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Joint contact loading in forefoot and rearfoot strike patterns during running

by

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Chapter 1. Introduction

Traditionally, running is associated with improved cardiovascular health but also increased risk of injury to the lower limbs. The incidence of lower extremity injury from running has been reported to range from 20% to 79% (Van Gent et al., 2007). The injury sites with the highest incidence are at the knee and lower leg. Many factors have been identified for increased risk of running injury, ranging from innate factors such as age and gender to modifiable factors such as lifestyle and running mileage. An additional risk factor may be foot strike pattern. Foot strike pattern was defined by Cavanagh and Lafortune (1980) as the point of initial contact of the foot with the supporting surface. The foot was sectioned into thirds moving from most posterior to anterior: rearfoot, midfoot, and forefoot. For the purpose of our study, one group of interest was composed of runners that practiced a rearfoot strike pattern (RFS). The habitual midfoot and forefoot strike runners were combined into a second group, forefoot strike pattern (FFS) because of the similar kinetics and kinematics associated with both.

The kinematics and kinetics of each foot strike pattern have been characterized but their association with injury is not well understood. FFS is characterized by an attenuated impact peak for the vertical ground reaction force (GRF) (Cavanagh & Lafortune, 1980; Lieberman et al., 2010). RFS is characterized by a larger impact peak and an increased loading rate of the vertical GRF. This evidence suggests that RFS runners are at a greater risk of injury because stress fractures have been associated with increased GRF rate of loading (Milner et al., 2006).

Minimalist footwear and barefoot running has become of interest because of its tendency to force runners to run with an adapted FFS. This minimalist approach does not

provide the heel cushion a typical running shoe provides. Therefore, de Wit et al. (2000) suggest that when running barefoot a runner will tend to run with a flatter foot, midfoot strike, in order to distribute the impact force over a greater area. Cavanagh and Lafortune (1980) described the midfoot strike as a distribution of the pressure between the heel and metatarsal-phalangeal joint creating a center of pressure (COP) near the middle of the foot. De Wit et al. (2000) also observed an effect on the kinematics of the knee and ankle during barefoot running. The knee was more flexed and the ankle was more plantar flexed in preparation for foot contact. This kinematic change suggests the body adjusts for better absorption of the shock.

Lieberman et al. (2010) showed the loading rate for FFS barefoot running was about seven times lower than barefoot RFS running. But even with the kinematic adjustments, FFS barefoot running has a similar vertical GRF loading rate to shod RFS running. This would suggest that benefits from FFS, with respect to loading rate of the vertical GRF, are cancelled out when running barefoot. Therefore, to decrease the loading rate with a FFS one would need to run shod.

Advocates for FFS pattern claim that a decreased impact peak and loading rate for the vertical GRF decreases the risk of injury. However, FFS has an increased active peak in the vertical GRF (Laughton et al., 2003), which may be due to increased muscle activity of the triceps surae during the propulsion phase of running. Laughton et al. (2003) also found a significantly greater tibial acceleration in FFS group, and Milner et al. (2006) associated increased tibial acceleration with increased risk of tibial stress fracture.

It is unclear whether the impact peak or the active peak of vertical GRF is more indicative of injury. In either case, GRF is an external load that is placed on the body. From

an injury perspective, it is typically the internal loads that damage the biological tissues. The internal joint load is composed of the joint reaction forces (JRF) and the muscle forces acting across the joint (Sasimontonkul et al., 2007). While GRF reaches magnitudes of approximately 2.5 times body weight during running, the joint contact forces can reach 8-15 body weights (Scott & Winter, 1990; Sasimontonkul et al., 2007).

Internal loading is difficult to measure without using invasive techniques. However, measuring internal forces in the lower extremity during running has been successful through the use of GRF and optimization techniques for muscle forces (Sasimontonkul et al., 2007; Scott & Winter, 1990; Glitsch & Baumann, 1997; Edwards et al., 2008). The majority of the internal loading is made up of muscle forces crossing the joints. Therefore, the peak internal loading occurs during propulsion of the stance phase when muscle activity is the greatest. This indicates that muscle activity, rather than GRF, has a greater influence on internal loading. Kleindienst et al. (2007) found that there was greater power absorption at the ankle and metatarsal-phalangeal joints for FFS running, suggesting increased muscle activity across the joints. In contrast, the RFS running demonstrated greater power absorption at the knee.

