure 1.3). Using force transducers imbedded within the plantar aponeurosis of cadaveric foot specimens, Cheung et al. (2006) demonstrated that plantar aponeurosis tension increases as the pull exerted on the calcaneal tendon increases. Similar increases in plantar aponeurosis tension have been estimated in other cadaveric and computer modeling studies (Carlson et al., 2000; Cheng, Lin, Wang, et al., 2008). A second mechanism through which the plantar aponeurosis may lengthen is by dorsiflexing the toes. Increases in plantar aponeurosis tension have been observed when the toes are dorsiflexed passively (Cheng, Lin, Wang, et al., 2008). Further, increases in plantar aponeurosis tension results from passive toe dorsiflexion occurring during walking in living participants (Gefen, 2003). Both *in vivo* and *ex vivo* experiments confirm that the plantar aponeurosis stores and releases elastic energy during walking (Gefen, 2003; Ker et al., 1987). Further, computer models suggest that the plantar aponeurosis may play a similar role in energy storage and release during running (Wager & Challis, 2016).

Lastly, the plantar aponeurosis is commonly purported to modulate the rigidity of the foot through the windlass mechanism. The windlass mechanism was first described by Hicks (1954) and who suggested that the plantar aponeurosis is wrapped around the metatarsal heads, like a cable wraps around a windlass, when the toes are being dorsiflexed. This results in a proximal translation of the metatarsal heads, effectively shortening the foot arch and increasing its height (Cheng, Lin, Chou, et al., 2008; Hicks, 1954; Thordarson et al., 1997; Thordarson et al., 1995). Such a compaction of the foot, together with the increased tension in the plantar aponeurosis is hypothesized to increase the rigidity of the foot arch, thereby providing a more rigid lever for the lower extremity muscles to act on.

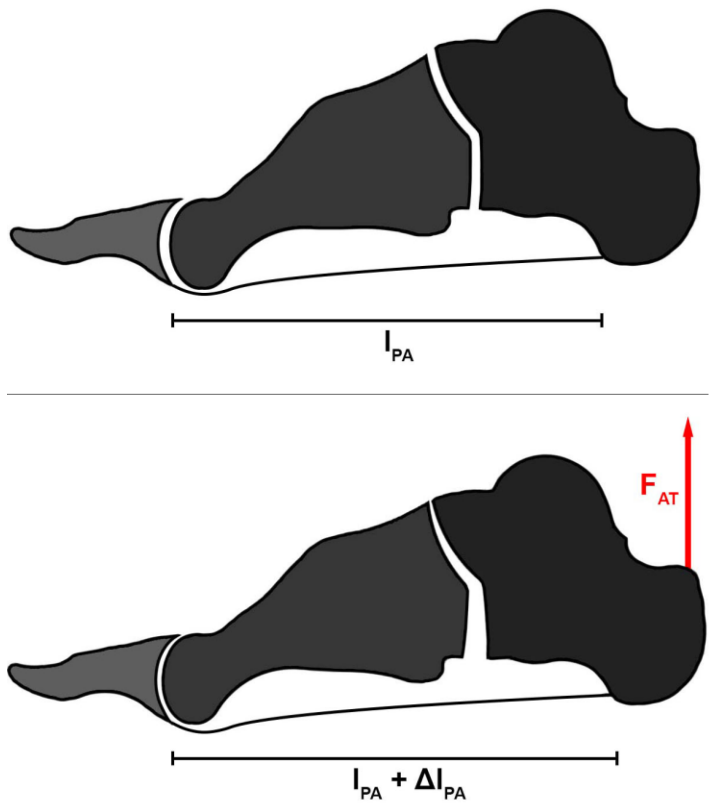

Figure 1.3. Illustration of how the plantar aponeurosis length (l_{PA}) changes from its resting position (top panel) when calcaneal plantar flexion is caused by a large Achilles tendon force (F_{AT}; bottom panel).

Previous research investigating the effect of combined toe dorsiflexion and calcaneal tendon pull observed that force transmission from the triceps surae to the forefoot improves due to the windlass mechanism (Cheung et al., 2006; Hicks, 1954; Thordarson et al., 1995). For example, Cheung et al. (2006) demonstrated that transmission of force from the calcaneal tendon to the forefoot decreased when the stiffness of the plantar aponeurosis was decreased in a computer model. This result was consistent when the toes of the model was in neutral and when dorsiflexed (Cheung et al., 2006). Further, the ability to transmit force from the calcaneal tendon to the forefoot is compromised following plantar fasciotomy (Erdemir & Piazza, 2004). Taken together, these results demonstrate that the plantar aponeurosis play a role in modulating the rigidity of the foot during locomotion. However, since the plantar aponeurosis consists exclusively of passive tissue, its ability to do so is entirely dependent on the configuration of the foot (Hicks, 1954). Moreover, recent research has found the foot to become less stiff as the toes extend during the push-off in gait (Welte et al., 2018). It can therefore be hypothesized that transmission of force from the calcaneal tendon to the forefoot would be suboptimal if the plantar aponeurosis was the only structure that can influence the rigidity of the foot arch. Thus, an active element is likely to assist in modulating foot arch rigidity (Caravaggi et al., 2009; Caravaggi et al., 2010; Pataky et al., 2008).

1.4.2 Active mechanisms

Selected extrinsic and intrinsic foot muscles have been hypothesized to resist arch deformation (Erdemir & Piazza, 2004; Jacob, 2001; Kelly et al., 2014; Kelly et al., 2012; Kelly et al., 2015; Thordarson et al., 1995). The four extrinsic muscles that have received most research attention is the fibularis longus, flexor digitorum longus, flexor hallucis longus, and tibialis posterior (Erdemir & Piazza, 2004; Jacob, 2001; Thordarson et al., 1995). These muscles originate from the lateral aspect of the fibula and fibular head, middle third of the posterior aspect of the tibia, distal two thirds of the posterior fibula, and proximal two thirds of the tibia and fibula, respectively (Gray, 1918). The flexor digitorum longus and flexor hallucis longus insert to the plantar surface of the base of the distal phalanx of the lateral four digits, and hallux, respectively, whereas the tibialis posterior inserts to all tarsal bones except for the calcaneus (Gray, 1918). Fibularis longus inserts to the base of the first metatarsal and the medial cuneiform (Gray, 1918).

The idea that these muscles may support the foot arch was first conceived by Basmajian and Stecko (1963). They observed electromyographical activity from these muscles when external loads were placed on the top of the thigh of seated participants, such that the load acted approximately axially through the leg segment (Basmajian & Stecko, 1963). More recent studies have also found indications that the extrinsic foot muscles may contribute to foot arch support. For example, Jacob (2001) estimated that the flexor hallucis longus and flexor digitorum longus exert forces equivalent to 52.4% and 8.8% of body weight during the push-off in gait. Another group used a forward dynamics model to estimate the effect of plantar fasciotomy on a simple heel raise exercise (Erdemir & Piazza, 2004). They found that the required flexor digitorum longus and flexor hallucis longus forces increase following plantar fasciotomy (Erdemir & Piazza, 2004). These results indicate that the flexor digitorum longus and flexor hallucis longus may contribute to foot arch support. However, these models are limited as they neglected either the plantar aponeurosis or intrinsic toe flexors (Erdemir & Piazza, 2004; Jacob, 2001). Lastly, Thordarson et al. (1995) observed that loading the tibialis posterior tendon in cadaveric specimens resulted in an increase in foot arch height and a decrease in foot arch length. This finding suggests that the tibialis posterior can play a role in supporting the foot arch and is further corroborated by clinical studies reporting development of *pes planus* deformity following posterior tibialis tendon rupture (Mann & Thompson, 1985). However, Okita et al. (2013) found that acute loss of tibialis posterior function did not affect the kinematics of the bones of the foot during *in vitro* simulated gait using cadaveric specimens. Regardless of the extrinsic foot muscles' role in supporting the foot arch, these muscles may not be well suited to prevent foot deformation due to calcaneal plantar flexion. Specifically, the extrinsic foot muscles do not attach to the calcaneus and can therefore be hypothesized to be poorly suited to oppose calcaneal plantar flexion (Gray, 1918).

In contrast to the extrinsic foot muscles, the intrinsic foot muscles have their origin and insertion within the foot. The intrinsic foot muscles are most often grouped into one dorsal and four plantar muscle layers (Soysa et al., 2012). The intrinsic muscles that are considered most important modulators of arch stiffness are found in the two most superficial plantar layers (Kelly et al., 2014; Kelly et al., 2015; Soysa et al., 2012). In the first layer just deep to the plantar aponeurosis are the abductor hallucis, flexor digitorum brevis, and abductor digiti minimi, which originate from the medial and lateral calcaneal tuberosities (Soysa et al., 2012). The abductor hallucis and abductor digiti minimi insert to the medial aspect of the first and lateral aspect of the fifth proximal phalanges, respectively (Gray, 2009). The flexor digitorum brevis inserts to the plantar aspect of the middle phalanx of the lateral four digits (Gray, 2009). In the second layer, quadratus plantae originates from the calcaneus and inserts to the distal phalanges through the tendons of the flexor digitorum longus (Soysa et al., 2012).

Early research investigating the role of these muscles for arch support dismissed them as unimportant, due to the absence of – or inability to detect – electromyographical activity of these muscles during standing (Basmajian & Stecko, 1963). Some authors have even gone as far as to suggest that the plantar intrinsic muscles have become obsolete with the diminished need for a prehensile foot, and as such, represents an incomplete evolution (Mann & Hagy, 1979). For this reason, much research has disregarded the effect of this support system. However, some research has given merit to the idea that the plantar intrinsic muscles play an important role in preventing collapse of the longitudinal foot arches when acted upon by external forces. For example, Headlee et al. (2008) observed a decrease in foot arch height following a toe flexor muscle fatiguing protocol, demonstrating that these muscles may contribute to foot arch support. However, since a protocol emphasizing toe flexion was used, the relative contribution of the extrinsic and intrinsic toe flexors to this function could not be assessed. Fiolkowski et al. (2003) administered a tibial nerve block at the tarsal tunnel, showing that inhibiting neural stimulation of the intrinsic, but not extrinsic, foot muscles increased deformation of the foot arches in weightbearing. Recently, the role of the intrinsic foot muscles for supporting the foot arches have been given more attention (Kelly et al., 2014; Kelly et al., 2019; Kelly et al., 2012; Kelly et al., 2015; Riddick et al., 2019; Takahashi et al., 2016; Takahashi et al., 2017). This research has revealed that the electromyographical activity of the plantar intrinsic muscles increase with both increasing postural demand, and with increases in axial load acting through the leg segment (Kelly et al., 2014; Kelly et al., 2012; Kelly et al., 2015). Further, electrical stimulation of the plantar intrinsic muscles increases arch height and decreases arch length (Kelly et al., 2014). Lastly, the intrinsic foot muscles have been found to contribute to energy generation and absorption (Riddick et al., 2019). These studies have demonstrated that the plantar intrinsic muscles in the first and second layer may play an important role in preventing collapse of the foot arch both in static and dynamic situations.

In addition to preventing collapse of the arch while acted upon by external forces, the plantar intrinsic muscles may play a major role in the active modulation of arch rigidity (Kelly et al., 2014; Kelly et al., 2015; Riddick et al., 2019; Takahashi et al., 2016). Consequently, sufficient force generation from the intrinsic foot muscles may be of importance for transmission of forces from the calcaneal tendon to the ground (Kelly et al., 2015; Takahashi et al., 2016). Specifically, Takahashi et al. (2016) provided proof of concept for this mechanism using insoles of differing rigidities to demonstrate that a stiffer foot increases the force exerted by the soleus muscle. Further, electromyographical activity of the abductor hallucis, flexor digitorum brevis, and quadratus plantae has been reported to increase with increasing locomotion demands, giving merit to the hypothesis that these may play a role in modulating foot arch rigidity (Kelly et al., 2015). However, there is still paucity in experimental research investigating this hypothesis.

1.5 Foot muscle strength and vertical jump performance

It is well established that the ankle plantar flexors, and in particular, the triceps surae muscles are important for vertical jump performance. This is evident if the ankle plantar flexor's ability to perform work during the vertical jump is reduced using restrictive devices or fatigue (Arakawa et al., 2013; Bobbert et al., 2011; Haguenauer et al., 2006; Virmavirta & Komi, 2001). However, resistance training studies using exercises that target the triceps surae muscles to improve vertical jump performance have provided conflicting results (Bangerter, 1968; Chiu et al., 2017). Further, changes in vertical jump height following triceps surae exercise combined with general resistance training has not been accompanied by increases in ankle plantar flexor work or net joint moment during jumping (Chiu et al., 2017). Therefore, it may be questioned whether the increases in vertical jump height can be attributed to an increase in triceps surae strength in this study (Chiu et al., 2017). One potential reason for this discrepancy is that an increase in the force exerted on the calcaneus by the triceps surae muscles may lead to an increase in foot arch deformation (Blackman et al., 2009; Cheung et al., 2006; Thordarson et al., 1995). This may in turn have affected how well this force was transferred to the forefoot, and thus have confounded the effect on vertical jump height. It may therefore be hypothesized that simultaneous strengthening of the ankle plantar flexor and the intrinsic foot muscles would be more effective for improving vertical jump height compared to ankle plantar flexor strengthening, alone.

Strength training protocols targeting the intrinsic toe flexors have been used in several research studies (Goldmann et al., 2011; Goldmann et al., 2013; Hashimoto & Sakuraba, 2014; Jung et al., 2011; Kokkonen et al., 1988; Mickle et al., 2016; Unger & Wooden, 2000). Most of these utilized an isometric exercise mode (Goldmann et al., 2011; Goldmann et al., 2013; Kokkonen et al., 1988; Unger & Wooden, 2000). Isometric strength-training protocols have proved effective for increasing toe flexor strength, with effect size increases of 1.4 to 3.0 standard deviations in isometric toe flexor strength following 6- to 12-week interventions (Goldmann et al., 2011; Goldmann et al., 2013; Kokkonen et al., 1988; Unger & Wooden, 2000). However, the effect of isometric toe flexor training on performance in dynamic tasks, such as walking, running, and vertical and horizontal jumping, appears to be of small magnitude. For example, Goldmann et al. (2013) observed no improvement in vertical jump height, and only small effect size improvements in horizontal jump distance following 7 weeks of toe flexor training 4 times per week. Moreover, no changes were observed in ankle plantar flexor or metatarsophalangeal joint moments during walking, running, vertical jumping, or horizontal jumping. In contrast, Unger and Wooden (2000) found that vertical and horizontal jump performance increased following 6 weeks of toe flexor training 3 times per week.

Few studies have utilized a dynamic strengthening protocol to exercise the toe flexors. Mickle et al. (2016) found improvements in hallux and lesser toe flexor strength, and balance following 12-weeks of resistance training targeting the toe flexors in older individuals (> 60 yrs. old).

Similarly, Jung et al. (2011) found a greater increase in hallux flexor strength and abductor hallucis cross-sectional area following eight weeks of combined use of foot orthoses and daily short foot exercise, compared to use of foot orthoses only, in individuals with *pes planus* deformity. Lastly, Hashimoto and Sakuraba (2014) found that toe flexor strength, vertical jump height, and horizontal jump distance increased, whereas 50 m sprint time decreased following eight weeks of toe curl exercise 3 times per week in healthy young males. Taken together, it appears that multiple exercise modalities may be used to increase toe flexor strength. However, the extent to which increased toe flexor strength affects performance in dynamic tasks varies between different exercise protocols.

A common factor to all previous research studies attempting to strengthen the plantar intrinsic muscles is that they have used exercises that primarily emphasizes a toe or metatarsal flexor action (Goldmann et al., 2011; Goldmann et al., 2013; Hashimoto & Sakuraba, 2014; Jung et al., 2011; Kokkonen et al., 1988; Mickle et al., 2016; Unger & Wooden, 2000). Such protocols for strengthening the plantar intrinsic muscles may be limited in two ways. First, it is likely that the extrinsic toe flexor muscles are used during the strength training exercises (Goldmann & Bruggemann, 2012). However, to what degree the extrinsic and intrinsic toe flexors contribute to the resultant moment during these exercises has not been determined. One research study attempted to address this issue by plantar flexing the ankle, so that flexor digitorum longus and flexor hallucis longus would operate on the ascending limb of their strength curves (Hashimoto & Sakuraba, 2014). Although both *in vivo* measurements of toe flexor strength and data from cadaveric specimens suggest that these muscles indeed would operate on the ascending limb of their force-length curves in this position, it remains unknown where exactly on the force-curve the extrinsic toe flexors are when the ankle is plantar flexed (Goldmann & Bruggemann, 2012; Klein Horsman et al., 2007). Ultimately this would depend on the sarcomere length of the muscles in the plantar flexed position relative to the optimal sarcomere length of approximately 2.7 µm (Walker & Schrodt, 1974). As no studies have reported the length change of the flexor digitorum longus and flexor hallucis longus with a given ankle joint excursion, it cannot be predicted what the sarcomere length of these muscles are when the ankle is plantar flexed. Contribution of the extrinsic toe flexor muscles may therefore not be eliminated as a confounding factor using this design.

A second limitation is that the toe flexor action used in previous studies may not be representative of the function of the plantar intrinsic muscles during typical human movement, such as locomotor tasks and jumping. Early humans evolved from arboreal beings to habitual bipedal terrestrial runners and the prehensile function of the toes has become greatly reduced (Latimer & Lovejoy, 1990a, 1990b). However, the intrinsic toe flexor muscles that originate on the calcaneus remain of a significant size and strength; the abductor hallucis, flexor digitorum brevis, quadratus plantae, and abductor digiti minimi collectively make up approximately 80% of the intrinsic foot muscles (Kura et al., 1997; Lachowitzer et al., 2007; Silver et al., 1985). Further, the combined cross-sectional area of the plantar intrinsic muscles is larger than that of flexor digitorum longus and flexor hallucis longus, comparable to that of tibialis posterior, and approximately 75% of that of the lateral gastrocnemius (Chang et al., 2012; Fukunaga et al., 1992; Silver et al., 1985). It thus appears that the plantar intrinsic toe flexors may have evolved to assume a new role as early hominids evolved into habitual terrestrial runners. Some have suggested that the role of these muscles is to control dorsiflexion of the toes as the center of pressure moves anterior to the metatarsophalangeal joints during gait (Rolian et al., 2009). However, since the plantar aponeurosis tension increases with increasing toe dorsiflexion due to the Windlass mechanism, it may be argued that the intrinsic toe flexors are not essentials in this phase of gait (Cheng, Lin, Chou, et al., 2008; Gefen, 2003). Further, considerable activation of the plantar intrinsic muscles has been observed well prior to heel-off during walking and running (Kelly et al., 2015). Collectively, these findings indicate that the plantar intrinsic muscles serve another purpose during locomotor tasks. Specifically, we hypothesize that the plantar intrinsic muscles act to restrict calcaneal plantar flexion during tasks where large calcaneal tendon forces are exerted, and thus improve transmission of force from the calcaneal tendon to the forefoot.

If this hypothesis is correct, another method of strengthening the plantar intrinsic muscles may be to use an exercise where calcaneal tendon force is large, while the toes remain in a fixed position. One exercise in which calcaneal tendon force may be high is the heel-raise exercise. Specifically, raising the body center of mass through foot plantar flexion requires large ankle plantar flexor moments, and the activation of the triceps surae muscles is large during heel-raise exercise (Chiu & Dæhlin, 2021; Flanagan et al., 2005; Fujisawa et al., 2015). A large triceps surae force, acting posterior to the medio-lateral calcaneal rotation axis would generate a large calcaneal plantar flexion moment (Klein et al., 1996). If the toes were simultaneously fixed against a support, the abductor hallucis, flexor digitorum brevis, quadratus plantae, and abductor digiti minimi could simultaneously act on the calcaneus, generating an opposing (i.e. calcaneal dorsiflexion) moment. However, a conventional heel-raise is typically performed on a block. When standing on a block, the toes will not remain fixed against the support while the ankle is dorsiflexing beyond neutral (Chiu & Dæhlin, 2021). While the ankle is dorsiflexed, any action of the intrinsic toe flexor muscles would therefore cause movement of the relatively unrestricted toes, and little force would be exerted on the calcaneus. This is significant as the triceps surae can generate larger force when the ankle is dorsiflexed compared to when plantar flexed (Gravel et al., 1990; Maganaris, 2003). An alternative to the conventional heel-raise is to perform the heel-raise exercise on an incline surface (incline heel-raise). In an incline heel-raise, the toes remain fixed against the incline throughout the entire movement (Chiu & Dæhlin, 2021). With the toes fixed, the intrinsic toe flexor muscles may be used to exert a dorsiflexion moment on the calcaneus. A recent study supports these hypotheses, as it found that both midfoot and ankle plantar flexor net joint moments were greater during heel-raises performed on an incline compared to when performed on a flat block (Chiu & Dæhlin, 2021). As this difference was particularly pronounced at the midfoot, these exercises may be good alternatives for investigating the effects of combined ankle plantar flexor and intrinsic muscle strengthening compared to ankle plantar flexor strengthening, alone.

1.6 Purpose and hypotheses

Study 1 (Chapter 2): Forefoot and heel take-off strategies result in different distribution of lower extremity work during landings

The primary purpose of this study was to investigate whether centre of pressure location during the propulsion phase of vertical jumping affects vertical jump height, as well as the work performed at the ankle, knee, and hip joint during the subsequent landing. This investigation sought to evaluate to which extent the work performed by the ankle plantar flexors during the propulsion phase of vertical jumping may affect jump height and/or the distribution of lower extremity work during landings.

We hypothesized that performing the propulsion phase with the centre of pressure close to the forefoot would result in more ankle plantar flexor work being performed during propulsion compared to jumps performed with the centre of pressure close to the heel. Further, we hypothesized that an increase in ankle plantar flexor work performed during jumping would result in participants jumping higher in the condition where the centre of pressure was located near the forefoot. Lastly, we hypothesized that more ankle plantar flexor work would be performed in landings following jumps performed with the centre of pressure close to the forefoot during propulsion compared to when located near the heel.

Study 2 (Chapter 3): Roles of the intrinsic and extrinsic foot muscles in supporting the medial longitudinal arch

The purpose of this study was to investigate the changes in midfoot angle and net joint moments in response to changes in external load with and without maximal voluntary toe flexor activation. A further purpose was to estimate the relative contribution of the extrinsic foot muscle, intrinsic foot muscles, and foot ligaments to the midfoot net joint moment in the various loading and voluntary toe flexor activation conditions. A musculoskeletal computer model was developed for the latter purpose.

We hypothesized that the forefoot would dorsiflex more relative to the rearfoot, effectively lowering the foot arch, with heavier compared to lighter loads. Further, we hypothesized that the midfoot net joint moment would increase with increasing external load. Lastly, we hypothesized that maximally voluntarily activating the toe flexor muscles would result in larger midfoot net joint moments, but smaller foot arch deformation as quantified by the midfoot angle.

Study 3 (Chapter 4): Effect of incline versus block heel-raise exercise on vertical jump performance – A randomized longitudinal study

The primary purpose of this study was to assess the effects of 11 weeks of heel-raise exercise performed on an incline plane versus on a flat block on vertical jump height, ankle plantar flexor strength, and toe flexor strength. A secondary purpose was to investigate how these two exercises would affect the work performed at the ankle, knee, and hip during vertical jumps.

We hypothesized that heel-raises performed on an incline would be more effective at increasing vertical jump height and toe flexor strength compared to the heel-raises performed on a flat block. Changes in the work performed at the ankle, knee, and hip were assessed to explore the potential mechanisms responsible for any change in vertical jump height. Therefore, no hypotheses were postulated with respect to the directions of changes in the work performed at the lower extremity joints.

1.7 Significance

Previous research has identified three groups of structures that may affect the ability generate a midfoot moment to meet the demand of being both compliant and rigid, namely the extrinsic foot muscles, intrinsic foot muscles, and plantar ligaments. However, the relative contribution of each of these to the midfoot moment is poorly understood. Moreover, little research has been done to understand how these mechanisms may interact with the rest of the lower extremity, and therefore performance during propulsive tasks, such as vertical jumping. The research undertaken in this thesis seeks to further our understanding of how the extrinsic and intrinsic foot muscles may contribute to the net joint moment acting at the midfoot, and whether increasing the force generating capacity of these muscles may affect performance in vertical jumping.

Improving vertical jump height is of interest to many athletes and coaches since vertical jumping is an integral part of many sports, such as volleyball and basketball. Further, vertical jump performance is often used as one of the physical performance criteria in selections to professional leagues (Ebben & Blackard, 2001; Ebben et al., 2004). This work contributes to our understanding of how training can be optimized to improve vertical jump performance, as it addresses a form of training that has received little attention with respect to its effect on vertical jump performance. Moreover, furthering our understanding of how the foot muscles may contribute to the net joint moment exerted at the midfoot may be of interest in a clinical setting. This research may inform how the foot muscles may be trained to affect foot conditions, such as dynamic flexible flat foot deformity.

1.8 Delimitations

Participants in the research undertaken in this thesis were healthy individuals between 15 and 40 years old. Further, only female ice hockey and volleyball players were recruited for study 1 (Chapter 2), while only female volleyball players were recruited for study 3 (Chapter 4). Moreover, it was not the goal of this study to understand the effects of different foot arch types on the outcomes, and as such, no selection was made for individuals with specific functional or structural arch types.

Chapter 2

Forefoot and heel take-off strategies result in different distribution of lower extremity work during landings

2.1 Introduction

Many sports, such as volleyball, handball, and basketball, involve jumping and landing. Further, jumping is commonly used as a training modality for athletes engaged in sports where jumping and landing do not typically occur during game play (Ebben et al., 2004). Due to the high occurrence of non-contact injuries during landings, their mechanics have received considerable research attention (Afifi & Hinrichs, 2012; Chiu & Moolyk, 2015; Edwards et al., 2010; Kovacs et al., 1999; Self & Paine, 2001).

Two types of landing that have been described and studied are forefoot and heel landings (Kovacs et al., 1999; Self & Paine, 2001). During forefoot landings more ankle plantar flexor, and less knee and hip extensor net joint moment (NJM) work is performed, compared to heel landings (Kovacs et al., 1999). Moreover, larger ankle plantar flexor NJM and Achilles tendon force are exerted during forefoot landings, while knee and hip extensor NJM are larger in heel landings (Kovacs et al., 1999; Self & Paine, 2001). These findings demonstrate that less energy is absorbed through ankle plantar flexor NJM work during heel landings, resulting in a larger mechanical load placed on the knee and hip extensors. The reduced ability to absorb impact energy with the ankle plantar flexors may increase risk of injury, as injury is hypothesized to occur if a tissue's maximal tolerance for energy absorption is exceeded (Myers & Hawkins, 2010; Self & Paine, 2001; Zhang et al., 2000). Moreover, heel landings have a detrimental effect on the time spent transitioning from landing to propulsion, as well as the work performed during propulsion when consecutive jumps are performed (Kovacs et al., 1999). Thus, heel landings may also impede performance in situations where two or more consecutive jumps are required in competition.

Kovacs et al. (1999) and Self and Paine (2001) employed a drop landing protocol, where participants dropped from an overhead bar or stepped off a box. However, drop landing protocols are not representative of landings following a jump. Foot plantar flexion, and anterior leg and pelvis rotation are greater, whereas peak vertical ground reaction force and loading rate are smaller during landings following a jump (jump landing) compared to drop landings (Afifi & Hinrichs, 2012; Chiu & Moolyk, 2015; Edwards et al., 2010; Harry et al., 2017). Consequently, it is not known if similar lower extremity mechanics are present for heel landings following a jump compared to dropping from a bar or stepping off a box.

The majority of lower extremity muscles are activated throughout the flight phase following a jump and preceding the landing, whereas muscle activation onset during drop landings occurs immediately prior to impact (Afifi & Hinrichs, 2012; Edwards et al., 2010; Santello, 2005). This indicates that different mechanics during take-off may affect the subsequent landing. For example, the amount of ankle plantar flexor NJM work performed during the jump take-off may be hypothesized to affect the subsequent landing. Previous research has used restrictive footwear and fatigued the ankle plantar flexors to manipulate the amount of ankle plantar flexor NJM work performed during jumping (Arakawa et al., 2013; Bobbert et al., 2011; Haguenauer et al., 2006; Virmavirta & Komi, 2001). However, to examine whether jump take-off mechanics affect the subsequent landings, it is important to not constrain foot and ankle mechanics following take-off. An alternative way of manipulating ankle plantar flexor NJM work during jumping is to have participants shift their weight to the forefoot or the heel, and push through this part of the foot during the jump (forefoot and heel jumps, respectively).

The purpose of this study was to investigate whether landing mechanics differ between jumps performed with forefoot versus heel take-off strategies. We hypothesised that using a heel take-off strategy would reduce the ankle plantar flexor NJM work performed during take-off, subsequently reducing the ankle plantar flexor NJM work performed during landing compared to forefoot jumps.

2.2 Methods

2.2.1 Design

A repeated measures design was used to investigate the effect of the independent variable: take-off strategy, on the primary dependent variables: sagittal plane ankle, knee, and hip NJM work during jumping and landing, as well as foot, leg, thigh, and pelvis segment angles at initial foot contact. The independent variable had two levels, forefoot take-off strategy and heel take-off strategy.

2.2.2 Participants

Female volleyball (n = 17, age = 18 ± 2 yrs., stature = 1.76 ± 0.07 m, mass = 67.1 ± 7.3 kg) and ice hockey (n = 19, age = 17 ± 2 yrs., stature = 1.64 ± 0.06 m, mass = 62.9 ± 7.8 kg) players volunteered to participate. Volleyball and ice hockey players were recruited to obtain a participant pool consisting of athletic individuals with different experience performing vertical jumps. All participants played volleyball or ice hockey at a competitive level and were free of disease or injury preventing them from safely jumping and landing. Written informed consent was obtained from all participants (age ≥ 18 yrs.) or their legal guardians (age < 18 yrs.) prior to participation. Participants under 18 yrs. provided written assent. The study protocol (Study ID: Pro00067479) was approved by a research ethics board at the authors' institution.

2.2.3 Procedures

Participants performed six maximal countermovement vertical jumps in each of two conditions: 1) forefoot take-off and 2) heel take-off. Unsuccessful trials were repeated. No participant required more than eight attempts to obtain six successful trials in either condition. Participants determined length of their own rest period between attempts. The forefoot and heel conditions were completed in a randomised order. For the forefoot jumps, participants were instructed to shift the weight to the ball of their foot prior to the jump, and push through the forefoot during the entire jump. For the heel jumps, participants were instructed to shift the weight to the heel of their foot, and push through the heel during the entire jump. No instructions were given about the landings, and upon inquiry the participants were asked to land in a way that felt natural to them. An overhead target was suspended above the force platforms, as use of an overhead target increase maximal jump height (Mok et al., 2017). The participants were instructed to hold their hands at shoulder level with their elbows flexed and subsequently reach for the overhead target with both hands during the jumps (Chiu et al., 2017). This arm position was used to prevent the arms from obstructing the view of reflective markers placed on the participants' pelvis (Figure 2.1).

During the jumps and subsequent landings, retro-reflective markers' trajectories were recorded using seven optoelectronic cameras (ProReflex MCU 240; Qualisys, Gothenburg, Sweden) sampling at 120 Hz. The participants jumped and landed with each foot on separate force platforms measuring ground reaction forces (OR6-6; AMTI, Watertown, MA, USA) at 1200 Hz. A six-degree of freedom marker set consisting of 21 calibration and 25 tracking markers was used (Figure 2.1). Calibration markers were adhered bilaterally to the shoe over the 1st and 5th metatarsal heads, and to the skin over the medial and lateral malleolus, medial and lateral tibial condyles, medial and lateral femoral epicondyles, greater trochanters, and iliac crests using adhesive tape. Tracking clusters of three or four markers affixed to rigid thermoplastic plates were secured to the feet, legs, and thighs with neoprene straps, respectively. The proximal calibration markers of the pelvis as well as a marker placed between the spinous processes of L5 and S1 served as tracking markers for this segment. All participants wore tight fitting shorts (e.g. spandex), a t-shirt, and their own indoor training shoes.

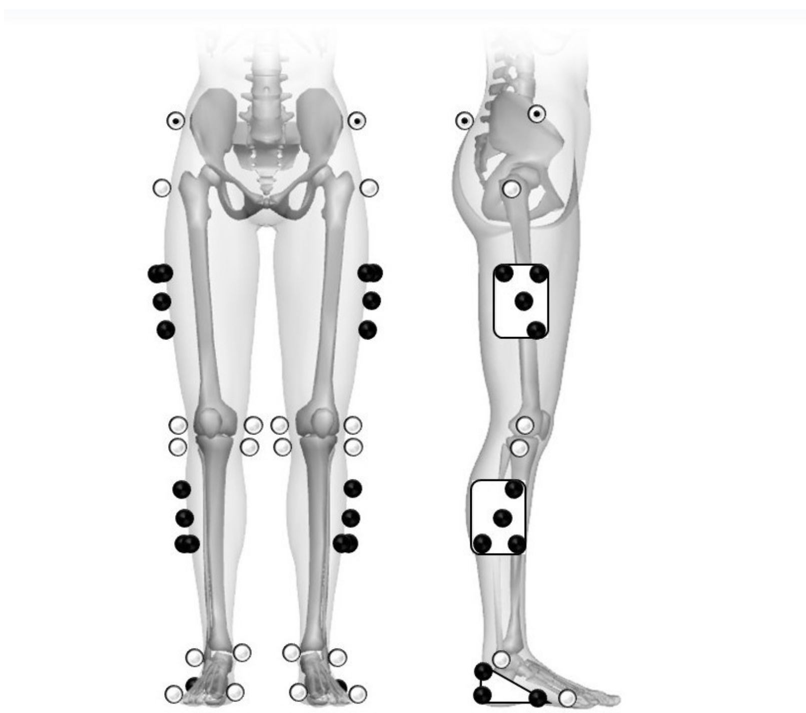

Figure 2.1 Frontal plane (left panel) and sagittal plane (right panel) view of the marker set used. Black markers served as tracking marker, white markers served as calibration markers, and white markers with black dot served as both tracking and calibration markers.

2.2.4 Data processing and reduction

Data from minimum five of the six trials were processed and used for analysis. The reason for excluding trials was technical difficulties, such as a marker not being captured during a recording. A rigid body model consisting of seven segments representing the pelvis and both thighs, legs, and feet was created using Visual 3D (Version 5.0; C-motion, Germantown, MD, USA). Marker and force data were filtered using a 4^{th} order recursive Butterworth filter with a 12 Hz cut-off frequency. Marker and force data were filtered using the same cut-off frequency as previous research has found that artefacts are introduced in NJM data when marker and force data are filtered using different cut-offs (Kristianslund et al., 2012). The filter cut-off frequency was selected based on a residual analysis of the marker data (Kristianslund et al., 2012; Wells & Winter, 1980). Segment and laboratory coordinate systems conformed to the right hand rule with the positive X-, Y-, and Z-axes pointing right, anteriorly, and up, respectively. Segment angles were calculated as the orientation of the segment coordinate systems relative to the laboratory coordinate system using a ZYX Cardan sequence. Joint angles were calculated as the orientation of the proximal relative to the distal segment coordinate system using an XYZ Cardan sequence. A Newton-Euler inverse dynamics procedure was used to compute ankle, knee, and hip NJMs, which were expressed in the coordinate system of the distal segments. Segments' inertial properties were determined based on segments having the shape of conical frusta. Segment masses were determined using anthropometric data from Dempster (1955). Ankle, knee, and hip power were computed as the dot product of the local sagittal plane NJM and joint angular velocity, and then integrated using the trapezoid rule to yield NJM work performed at the respective joints. Only sagittal plane kinematics and kinetics were extracted for analysis. Jump height was determined using the pelvic kinematic method (Chiu & Salem, 2010). The anterior-posterior location of the participants' centre of pressure was calculated and expressed as a percentage of the distance between the distal and proximal end of the foot segment. The distal and proximal ends of the foot were defined as the midpoint between the 1st and 5th metatarsal heads and the midpoint between medial and lateral malleolus, respectively. The distal and proximal ends of the foot were defined as 0% and 100% of foot length respectively, and negative percentages indicate that the centre of pressure was located anterior to the midpoint between the 1st and 5th metatarsal heads. The average anterior-posterior position of the centre of pressure during the propulsion phase (defined below) was used for further analyses. Previous research has found vertical jumping and landing to be bilaterally symmetrical tasks (Harry et al., 2017). Centre of pressure and segment kinematics were therefore averaged across limbs, whereas joint kinetics were summed between limbs and normalised to body mass. Subsequently, the variables of interest were averaged across the six trials performed in each condition.

The events and phases of interest were the propulsion phase, initial foot contact, and landing. The propulsion phase was defined to begin the instant the vertical velocity of the pelvis' centre of mass became negative and ended the instant the vertical ground reaction force dropped below 10 N as the foot left the force platform. Initial foot contact was defined as the instant vertical ground reaction force increased above 10 N as the foot contacted the force platform. The landing was divided into two separate phases, the first starting at initial foot contact and ending when sagittal plane angular velocity of the foot segment became zero (foot flat). The second landing phase was defined to begin at foot flat and ended when knee flexion angle was maximal (peak knee flexion; 0° defined as fully extended). The foot flat event dividing these phases is important, as the foot is no longer free to rotate, which affects the ankle joint excursion and ankle plantar flexor NJM work that may be performed following this event.

2.2.5 Statistical analyses

Multivariable ANOVAs with repeated measures were used to compare the effect of forefoot and heel take-off on the dependent variables: Sagittal plane ankle, knee, and hip NJM work performed during the propulsion and the first and second landing phases, as well as foot, leg, thigh, and pelvis angles at initial foot contact. Sagittal plane ankle, knee, and hip NJM during these phases, as well as sagittal plane foot, leg, thigh, and pelvis angles at the event defining their end were assessed to explain any differences in lower extremity NJM work. Therefore, these analyses were performed as *post hoc* tests only when differences in NJM work were statistically significant in order to avoid multiplicity issues (Knudson, 2009). The ankle, knee, and hip were used as multivariate levels for analyses of NJM work and NJM, whereas the foot, leg, thigh, and pelvis were used as multivariate levels for analyses of segment angles. Univariate analyses were only considered when the multivariate test was significant using Wilk's λ. The data was normally distributed as assessed by visual inspection of Q-Q plots. Mauchly's test was used to assess sphericity. In instances where the sphericity assumption was violated, Greenhouse-Geisser correction was used. Mean ± standard deviation (SD) and 95% confidence intervals (CI) of simple effect sizes are reported (Baguley, 2009). The level of significance was set *a priori* to 0.05. Statistical analyses were conducted in SPSS (Version 24; SPSS Inc., Chicago, IL, USA).

2.3 Results

Participants jumped higher during forefoot (38.6 ± 5.1 cm) compared to heel (29.5 ± 7.7 cm; $p < 0.01$; CI = [−0.12 −0.06]) jumps, and the centre of pressure was located at −3.1 ± 11.1% and 75.7 ± 15.4% of their foot length during forefoot and heel jumps, respectively (Figure 2.2). Net joint moment power curves from a representative forefoot and heel jump trial are presented in Figure 2.4. Further, more ankle plantar flexor and knee extensor NJM work was performed during the propulsion phase of forefoot (1.30 ± 0.26 $J \cdot kg^{-1}$ and 0.99 ± 0.40 $J \cdot kg^{-1}$, respectively) compared to heel (0.80 ± 0.47 $J \cdot kg^{-1}$; $p < 0.01$; CI [−0.74 −0.28] and 0.78 ± 0.32 $J \cdot kg^{-1}$; $p < 0.01$; CI [−0.36 −0.05], respectively; Figure 2.3) jumps. There were no differences in hip extensor NJM work during the propulsion phase ($p > 0.05$; CI [−0.23 0.06]). The average ankle plantar flexor NJM was greater, whereas knee extensor NJM was smaller during the propulsion phase in forefoot compared to heel jumps ($p < 0.01$; Table 2.1). The foot plantar flexion, anterior pelvis, and posterior thigh rotation angles were greater at take-off in forefoot compared to heel jumps ($p < 0.01$; Table 2.1).

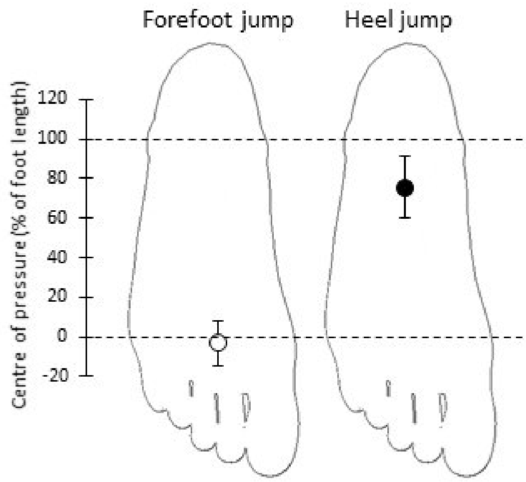

Figure 2.2. Average anterior-posterior location of the participants' centre of pressure from initiation of the countermovement until take-off during forefoot (white circle) and heel (black circle) jumps. The centre of pressure location was expressed as a percentage of foot length (see text for definition) where 0% represents the midpoint between the 1st and 5th metatarsal heads, and 100% represents the midpoint between the medial and lateral malleolus.

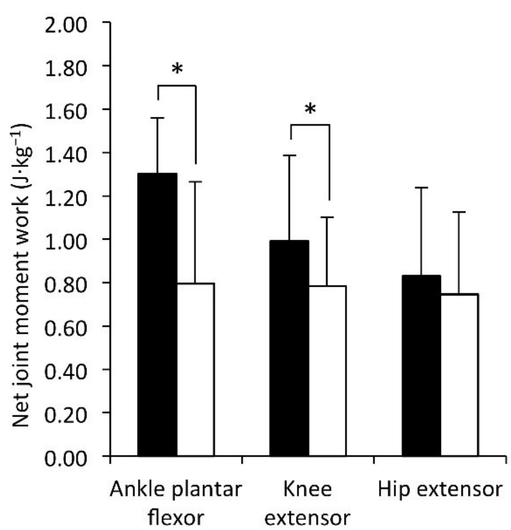

Figure 2.3. Ankle plantar flexor, knee extensor, and hip extensor net joint moment work performed during forefoot (black bars) and heel (white bars) take-offs.
* Significant difference between forefoot and heel jumps, $p < 0.05$.

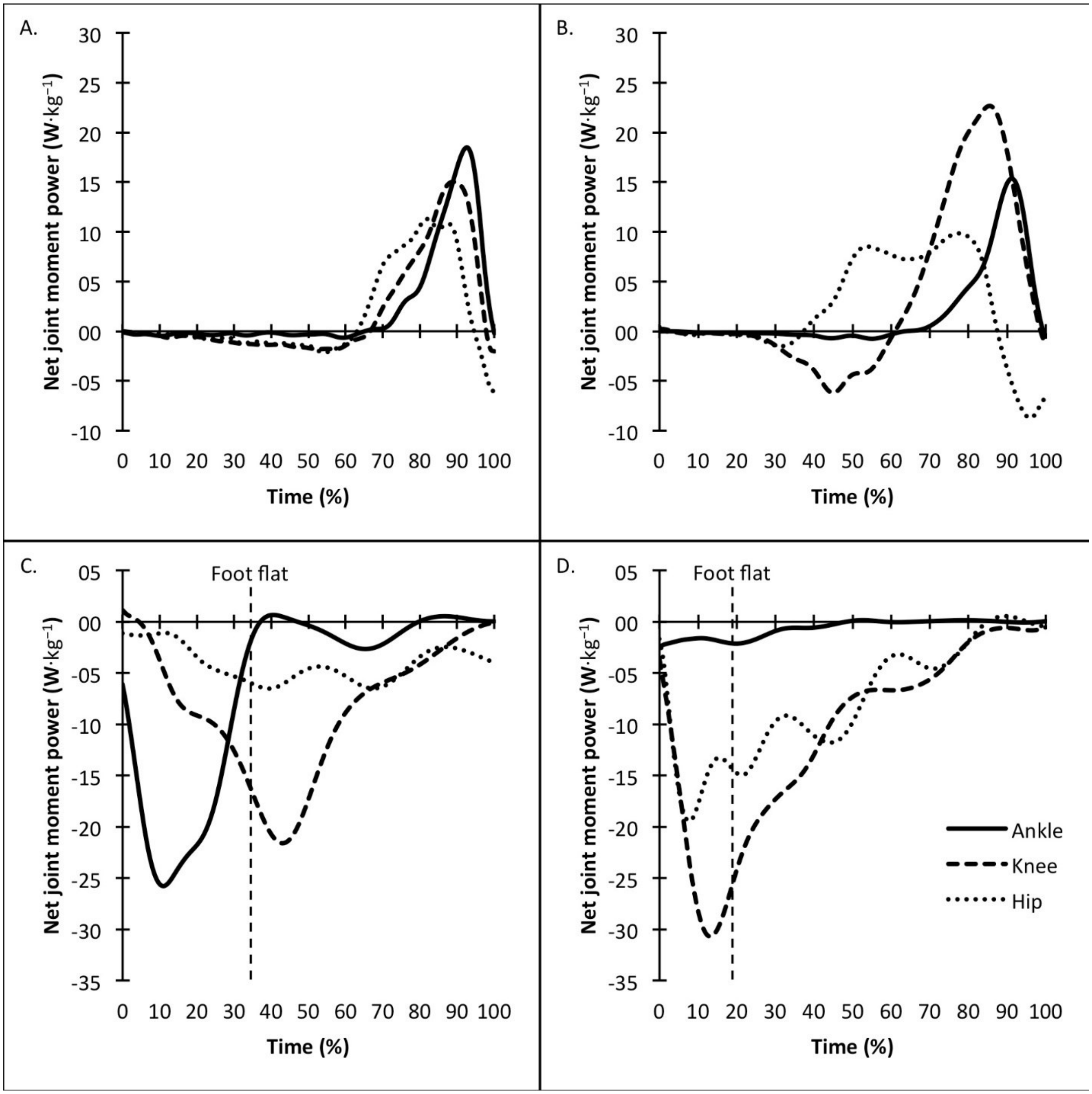

Figure 2.4. Ankle (solid lines), knee (dashed lines), and hip (dotted lines) net joint moment power during jumping (panel A and B) and landing (panel C and D) from a representative forefoot (panel A and C) and heel (panel B and D) jump trial. Time during jumps and landings are expressed as a percentage of the duration from countermovement initiation to take-off and from initial foot contact to peak knee flexion, respectively. In panel C and D, the instant of foot flat is indicated with a vertical dashed line.

Table 2.1. Average lower extremity net joint moment (NJM) exerted during propulsion, from initial foot contact to foot flat (landing phase 1), and from foot flat to peak knee flexion (landing phase 2) and lower extremity segment angles at take-off, foot flat, and peak knee flexion for the forefoot and heel jump conditions.