Differences in GRF and kinematics between RFS and FFS patterns during running have been established (Cavanagh & Lafortune, 1980; Kleindienst et al., 2007). However, it is not well understood how these strike patterns affect the internal loading of the lower extremity. The purpose of this study is to compare the internal joint loading and rates of loading at the ankle, knee and hip of habitual RFS runners to habitual FFS runners. Williams et al. (2000) found that RFS runners could convert to FFS with limited practice due to the mechanical coupling at the joints. Therefore, regardless of which strike pattern participants

are practiced at they will run both FFS and RFS. The FFS is expected to have greater internal joint loading at the ankle because of the increased activation of the triceps surae. However, the internal joint loading at the knee and hip are expected to be greater for the RFS pattern. The loading rate at the joints is not expected to differ much between groups because of the limited contribution of the external GRF.

Chapter 2. Literature Review

There are many different forms of running, one could run barefoot with pose technique or they could run shod with chi technique. It would be impossible to compare all the different combinations of running styles and techniques. So the specific variable of interest for this study is the strike pattern, or the point of initial contact of the foot with the supporting surface. Cavanagh and Lafortune (1980) segmented the foot into thirds in order to define rearfoot strike, midfoot strike and forefoot strike. This review of literature will detail the findings for FFS versus RFS, barefoot and/or minimalist running, and internal loading of the lower extremity.

The majority of runners perform a RFS when they run. Hasegawa et al. (2007) analyzed 248 runners at the 15 km mark of a half marathon. Twenty-three percent of runners ran with a midfoot strike and only 1.4% ran with forefoot strike. This leaves more than three quarters of the population falling in the RFS category. It was interesting, however, that the more elite the runners were, the higher the percentage of midfoot and forefoot strike runners. Of the top 50 runners, 38% of them exhibited midfoot strike and forefoot strike, while the percentage decreased to 21% for the last 48 runners.

Since there are considerably fewer habitual FFS runners, Williams et al. (2000) mechanical linking of the lower extremity has provided justification for using a habitual RFS and converting them to FFS for a FFS to RFS comparison. This was achieved by comparing the kinematics and kinetics of habitual FFS runners to converted FFS runners. There were no significant differences for most of the variables such as initial plantarflexion and inversion angles, dorsiflexion and eversion excursions, and vertical load rate of the GRF. However, the

vertical GRF was 0.39 BWs lower for the converted FFS. Also the ankle plantarflexion moment was 0.42 Nm/(kg*ht) greater for the converted FFS.

Epidemiological evidence is lacking concerning injury rates of forefoot/midfoot versus rearfoot strikers. There are, however, differences in the loading patterns between the different foot strikes. In particular, the kinematic differences are obvious even to the naked eye. In order to run with FFS pattern, the ankle needs to be in a plantarflexed position to avoid making initial contact with the heel (Laughton et al., 2003, Kleindienst et al., 2007). In a RFS pattern, it is just the opposite; the ankle is in a dorsiflexed position so that the heel can make the initial contact. The FFS has a greater dorsiflexion excursion because it starts out plantarflexed then gradually dorsiflexes as the heel is lowered and then plantarflexes during the propulsion phase.

There are also more subtle differences in the frontal plane ankle kinematics. The FFS starts out more inverted than the RFS (Stackhouse et al., 2003). This initial inversion of the FFS allows for a greater eversion excursion. Stackhouse et al. (2003) used habitual RFS runners and converted them to a FFS for comparison. The authors found an increased eversion excursion of 2.73° for the converted FFS. The converted FFS also had a faster eversion velocity, $270.6°s^{-1}$ compared to $190.91°s^{-1}$ for the RFS. Increased eversion velocity is linked with increased risk of shin splints (Clemente et al., 1981). However, RFS runners are characterized by a larger eversion peak angle (Kleindienst et al., 2007). Excessive rear-foot eversion is related to increase achilles stress (Clemente et al., 1981).

The kinematics at the knee for FFS and RFS have also been well established. Converted FFS has a more flexed knee at contact and lower peak knee flexion angle compared to RFS (Laughton et al., 2003). Therefore, the knee flexion excursion has been

reported to be 12% lower for the converted FFS. RFS also has an increased knee flexion velocity, because increased excursion occurs in a similar stance time.

Most of the research involving FFS versus RFS has focused on kinetic differences, especially the GRF, associated with the two strike patterns. The vertical GRF is typically broken down into an impact peak and an active peak. The impact peak occurs at the initial contact of the foot with the ground, while the active peak occurs around mid-stance during propulsion of the body (Cavanagh & Lafortune, 1980). It is fairly well accepted that a FFS has an attenuated impact peak (Cavanagh & Lafortune, 1980; Lieberman et al., 2010; Arendse et al., 2004). Lieberman et al. (2010) found that habitual FFS runners' impact was three times smaller than habitual RFS runners. FFS may have a decreased impact peak but it may be characterized by an increased overall active peak of the vertical GRF (Laughton et al., 2003; Kleindienst et al., 2007). The vertical GRF active peak was 2.64 and 2.48 BW for converted FFS and habitual RFS respectively (Laughton et al., 2003). However, other studies have found no difference in the active peak between FFS and RFS (Cavanagh & Lafortune, 1980; Arendse et al., 2004), but there are no studies that suggest a decrease in the active peak for the FFS.