Phase/event		**Forefoot take-off**	**Heel take-off**	**95% CI**
Propulsion phase:				
Ankle NJM *	($Nm{\cdot}kg^{-1}$)	−1.30 ± 0.22	−0.27 ± 0.23	[0.89 1.17]
Knee NJM *	($Nm{\cdot}kg^{-1}$)	1.05 ± 0.24	1.19 ± 0.18	[0.05 0.24]
Hip NJM	($Nm{\cdot}kg^{-1}$)	−1.49 ± 0.33	−1.47 ± 0.28	[−0.07 0.10]
Take-off:				
Foot angle *	(°)	−33.0 ± 3.6	−16.0 ± 7.8	[12.4 21.6]
Leg angle	(°)	−0.5 ± 2.4	−1.6 ± 4.4	[−2.7 0.5]
Thigh angle *	(°)	0.2 ± 2.2	8.5 ± 4.1	[6.8 9.7]
Pelvis angle *	(°)	−13.2 ± 5.3	−11.5 ± 6.6	[0.5 2.9]
Landing phase 1:				
Ankle NJM *	($Nm{\cdot}kg^{-1}$)	−1.89 ± 0.39	−1.00 ± 0.73	[0.58 1.17]
Knee NJM	($Nm{\cdot}kg^{-1}$)	1.32 ± 0.50	1.58 ± 0.81	[−0.02 0.56]
Hip NJM *	($Nm{\cdot}kg^{-1}$)	−0.18 ± 0.41	−0.61 ± 0.76	[−0.74 −0.11]
Foot flat:				
Foot angle *	(°)	−4.3 ± 3.2	−1.3 ± 2.4	[1.8 4.1]
Leg angle *	(°)	−32.3 ± 4.1	−26.4 ± 4.4	[4.0 7.8]
Thigh angle *	(°)	27.8 ± 6.5	31.8 ± 9.1	[1.0 7.0]
Pelvis angle	(°)	−21.5 ± 7.2	−21.6 ± 8.0	[−1.8 1.6]
Landing phase 2:				
Ankle NJM	($Nm{\cdot}kg^{-1}$)	−2.01 ± 0.45	−1.08 ± 0.66	[0.66 1.19]
Knee NJM	($Nm{\cdot}kg^{-1}$)	2.40 ± 0.52	2.65 ± 0.62	[0.03 0.47]
Hip NJM	($Nm{\cdot}kg^{-1}$)	−1.82 ± 0.77	−1.92 ± 0.90	[−0.41 0.20]
Peak knee flexion:				
Foot angle	(°)	−5.2 ± 4.0	−1.4 ± 2.6	[2.4 5.4]
Leg angle	(°)	−38.8 ± 4.5	−31.5 ± 4.9	[5.4 9.1]
Thigh angle	(°)	50.5 ± 13.9	52.4 ± 14.5	[−0.1 4.1]
Pelvis angle	(°)	−24.9 ± 10.0	−24.8 ± 9.6	[−1.3 1.4]

Positive values indicate ankle dorsiflexor, knee extensor, and hip flexor NJM, and foot dorsiflexion, and posterior leg, thigh, and pelvis rotation angles.
* Statistically significant difference between forefoot and heel jumps, $p < 0.05$.
95% CI = 95% confidence interval

At initial foot contact, the participants' feet were more plantar flexed following forefoot compared to heel jumps ($p < 0.01$; CI [17.1 27.7]; Figure 2.5). In contrast, anterior leg rotation ($p < 0.01$; CI [−3.9 −1.3]), anterior pelvis rotation ($p < 0.01$; CI [3.0 5.7]), and posterior thigh rotation ($p < 0.01$; CI [−4.1 −0.5]) angles were smaller following forefoot compared to heel jumps (Figure 2.5).

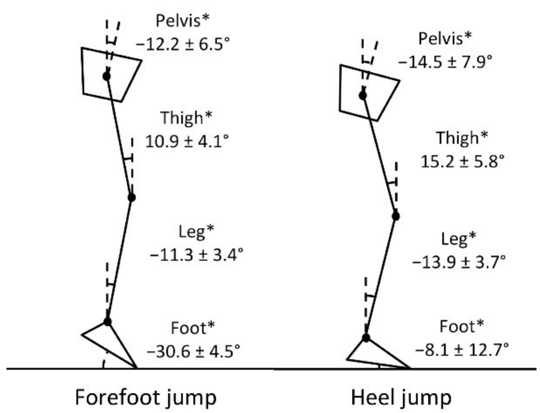

Figure 2.5. Foot, leg, thigh, and pelvis segment angles at initial foot contact following forefoot and heel take-offs.
* Significant difference between forefoot and heel jumps, $p < 0.05$.

Participants performed more ankle plantar flexor NJM work from initial foot contact to foot flat following forefoot (-1.33 ± 0.38 J·kg^{-1}) compared to heel (-0.46 ± 0.48 J·kg^{-1}; $p < 0.01$; CI [0.65 1.09]; Figure 2.6) jumps. In contrast, less hip extensor NJM work was performed from initial foot contact to foot flat following forefoot (-0.07 ± 0.20 J·kg^{-1}) compared to heel (-0.33 ± 0.39 J·kg^{-1}; $p < 0.01$; CI [−0.43 −0.08]; Figure 2.6) jumps. There was no difference in knee extensor NJM work from initial foot contact to foot flat ($p > 0.05$; CI [−0.28 0.14]). Average ankle plantar flexor NJM was greater, whereas average hip extensor NJM was smaller from initial foot contact to foot flat following forefoot compared to heel jumps ($p < 0.01$; Table 2.1). Foot plantar flexion and anterior leg rotation angles were greater, whereas posterior thigh rotation angles were smaller at foot flat in forefoot compared to heel jumps ($p < 0.01$; Table 2.1).

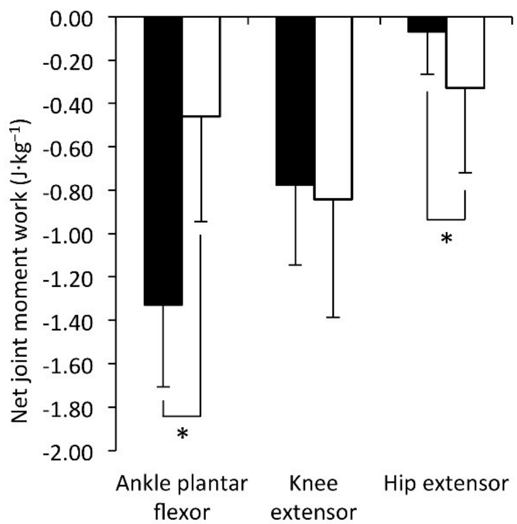

Figure 2.6. Ankle plantar flexor, knee extensor, and hip extensor net joint moment work performed from initial foot contact to foot flat during forefoot (black bars) and heel (white bars). * Significant difference between forefoot and heel jumps, $p < 0.05$.

There were no differences in the NJM work performed at the ankle, knee, or hip between the forefoot and heel jumps from foot flat until peak knee flexion (Wilk's $\lambda = 0.84$; $p = 0.06$; Figure 2.7). Consequently, lower extremity NJM from foot flat to peak knee flexion or segment angles at peak knee flexion were not statistically compared between conditions (Table 2.1).

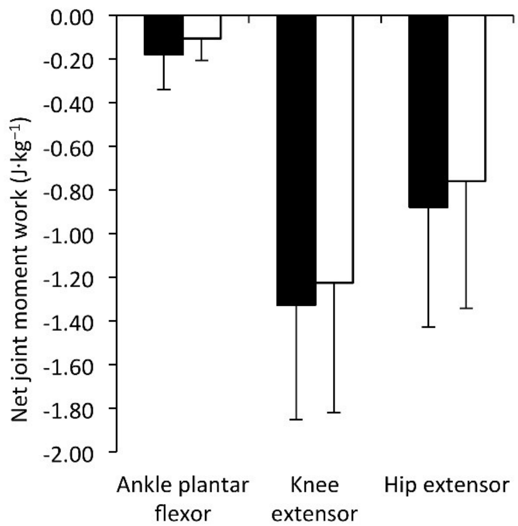

Figure 2.7. Ankle plantar flexor, knee extensor, and hip extensor net joint moment work performed from foot flat to peak knee flexion during forefoot (black bars) and heel (white bars).

2.4 Discussion

The present study investigated whether manipulating the location of the centre of pressure during jump take-off affect the subsequent landing. Our main finding is that maintaining the centre of pressure under the rearfoot, as opposed to the forefoot during jump take-off affects the distribution of lower extremity NJM work performed during landing. Specifically, more negative ankle plantar flexor NJM work was performed to absorb the impact energy during landings following forefoot jumps, whereas more negative hip extensor NJM work was performed during landings following heel jumps. These differences in NJM work distribution occurred between initial foot contact and foot flat, whereas no differences in lower extremity NJM work distribution were observed following foot flat.

More ankle plantar flexor NJM work was performed during the jump take-off using a forefoot compared to heel strategy. This larger NJM work was performed due to both larger ankle plantar flexor NJM and larger foot angular excursion during jump take-off. The differences observed in ankle plantar flexor NJM work, NJM, and foot excursion between forefoot and heel jumps are comparable to those between jumps performed with and without footwear that restricts ankle plantar flexion (Arakawa et al., 2013; Haguenauer et al., 2006). It should be noted that some foot plantar flexion occurred also during heel take-offs. However, this plantar flexion may be occurring passively due to gravity acting on the foot. Specifically, the force of gravity, acting on the foot centre of mass, would cause some plantar flexion of the foot as it comes off the ground, regardless of how much ankle plantar flexor moment is exerted. Arakawa et al. (2013) made a similar observation despite using a customized shoe that restricted ankle plantar flexion during jumping. Nonetheless, maintaining the centre of pressure closer to the heel throughout the propulsion phase of jumping resulted in a reduction of ankle plantar flexor work performed during take-off, as we had hypothesised.

Knee extensor NJM work performed during the propulsion phase was also greater using a forefoot compared to heel take-off strategy. However, knee extensor NJM was lower when using a forefoot versus heel take-off strategy. The smaller knee extensor NJM in the forefoot strategy can be explained by the larger ankle plantar flexor NJM, as both moments act on and tend to rotate the leg segment in the same direction (Flanagan & Salem, 2008). Consequently, the knee extensor NJM is smaller in the forefoot jumps where ankle plantar flexor NJM is large, compared to the heel jumps where ankle plantar flexor NJM is smaller. Due to the smaller knee extensor NJM, participants had to go through a larger knee angular excursion as more knee extensor NJM work was performed in forefoot compared to heel jumps. This was achieved by extending the thigh more prior to take-off in the forefoot jumps. This finding supports the hypothesis that minimising foot plantar flexion during jumping will prevent the knee from fully extending at take-off (van Ingen Schenau et al., 1987). Taken together with results from previous research investigating jumping with limited foot plantar flexion, it appears that foot plantar flexion is necessary to achieve full knee extension at take-off (Arakawa et al., 2013; Haguenauer et al., 2006).

The larger ankle plantar flexor and knee extensor NJM work performed during the propulsion phase of forefoot jumps, resulted in approximately 30% greater vertical jump height compared to heel jumps. Consequently, the participants possessed more kinetic energy at impact following forefoot, compared to heel take-offs. It is therefore not surprising that more total lower extremity NJM work was performed to absorb energy following forefoot (-4.6 J·kg^{-1}) compared to heel take-offs (-3.7 J·kg^{-1}). However, from initial foot contact until foot flat, the larger NJM work performed was not uniformly distributed between the lower extremity joints. Specifically, more ankle plantar flexor and less hip extensor NJM work was performed during this phase following the forefoot compared to heel take-off. Similarly, participants exerted larger ankle plantar flexor and smaller hip extensor NJM from initial foot contact to foot flat following the forefoot compared to heel take-off. These differences in lower extremity NJM work and NJM distribution are similar to those reported by Kovacs et al. (1999). Kovacs et al. (1999) observed that ankle plantar flexor NJM work and NJM were greater during forefoot landings, whereas knee and hip extensor NJM work and NJM were greater during heel landings. Our findings demonstrate that using a forefoot take-off strategy results in landing mechanics similar to those observed during forefoot drop landings, whereas use of a heel take-off strategy results in landing mechanics similar to those observed during heel drop landings. Researchers and practitioners evaluating an individual's landing mechanics should consider the take-off strategy used.

The heel take-off used in the present study represents an artificial manipulation of jumping technique that may or may not be employed by athletes. Nonetheless, the results of this study demonstrate proof of concept for the effect of take-off strategy on the subsequent landing. Moreover, situations in which a take-off strategy similar to that of the heel take-off have been described. One notable example is when the ankle plantar flexors are fatigued. Specifically, ankle plantar flexor NJM work and activation is greatly reduced during jumping when these muscles are fatigued (Bobbert et al., 2011). Ankle plantar flexor fatigue may be induced during running or repeated jumping during participation in sports such as volleyball, handball, and basketball (Perrey et al., 2010). Thus, take-off strategies resembling that of the heel take-off could be postulated to occur in game situations where the athlete is fatigued, such as following intense bouts of sprinting or near the end of a game. Another situation in which a take-off strategy similar to the heel take-off may occur is when the ankle is taped or braced. Ankle taping and bracing have been found to affect passive ankle range of motion (Myburgh et al., 1984). Moreover, taping and bracing affects ankle plantar flexor NJM work and ankle range of motion during drop landings (Cordova et al., 2010; Gardner et al., 2012). Thus, it seems reasonable to hypothesise that ankle taping or certain ankle braces may sufficiently restrict ankle plantar flexor NJM work during take-off to affect the subsequent landings. Future research should determine whether take-off strategies similar to the heel take-off occur naturally in situations such as those described above.

2.5.1 Conclusion

In support of our hypotheses, a heel take-off strategy reduces ankle plantar flexor work performed during jumping, leading to a smaller jump height in comparison to a forefoot take-off strategy. Moreover, the reduced ankle plantar flexor work during jumping results in less ankle plantar flexor, but more hip extensor NJM work being performed from initial foot contact to foot flat. Future research is required to determine whether fatigue, sport specific footwear, or ankle taping and bracing may cause a heel take-off strategy to be used.

Conflict of interest: Authors declare no conflict of interest

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Chapter 3

Roles of the intrinsic and extrinsic foot muscles in supporting the medial longitudinal arch

3.1 Introduction

The feet are the only point of contact with the ground in many tasks, such as walking, running, and jumping. In such tasks, force is exerted on the rearfoot by the triceps surae muscles, transmitted to the forefoot, and subsequently the ground (Aronow et al., 2006; Blackman et al., 2009). However, the foot is not a rigid structure, but may be deformed by internal and external forces acting on it. Notably, the force exerted on the rearfoot by the triceps surae, the joint reaction force at the ankle, and the vertical ground reaction forces acting on the fore- and rearfoot all tend to deform the foot (Blackman et al., 2009; Cheung et al., 2006; Kelly et al., 2014). This deformation may be desirable in activities where the foot absorbs impact energy or stores elastic energy for release later in the movement cycle (Ker et al., 1987; Riddick et al., 2019). However, during many propulsive tasks, greater arch rigidity is desirable as it allows rapid transmission of energy from the triceps surae muscles to the ground (Kelly et al., 2015; Takahashi et al., 2016). To meet this seemingly paradoxical requirement of being both compliant and rigid, the foot must modulate its rigidity.

Multiple theories for how the foot may achieve this modulation have been postulated. One theory suggests that foot rigidity is modulated through deformation of passive tissues as the alignment of the foot skeleton changes during movement (Hicks, 1954). Specifically, Hicks (1954) theorized that plantar aponeurosis tension increases when it is wound around the metatarsal heads during toe extension, which in turn would increase the rigidity of the foot arch. Hicks' theory has later been corroborated by cadaveric studies demonstrating that plantar aponeurosis tension and foot arch height increases, while foot arch length decreases when the toes are extended (Thordarson et al., 1997). However, as this mechanism is passive, it may only modulate foot rigidity as a function of the relative orientation of the foot bones. Moreover, it contrasts with observations made during gait, where the foot arch is recoiling and becoming less stiff as the toes are extended (Welte et al., 2018). The latter observations suggests that an active mechanism may contribute to modulating foot arch rigidity.

One active mechanism that has been theorized is that extrinsic foot muscle activation is responsible for modulating foot arch rigidity. Specifically, it is purported that the tibialis posterior, fibularis longus, flexor hallucis longus, and flexor digitorum longus muscle all contribute to increasing foot arch rigidity as they cross several of the intertarsal joints (Basmajian & Stecko, 1963). The first observations supporting this theory were made by Basmajian and Stecko (1963) who found that activation of the tibialis posterior, fibularis longus, and flexor digitorum longus muscles increased as they increased external loads applied to the foot. Later, Thordarson et al. (1995) observed an increase in foot arch height when pull was exerted on the tibialis posterior tendon, corroborating this theory. However, the triceps surae can cause foot arch deformation due to pulling the calcaneus into plantar flexion (Blackman et al., 2009; Thordarson et al., 1995). As none of the extrinsic foot muscles act on the calcaneus, their ability to oppose calcaneal plantar flexion may be limited. Therefore, it is likely that another active mechanism is responsible for opposing this form of arch deformation.

In contrast to the extrinsic foot muscles, several of the intrinsic foot muscles act directly on the calcaneus and have a relatively large moment arm with respect to the intertarsal joints (Farris et al., 2019). For this reason, the intrinsic foot muscles are theorized to contribute to modulation of foot arch rigidity (Kelly et al., 2014; Kelly et al., 2012; Kelly et al., 2015). Fiolkowski et al. (2003) provided support for this theory, demonstrating that foot deformation increases when disabling the intrinsic foot muscles using a tibial nerve block. Other studies have found that activation of the abductor hallucis, flexor digitorum brevis, and quadratus plantae muscles increases with increasing external loads acting on the foot (Kelly et al., 2014; Kelly et al., 2015). Moreover, using intramuscular electrical stimulation, Kelly et al. (2014) demonstrated that the aforementioned muscles can increase foot arch height and shorten its length despite the presence of large arch deforming external loads.

Taken together, there are three prevailing theories for how the foot arch rigidity may be modulated to meet the demands of a given task. Although supporting evidence exists for each of these theories, little has been done to understand the relative contribution of each mechanism to foot arch support, or how they interact. The purpose of this study was to estimate the relative contributions of the extrinsic muscles, intrinsic muscles, and foot ligaments/aponeuroses to foot arch support in loading conditions similar to those experienced during daily tasks.

3.2 Methods

3.2.1 Study design

A two-way repeated measures design was employed to explore the theoretical potential for the extrinsic foot muscles, intrinsic foot muscles, and passive elastic foot structures (i.e., plantar aponeurosis and foot ligaments) to contribute to foot arch support. The independent variables that were manipulated was the external load applied to the foot and whether maximal voluntary toe flexor muscle contractions were performed or not. An available open-source musculoskeletal model was modified for the present study and individually scaled to represent each of 12 participants. Inverse solutions for eight different combinations of external loading and voluntary toe flexor muscle activation were computed using the scaled models and motion capture data. Subsequently, static optimization was performed to estimate the distribution of observed net joint moments (NJM) between the muscles and ligaments of the models.

3.2.2 Participants

Six healthy females (age: 23 ± 3 yrs., stature: 1.62 ± 0.05 m, mass: 57.3 ± 5.6 kg) and six healthy males (age: 27 ± 6 yrs., stature: 1.78 ± 0.08 m, mass: 78.7 ± 8.8 kg) volunteered to participate in the study. Participants were free of injury to the right foot and ankle, foot deformities (such as hallux valgus), and systemic disease that may affect the function of the foot and/or ankle. Participants were informed of the study procedures and provided written consent prior to participation. The study protocol was approved by a Research Ethics Board at the authors' institution (Study ID: Pro00086659).

3.2.3 Data acquisition

Participants were instrumented with 11 retroreflective markers place on their right leg and foot. Specifically, markers were placed on the medial and lateral tibial condyles, medial and lateral malleoli, most posterior aspect of the calcaneal tuberosity, navicular tuberosity, cuboid, and the 1^{st} and 5^{th} metatarsal heads. Additionally, a tracking marker was placed on each of the medial and lateral surfaces of the calcaneus. Marker trajectories were recorded with 6 optoelectronic cameras (ProReflex MCU 240, Qualisys, Gothenburg, Sweden) sampling at 100 Hz. The ground reaction force acting under the right forefoot was simultaneously recorded with a Bertec force platform (Bertec FP-4060-10-2000, Bertec, Colombus, OH, USA) sampling at 1000 Hz.

Participants were recorded in two experimental conditions with four different loads in each condition. Both conditions resemble that used by Kelly et al. (2019), and involved sitting on an adjustable height seat with the hip and knee flexed to approximately 90 degrees and the forefoot resting on the edge of the force platform (Figure 3.1). The force platform was raised approximately 10 cm from the surrounding floor. The rearfoot was suspended over the floor and participants were instructed to engage their ankle plantar flexors to maintain the sole of the foot parallel to the ground for the entire duration of each trial. A wooden block with a rectangular sponge glued on top, and that was flush with the surface of the force platform, was used to provide tactile feedback to assist the participants in maintaining the foot parallel to the ground. In one of the two conditions, the participants were instructed to relax their toe flexor muscles as much as possible while engaging the ankle plantar flexors only to the extent necessary to maintain the prescribed position while external loads were lowered onto their distal thigh. In the second condition, participants were instructed to maximally voluntarily contract (MVC) their toe flexors muscles by pressing their toes as hard as possible against the force platform while the load was applied. The loads used were 0%, 50%, 100%, and 150% of the participants' body mass, and were gently lowered onto the participants' distal thigh by the investigators. These loads were selected as they would yield ground reaction forces representative of those observed during bipedal stance, unipedal stance, and gait (Kelly et al., 2019). Participants were recorded with the motion capture system for at least 3 seconds while the load was resting on their thigh. The two conditions were performed in a randomized order, while heavier external loads always followed lighter loads within each of the two conditions.

Figure 3.1. Illustration of the experimental data setup. The illustration does not show the sponge used for tactile feedback or the padding that was placed between the barbell and the participants' thigh.

3.2.4 Musculoskeletal model

The musculoskeletal model used in the present study was created by modifying a previously described open-source musculoskeletal model (Malaquias et al., 2017). The model editing was performed using OpenSim's MATLAB (R2019a, MathWorks, Inc., Natick, MA, USA) application programming interface and all simulations were performed in OpenSim 4.1 (Delp et al., 2007). The edited model consisted of 6 segments and had 8 degrees of freedom. Further, the model was actuated by 2 torque actuators and 46 hill-type muscle actuators and had 26 passive elastic elements representing ligaments and aponeuroses (Figure 3.2).

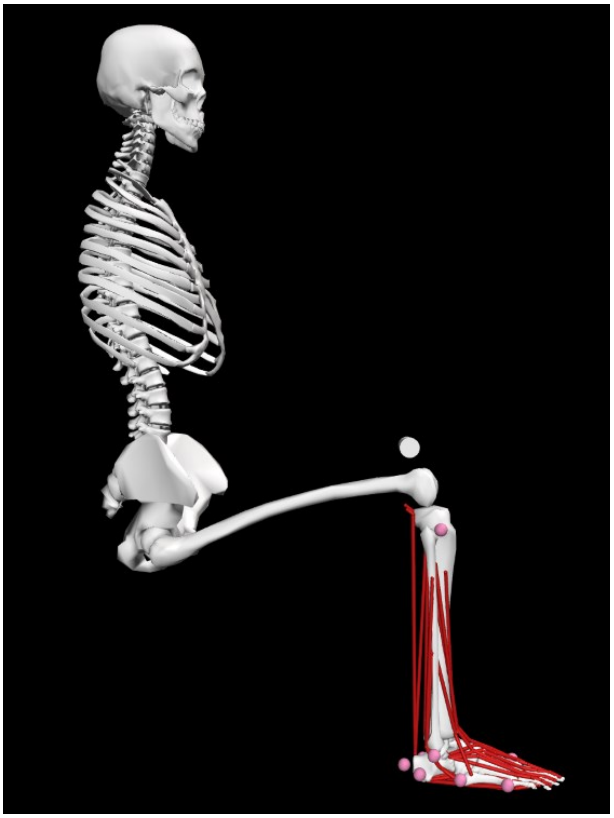

Figure 3.2. The musculoskeletal model used in the present study.

3.2.4.1 Segments and joints

A two-dimensional model of the right lower extremity as well as head, arms, and trunk (HAT) was used in the present investigation. The model was created by removing the left lower extremity, as well as all non-sagittal translational and rotational degrees of freedom from the Malaquias et al. (2017) musculoskeletal model. Further, the joints connecting the torso and pelvis, talus and calcaneus, and midfoot and forefoot segments were replaced by weld joints, effectively reducing the number of rigid bodies making up the model. A rigid body representing the external load used during data acquisition was added to the model and connected to the distal thigh with a weld joint. The result of these modifications was a planar model consisting of 6 rigid segments representing the HAT, right thigh with barbell, right leg, right rearfoot, right forefoot, and right toes. The model had 8 degrees of freedom, allowing anterior-posterior and vertical translation, and sagittal plane rotation of the HAT segment. Further, the model allowed flexion/extension at the hip and knee, and dorsiflexion/plantarflexion at the ankle, transverse tarsal, and metatarsophalangeal joints. All joints were idealized as pin joints. Thus, the foot arch was represented by the rigid rearfoot and forefoot, connected by a pin joint at the approximate location of the talonavicular and calcaneocuboid joints (Malaquias et al., 2017). This is consistent with several previous musculoskeletal models used to investigate arch function (Erdemir & Piazza, 2004; Simkin & Leichter, 1990).

3.2.4.2 Torque actuators

All muscles crossing the hip and knee joint (except the medial and lateral gastrocnemius) were replaced by a single torque actuator at each joint. The torque of each actuator was linearly dependent on the activation of the actuator and may be considered equivalent to the NJM at the hip, and the NJM without the contribution of medial and lateral gastrocnemius at the knee, respectively. The optimal force of these torque actuators was set to 300 Nm for the knee and 250 Nm for the hip (Krantz et al., 2020; O'Brien et al., 2010). This simplification was made to reduce model complexity, as well as the number of input parameters that needed to be estimated. This simplification is considered justified as muscles crossing the hip and knee joint had no direct bearing on the research question.

3.2.4.3 Muscle-tendon models

The muscles of the triceps sura, extrinsic foot muscles, and intrinsic foot muscles were represented by Hill-type muscle models available in OpenSim (Thelen, 2003; Zajac, 1989). Briefly, each muscle model consists of a length and velocity dependent contractile element in parallel with an exponential spring element, representing the active contractile and passive elastic properties of muscle fibres, respectively. These are arranged in series with a spring element which represents the elastic properties of tendons and intramuscular connective tissue of the muscle, and has an exponential toe region and is linear beyond this region (Thelen, 2003). Each of the muscle models requires the specification of the maximal isometric force the muscle can generate, its optimal muscle fibre length, its tendon slack length, and the muscle pennation angle at optimal fibre length (Zajac, 1989). In addition, the geometry of the muscle path is required to compute each muscles' effect on the dynamics of each model segment.

For the triceps sura and extrinsic foot muscles, muscle geometries and properties were retained from the Malaquias et al. (2017) model. However, this model does not represent the intrinsic foot muscles as Hill-type muscles. Therefore, we replaced the linear actuators representing the intrinsic foot muscles in the Malaquias et al. (2017) model with Hill-type models with geometry and muscle parameters taken from (Lachowitzer et al., 2007; Figure 3.3).

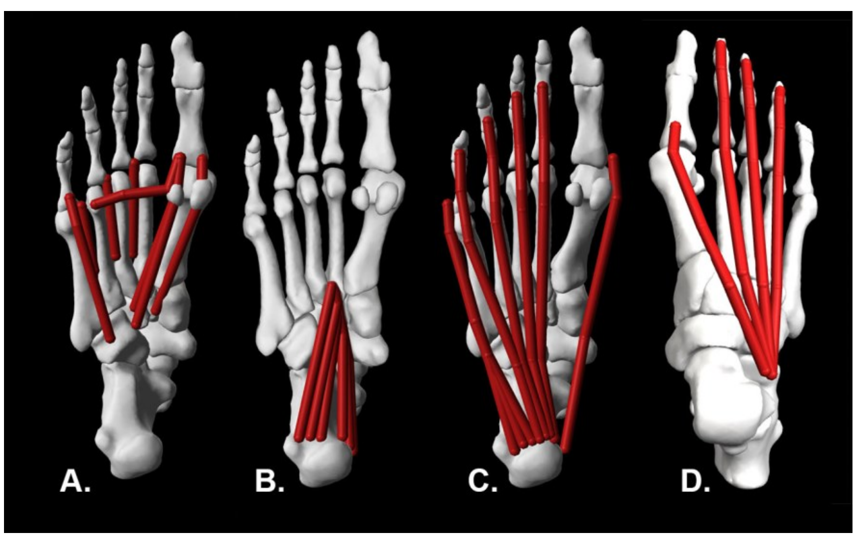

Figure 3.3. The muscle models representing the intrinsic foot muscles in the deep (A.), middle (B.), and superficial (C.) plantar layers, and the dorsal (D.) layer.

3.2.4.4 Ligament models

Foot and ankle ligaments crossing the ankle, transverse tarsal, and metatarsophalangeal joints, as well as the plantar aponeurosis were modelled as non-linear springs. All ligament models, except for the plantar aponeurosis model was retained from the Malaquias et al. (2017) model, whereas the force-length characteristics of the plantar aponeurosis was modelled as described by Wager and Challis (2016). The resting length of the ligaments were assumed to be equal to the computed ligament lengths during a static trial where participants were seated with the entire foot supported on the force platform, and without external load applied on the thigh.

3.2.4.5 Foot-ground interaction

Since both the forefoot and toe segments were in contact with the force platform during experimental data collection, the measured ground reaction force captured the resultant of the forces acting on both segments. To distribute the ground reaction force between the forefoot and toe segments, the force applied to the forefoot was assumed to act at a point representing the midpoint between the 1st and 5th metatarsal heads projected onto the surface of the force platform. Similarly, the force acting on the toe segment was assumed to be located at the distal phalanx of the third digit projected onto the surface of the force platform. The vertical force acting on the toe segment was determined by summing the moments about a medio-lateral axis passing through the midpoint between the 1st and 5th metatarsal heads, using the measured centre of pressure and vertical ground reaction force as input. The vertical force acting on the forefoot was subsequently calculated by subtracting the vertical force acting on the toe segment from the total vertical ground reaction force. The anterior-posterior ground reaction force was distributed between the forefoot and toe segments in proportion to the vertical ground reaction force acting on each segment. That is, the coefficient of friction between the foot and ground was assumed to be constant and equal at the two contact locations.

3.2.4.6 Model scaling

The generic musculoskeletal model was individually scaled to represent each participant using the OpenSim scale tool (Delp et al., 2007). Specifically, the leg, rearfoot, and forefoot segments were scaled based on marker data from the static trial described above, whereas the remaining segments were scaled manually based on anthropometric data reported by Dempster (1955).

Maximal isometric muscle forces were scaled based on volume fractions of the leg and foot muscles (Handsfield et al., 2014; Silver et al., 1985). Specifically, total lower extremity muscle volume without the intrinsic foot muscles was estimated based on the regression equation provided by Handsfield et al. (2014). Subsequently, this volume was adjusted based on the fraction of lower extremity muscle volume that is attributed to leg musculature and the volume of the intrinsic foot muscles relative to total leg muscle mass (Handsfield et al., 2014; Silver et al., 1985). The total leg and foot muscle volume was distributed between the triceps surae, extrinsic, and intrinsic foot muscles according to the mass fractions reported by Silver et al. (1985), divided by a muscle density of 1.056 g/cm^3 to yield muscle volumes (Ward et al., 2009). The muscle volumes were divided by the optimal fibre length and multiplied by specific tension to yield maximal isometric muscle force. Reported values for specific tension varies between 15 N/cm^2 and 100 N/cm^2 (Buchanan, 1995; Maganaris et al., 2001). We performed static optimization using two different specific tension values from this range, 30 N/cm^2 and 60 N/cm^2 (Cristea et al., 2008; O'Brien et al., 2010).

3.2.4.7 Simulations

The OpenSim inverse kinematics tool was used to estimate the model pose by minimizing the squared distance between the experimental markers and markers fixed in the local coordinate systems of their corresponding model segments (Delp et al., 2007). The net forces and moments required at each of the model's generalized coordinates were calculated using the OpenSim inverse dynamics tool.

Lastly, the OpenSim static optimization tool was used to estimate the contribution of each individual muscle crossing a generalized coordinate to the net force and moment acting at/about it. This was achieved by minimizing the sum of required muscle activations squared (Crowninshield & Brand, 1981). To ensure that the model was able to generate the required force and moments, reserve actuators were added at each joint. The optimal force of the reserve actuators was set to 1 N and 1 Nm for linear and rotational actuators, respectively, thus requiring a large activation to generate an appreciable force or moment. As the static optimization routine minimizes the squared activation of the actuators, it would preferentially activate muscles, whose activation ranged from 0 to 1, before reserve actuators.

3.2.5 Data reduction and Statistical analysis

The contribution of each of the extrinsic foot muscles and intrinsic foot muscles to the transverse tarsal NJM during motion analysis trials was computed as the product of the muscle forces estimated using static optimization and their respective midfoot moment arms for each instant in time. The total moment contributed by each group, as well as any residual moment applied by the reserve actuators was used for analysis. In addition, midfoot joint angles and NJM were extracted for analysis. Since an approximately static pose was maintained throughout each trial, data was averaged over a 50-sample (0.5 sec) window centred on the halfway point of each trial to provide a single data point for each variable of interest per participant per condition.

Statistical analyses were performed for data that served as input to the model. However, as the model outputs must be considered theoretical estimates of muscle forces, they were not compared statistically. Midfoot NJMs and angles were compared across conditions and external loads with 2-by-4 repeated measures ANOVAs. All data were found to be normally distributed based on inspection of Q-Q plots. The sphericity assumption was assessed using Mauchly's test and Greenhouse-Geisser corrections were used in instances where sphericity was violated. Bonferroni corrected pairwise comparisons were computed when the omnibus was significant. Mean ± standard deviations are reported. Alpha level was set *a priori* to 0.05. All statistical analyses were conducted in R (version 4.1.1; R Foundation for Statistical Computing, Vienna, Austria) using the "car", "pastecs", and "ez" packages.

3.3 Results

3.3.1 Midfoot net joint moment and angle

A significant main effect for external load was observed for midfoot NJMs ($p < 0.001$; Table 3.1). Post hoc comparisons revealed that midfoot NJM increased significantly from 0% to 50% (-0.55 ± 0.07 $Nm \cdot kg^{-1}$ vs. -0.14 ± 0.06 $Nm \cdot kg^{-1}$, $p < 0.001$), 50% to 100% (-0.92 ± 0.09 $Nm \cdot kg^{-1}$ vs. -0.55 ± 0.07 $Nm \cdot kg^{-1}$, $p < 0.001$), and 100% to 150% (-1.27 ± 0.12 $Nm \cdot kg^{-1}$ vs. -0.92 ± 0.09 $Nm \cdot kg^{-1}$, $p < 0.001$) of body mass. Further, a significant main effect for condition revealed that midfoot moments were greater with toe flexor MVC compared to without (-0.76 ± 0.43 $Nm \cdot kg^{-1}$ vs. -0.68 ± 0.43 $Nm \cdot kg^{-1}$, $p < 0.001$).

Table 3.1. Ankle, midfoot, and metatarsophalangeal joint net joint moments for each load used in the relaxed and toe flexor maximal voluntary contraction conditions. Ankle and metatarsophalangeal joint data area presented for descriptive purposes, only.

Variable/load		**Relaxed**	**Toe flexor MVC**
Ankle NJM:			
0% Body mass	($Nm \cdot kg^{-1}$)	−0.12 ± 0.04	−0.22 ± 0.06
50% Body mass	($Nm \cdot kg^{-1}$)	−0.63 ± 0.08	−0.69 ± 0.08
100% Body mass	($Nm \cdot kg^{-1}$)	−1.09 ± 0.12	−1.13 ± 0.12
150% Body mass	($Nm \cdot kg^{-1}$)	−1.50 ± 0.16	−1.58 ± 0.14
Midfoot NJM:[b]			
0% Body mass[a]	($Nm \cdot kg^{-1}$)	−0.10 ± 0.03	−0.18 ± 0.05
50% Body mass[a]	($Nm \cdot kg^{-1}$)	−0.52 ± 0.06	−0.57 ± 0.07
100% Body mass[a]	($Nm \cdot kg^{-1}$)	−0.90 ± 0.09	−0.94 ± 0.10
150% Body mass[a]	($Nm \cdot kg^{-1}$)	−1.22 ± 0.12	−1.32 ± 0.11
Metatarsophalangeal NJM:			
0% Body mass	($Nm \cdot kg^{-1}$)	−0.02 ± 0.01	−0.05 ± 0.03
50% Body mass	($Nm \cdot kg^{-1}$)	−0.09 ± 0.05	−0.13 ± 0.06
100% Body mass	($Nm \cdot kg^{-1}$)	−0.14 ± 0.07	−0.19 ± 0.08
150% Body mass	($Nm \cdot kg^{-1}$)	−0.15 ± 0.10	−0.26 ± 0.10
Ankle angle:			
0% Body mass	(°)	−1.7 ± 5.1	−3.8 ± 3.3
50% Body mass	(°)	−2.7 ± 5.8	−4.8 ± 3.4
100% Body mass	(°)	−2.9 ± 6.2	−5.1 ± 4.3
150% Body mass	(°)	−1.6 ± 5.3	−3.9 ± 3.3
Midfoot angle:			
0% Body mass	(°)	1.3 ± 5.7	1.8 ± 4.7
50% Body mass	(°)	4.3 ± 4.6	2.1 ± 4.6
100% Body mass[c]	(°)	4.6 ± 4.4	3.4 ± 4.6
150% Body mass	(°)	5.5 ± 3.4	3.6 ± 3.7
Metatarsophalangeal angle:			
0% Body mass	(°)	1.4 ± 4.2	1.9 ± 4.8
50% Body mass	(°)	−1.1 ± 3.7	1.7 ± 3.7
100% Body mass	(°)	−2.1 ± 4.0	0.1 ± 4.9
150% Body mass	(°)	−3.8 ± 3.9	−1.2 ± 3.9

[a]Significant with respect to the next lighter load.
[b]Significant main effect for condition.
[c]Significant with respect to 0% body mass load.
NJM = net joint moment, MVC = maximal voluntary contraction

A significant main effect for external load was also observed for midfoot angles ($p = 0.02$, Table 3.1). Post hoc comparisons revealed that foot arch deformation was greater at external loads of 100% of body mass compared to 0% of body mass (4.0° ± 4.4° vs. 1.6° ± 5.1°, $p = 0.02$). In addition to midfoot NJM and angles, Table 3.1 provides ankle and metatarsophalangeal NJM and angles for descriptive purposes.

3.3.2 Specific tension

The model was unable to solve the static optimization problem without considerable use of the reserve actuators (>10% of total NJM) for most conditions and external loads when a specific tension of 30 N/cm^2 was used (Figure 3.4). Specifically, the relative contributions from reserve actuators were greater than 10% of the midfoot NJM in 4, 6, 11, and 12 of the 12 participants with loads of 0%, 50%, 100%, and 150% of body mass and toe flexor muscles relaxed, respectively. The corresponding numbers for the toe flexor MVC condition were 1, 10, 10, and 12, respectively.

Once specific tension was increased to 60 N/cm^2, the model was able to solve the static optimization problem for all participants both with and without toe flexor MVC when external loads were 50% of body mass or smaller (Figure 3.5). However, reserve actuation greater than 10% of the total NJM was still necessary for 3 and 7 participants, respectively, in the 100% and 150% loading conditions when participants relaxed their toe flexors. With toe flexor MVC, the corresponding numbers increased to 6 and 10 participants for the 100% and 150% body mass loads, respectively.

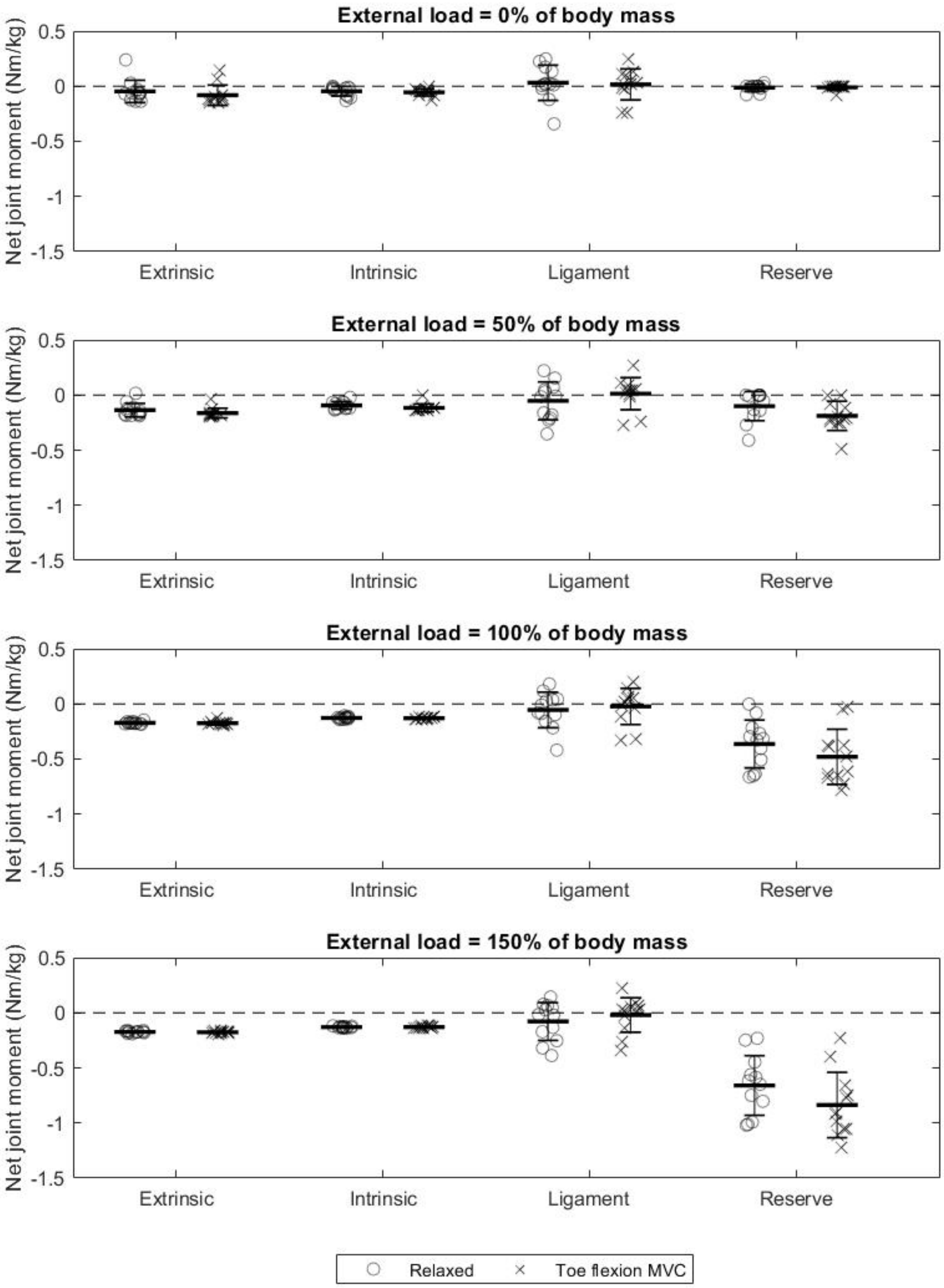

Figure 3.4. Midfoot net joint moment contributed by the extrinsic muscles, intrinsic muscles, ligaments, and reserve actuators for simulations with a specific tension of 30 N/cm^2.

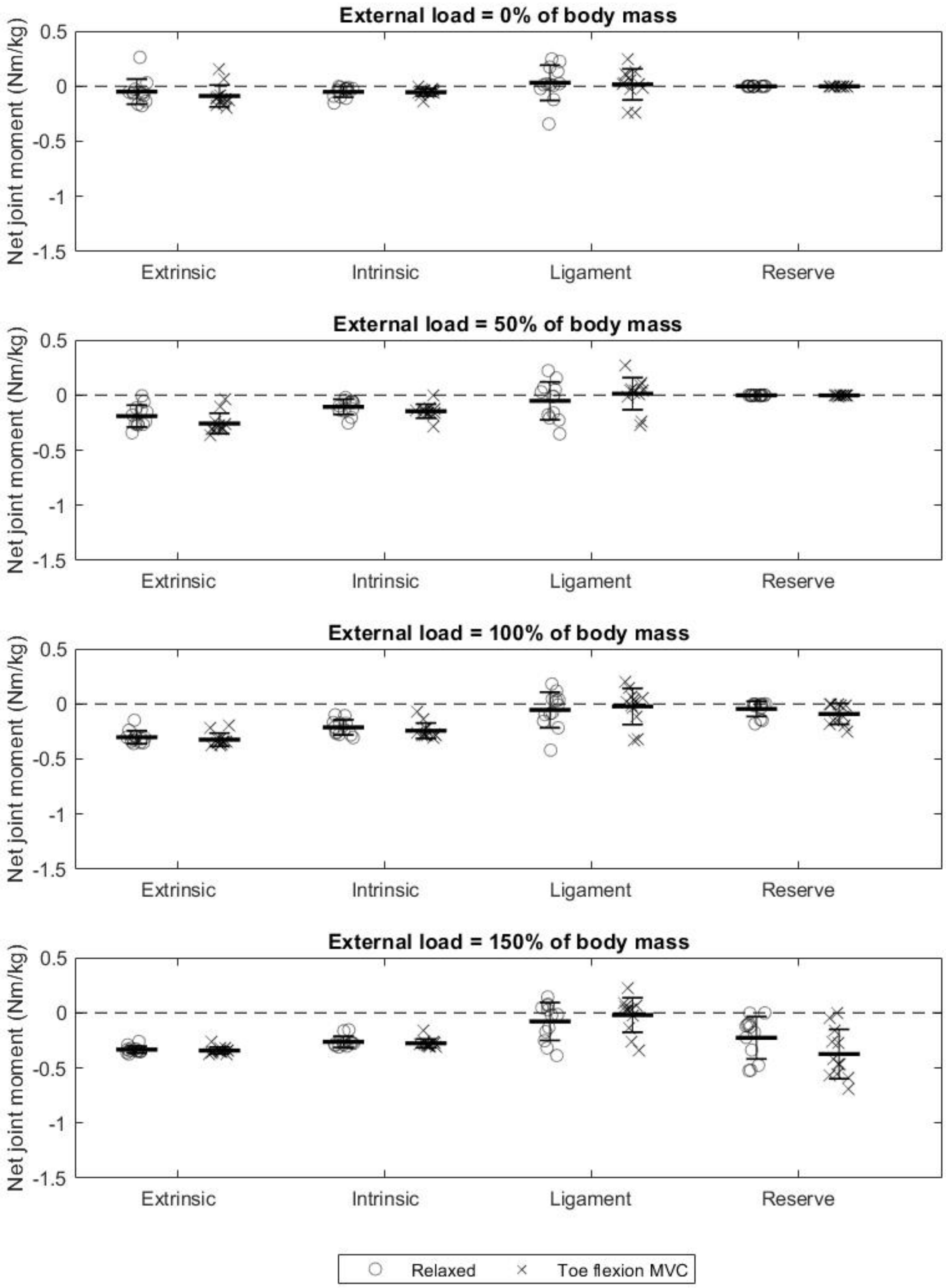

Figure 3.5. Midfoot net joint moment contributed by the extrinsic muscles, intrinsic muscles, ligaments, and reserve actuators for simulations with a specific tension of 60 N/cm^2.