The loading rate of the vertical GRF is also considered to be a possible indicator of injury risk to the lower extremity (Milner et al., 2006). It is suggested the vertical loading rate is decreased with a FFS (Lieberman et al., 2010; Williams et al., 2000). Lieberman et al. (2010) calculated the average loading rate from 0 to $6.2 \pm 3.7\%$ of stance; this was based on the time to the impact peak for the RFS runners. The authors found the loading rate when barefoot was decreased by seven times for habitual FFS versus RFS. Laughton et al. (2003) did not find a difference in loading rates between converted FFS and habitual RFS. They

determined their rate of loading by taking an average from 20 to 80% of the first slope until there was a noticeable change in the slope. For the RFS, this change occurred at the impact peak, while the FFS runners had a slight shift in the curve at a similar point to the impact peak for the RFS.

The anterior-posterior (AP) GRF also has some distinguishing characteristics between the RFS and the FFS. Both AP GRF curves are characterized by an initial braking force followed by and propulsion force over the second half of stance. The greatest differences in the curves occur in the initial braking force. While the RFS has smooth unimodal braking force, the FFS has bimodal braking force (Cavanagh & Lafortune, 1980; Kleindienst et al., 2007). This bimodal braking force is explained as an initial peak at contact of the forefoot with the ground and a slight decrease as the COP moves posteriorly until the second braking peak. There is some evidence that rate of loading (ROL) for the AP GRF is greater for FFS (Laughton et al., 2003). The medial-lateral GRF is very inconsistent between subjects and no noticeable patterns have been identified (Cavanagh & Lafortune, 1980).

Research focused on differences in joint moments between FFS and RFS is less detailed. Kleindienst et al. (2007) found converted FFS runners have a 13% decrease in knee extension moment at midstance and a 33% increase in external rotation after midstance. The increased internal rotation was 22% greater for the converted FFS at contact of the foot with the ground. The knee abduction moment was greater for FFS at foot contact and during early stance. Knee abduction moments and external rotation has been associated with patellofemoral pain syndrome. Kleindienst et al. (2007) also found increased peak plantarflexion moment at the ankle for FFS just after mid stance.

Laughton et al. (2003) looked at sagittal plane joint stiffness in FFS and RFS runners. The authors found converted FFS had 14.7% greater stiffness at the knee, while habitual RFS had 99.6% greater stiffness at the ankle. The decreased ankle stiffness in the FFS was attributed to increased dorsiflexion angle excursion, and the increased knee stiffness to the decreased knee flexion excursion.

The lower extremity stiffness is believed to be most influenced by the knee joint (Hamill et al., 2000; Laughton et al., 2003). Laughton et al. (2003) attributed a 22.5% increase in lower extremity stiffness in converted FFS runners to the increased knee stiffness. In turn, they found that FFS had increased tibial accelerations opposed to the RFS. The increased lower extremity stiffness is believed to partially account for this increased tibial acceleration. Greater peak vertical and AP GRF for the FFS were also believed to have an influence on the increased tibial acceleration. However, the change in acceleration is most likely due to a decrease in effective mass for FFS runners (Lieberman et al., 2010). Milner et al. (2006) looked at runners with a history of tibial stress fractures and found that these individuals had increased tibial acceleration when compared to a control group without a history of stress fractures. There was no determination of foot strike pattern in this study, so it may not hold true for FFS runners.

There is a growing trend for people to run in minimalist shoes or even barefoot. This has been encouraged by studies such as Lieberman et al. (2010) and popular books such as "Born to Run" by Christopher McDougall. At its core, this trend is based on the notion that humans evolved without footwear, and their continued use leads to increased injury risk. Barefoot impact forces occur at a very high rate and magnitude. Thus, the minimalist approach to running leads to the adoption of midfoot (De Wit et al., 2000) and forefoot strike

pattern (Lieberman et al., 2010). The midfoot strike pattern distributes the impact over a greater surface area. Whereas, both forefoot and midfoot strike patterns increase the activation of the plantar flexor muscles thus gradually lowering the heel to the ground and attenuating the impact (Cavanagh & Lafortune, 1980). Using EMG data on the triceps surae, electrodes were placed on the medial and lateral gastrocnemius and the soleus, barefoot running had increased muscle pre-activation in preparation for impact (Divert et al., 2005).