3.3.3 Midfoot moment distribution

The distribution of the midfoot moment between the extrinsic muscles, intrinsic muscles, and foot ligaments followed a similar pattern for simulations conducted with both specific tensions, although maximal midfoot NJM contributed by the two muscle groups was reached with a smaller external load when a specific tension of 30 N/cm^2 was used. Numeric values are provided based on simulations performed with a specific tension of 60 N/cm^2.

The mean extrinsic foot muscle contribution to the midfoot NJM was larger than that of the intrinsic muscles and foot ligaments with all external loads (Figure 3.5). Specifically, the extrinsic muscles exerted -0.05 ± 0.11 $Nm \cdot kg^{-1}$ at 0% of body mass, -0.19 ± 0.10 $Nm \cdot kg^{-1}$ at 50% of body mass, -0.30 ± 0.06 $Nm \cdot kg^{-1}$ at 100% body mass, and -0.33 ± 0.03 $Nm \cdot kg^{-1}$ at 150% of body mass in the relaxed condition. In the toe flexor MVC condition, the extrinsic muscles exerted -0.09 ± 0.10 $Nm \cdot kg^{-1}$ at 0% of body mass, -0.26 ± 0.09 $Nm \cdot kg^{-1}$ at 50% of body mass, -0.33 ± 0.06 $Nm \cdot kg^{-1}$ at 100% body mass, and -0.34 ± 0.03 $Nm \cdot kg^{-1}$ at 150% of body mass.

In comparison, the intrinsic muscles exerted -0.05 ± 0.05 $Nm \cdot kg^{-1}$ at 0% of body mass, -0.11 ± 0.07 $Nm \cdot kg^{-1}$ at 50% of body mass, -0.21 ± 0.07 $Nm \cdot kg^{-1}$ at 100% body mass, and -0.26 ± 0.05 $Nm \cdot kg^{-1}$ at 150% of body mass in the relaxed condition (Figure 3.5). The corresponding moments during the toe flexor MVC condition were -0.06 ± 0.03 $Nm \cdot kg^{-1}$ at 0% of body mass, -0.15 ± 0.06 $Nm \cdot kg^{-1}$ at 50% of body mass, -0.24 ± 0.07 $Nm \cdot kg^{-1}$ at 100% body mass, and -0.28 ± 0.04 $Nm \cdot kg^{-1}$ at 150% of body mass.

The mean ligament moment exerted at the midfoot were smaller than those exerted by the extrinsic and intrinsic foot muscles (Figure 3.5). However, they also displayed considerably greater variability. Specifically, ligaments exerted 0.03 ± 0.16 Nm·kg^{-1} at 0% of body mass, -0.05 ± 0.17 Nm·kg^{-1} at 50% of body mass, -0.06 ± 0.16 Nm/kg at 100% body mass, and -0.08 ± 0.17 Nm·kg^{-1} at 150% of body mass in the relaxed condition. In the toe flexor MVC condition, ligaments exerted 0.02 ± 0.14 Nm·kg^{-1} at 0% of body mass, 0.01 ± 0.15 Nm·kg^{-1} at 50% of body mass, -0.02 ± 0.16 Nm·kg^{-1} at 100% body mass, and -0.02 ± 0.16 Nm·kg^{-1} at 150% of body mass.

3.4 Discussion

In the present study, an existing open-source musculoskeletal foot model was modified for the purpose of investigating the relative contributions of the extrinsic muscles, intrinsic muscles, and foot ligaments to foot arch support. Despite using available data from MRIs of healthy young individuals to estimate muscle strength, skeletal geometry based on CT scans, and muscle parameters from cadaveric investigations, the model was unable to solve the static optimization problem for the simulated task without 1) relying considerably on reserve torque actuators (>10% of total midfoot NJM), or 2) increasing the specific tension used for calculation of maximal muscle force to unrealistically high values (60 Nm/cm^2). Nonetheless, the model demonstrated a consistent pattern with regards to the relative contribution of the extrinsic muscles, intrinsic muscles, and foot ligaments to foot arch support. This pattern suggests that the extrinsic muscles may have the greatest potential to support the foot arches, followed by the intrinsic muscles, and then ligaments. Since the force produced by both muscles and ligaments depend on their length, this pattern may differ for tasks where the toes are notably dorsiflexed.

Regardless of the specific tension used, the model predicted a consistent pattern of extrinsic muscles, intrinsic muscles, and foot ligament contribution to the total midfoot NJM. Specifically, the extrinsic muscles made the largest relative contribution, followed by the intrinsic muscles, and then ligaments. That the extrinsic foot muscles make a greater contribution to foot arch support is not surprising given their relative size compared to the intrinsic muscles (Chang et al., 2012; Silver et al., 1985). For example, the total volume of the intrinsic foot muscles is approximately the same as that of tibialis posterior, or the combined volume of flexor digitorum longus and flexor hallucis longus (Chang et al., 2012). Similarly, the mass fractions reported by Silver et al. (1985) suggest that the intrinsic foot muscles have a total mass of approximately one quarter of the extrinsic foot muscles. In the present study, the midfoot moments exerted by the intrinsic muscles was approximately 70% of that exerted by the extrinsic muscles. However, the intrinsic muscle moment arm at the midfoot is greater than that of the extrinsic muscles (Angin et al., 2014; Farris et al., 2019). Therefore, it appears that the relative contributions of these muscle groups to the midfoot moments are realistic, despite the uncertainty regarding their absolute magnitudes. The smallest contribution to foot arch support was from the foot ligaments, including the plantar aponeurosis. The conditions simulated in this study involved minimal metatarsophalangeal extension and deformations of the foot arch quantified by the midfoot joint angle were less than 5 degrees. The strain of the plantar aponeurosis and foot ligaments was therefore small. Since the force exerted by ligaments depend on their length (Wager & Challis, 2016), it is not surprising that the foot ligaments did not make a large contribution to the midfoot moment. Although the magnitudes of the moments exerted by the extrinsic muscles and intrinsic muscles were dependent on the specific tension chosen, the present model gives a consistent estimate of how much each of these structures may contribute to foot arch support relative to each other.

The extent to which the model had to rely on its reserve actors to solve the static optimization problem differed between the two specific tensions used. Specifically, the reserve actuators contributed more than 10% of the midfoot NJM even for the lighter loads when a specific tension of 30 N/cm^2 was used. However, the maximal ankle plantar flexor and metatarsophalangeal moments the model could generate when this specific tension was used were 3.41 ± 0.29 and 0.16 ± 0.008 Nm/kg, respectively. These values are in reasonable agreement with maximal ankle plantar flexor and metatarsophalangeal moments reported during maximal isometric contractions (Ekblom, 2010; Goldmann & Bruggemann, 2012; Shima et al., 2002). Further, the estimated muscle forces of the flexor hallucis longus, flexor hallucis brevis, fibularis longus, flexor digitorum longus, and flexor digitorum brevis were $54.9 \pm 4.5\%$, $9.7 \pm 11.0\%$, $76.0 \pm 5.5\%$, $21.4 \pm 1.3\%$, and $20.5 \pm 2.4\%$ of body weight during the relaxed condition with an external load 100% body mass if a specific tension of 30 N/cm^2 was used. The external load during this condition is similar to that used by Jacob (2001) to estimate the forces in these muscles during gait. Jacob (2001) estimated forces of 52.4%, 35.5%, 57.8%, 8.8%, and 12.5% of body weight for the flexor hallucis longus, flexor hallucis brevis, fibularis longus, flexor digitorum longus, and flexor digitorum brevis, respectively. Generally, the forces estimated by the model in the present study agree well with those provided by Jacob (2001) if a specific tension of 30 N/cm^2 is used.

In contrast, using a greater specific tension allowed the model to find viable solutions to the static optimization problem without relying heavily on reserve actuators in most conditions. However, in his case, the maximal ankle plantar flexor and toe flexor strength well exceeded values observed during *in vivo* experiments (Ekblom, 2010; Goldmann & Bruggemann, 2012; Shima et al., 2002). This challenge is not unfamiliar in the literature describing musculoskeletal simulation models. For example, Rajagopal et al. (2016) found that using a specific tension lower than 60 N/cm^2 made it impossible to generate forwards simulations of walking and running. However, using a specific tension of 60 N/cm^2 resulted in maximal ankle, knee, and hip moments that were considerably larger than those reported in living participants (Rajagopal et al., 2016). The difficulty in obtaining viable model solutions using lower specific tensions are also consistent with the findings of Bruno et al. (2015) who reported that spine biomechanics of even low intensity tasks were impossible to simulate using specific tension values lower than 40 N/cm^2. For some of the higher intensity tasks they simulated, it was necessary to use specific tensions of up to 140 N/cm^2 (Bruno et al., 2015). Taken together, it appears that using the lower specific of 30 N/cm^2 results in the most realistic muscle forces. However, as muscular moments could only account for a small proportion of the total midfoot NJM when this specific tension was used, it is likely this model fails to account for some structure(s), that in the case of the foot, contributes a substantial proportion the observed NJM.

One limitation of the model used in this study is that it does not capture the contribution of joint reaction forces to the NJMs. Specifically, joints in the current model are idealized as pin joints, where pure rotations occur about the joint axes. Although this idealization may be appropriate for some joints, such as the knee and elbow, it may not accurately represent the function of articulations found in the foot. For example, a cadaveric investigation reported that the foot arch retained 63% of its stiffness even after severing the plantar aponeurosis, spring ligament, and long and short plantar ligaments (Huang et al., 1993). Bone-on-bone reaction forces or the elastic behavior of the joint capsular tissues could potentially explain this phenomenon. Since neither mechanism is captured in the present model, it is possible that these mechanisms explain the large reliance on the reserve actuator to generate adequate midfoot NJM. Another mechanism that has been proposed to contribute to arch support is mid-tarsal locking, which occurs when the rotation axes of the calcaneocuboid and talonavicular joints become increasingly misaligned with calcaneal eversion (Blackwood et al., 2005). The model represents the talonavicular and calcaneocuboid joints as a single pin joint and can therefore not account for the midtarsal locking mechanism. However, there is controversy regarding whether such a mechanism truly opposes midfoot joint motion (Okita et al., 2014). Therefore, the author is hesitant to suggest that this mechanism can explain the need for reserve actuators or large specific tension values to account for the midfoot NJM in the present study. Lastly, the mechanical properties of the intertarsal ligaments in the model were retained from the model of Malaquias et al. (2017). However, they suggest that these ligaments "require refinement", indicating that these ligament models may not accurately represent the behavior of the intertarsal ligaments (Malaquias et al., 2017). Therefore, ligament models representing the plantar aponeurosis were replaced with a model based on data derived from cadaveric experiments (Erdemir et al., 2004; Wager & Challis, 2016). However, suitable data from cadaveric investigations of other relevant foot ligaments could not be found. Consequently, the remaining ligament models already implemented in the Malaquias et al. (2017) model were not modified. Nonetheless, it appears unlikely that the foot ligaments, without the plantar aponeurosis, can explain the reserve moments observed when the lower specific tension values were used, given that their contribution to foot arch support is small compared to that of the plantar aponeurosis (Huang et al., 1993).

Previous research has theorized that the extrinsic muscles, intrinsic muscles, or foot ligaments may modulate foot arch rigidity to meet the paradoxical requirement of being both rigid and compliant. However, these theories all assume that an increase in foot arch rigidity is achieved by increasing muscle or ligament tension, ultimately exerting a moment about the intertarsal joints of the foot (Hicks, 1954; Kelly et al., 2014; Thordarson et al., 1995). An alternative explanation for how the foot muscles and ligaments may modulate arch rigidity is by acting as "slings" or "cushions" that prevent the tarsal bones from translating inferiorly. Specifically, as the muscle bellies, muscle tendons, and ligaments wrap underneath the tarsal bones, they can be hypothesized to exert a pressure on the inferior surfaces of these bones. The resultant force of this pressure would exert an approximately superiorly directed force on the foot bones and may therefore contribute to elevate the foot arch. Moreover, if the tension of the any of these structures are increased through active muscle contraction the pressure exerted on the inferior surface of the tarsal bones would increase (Ryan et al., 2020). This mechanism could therefore be hypothesized to contribute to the foots' ability to modulate arch rigidity. Structures acting as slings to support the foot arch is not an entirely new concept. Specifically, the inferior-lateral and inferior-medial calcaneonavicular ligaments, have long been purported to act as a "sling" supporting the head of the talus (Davis et al., 1996). The plantar intrinsic muscles may be particularly well suited for foot arch support through this mechanism for two reasons. First, recent research has demonstrated that in addition to exerting a longitudinal force, muscles also exert an outwards directed pressure perpendicular to their long axis due to muscle bulging (Ryan et al., 2020). Second, if muscles truly are isovolumetric, they cannot compress, but only deform as the arch deforms. Since several of the plantar intrinsic muscles are housed within a dense connective tissue framework they would have limited room to deform, thus preventing the bones of the midfoot translating inferiorly. Future research should determine whether this proposed mechanism may contribute to foot arch support.

In conclusion, the extrinsic muscles seem to have the greatest ability to modulate foot arch rigidity through their longitudinal actions, followed by the intrinsic muscles and foot ligaments. However, it seems unlikely that the extrinsic muscles, intrinsic muscles, and foot ligaments can generate a sufficient moment at the transverse tarsal joint to prevent foot arch collapse. It is theorized that an additional mechanism is required to support the arch. The pressure exerted on the inferior surface of the tarsal bones by the extrinsic foot muscle tendons, intrinsic foot muscle bellies and tendons, and plantar ligaments is hypothesized be a contributor to foot arch support.

Chapter 4

Effect of incline versus block heel-raise exercise on vertical jump performance – A randomized longitudinal study

4.1 Introduction

The ankle plantar flexors are one muscle group that strongly influences vertical jump performance. This is exemplified by the considerable decrease in vertical jump height observed when the ankle plantar flexors' ability to perform work is reduced experimentally or by fatigue during jumps (Arakawa et al., 2013; Bobbert et al., 2011; Dæhlin & Chiu, 2019). Conversely, strengthening the ankle plantar flexors through resistance training improves vertical jump performance (Chiu et al., 2017). However, to take complete advantage of an increase in ankle plantar flexor strength, efficient transmission of the energy generated by these muscles to the ground is necessary.

Ankle plantar flexor work performed during the propulsion phase of jumping is transmitted to the ground through the non-rigid lever provided by the foot. However, by acting on the calcaneus, the strongest ankle plantar flexors, the triceps surae, can cause considerable deformation of the foot arches (Cheung et al., 2006; Thordarson et al., 1995). This is notable as excessive deformation of the foot could reduce the efficiency of this energy transmission (Takahashi et al., 2016). In contrast, several extrinsic and intrinsic toe flexor muscles have been demonstrated to limit foot arch deformation (Kelly et al., 2014; Thordarson et al., 1995). It may therefore be hypothesized that concurrent strengthening of the ankle plantar flexors and toe flexors would allow for more ankle plantar flexor work to be performed without compromising the ability to transmit this energy efficiently through the foot to the ground.

Previous research investigating the effects of toe flexor strengthening on vertical jump performance have reported conflicting results (Hashimoto & Sakuraba, 2014; Kokkonen et al., 1988; Unger & Wooden, 2000). For example, Goldmann et al. (2013) found no improvement in vertical jump height following seven weeks of toe flexor training. In contrast, Hashimoto and Sakuraba (2014) found vertical jump height to increase following eight weeks of toe flexor training. Common to these and other studies investigating toe flexor strengthening protocols, is that they performed exercises emphasizing toe flexion (Hashimoto & Sakuraba, 2014; Kokkonen et al., 1988; Unger & Wooden, 2000). Further, the toe flexor exercises were not performed in combination with ankle plantar flexor strengthening, and the contraction mode was either isometric or concentric (Hashimoto & Sakuraba, 2014; Kokkonen et al., 1988; Unger & Wooden, 2000). However, the toe flexor muscles are thought to act eccentrically, resisting the foot deformation caused by the force exerted on the calcaneus by the triceps surae during tasks that involve elevating the body centre of mass (Chiu & Dæhlin, 2021). It may be hypothesized that an exercise where the ankle plantar flexor and toe flexor muscles are trained simultaneously would be more suitable for improving vertical jump height. This may be achieved by having the toe flexor muscles resist arch deformation caused by a large triceps surae force.

One exercise that creates this condition is a heel-raise performed on an incline plane (incline heel-raise; Chiu & Dæhlin, 2021). Specifically, a large triceps surae force acting on the calcaneus to raise the system mass against gravity would cause foot arch deformation, while the toe flexor muscles simultaneously may act eccentrically to prevent arch deformation (Chiu & Dæhlin, 2021; Riddick et al., 2019). A recent study provides support for this mechanism demonstrating that midfoot net joint moments (NJM) are greater during incline heel-raises compared to heel-raises performed on the edge of a flat block (block heel-raises; Chiu & Dæhlin, 2021).

The purpose of this study was to assess the effect of incline versus block heel-raise exercise on vertical jump height, ankle plantar flexor strength, and toe flexor strength. We hypothesize that incline heel-raise exercise would result in greater increases in vertical jump height and toe flexor strength compared to block heel-raise exercise. In addition, changes in NJM work performed at the ankles, knees, and hips were investigated as secondary outcomes to explain the mechanism responsible for any increase in vertical jump height.

4.2 Methods

4.2.1 Experimental design

A randomized parallel longitudinal design with three repeated measures was used to investigate our hypotheses. We used motion analysis techniques and dynamometry to assess vertical jump mechanics and strength of competitive volleyball players before (pre), after 7 weeks (mid), and after 11 weeks (post) of heel-raise exercise with 3 weekly training sessions. All participants completed the same vertical jump practice protocol and heel-raise training program, except that one training group performed incline heel-raises, while the other training group performed block heel-raises. Changes in vertical jump height, ankle plantar flexor strength, and toe flexor strength between pre-, mid-, and posttest were compared between groups as primary outcome variables. Additionally, changes in ankle, knee, and hip NJM work were assessed as secondary (explanatory) outcomes.

4.2.2 Participants

Thirty-three competitive female volleyball players between 15 and 22 years old were recruited for participation (Figure 4.1). Previous research has reported 0.7 and 0.5 standard deviation improvements in vertical jump height when ankle plantar flexor or toe flexor exercise is added to a general resistance training program in trained individuals, respectively (Chiu et al., 2017; Kokkonen et al., 1988). Using an effect size estimate of 0.5 standard deviations and a conservative repeated measures correlation of 0.5, an *a priori* power analysis conducted in G*Power (Faul et al., 2007) suggested that a total sample size of 28 participants would be sufficient to detect time-by-group interactions.

To be eligible for the study, participants had to be free from current or previous injuries preventing them from safely performing jumping and resistance training. Moreover, participants were required to play volleyball competitively to be eligible. Participants, and legal guardians of players under 18 yrs. old, were informed of the potential risks and benefits of the study and provided written informed consent/assent. Participants over 18 yrs. old provided consent on their own behalf, whereas legal guardians provided consent on behalf of participants under 18 yrs. old. The latter participants provided written assent. The study protocol was approved by the regional ethics board at the authors' institution (Study id: Pro00103415).

Participants were stratified by playing level (high-school, varsity rookies, and varsity non-rookies) and randomly allocated to an incline (n = 17) or block (n = 16) exercise group in a one-to-one ratio (Figure 4.1). A custom written MATLAB (MathWorks®, Natick, MA, USA) program was used to generate separate random allocation sequences for each stratification level and placed in opaque sequentially numbered envelopes before the start of recruitment. Each sequence used a block randomization scheme with block size of 6 participants. The investigator recruiting and allocating participants was blinded to the allocation sequence.

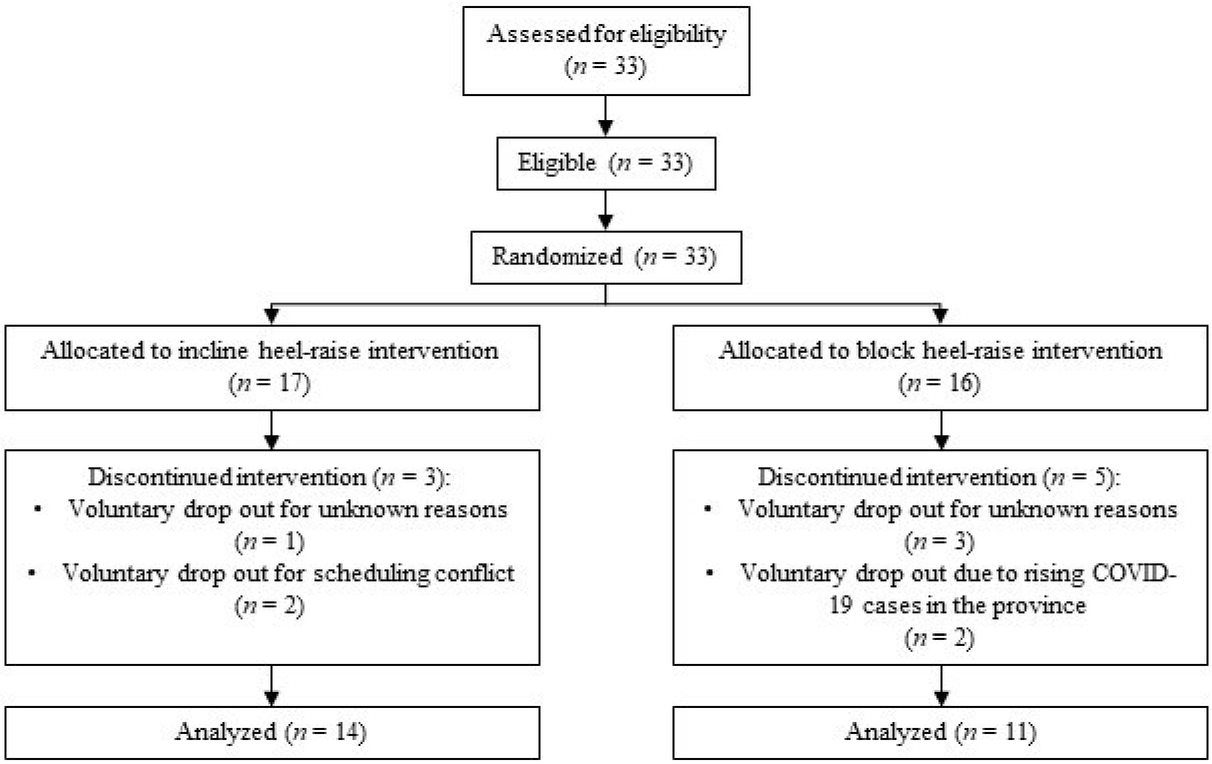

Figure 4.1. Flowchart of participant screening, enrollment, and retention.

4.2.3 Training intervention

Participants completed a heel-raise exercise and jumping practice program 3 times per week throughout the 11-week intervention, except for week 7 when mid-test was performed in place of the third training session. In addition, all participants completed their regular team practices and resistance training. Team resistance training was the same for all participants within each level used for stratification (appendix A).

The heel-raise training program consisted of three training blocks. The first block emphasized neural adaptation and familiarization with the exercises (weeks 1 and 2), the second emphasized muscle hypertrophy (weeks 3-7), and the third emphasized maximal strength (weeks 8-11; Table 4.1). Specifically, low, moderate, and high intensities and high, moderate, and moderate volumes were used for the first, second, and third block of the heel-raise training program, respectively. One notable aspect of the training program is that the participants completed a 10, 8, or 5 repetition maximum (RM) test at day three of each training week, during the first, second and third training blocks, respectively. The result from this test was used to set the training intensity for the next training week, thus ensuring that strength gains were accompanied by an increase in intensity for the heel-raise exercise. All training sessions were supervised by an investigator who provided guidance in proper exercise technique and monitored whether the prescribed exercise was performed. Neither the participants nor investigators were blinded to group allocation, and participants completed the training sessions in groups of 3-6 players. Participants also completed training diaries where they registered the resistance used for each set, the number of successful repetitions and sets, and rating of perceived exertion on Borg's CR-10 scale (Borg, 1982).

Table 4.1. The 11-week block periodized heel-raise and jumping practice program.

	Week 1-2			**Week 3-7**			**Week 8-11**		
	Day 1	Day 2	Day 3	Day 1	Day 2	Day 3	Day 1	Day 2	Day 3
Heel-raise: (set × repetitions)									
50% nRM	3 × 10	3 × 10	1 × 10	2 × 8	2 × 8	1 × 8	2 × 5	1 × 5	1 × 5
70% nRM	3 × 10	3 × 10	1 × 10	4 × 8	2 × 8	1 × 8	2 × 5	1 × 5	1 × 5
80% nRM			1 × 10			1 × 8	3 × 5	1 × 5	1 × 5
90% nRM			1 × 10		2 × 8	1 × 8		1 × 5	1 × 5
nRM			1 × 10			1 × 8		3 × 5	1 × 5
Jumping: (set × repetitions)									
Countermovement jump	3 × 4		3 × 4	3 × 6	3 × 6	3 × 6	4 × 5	4 × 5	4 × 5
Box jump	3 × 4		3 × 4	3 × 6	3 × 6	3 × 6	4 × 5	4 × 5	4 × 5
Drop landing	3 × 4		3 × 4	3 × 6	3 × 6	3 × 6	4 × 5	4 × 5	4 × 5

Participants performed the heel-raises standing on one leg with the hip and knee extended. A barbell attached to a linear slide-rail allowing vertical-only motion was used to apply external resistance. The prescribed number of repetition and sets were completed on each lower extremity. The incline group performed heel-raises on a plane sloped 22° from posterior to anterior, while the block group performed heel-raises with the forefoot on a platform elevated 10 cm from the surrounding floor and the rearfoot freely suspended over its edge (Figure 4.2 and 4.3). During both exercises, participants lowered their center of mass by dorsiflexing the ankle until reaching maximal dorsiflexion. Subsequently, participants elevated their centre of mass by plantar flexing the ankle until maximal plantar flexion was reached. Vertical countermovement jumps, box jumps, and drop landings were used for the jumping practice protocol and performed as described by Powers (1996) and (Chiu & Moolyk, 2015).

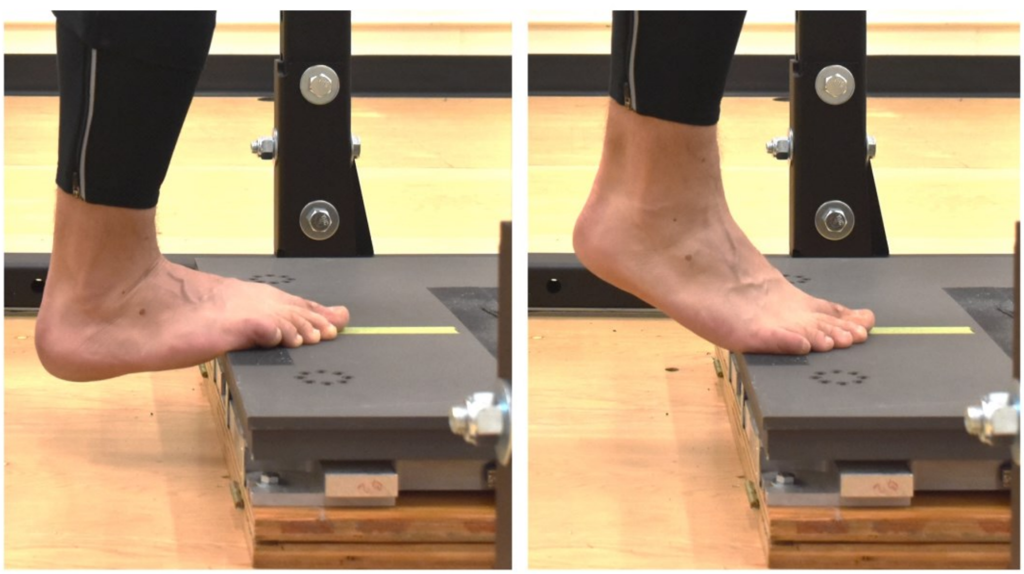

Figure 4.2. The bottom position (left panel) and top position (right panel) of the foot and ankle during the block heel-raise exercise.

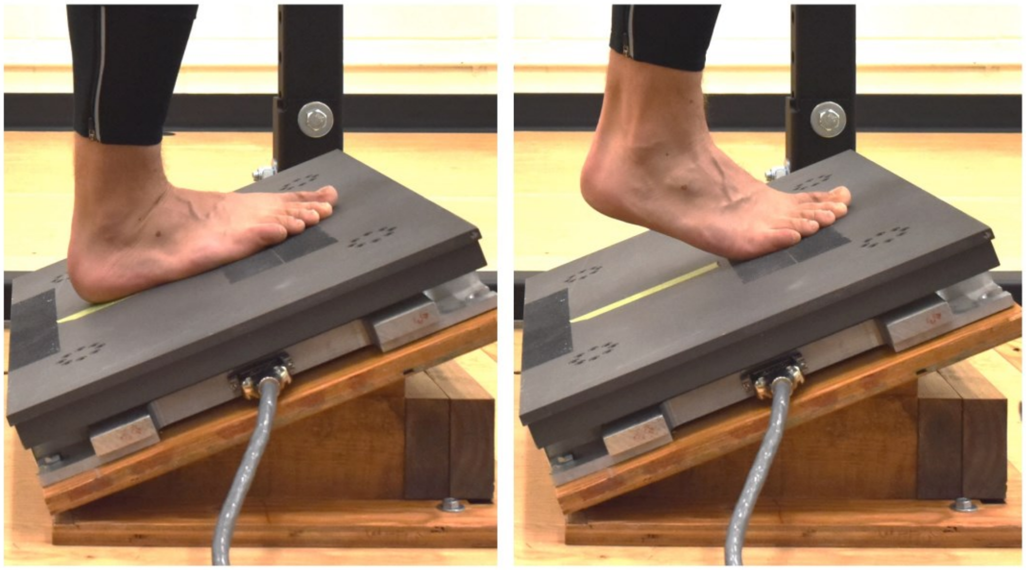

Figure 4.3. The bottom position (left panel) and top position (right panel) of the foot and ankle during the incline heel-raise exercise.

4.2.4 Test procedures

Assessments of outcome variables were completed in the sports biomechanics laboratory before, after 7 weeks, and after 11 weeks of heel-raise exercise. To examine whether heel-raise variation would affect jump performance, vertical countermovement and 3-step approach jump tests were performed using motion analysis techniques. Further, to examine whether heel-raise variation would affect toe flexor strength or ankle plantar flexor strength, maximum voluntary toe flexor and ankle plantar flexor strength tests were performed using dynamometry. Pre-test was conducted within seven days of the first training session, while posttest was completed two to nine days following the last training session. The order of all tests was the same at pre-, mid-, and posttest.

4.2.4.1 Jumping mechanics

Participants were instrumented with 55 retroreflective markers for the assessments of vertical countermovement and approach jump mechanics. The markers were affixed to the participants' pelvis, and both thighs, legs, and feet with adhesive tape and neoprene straps. Specifically, anatomical markers were placed bilaterally on the anterior superior iliac spines, posterior superior iliac spines, iliac crests, greater trochanter, medial and lateral femoral epicondyles, medial and lateral malleoli, posterior calcaneal tuberosity, navicular tuberosity, cuboid, and first and fifth metatarsal heads (Cappozzo et al., 1995). Additionally, clusters of markers placed in a non-colinear configuration on rigid thermoplastic plates were used to track the pelvis (3 markers), thigh (5 markers), and leg (5 markers) segments during dynamic trials. Markers were also placed on the medial and lateral aspects of the calcaneus (Chizewski & Chiu, 2012; Stamm & Chiu, 2016). Trajectories of the retro-reflective markers were recorded using 8 optoelectronic cameras (Miqus 3, Qualisys, Gothenburg, Sweden) sampling at 200 Hz. Simultaneously, ground reaction forces were measured using two force platforms (OR6-6, AMTI, Watertown, MA, USA) sampling at 1000 Hz.

Participants performed 4 maximal vertical countermovement jumps and 4 maximal 3-step approach jumps with each limb as lead limb. For the vertical countermovement jumps, participants started standing with one foot on each of two force platforms imbedded in the laboratory floor. Subsequently, they performed a countermovement to a self-selected depth and jumped vertically with maximal effort. The 3-step approach jumps were performed as described by Chiu et al. (2017). Briefly, participants started behind a 10 cm high platform firmly mounted to the floor 30 cm behind the force platforms. The force platforms were mounted side-by-side and flush with the floor. Participants initiated the approach by stepping forwards onto the raised platform with their trailing limb. Second, they stepped onto the first force platform with their leading limb, before stepping down onto the second force platform with their trailing limb. Subsequently, they performed a maximal jump. The entire 3-step approach jump was performed in one fluid motion, and the lead limb was altered between trials until 4 successful trials were obtained for each side. The limb participants led with during the first jump was determined according to a random sequence generated *a priori*, but maintained consistent within participants for pre-, mid-, and posttest.

For all jumps, participants reached for an overhead target suspended above the force platforms (Mok et al., 2017). Further, participants performed all jumps with a self-selected arm swing, and were required to land with each foot fully contacting the force platform they were on at take-off to be considered successful. Unsuccessful trials were repeated.

4.2.4.2 Maximal voluntary isometric contractions

Ankle plantar flexor strength was tested via maximal voluntary isometric contractions (MVIC) performed on a custom-built dynamometer (unpublished to date). The dynamometer consists of a rigid frame with a car seat at its rear end and an adjustable ankle brace mounted on a trolley that move along the frame and can be fixed at any position forward of the seat. The angle of the ankle brace relative to the bracket affixing it to the trolley is adjustable and has an inset footplate which can rotate away from the ankle brace about an axis running medio-laterally relative to the participant’s foot. The foot plate can only rotate in the direction that heel-lift may occur during plantar flexor contractions. Two single-axis load cells (MLP-300; Transducer Techniques, Temecula, CA, USA) are mounted on the undersurface of the foot plate, one aligned with its medio-lateral rotation axis and one at a fixed distance from the first in the direction of the participant’s heel. This allowed the forces acting under both the ball of the foot and heel to be measured during the maximal ankle plantar flexor efforts. Additionally, the medio-lateral axis of the foot plate and the ankle brace’s rotation axis with respect to its mounting bracket are instrumented with potentiometers allowing the ankle angle to be monitored and corrected for any heel-lift occurring during maximal plantar flexor efforts. The dynamometer produces reliable measurements of ankle plantar flexor torques at ankle angles between 30° of plantar flexion and 10° of dorsiflexion (ICC(2, k) = 0.63-0.87, SEM = 0.2-6.1 Nm; unpublished to date).

During tests of ankle plantar flexor MVIC, participants were seated in the car seat with a fixed back angle of approximately 80°. The measurement unit was fixed at a position such that the participant’s knee was extended, and the foot was secured to the foot plate with nylon straps. The metatarsal heads were aligned with the medio-lateral axis of the foot plate. Participants performed two 3-second maximal efforts for each of three ankle positions, −15° (plantar flexed), 0° (neutral), and maximally dorsiflexed, on each leg. The two maximal efforts were separated by approximately 15 seconds, and participants rested for 2 minutes between each angular configuration.

Toe flexor strength was assessed via MVIC performed using a force platform (Bertec 4060, Columbus, OH, USA) sampling at 1000 Hz. The force platform was elevated approximately 10 cm from the surrounding floor and mounted on a bracket that allowed it to be set to two different angular configurations: 0° or 22° inclination. During the tests, participants were seated on a bench with the knee and hip flexed to approximately 90°. Their forefoot and rearfoot rested on a wooden block that was flush with the force platform and separated from it by a narrow gap. The participants' metatarsal heads were aligned with the edge of this block and allowed them to reach across the narrow gap to the force platform with their toes. Toe flexor MVIC were performed by pressing their toes as hard as possible against the force platform. The hallux and the lateral toes were tested separately for each leg, and tests were performed with the force platform at 0° and 22° of inclination. Two maximal 3-second efforts were performed in each configuration, with approximately 15 seconds between contractions. As separate muscles control the hallux and the lateral digits, participants did not rest between different configurations (Fukunaga et al., 1992; Soysa et al., 2012). The reliability of the resultant force exerted on the force platform was found to be good during pilot testing (ICC(2, 1) = 0.81 to 0.90, SEM = 14.1 to 25.4 N).

Loud verbal encouragement was provided during all MVIC trials. Further, the order of both angular configurations and side was randomized between participants prior to pre-test. The same order was maintained within participants for the mid- and posttest.

4.2.5 Data processing and reduction

Data collected during countermovement and approach jumps were exported from Qualisys Track Manager (Qualisys, Gothenburg, Sweden) and processed using a custom written MATLAB (R2019b, MathWorks®, Natick, MA, USA) program. Marker trajectories and force data was filtered with a fourth-order zero phase-lag Butterworth filter with a 10 Hz cut-off frequency. The cut-off frequency was determined based on a residual analysis (Wells & Winter, 1980). Marker trajectories and forces were filtered at the same cut-off frequency to avoid introduction of artifacts in calculated NJMs (Bisseling & Hof, 2006; Kristianslund et al., 2012).

Marker trajectories from a static trial where participants were standing quietly for 10 seconds were used to create a 7-segment rigid body model. The model segments represented the pelvis, and left and right thighs, legs, and feet. The segments were modelled as having the shape of conical frusta with proximal and distal ends calculated as detailed in Table 4.2. The length of each frustum was defined as the distance between proximal and distal segment ends, and radii were calculated as half the Euclidian norm between the two markers on each segment end, except for the proximal thigh where the distance between the greater trochanter and hip joint centre was used as radius. The hip joint centre was calculated based on the pelvis markers as described by Harrington et al. (2007). Inertial properties of the segments were determined from their geometric shape, and their mass relative to total body mass were calculated based anthropometric data from Dempster (1955). Laboratory and segment coordinate systems conformed to the right-hand rule, with the X-, Y-, and Z-axis oriented to the right, anteriorly, and up, respectively. Segment coordinate systems were defined as described in Wu et al. (2002) with the necessary relabeling of axes to conform to the convention given above.

Table 4.2. Definitions of proximal and distal segments ends for inverse dynamics calculations.

Segment	**Proximal end**	**Distal end**
Pelvis	Midpoint of iliac crest markers	Midpoint of hip joint centres
Thigh	Hip joint centre	Midpoint of med. and lat. femoral epicondyles
Leg	Midpoint of med. and lat. femoral epicondyles	Midpoint of med. and lat. malleolus
Foot	Midpoint of med. and lat. malleolus	Midpoint of 1st and 5th metatarsal head

A least square pose estimation algorithm was used to track the motion of local coordinate systems during moving trials (Cappozzo et al., 1997). Segment angles were computed as the orientation of segment coordinate systems relative to the laboratory coordinate system using a ZYX Cardan sequence. Joint angles were computed as the orientation of the distal relative to the proximal segment coordinate system using an XYZ Cardan sequence. This is equivalent to the joint coordinate systems convention set forth by the International Society of Biomechanics (Grood & Suntay, 1983; Wu et al., 2002). Newton-Euler iterative inverse dynamics was used to calculate the NJM acting at the ankle, knee, and hip joints and expressed in the local coordinate system of the distal segment. Net joint power was calculated as the dot product between the local sagittal plane NJM and joint angular velocity and was subsequently integrated using the trapezoid rule to yield the work performed by the NJMs. Jump height was calculated using the pelvic kinematic method for both vertical countermovement and approach jumps (Chiu & Salem, 2010). This method was chosen as it allows vertical jump height to be determined 1) with the use of a lower extremity marker set, and 2) for jumping tasks where the participants do not begin the task on the force platforms. As the purpose of this study was to investigate changes in lower extremity strength and jumping mechanics, the additional time required to place reflective markers for a full body marker set was not considered to be justified.

For the vertical countermovement jumps, bilateral symmetry was assumed, and segment kinematics were averaged across limbs while joint kinetics summed across limbs (Dæhlin & Chiu, 2019). In contrast, outcome variables were analyzed separately for the left and right lower extremities for the approach jumps. The phases and events of interest during the countermovement and approach jumps were defined as described by Dæhlin and Chiu (2019), and Chiu et al. (2017), respectively. Briefly, initiation of the propulsion phase was defined as the instant the vertical velocity of the pelvis centre of mass became negative for the countermovement jump, and as the instant the vertical ground reaction force increased above 15 N for each limb separately for the approach jumps. The end of the propulsion phase was defined by the instant vertical ground reaction force decreased below 15 N as the participants' feet left the ground. For the vertical countermovement jump, the average of the two instants detected for each limb was used, while the events were detected separately for each lower extremity during approach jumps.

MVIC data was processed using custom written MATLAB programs. Raw data were filtered with a fourth-order zero phase-lag Butterworth filter with a 10 Hz cut-off frequency. Ankle plantar flexor moment during MVIC trials was calculated by summing moments about the ankle using the forces recorded with the load cells, the known distance between them, and the moment arm between the metatarsal heads and the ankle. The latter was determined by finding the horizontal distance between the midpoint of the malleolus markers and the midpoint of the 1st and 5th metatarsal head markers from the static motion analysis trial. Additionally, the ankle angle during the maximal efforts was calculated as the difference between the ankle potentiometer and foot plate potentiometer. For maximal voluntary toe flexor actions, the resultant force exerted during maximal efforts was calculated as the Euclidian norm of the force vector recorded by the force platform. For both tests, the maximal value of the average moment or resultant force over a 50 ms moving window was used for statistical analyses.

4.2.6 Statistical analyses

A 2-by-3 mixed ANOVA with group as between- and time as within-participant factors were used to compare changes in vertical countermovement jump height between the two groups over time. Similarly, a 2-by-2-by-3 mixed ANOVA with group as between-, and side and time as within-participant factors were used to compare changes in jump height in approach jumps between the two groups over time performed with the dominant and non-dominant limb as lead limb. The assumptions of normality and homogeneity of variance were met as assessed by Shapiro-Wilk's test and Levene's test, respectively. The assumption of sphericity was assessed using Mauchly's test, and Greenhouse-Geisser corrections were used for instances where the sphericity assumption was violated.

Group-by-side-by-time MANOVAs were used to analyze the ankle plantar flexor and toe flexor MVICs. For each test, the angles were used as multivariate levels. Similarly, a group-by-time MANOVA was used to analyze the work performed at the ankle, knee, and hip during vertical countermovement jumps, and group-by-side-by-time MANOVAs were used to analyze these effects for the approach jumps. For the jumps, the ankle, knee, and hip were used as multivariate levels. The multivariate omnibus was assessed using Pillai's trace (Pillai's V) and follow-up univariate tests were only performed in the event of a significant omnibus test. Multivariate normality was assessed using Shapiro-Wilk's multivariate normality test.

As it was not an aim of this study to assess bilateral difference, any significant main effects for side were ignored. All significant univariate and main effects were followed-up with multiple Bonferroni corrected t-test to compare the outcomes at mid- and posttest to those at pretest. Pairwise comparisons between mid- and posttests were not performed. Means ± standard deviations (SD) and 95% confidence intervals (CI) of simple effect sizes are reported (Baguley, 2009). An α-level of 0.05 was set *a priori*. All statistical analyses were conducted in R (version 4.1.1; R Foundation for Statistical Computing, Vienna, Austria) using the "car", "pastecs", and "ez" packages.

4.3 Results

4.3.1 Attrition and compliance

Three participants allocated to the incline group and five participants allocated to the block group withdrew from the study prior to completion (Figure 4.1). The reasons for withdrawal were increasing number of active COVID-19 cases in the province research was conducted (n = 2), scheduling conflicts (n = 3), or not provided (n = 3). Of the remaining 25 participants, 14 participants were in the incline group and 11 participants were in the block group (Table 4.3). Compliance with the training program was equal in both groups, as measured by number of sessions completed (Table 4.3).

Table 4.3. Participant characteristics

Characteristic		**Block (n = 11)**	**Incline (n = 14)**
Age	(yrs.)	17 ± 2	16 ± 1
Stature	(m)	1.80 ± 0.07	1.77 ± 0.08
Mass	(kg)	70.3 ± 7.2	67.1 ± 11.1
Sessions completed	(of 32)	31 ± 1	31 ± 1
Playing level	(HS:R:NR)	9:1:1	12:2:0

HS = High school, R = Varsity rookie, NR = Varsity non-rookie.

4.3.2 Jump height

The omnibus did not indicate any significant time-by-group interaction ($F(2,46) = 1.184$, $p = 0.315$) or group main effect ($F(1,23) = 0.347$, $p = 0.562$) on vertical countermovement jump height. However, it did indicate a significant main effect for time ($F(2,46) = 5.016$, $p = 0.011$). Specifically, post-hoc tests revealed that participants improved their vertical countermovement jump height from pre-test to posttest (change = 1.24 ± 2.25 cm; 95% CI = [0.31 2.17] cm; $p = 0.033$; Figure 4.4 and 4.5).

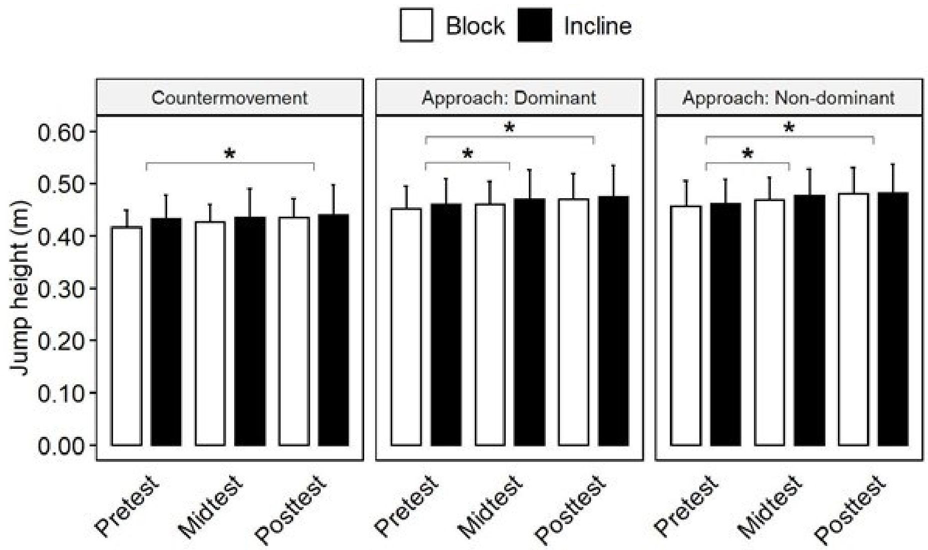

Figure 4.4. Jump heights in the block and incline groups at pre- mid- and posttest during vertical countermovement and 3-step approach jumps. Data is presented separately for approach jumps performed with the doninant and non-dominant limb as lead limb.
* Main effect for time: Significant with respect to pretest ($p < 0.05$).

Similarly, the omnibus test for approach jump height did not indicate a significant time-by-group interaction ($F(2,46) = 0.208$, $p = 0.764$) or group main effect ($F(1,23) = 0.099$, $p = 0.755$), but did indicate a significant main effect for time ($F(2,46) = 8.861$, $p = 0.001$; Figure 4.4 and 4.5). Post-hoc tests revealed that approach jump height increased from pre- to mid-test (change = 1.12 ± 2.58 cm, 95% CI = [0.46 1.93] cm, $p = 0.006$), mid- to posttest and pre- to posttest (change = 1.89 ± 2.64 cm, 95% CI = [1.14 2.64] cm, $p < 0.001$).

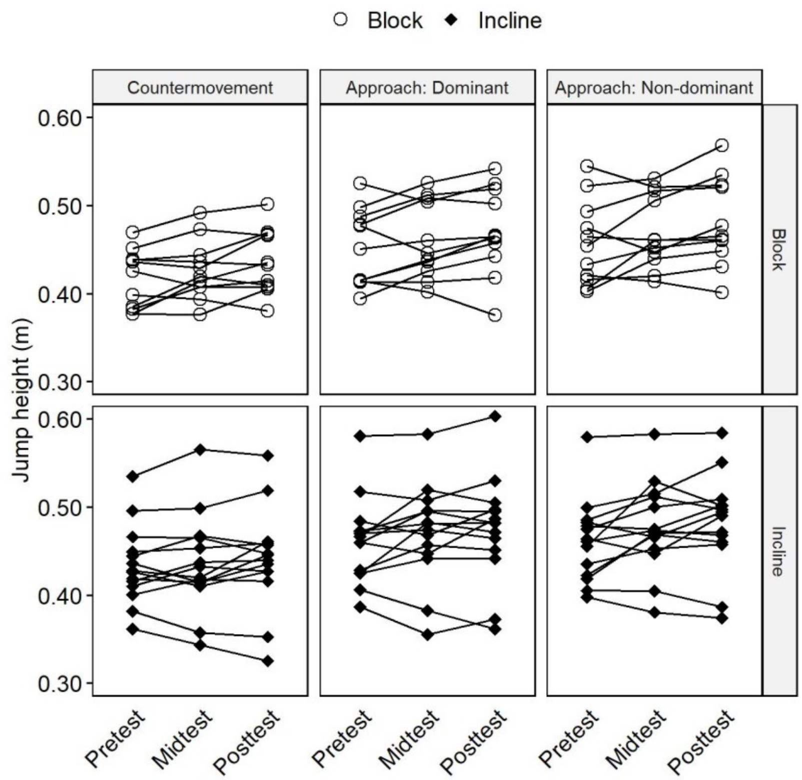

Figure 4.5. Individual jump heights of participants in the block and incline groups at pre- mid- and posttest during vertical countermovement and 3-step approach jumps. Data is presented separately for approach jumps performed with the doninant and non-dominant limb as lead limb.

4.3.3 Maximal voluntary isometric strength

The multivariate test for toe flexor strength did not indicate a significant time-by-group (Pillai's V = 0.040, F(2,22) = 0.465 p = 0.634) or group main effect (Pillai's V = 0.083, F(1,23) = 2.069 p = 0.164). However, main effect for time was discovered on the multivariate test (Pillai's V = 0.264, F(2,22) = 3.950 p = 0.034). Univariate follow-up tests indicated that the time main effects were significant for hallux strength in the flat (F(2,46) = 7.038, p = 0.002) and incline (F(2,46) = 5.085, p = 0.010) configurations (Table 4.4). Pairwise comparisons revealed that hallux strength increased from pre- to mid-test in both the flat (change = 0.23 ± 0.45 N·kg^{-1}, 95% CI = [0.10 0.36] N·kg^{-1}, p = 0.002) and incline configuration (change = 0.20 ± 0.42 N·kg^{-1}, 95% CI = [0.08 0.32] N·kg^{-1}, p = 0.005). This difference was significant with respect to pre-test also at post-test for both the flat (change = 0.29 ± 0.49 N·kg^{-1}, 95% CI = [0.15 0.42] N·kg^{-1}, $p < 0.001$) and incline configuration (change = 0.25 ± 0.51 N·kg^{-1}, 95% CI = [0.11 0.40] N·kg^{-1}, $p = 0.003$).

No significant time-by-group interaction or time or group main effects were observed for ankle plantar flexor strength ($p > 0.05$; Table 4.5).

Table 4.4. Maximal voluntary isometric hallux and 2nd through 5th digit (II-V) flexor strength in the incline and block heel-raise groups at pre-, mid-, and posttest.

Digits/group		Pretest	Midtest	Posttest
Hallux dominant leg: 22°			#	#
Block	(N·kg^{-1})	2.36 ± 1.10	2.40 ± 1.02	2.46 ± 1.17
Incline	(N·kg^{-1})	1.83 ± 0.57	2.05 ± 0.77	2.21 ± 0.79
Hallux dominant leg: 0°			#	#
Block	(N·kg^{-1})	1.83 ± 0.77	2.17 ± 0.99	2.04 ± 1.08
Incline	(N·kg^{-1})	1.59 ± 0.50	1.71 ± 0.60	1.83 ± 0.67
II-V dominant leg: 22°				
Block	(N·kg^{-1})	2.27 ± 0.92	2.13 ± 0.63	2.19 ± 0.81
Incline	(N·kg^{-1})	1.70 ± 0.56	1.69 ± 0.63	1.93 ± 0.77
II-V dominant leg: 0°				
Block	(N·kg^{-1})	1.75 ± 0.62	2.06 ± 0.64	1.94 ± 0.67
Incline	(N·kg^{-1})	1.54 ± 0.39	1.62 ± 0.61	1.71 ± 0.58
Hallux non-dominant leg: 22°			#	#
Block	(N·kg^{-1})	2.21 ± 0.87	2.39 ± 0.77	2.53 ± 1.02
Incline	(N·kg^{-1})	1.70 ± 0.50	2.01 ± 0.71	1.89 ± 0.58
Hallux non-dominant leg: 0°			#	#
Block	(N·kg^{-1})	1.58 ± 0.66	1.98 ± 0.72	1.98 ± 0.76
Incline	(N·kg^{-1})	1.41 ± 0.38	1.53 ± 0.55	1.70 ± 0.74
II-V non-dominant leg: 22°				
Block	(N·kg^{-1})	2.04 ± 0.92	2.16 ± 0.38	2.19 ± 0.78
Incline	(N·kg^{-1})	1.69 ± 0.69	1.78 ± 0.68	1.94 ± 0.76
II-V non-dominant leg: 0°				
Block	(N·kg^{-1})	1.68 ± 0.75	1.92 ± 0.44	1.89 ± 0.80
Incline	(N·kg^{-1})	1.52 ± 0.41	1.53 ± 0.70	1.69 ± 0.71

Main effect for time: Significant with respect to pre-test ($p < 0.05$)

Table 4.5. Maximal voluntary isometric ankle plantar flexor strength in the incline and block heel-raise groups at pre-, mid-, and posttest.

Side/angle		Pretest	Midtest	Posttest
Dominant leg: -15°				
Block	($Nm \cdot kg^{-1}$)	−1.06 ± 0.43	−0.94 ± 0.31	−0.92 ± 0.31
Incline	($Nm \cdot kg^{-1}$)	−0.83 ± 0.29	−0.84 ± 0.27	−0.86 ± 0.26
Dominant leg: 0°				
Block	($Nm \cdot kg^{-1}$)	−1.21 ± 0.45	−1.18 ± 0.41	−1.22 ± 0.45
Incline	($Nm \cdot kg^{-1}$)	−1.15 ± 0.37	−1.13 ± 0.33	−1.14 ± 0.33
Dominant leg: Max DF				
Block	($Nm \cdot kg^{-1}$)	−1.41 ± 0.49	−1.41 ± 0.45	−1.49 ± 0.50
Incline	($Nm \cdot kg^{-1}$)	−1.35 ± 0.39	−1.36 ± 0.37	−1.40 ± 0.36
Non-dominant leg: -15°				
Block	($Nm \cdot kg^{-1}$)	−1.06 ± 0.44	−0.88 ± 0.32	−0.94 ± 0.29
Incline	($Nm \cdot kg^{-1}$)	−0.80 ± 0.29	−0.84 ± 0.31	−0.82 ± 0.29
Non-dominant leg: 0°				
Block	($Nm \cdot kg^{-1}$)	−1.16 ± 0.40	−1.15 ± 0.42	−1.25 ± 0.35
Incline	($Nm \cdot kg^{-1}$)	−1.08 ± 0.39	−1.05 ± 0.36	−1.11 ± 0.38
Non-dominant leg: Max DF				
Block	($Nm \cdot kg^{-1}$)	−1.36 ± 0.50	−1.37 ± 0.45	−1.46 ± 0.45
Incline	($Nm \cdot kg^{-1}$)	−1.33 ± 0.35	−1.34 ± 0.45	−1.35 ± 0.39

4.3.4 Jumping mechanics

No significant group-by-time interaction nor time or group main effects were observed in NJM work performed at the ankle, knee, or hip during vertical countermovement jumps or 3-step approach jumps performed with the dominant limb as leading limb ($p > 0.05$; Table 4.6). For the 3-step approach, the omnibus test did not indicate a significant group-by-time interaction (Pillai's V = 0.181, $F(2,22) = 2.429$, $p = 0.111$) or group main effect (Pillai's V = 0.008, $F(1,23) = 0.183$, $p = 0.673$). However, there was a significant main effect for time during 3-step approach jumps performed with the non-dominant limb as lead limb (Pillai's V = 0.264, $F(2,22) = 4.356$, $p = 0.026$). However, univariate follow-up test did not reveal any significant main effects for time at the ankle, knee, or hip ($p > 0.05$; Table 4.6).

Table 4.6. Net joint moment work performed at the ankle, knee, and hip during vertical countermovement jump and 3-step approach jumps. The sum of left and right limb work is presented for the vertical jump, while lead and trail limb values are presented separately for approach jumps.

Jump type/group		**Pretest**	**Midtest**	**Posttest**
Vertical jump NJM work: Block				
Ankle	($J \cdot kg^{-1}$)	2.11 ± 0.56	2.07 ± 0.60	1.91 ± 0.62
Knee	($J \cdot kg^{-1}$)	0.83 ± 0.82	0.78 ± 0.62	0.83 ± 0.98
Hip	($J \cdot kg^{-1}$)	1.27 ± 0.61	1.52 ± 0.52	1.43 ± 0.73
Vertical jump NJM work: Incline				
Ankle	($J \cdot kg^{-1}$)	2.16 ± 0.54	2.24 ± 0.54	2.42 ± 0.52
Knee	($J \cdot kg^{-1}$)	1.08 ± 0.60	0.99 ± 0.59	0.79 ± 0.45
Hip	($J \cdot kg^{-1}$)	1.51 ± 0.51	1.58 ± 0.49	1.61 ± 0.62
Dominant approach NJM work: Block				
Ankle – Lead limb	($J \cdot kg^{-1}$)	0.82 ± 0.20	0.92 ± 0.25	1.02 ± 0.35
Knee – Lead limb	($J \cdot kg^{-1}$)	0.39 ± 0.53	−0.03 ± 0.72	−0.08 ± 0.44
Hip – Lead limb	($J \cdot kg^{-1}$)	0.52 ± 0.32	0.98 ± 0.71	1.07 ± 0.67
Ankle – Trail limb	($J \cdot kg^{-1}$)	0.48 ± 0.32	0.46 ± 0.40	0.63 ± 0.37
Knee – Trail limb	($J \cdot kg^{-1}$)	0.30 ± 0.58	0.35 ± 0.66	0.15 ± 0.75
Hip – Trail limb	($J \cdot kg^{-1}$)	0.55 ± 0.35	0.50 ± 0.40	0.74 ± 0.46
Dominant approach NJM work: Incline				
Ankle – Lead limb	($J \cdot kg^{-1}$)	0.91 ± 0.37	1.05 ± 0.26	0.96 ± 0.30
Knee – Lead limb	($J \cdot kg^{-1}$)	0.11 ± 0.63	0.16 ± 0.50	0.25 ± 0.47
Hip – Lead limb	($J \cdot kg^{-1}$)	0.70 ± 0.38	0.81 ± 0.54	0.89 ± 0.50
Ankle – Trail limb	($J \cdot kg^{-1}$)	0.42 ± 0.30	0.65 ± 0.27	0.57 ± 0.29
Knee – Trail limb	($J \cdot kg^{-1}$)	0.45 ± 0.62	0.12 ± 0.36	0.28 ± 0.39
Hip – Trail limb	($J \cdot kg^{-1}$)	0.59 ± 0.41	0.72 ± 0.33	0.59 ± 0.28
Non-dominant approach NJM work: Block				
Ankle – Lead limb	($J \cdot kg^{-1}$)	0.85 ± 0.29	0.80 ± 0.22	0.71 ± 0.24
Knee – Lead limb	($J \cdot kg^{-1}$)	0.49 ± 0.24	0.54 ± 0.63	0.72 ± 0.52
Hip – Lead limb	($J \cdot kg^{-1}$)	0.57 ± 0.32	0.80 ± 0.36	0.74 ± 0.39
Ankle – Trail limb	($J \cdot kg^{-1}$)	0.76 ± 0.26	0.97 ± 0.28	0.89 ± 0.30
Knee – Trail limb	($J \cdot kg^{-1}$)	0.12 ± 0.31	0.22 ± 0.24	0.35 ± 0.35
Hip – Trail limb	($J \cdot kg^{-1}$)	0.78 ± 0.24	1.04 ± 0.25	1.07 ± 0.27
Non-dominant approach NJM work: Incline				
Ankle – Lead limb	($J \cdot kg^{-1}$)	0.87 ± 0.20	0.96 ± 0.23	0.92 ± 0.22
Knee – Lead limb	($J \cdot kg^{-1}$)	0.67 ± 0.47	0.63 ± 0.33	0.66 ± 0.38
Hip – Lead limb	($J \cdot kg^{-1}$)	0.98 ± 0.26	0.96 ± 0.31	1.00 ± 0.34
Ankle – Trail limb	($J \cdot kg^{-1}$)	0.93 ± 0.28	1.04 ± 0.31	1.06 ± 0.30
Knee – Trail limb	($J \cdot kg^{-1}$)	0.33 ± 0.53	0.33 ± 0.51	0.36 ± 0.37
Hip – Trail limb	($J \cdot kg^{-1}$)	1.04 ± 0.35	1.00 ± 0.41	1.10 ± 0.54

4.4 Discussion

The main finding of this experiment was that vertical jump performance improved in both training groups over the 11-week training period. Moreover, the magnitude of improvement in jump height was comparable to those reported in a previous resistance training interventions aiming to improve vertical jump performance by strengthening the ankle plantar flexors (Chiu et al., 2017). The vertical jump improvements were accompanied by an increase in hallux flexor strength. However, there was no difference in the effectiveness of incline compared to block heel-raises in eliciting these improvements. Nor was any change observed in ankle plantar flexor MVIC or NJM work performed at the ankle, knee, or hip joint during jumping. The latter findings were surprising as the 8RM and 5RM heel-raise strength increased by 12.3 ± 7.8 kg and 14.1 ± 4.1 kg, respectively, in the block group, and 13.6 ± 6.4 kg and 16.9 ± 4.2 kg, respectively in the incline group (appendix B).

Previous research has reported increases in both vertical jump height and ankle plantar flexor strength following heel-raise exercise (Bangerter, 1968; Weiss et al., 1988). Specifically, strength gains between 13.2% and 40.0% have been reported following exercise with various variations of heel-raises (Weiss et al., 1988). However, the improvement in jumping performance in this study was greater than following heel-raise training alone, but smaller than those reported following lower extremity exercise programs that include heel-raises (Bangerter, 1968; Fatouros et al., 2000). In the preset study, participants increased their 8RM and 5RM in the heel-raise exercise, but no improvements were observed in ankle plantar flexor strength as measured by dynamometry. Participants were instructed to perform the heel-raises with the knees and hips extended (neutral) and to go through their maximal dorsiflexion/plantar flexion range of movement. Moreover, all exercise sessions were monitored by a member of the study team who ensured that the heel-raises were performed as prescribed. It therefore seems unlikely that other muscle groups than the ankle plantar flexors could be responsible for the improvements in heel-raise 8RM and 5RM. However, this begs the question of why the participants were not able utilize this strength during the ankle plantar flexor MVIC performed in the dynamometer. Similar discrepancies between strength adaptations to dynamic versus isometric exercises have been reported (Hartmann et al., 2012; Wilson & Murphy, 1996). These provide lack of conformity between the training exercise and test condition as the most likely explanation for such discrepancies, given the poor associations observed between dynamic exercises and isometric tests (Wilson & Murphy, 1996). Thus, it seems likely, that a lack of similarity between the dynamic heel-raises and the isometric ankle plantar flexor tests is the most likely explanation for the discrepancy between the 8RM and 5RM heel-raise and MVICs. Nonetheless, as ankle plantar flexor NJM work during jumps did not change over the course of the intervention, the increase in heel-raise strength does not provide a compelling explanation for the observed improvement in vertical jump performance.

One potential reason that changes in the work performed at the ankle, knee, or hip during jumps were not observed is that inter-individual differences in adaptations to the resistance training may have masked these changes. Specifically, since all athletes completed their team's general resistance training, which included squatting variations and weightlifting exercises, the athletes could have improved their vertical jump performance by performing more work at either of the lower extremity joints (appendix A). However, an improvement in vertical jump height due to the increase in ankle plantar flexor strength would only be expected if the athlete's ankle plantar flexor strength was the factor limiting vertical jump performance. This phenomenon may be explained in terms of the ratio of the NJMs during jumping to the maximal voluntary isometric strength, commonly referred to as relative muscular effort (RME; Chiu, 2018). For example, the vertical jump performance of an athlete with a large ankle plantar flexor RME compared to hip and knee RME during jumps would likely be limited by the ankle plantar flexor strength. In contrast, the vertical jump performance of an athlete whose ankle plantar flexor RME during jumping was low may instead be limited by hip or knee extensor strength. It is likely that the effect of strengthening the ankle plantar flexors would be dependent on whether the ankle plantar flexors were operating at a high or low RME (Milot et al., 2007; Milot et al., 2008). Thus, the former athlete in the example above may have improved vertical jump by increasing the ankle plantar flexor NJM and therefore likely NJM work during vertical jumps. However, the latter athlete would be more likely to increase vertical jump height by increasing the work performed at the hip or knee. Future resistance training studies should quantify the RME of the muscle group(s) involved if the aim is to understand how the training affects joint kinetics in a specific task.

An alternative explanation is that an increase in toe flexor strength contributed to the increase in vertical jump height. Previous research investigating the effects of toe strengthening protocols on vertical jump performance provide some experimental support to this hypothesis (Hashimoto & Sakuraba, 2014; Kokkonen et al., 1988; Unger & Wooden, 2000). Moreover, an increase in hallux strength from pre- to posttest was observed in the present study. However, as only two force platforms and no pressure mats or insoles were available for capturing ground reaction forces in the present study, it was not feasible to use a multi-segment kinetic foot model.

Whether the improvement in hallux strength could be responsible for the observed increases in jump height remains unclear.

Although it cannot be ascertained whether the improvement in vertical jump height was tied to the improvements in hallux flexor strength, the latter still provides insight to the utility of heel-raise exercises for improving foot function. Previous research has used toe grip cable curl, short foot, and towel grip exercises to strengthen the toe flexors (Hashimoto & Sakuraba, 2014; Jung et al., 2011; Lynn et al., 2012). However, these exercises have some limitations that the heel-raise exercises used in the present study do not. For example, the toe grip cable curl requires a special apparatus specially designed for toe flexor exercise, while increasing exercise intensity of short foot and towel grip exercises is difficult to achieve. In contrast, heel-raise exercises may be performed with a standard smith-machine, which is available in most commercial gyms. Additionally, incline and flat boxes used as platform for these exercises may be fashioned inexpensively out of wood. Alternatively, existing implements commonly found in gyms, such as a bumper plate may be used as platform for this exercise. Increasing the intensity of the heel-raise exercises can be achieved by adding weight to the barbell of the smith-machine. Given these advantages of heel-raise exercise compared to toe flexion-based and short-foot exercises, a comparison between these exercises to investigate whether they yield similar adaptations in foot function would be of interest.

Although the present study provides insight to how heel-raise exercise may improve foot function, the study also has limitations. First, the present study involves a relatively smally number of participants and has relatively high attrition (24%). Two factors contributed to these limitations. First, the pool of potential participants, which consisted of high-level female volleyball players, was relatively small. Second, capacity limits at training facilities and individuals' reservations towards undertaking in-person training during the COVID-19 pandemic exacerbated the challenge of recruiting and retaining research participants. Another limitation is that neither participants nor the investigators were blinded to the exercise protocol they completed, which may increase the risk of performance and detection bias. However, blinding the participants or accessors to which exercise variation that is being performed is impractical, if not impossible. Unfortunately, funds were not available to hire separate staff to perform either the training or the assessments. Therefore, blinding the assessors to group allocation was not possible. These potential risks of bias should be considered when interpreting the present findings. Lastly, the heel-raise exercise represented a relatively small proportion of the participants' total weekly training time. Specifically, the participants completed their team volleyball practices and general resistance training throughout the intervention. Although this likely have made it more difficult to isolate the effects of the heel-raise exercise, it must also be considered that the athletes' performance likely would deteriorate if their regular volleyball and resistance training had been discontinued (Bosquet et al., 2013). A loss of performance due to discontinuing team organized training could potentially have offset the benefits of the heel-raise training intervention. Moreover, asking the athletes to discontinue their training would not represent an ecologically valid situation for most trained individuals. Nonetheless, the improvements in hallux strength gives merit to the inclusion of the heel-raise exercise if one seeks to improve toe flexor strength or foot function.

In conclusion, 11-weeks of combined heel-raise exercise, jumping practice, and general resistance training is effective for improving vertical jump performance. However, the present study could not determine whether an increase in ankle plantar flexor strength or toe flexor strength contributed to the improvements in jump height. Nonetheless, the present results indicate that either incline or block heel-raise exercises may be effective for improving toe flexor strength.

4.4.1 Practical applications

Incline and block heel-raise exercise appears to be effective for improving toe flexor strength. Practitioners should therefore consider these heel-raise variations if they seek to improve an individual's toe flexor strength. As participants who performed both incline and block heel-raises improved their hallux flexor strength at a similar rate, practitioners may choose either exercise variation for this purpose. Nonetheless, practitioners should be aware that it remains to be elucidated how these heel-raise variations compare to other toe flexor strengthening exercises described in the literature. The results of the present study neither supports nor rejects the use of these heel-raise variations for the purpose of improving vertical jump height. However, given the relatively amount small time needed to include either incline or block heel-raises to a general resistance training program, it may prove worthwhile in situations where time is not a limiting factor.

Chapter 5

Discussion

This thesis was undertaken with the goal of furthering our understanding of how the extrinsic and intrinsic foot muscles may actively modulate foot arch rigidity. Specifically, this work sought to understand how foot arch rigidity may mediate the effect of ankle plantar flexor work on vertical jump performance as a propulsive task.

5.1 Ankle plantar flexor work affect vertical jump performance

Mechanical work performed at the ankle has received considerable research attention in an effort to understand the mechanical determinants of vertical jump performance (Arakawa et al., 2013; Bobbert et al., 2011; Chiu et al., 2017; Haguenauer et al., 2006; Virmavirta & Komi, 2001). The current work adds to this body of knowledge, demonstrating that the ankle plantar flexors perform more net lower extremity work during vertical jumping than the knee and hip extensors (Dæhlin & Chiu, 2019). Further, this research corroborates previous work demonstrating that restricting the work performed at the ankle during vertical jumping leads to a decrease in jump height (Arakawa et al., 2013; Bobbert et al., 2011; Haguenauer et al., 2006). However, a novel contribution from this work is that modulating an individual's jumping technique may affect the ankle plantar flexor work performed. This differs from previous work where restrictive devices or fatigue has been used to decrease the ankle plantar flexor's ability to perform work (Arakawa et al., 2013; Bobbert et al., 2011; Haguenauer et al., 2006; Virmavirta & Komi, 2001). Since reducing ankle plantar flexor work performed during the propulsion phase of jumping through a variety of mechanisms have a similar effect on vertical jump height, it seems appropriate to ascribe this effect to the ankle plantar flexor work rather than some other mediating mechanism.

As the ankle plantar flexor work performed during the propulsion phase of jumping affects vertical jump height, it is arguably of interest to practitioners to understand whether the work performed at the ankle may be increased through exercise. A limited number of studies have been undertaken to investigate the effects of strengthening the ankle plantar flexors on vertical jump performance. Bangerter (1968) investigated the effects of 8 weeks of heel-raise exercise on vertical jump performance but did not observe a significant change in vertical jump height. In contrast, Chiu et al. (2017) used the glute-ham-gastroc raise exercise to strengthen the gastrocnemius and demonstrated a positive effect on vertical jump performance. However, the ankle plantar flexor work performed during vertical jumps did not change following exercise (Chiu et al., 2017). In the present work two different variations of the heel-raise exercise were used to investigate the effects of combined ankle plantar flexor and toe flexor strengthening on vertical jump height. Both heel-raise variations resulted in similar improvements in vertical jump performance. However, like in Chiu et al. (2017), these improvements were not accompanied by an increase in the ankle plantar flexor work performed during jump propulsion. Thus, the mechanism through which an increase in ankle plantar flexor strength improves vertical jump performance remains to be elucidated.

5.2 Role of the plantar foot muscles in improving vertical jump performance

The strongest of the ankle plantar flexor muscles, the triceps surae, may deform the foot arch due to their action on the calcaneus (Blackman et al., 2009; Cheung et al., 2006; Thordarson et al., 1995). Excessive deformation of the foot arch may in turn lead to less efficient transfer of force from the triceps to the ground (Kelly et al., 2015; Takahashi et al., 2016). Thus, foot arch rigidity may act as a mediator that influences the effects of increasing ankle plantar flexor strength on vertical jump performance (Riddick et al., 2019).

The extrinsic foot muscles, intrinsic foot muscles, and plantar ligaments have been purported to modulate foot arch rigidity (Cheung et al., 2006; Farris et al., 2019; Kelly et al., 2014; Kelly et al., 2019; Kelly et al., 2015; Riddick et al., 2019; Thordarson et al., 1995). For example, plantar aponeurosis tension caused by toe extension may shorten and elevate the foot arch. (Blackman et al., 2009; Cheung et al., 2006; Thordarson et al., 1995) Similarly, both the extrinsic and intrinsic foot muscles have been found to contribute to foot arch elevation (Kelly et al., 2014; Thordarson et al., 1995). Despite the large number of studies undertaken to understand how each of these structures may contribute to foot arch support, this work is the first to investigate the relative contribution of the extrinsic muscles, intrinsic muscles, and plantar ligaments to the arch supporting moment. Specifically, a musculoskeletal model was used to demonstrate that the extrinsic and intrinsic muscles can generate a considerable moment at the midfoot, while the foot ligaments made a markedly smaller contribution (Chapter 3). Although this model demonstrated that the magnitudes of these muscle moments are significant, there is a large proportion of the moment acting at the midfoot that must be attributed to mechanisms that were not captured by this model. Likely mechanisms include bone-on-bone contact forces, as well as forces exerted on the plantar surface of the tarsal bones due to soft tissue contact (Chapter 3). Regardless of mechanism at play, this work supports the notion that the extrinsic and intrinsic foot muscles may influence foot arch rigidity by exerting a moment at the midfoot.

Another aim of this work was to understand whether modulation of foot arch rigidity can act as a mediator for the effects of ankle plantar flexor strengthening on vertical jump performance. Although several previous studies have investigated the effects of toe flexor strengthening on vertical jump performance, these studies have been designed based on the hypothesis that toe flexor strengthening may improve vertical jump height by increasing the work performed at the metatarsophalangeal joint (Goldmann et al., 2013; Hashimoto & Sakuraba, 2014; Kokkonen et al., 1988; Unger & Wooden, 2000). Although some of these studies have been successful in eliciting improvements in vertical jump performance and toe flexor strength, the improvements in vertical jump height have not been successfully connected to an increase in work performed at the metatarsophalangeal joint (Hashimoto & Sakuraba, 2014; Kokkonen et al., 1988). In contrast, this work used a resistance training protocol that was designed based on the hypothesis that the toe flexor muscles may affect vertical jump performance predominantly due to their effect on the foot arch (Chapter 4). A heel-raise performed on an incline plane was used in an attempt to strengthen the toe flexor muscles in a task where they predominantly would oppose foot arch deformation (Chiu & Dæhlin, 2021). This exercise was compared to a conventional heel-raise, which was expected to strengthen the ankle plantar flexors, but have little effect on the toe flexor muscles (Chiu & Dæhlin, 2021). Surprisingly, both exercise variations were equally effective at increasing both vertical jump height and toe flexor strength (Chapter 4). No changes were observed in the work performed at the lower extremity joints during the propulsion phase of the jumps. As such, the mechanisms responsible for the improvement in vertical jump height could not be established. Nonetheless, this research provides experimental evidence that the extrinsic and intrinsic foot muscles may act to resist foot deformation caused by large calcaneal tendon forces, as toe flexor strength improved following the heel-raise training (Chapter 4). Further, it is possible that the extrinsic and intrinsic muscles' ability to modulate foot arch rigidity acted as a mediator to improve vertical jump height. Specifically, if the improvement in toe flexor strength could reduce the foot arch deformation that occurred during the vertical jumps, more of the energy generated through ankle plantar flexor work could be used to increase the foot's potential energy, kinetic energy, or both at take-off.

Calcaneal plantar flexion has been demonstrated to occur in tasks that involve large ankle plantar flexor moments, such as prior to heel-off in gait and partial squats (Chizewski & Chiu, 2012; Stamm & Chiu, 2016). However, plantar flexion of the rearfoot segment alone does not contribute to elevating the body centre of mass (Riddick et al., 2019). In contrast, if both the rearfoot and forefoot segments are rotated simultaneously about the metatarsophalangeal joint, the body centre off mass would be elevated given that the angles of the remaining lower extremity segments remain unchanged (Iwanuma et al., 2011; van Ingen Schenau, 1989; van Ingen Schenau et al., 1987). The elevation of the body centre of mass attributed to ankle plantar flexion prior to take-off is an important contributor to vertical jump height (Caia et al., 2016; Chiu & Dæhlin, 2020). Thus, if more of the ankle plantar flexor work could be utilized to simultaneously plantar flex the rearfoot and forefoot, rather than plantar flexing the rearfoot alone, it would be reasonable to expect that this may have affected the participants' vertical jump height. Coincidently, this argument also sheds light on how different foot arch types may affect vertical jump performance. Specifically, a *pes cavus* deformity, or high arch, would result in the foot effectively becoming a shorter lever on which the body is elevated (Arangio et al., 1998). The shorter lever provided by the foot would result in a smaller increase in centre of mass height for a given angular excursion, thus resulting in less elevation of the centre of mass prior to take-off. In contrast, with a dynamic flexible flat foot, the lever provided by the foot would be longer but less rigid (Arangio et al., 1998). Thus, more calcaneal plantar flexion would result from a given moment exerted on the rearfoot. This may be hypothesized to result in a smaller plantar flexion excursion for the rearfoot and forefoot combined. Similar to the *pes cavus* foot, this would result in a smaller increase in elevation of the body centre of mass prior to take-off. As all the above scenarios describe situations that would affect the elevation of the foot, and ultimately body, centre of mass prior to take-off, they all represent ways in which the foot rigidity may affect the potential energy of the foot at take-off for a given ankle plantar flexor work.

A similar argument can also be made for the kinetic energy of the foot segment. Specifically, if the ability to resist foot arch deformation was improved due to an improvement in toe flexor strength, more of the ankle plantar flexor work performed could contribute to increasing the linear and angular velocity of the combined rearfoot and forefoot segments. An increase in the angular and linear velocities of the combined rearfoot and forefoot would in turn increase the linear velocity of the body centre of mass at take-off, given that the kinematics of the remaining lower extremity remain unchanged. This is significant as an increase in take-off velocity increases the distance the body centre of mass is elevated after take-off, commonly referred to as the flight height (Chiu & Dæhlin, 2020; Vanrenterghem et al., 2001). In contrast, if more of the work performed is used to rotate only the rearfoot, which may be hypothesized to occur with a less rigid arch, a smaller increase in the vertical velocity of the body centre of mass would result. As such the flight height would be reduced, potentially resulting in a smaller vertical jump height. Future research should investigate whether an increase in foot muscle strength could improve vertical jump performance by influencing the change in foot segment potential and kinetic energy at take-off.

5.3 Limitations

Although the present work provides insight to how ankle plantar flexor and foot muscle function may contribute to vertical jump performance as a propulsive task, it also has several limitations. First, the foot was modelled as consisting of 1 (Chapter 2 and 4) or 3 (Chapter 3) rigid bodies. Further, in the multi-body foot model, the rigid bodies were assumed to be connected by 1 degree of freedom pin joints (Chapter 3). In reality, the foot consists of 26 bones that all may move relative to each other. Moreover, the relative motion of these bones cannot accurately be represented as occurring about a single axis of rotation. However, marker-based motion capture techniques, such as those used in this work, requires a minimum of three markers placed on each rigid body to quantify their motion in three dimensions (Cappozzo et al., 2005). Since many of the bones of the foot are small, the size of reflective markers, as well as the number of reflective markers that accurately can be distinguished in a small space by the cameras of the system, place constraints on how many bodies may feasibly be used to represent the bones of the foot (Bruening et al., 2012). Although more detailed motion of individual bones of the foot may be possible using invasive methods, such as bone pins, or methods that expose participants to ionizing radiation, such as video fluoroscopy, stereophotogrammetry is still the preferrable option for quantifying motion of the foot (Cappozzo et al., 1996; Gefen, 2003). This means that a trade-off exists between accurately representing the foot and the feasibility of accurately quantifying the motion of many small segments.

Another factor that complicates this matter is that computation of kinetics requires knowledge of the external forces acting on each rigid body. During vertical jumping, the ground reaction forces is the only external force acting on the system (Chiu & Dæhlin, 2020; Vanrenterghem et al., 2001). This force can be easily quantified for the entire foot using force platforms. However, partitioning this force between multiple foot segments poses a considerable challenge. Most previous studies that distribute the ground reaction force between multiple segments have used pressure mats or insoles to quantify the pressure under separate regions of the foot during motion trials (Bruening et al., 2010; MacWilliams et al., 2003). The pressure has subsequently been used to partition the ground reaction force between multiple segments. Unfortunately, this technology was not available for the present work. An alternative that has been investigated for the purpose of partitioning the ground reaction force between multiple foot segments is to use the centre of pressure location together with the location of joints connecting the foot segments to determine which segment the ground reaction force is acting on (Stefanyshyn & Nigg, 1997). However, as the ground reaction force measured by a force platform represents the resultant of a distributed force system acting at all portions of the foot in contact with the ground, this approach does not accurately represent the true nature of the force distribution between multiple foot segments. It is therefore not surprising that this method has poor validity (Bruening & Takahashi, 2018). For the musculoskeletal model, a third alternative was used. Specifically, the ground reaction force was distributed between the forefoot and toe segments by assuming the location at which the ground reaction force would be acting on each segment and using static equilibrium to solve for the distribution for each instance in time (Chapter 3). This was possible for the tasks used in this study as 1) only two segments (the forefoot and toes) were in contact with the force platform, 2) the contact area of each segment was small and could reasonably be located to the ball of the foot and the distal end of the toes, and 3) the task was nearly static. As these conditions cannot not be assumed for vertical jump tasks, it was not feasible to use this approach for the studies investigating vertical jump performance.

Taken together, the methods used in the present study were only suited to quantify the motion of a small number of foot segments. Moreover, the kinetics of these foot segments were only feasible to quantify in one of the studies undertaken. Despite these limitations, the present work was able to quantify the midfoot net joint moment and angles during static tasks and use these as input to estimate the relative contribution of the extrinsic foot muscles, intrinsic foot muscles, and foot ligaments to foot arch support.

A second limitation to the present work is that foot arch type was not quantified. As discussed above it is conceivable that foot arch type may affect the relationship between ankle plantar flexor work performed during the propulsion phase of jumping and jump height. Moreover, foot arch type may influence the extent to which the foot deforms under a given load. For example, individuals with a rigid arch type would be expected to see less foot arch deformation (Stamm & Chiu, 2016). It is therefore possible that some of findings, such as the large variability in ligament contributions to midfoot moments predicted by the musculoskeletal model may be explained in terms of foot arch type. However, this does not detract from the finding that the extrinsic and intrinsic foot muscles may generate a considerable midfoot moment.

Lastly, for the studies investigating vertical jump performance (Chapter 2 and 4), only young females engaging in competitive sports were included as participants. It is therefore not clear whether the findings of these studies generalize to male, untrained, or clinical populations. Caution should therefore be taken in extrapolating the current findings to other populations, such as those with foot arch deformities or other foot pathologies.

5.4 Conclusions

This thesis has investigated the roles of the extrinsic and intrinsic foot muscles in foot arch support, and how they may act synergistically with the ankle plantar flexors during propulsive tasks to improve vertical jump performance. This work provides evidence that the extrinsic and intrinsic foot muscles have the potential to modulate foot arch rigidity. Moreover, these muscles may be strengthened using exercises in which their primary purpose is to resist foot arch deformation caused by the triceps surae muscles. The work undertaken during this thesis also demonstrates that ankle plantar flexor work performed during jump propulsion has a large effect on jump height. Although resistance training that includes different heel-raises is effective for improving vertical jump height and toe flexor strength, the mechanism through which this form of exercise contributes to an increase in vertical jump performance remain unclear.

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Appendix A

The general resistance training program followed by the high-school volleyball players in study 3 (Chapter 4) for each training week. Resistance used was either low, medium, high, or self selected.

Weeks 1-2

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
Plate squat	L/6*2	Plate squat	L/6*2	Plate squat	L/6*2
Front squat	L/6*4	Hang-clean	L/3*6	Overhead squat	L/3*4
Russian twist	X/6*3	Overhead press	X/3*3		

Weeks 3-4

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
Hang-snatch	Light/3*5	Hang-clean	L/3*2, M/2*2	Plate squat	L/6*2
Front squat	L/6*2, M/4*2	Deadlift	L/4*2, M/4*2	Back squat	L/6*4
Russian twist	X/8*3	Overhead press	X/5*3	Inverted row	BW/4*3

Weeks 5-6

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
Hang-snatch	L/3*2, M/2*3	Hang-clean	L/3*2, M/2*2	Plate squat	L/6*2
Front squat	L/6*2, M/4*2	Deadlift	L/4*2, M/4*2	Back squat	L/6*2, M/4*2
Russian twist	X/10*3	Overhead press	X/5*3	Inverted row	BW/5*3

Week 7

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
Hang-snatch	L/3*5	Clean	L/3*6		
Front squat	L/4*2, M/2*2	Overhead press	X/3*3	Midtest	
Russian twist	X/6*3				

Weeks 8-9

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
Hang-snatch	L/3*2, M/2*4	Clean	L/3*6	Hang-Clean	L/3*2, M/2*2
Front squat	L/5*1, M/3*1, H/3*2	Deadlift	L/4*2, M/4*2	Back squat	L/5*1, M/4*1, H/4*2
Russian twist	X/8*3	Overhead press	X/4*3	Inverted row	BW/5*3

Weeks 10-11

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
Hang-snatch	L/3*2, M/2*4	Clean	L/3*2, M/2*4	Hang-Clean	L/3*2, M/2*2
Front squat	L/5*1, M/3*1, H/3*2	Deadlift	L/4*2, M/4*2	Back squat	L/5*1, M/4*1, H/4*2
Russian twist	X/8*3	Overhead press	X/4*3	Inverted row	BW/5*3

W/R*S = Weight/Repetitions*Sets

L = Low, M = Medium, H = high, X = self selected

The general resistance training program followed by the varsity rookie and non-rookie volleyball players in study 3 (Chapter 4) for each training week. Resistance used was either low, medium, high, body weight, or self selected. The general resistance training was discontinued after 10 weeks due to COVID-19 related gym closures.

Week 1

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
Hang-snatch	L/3*2, M/2*3	Split snatch	L/4*3, M/2*4	Snatch	L/3*3, M/2*5
Power snatch + snatch	L/1+2, M/1+1*2, H/1+1*3	Hang-clean	L/3*2, M/2*3	Clean	L/2*2, M/2*5
Hang-clean	L/3*2, M/2*5	Power-clean + clean	L/1+2, M/1+1*2, H/1+1*3	Snatch-grip deadlift	L/5, M/5, H/5
Front squat	L/5, M/5, H/3	Back squat	L/8, M/5, H/3	Front squat	L/5, M/5, H/5
Snatch-grip deadlift[a]	L/5, M/5*3	Round back deadlift[a]	L/8, M/8*2	Sots press[a]	L/5, M/5*2
Eccentric pull-up[a]	BW/5*4	Inverted row[a]	BW/8*3	Eccentric pull-up[a]	BW/5*4
Sots press[b]	L/5, M/5*2	Press[b]	L/5, M/5*2		
Russian twist[b]	X/6*3	Russian twist[b]	X/6*3		

Week 2

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
		Power snatch + snatch	L/1+2*2,1 M/1+1*2, H/1+1*3	Snatch	L/3*2, M/2*2, H/1*3
		Power clean + clean	L/1+2*2, M/1+1*2, H/1+1*3	Clean	L/3*2, M/2*2, H/1*3
No workout		Front squat	L/5. M/3, H/3	Back squat	Max/5
		Round back deadlift[a]	L/8, M/8*2	Snatch-grip deadlift	L/5, M/5*3
		Inverted row[a]	BW/8*3	Sots press[a]	L/5, M/5*2
		Press[b]	L/5, M/5*2	Eccentric pull-up[a]	BW/5*4
		Russian twist[b]	X/6*3		

Weeks 3-5

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
Snatch	L/3*2, M/3*5	Back squat	L/8, M/8, H/8*2	Snatch	L/3*2, M/3*5
Clean	L/3*2, M/3*5	Split snatch	L/4*2, M/2*4	Clean	L/3*2, M/3*5
Front squat	L/5, M/5, H/5*2	Power clean	L/3*2, M/2*4	Front squat	L/5, M/5, H/3*2
Snatch-grip deadlift[a]	L/5, M/5, H/5	Round back deadlift[a]	L/8, M/8, H/8	Snatch-grip deadlift[a]	L/5, M/5, H/3
Eccentric pull-up[a]	BW/8*3	Inverted row[a]	BW/8*3	Eccentric pull-up[a]	BW/8*3
Sots press[b]	L/5, M/5*2	Press[b]	L/5, M/5*2	Sots press[b]	L/5, M/5*2
Russian twist[b]	X/6*3	Russian twist[b]	X/6*3	Russian twist[b]	X/6*3

Week 6

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
Back squat	Max/5			Snatch	L/3*2, M/2*2, H/1*3
Split snatch	L/4*2, M/2*4			Clean	L/3*2, M/2*2, H/1*3
Power clean	L/2*2, M1*4			Front squat	Max/3
Snatch-grip deadlift	Max/3	No workout		Round-back deadlift[a]	L/8, M/8*2
Pull-up	BW/Max			Inverted row[a]	BW/8*3
Sots press[a]	L/5, M/5*2			Press[b]	L/5, M/5*2
Russian twist[a]	X/6*3			Russian twist[b]	X/6*3

Weeks 7-9

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
Snatch	L/2*2, M/1*2, H/1*4	Front squat	L/5, M/3, H/3	Snatch	L/2*2, M/1*2, H/1*3
Clean	L/2*2, M/1*2, H/1*4	Split snatch	L/4*2, M/2*5	Clean	L/2*2, M/1*2, H/1*3
Back Squat	L/5, M/3*3	Power clean	L/3*2, M/2*5	Front squat	L/5, M/3, H/1
Eccentric pull-up[a]	BW/8*3	Press[a]	L/5, M/3, H/3	Eccentric pull-up[a]	BW/8*3
Sots press[a]	L/5, M/5*2	Inverted row[a]	BW/8*3	Sots press[a]	L/5, M/5*2
Russian twist	X/6*3	Russian twist	X/8*3	Russian twist	X/6*3

Week 10

Day 1		Day 2		Day 3	
Exercise	*W/R*S*	*Exercise*	*W/R*S*	*Exercise*	*W/R*S*
Snatch	L/2*2, M/1*5	Front squat	L/5, M/3*2	Snatch	Max/1
Clean	L/2*2, M/1*5	Split snatch	L/4*2, M/2*3	Clean	Max/1
Back Squat	L/5, M/3*2	Power clean	L/3*2, M/2*3	Front squat	Max/1
Eccentric pull-up[a]	BW/8*3	Press[a]	L/5, M/3, H/3	Eccentric pull-up[a]	BW/8*3
Sots press[a]	L/5, M/5*2	Inverted row[a]	BW/8*3	Sots press[a]	L/5, M/5*2
Russian twist	X/6*3	Russian twist	X/8*3	Russian twist	X/6*3

W/R*S = Weight/Repetitions*Sets

L = Low, M = Medium, H = high, BW = body weight, X = self selected

[a, b, …] = Exercises were performed as super sets

Appendix B

Mean ± standard deviation for 10 repetition maximum (RM), 8 RM, 5 RM, as well as estimated (est.) 1 RM results recorded in participants' training diaries. There was no n RM test in week 7 due to midtest. All results are given in kilograms.

		Block		Incline	
Week	**Repetitions**	**nRM**	**Est. 1RM**	**nRM**	**Est. 1RM**
1	10 RM	29 ± 3	35 ± 6	24 ± 4	30 ± 5
2	8 RM	36 ± 8	42 ± 10	33 ± 9	38 ± 8
3	8 RM	41 ± 4	48 ± 8	40 ± 6	47 ± 6
4	8 RM	45 ± 4	53 ± 9	43 ± 5	50 ± 5
5	8 RM	47 ± 6	55 ± 11	45 ± 5	53 ± 6
6	8 RM	49 ± 6	57 ± 11	47 ± 5	55 ± 7
8	5 RM	61 ± 8	67 ± 12	60 ± 5	67 ± 8
9	5 RM	66 ± 9	73 ± 14	66 ± 6	74 ± 10
10	5 RM	72 ± 8	81 ± 12	71 ± 6	79 ± 11
11	5 RM	76 ± 9	88 ± 10	77 ± 5	88 ± 6