The minimalist approach would expect decreased impact peak and rate of loading because of its association with FFS. There is only mixed evidence that barefoot running does this. Divert et al. (2005) found that there was decrease impact and active peak vertical GRF for barefoot condition. During the barefoot condition, the participants' stride duration and contact time were significantly decreased, suggesting an increase in stride frequency. Since running velocity was controlled, the increased stride frequency indicates a decreased stride length. In turn, there would be less vertical displacement and velocity which could be an explanation for the decrease GRF in the barefoot condition. On the other hand, Lieberman et al. (2010) looked at a variety of different running conditions including barefoot FFS to shod RFS. They found that the loading rate of the vertical GRF was not significantly different but the impact peak was significantly decreased for the barefoot FFS. Another study compared barefoot and shod, allowing the participants to adjust their gait at will as long as they ran at a specific velocity (de Wit et al., 2010). This study showed that the participants decreased their stride length and landed more midfoot strike for the barefoot condition but still had increased impact peak and loading rate.

The kinematics involved with FFS running appear to be a protective measure that is employed when running barefoot. Adjustments are made even prior to contact with the

ground. De Wit et al. (2000) found when running barefoot the knee is more flexed 0.02 seconds prior to contact and the foot is more plantarflexed 0.03 seconds prior. Lieberman et al. (2010) looked at barefoot FFS versus RFS and there was an increase in impact peak and loading rate for the RFS runners. This would suggest that the adjustment to FFS during barefoot running do influence the vertical GRF.

The rearfoot eversion and tibial internal rotation excursion have also been detailed in a barefoot to shod comparison. Using intracortical bone pins Stacoff et al. (2000) did not find a difference between barefoot and shod. They did see increased eversion velocity in some shod conditions, when there was lateral heel flare.

To truly get an idea of what damage is being done to the internal biological structures, the internal loads on these tissues need to be calculated. These internal loads are calculated as a vector sum of the joint reaction forces and muscle forces. Inverse dynamics are used to determine the joint reaction forces from the GRF (Scott & Winter, 1990). The internal forces include muscles, ligaments, and bony structures. However, it is assumed the joint moments are, for the most part, the result of muscle forces crossing the joint times their moment arms. From this assumption, the net moment about the hip, knee and ankle joints are used as a constraint for calculating internal muscle forces (Glitsch & Baumann, 1997; Sasimontonkul et al., 2007, Edwards et al., 2008; Burdett, 1982). This is a an optimization methods that uses a cost function that finds a combination of muscle forces that produce the calculated net joint moment and either minimizes individual muscle forces or more commonly stress.

There are several studies looking at the internal loading of the lower extremity during running. These studies have found the same key result that most of internal loading environment is due to muscle forces. Peak vertical GRF during running are typically 2 to 2.5

BW, while ankle joint contact forces are between 10 and 15 BW in the axial direction (Scott & Winter, 1990; Glitsch & Baumann, 1997; Sasimontonkul et al., 2007). In the case of axial joint loading, both joint reaction forces and muscle force compress the joint. Glitsch and Baumann (1997) provided evidence that as the joint contact forces move distal to proximal there is an increase of muscle activity across the joint and thus increased compressive loading. However, Edwards et al. (2008) found that the compressive force at the knee was 26.9% greater than that of the hip. It has been well established that the peak compressive and shear loading of the joints occur at midstance; this would suggest the biological structures would not experience the greatest stress at impact.

Many parameters of FFS and RFS have been documented, but from the evidence provided, there is still ambiguity as to which one is most likely to cause injury. Part of this ambiguity may stem from the use surrogate measures of internal loading. Ground reaction force, net joint moments and kinematics all have drawbacks. Appropriate internal loading variables have been calculated for typical running, but FFS and RFS have not been systematically compared

This study is designed to compare the FFS and RFS on joint contact loading at the ankle, knee and hip. In order to accomplish this, we will use habitual as well as converted forefoot and rearfoot strikers. The main variable of interest for this study is the compressive loading and loading rates of the ankle, knee and hip. We expect that FFS will have greater compressive load at the ankle compared to the RFS, and the RFS will have greater compressive load at the knee and hip.

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Chapter 3. Joint contact loading in forefoot and rearfoot strike patterns during running

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Brandon D. Rooney and Timothy R. Derrick

Abstract

Research concerning forefoot strike pattern (FFS) versus rearfoot strike pattern (RFS) during running has focused on the ground reaction force. The main purpose of this study was to determine the internal loading of the joints for each strike pattern. A secondary purpose was to determine if a converted runner can adequately represent the habitual population for both FFS and RFS. Using inverse dynamics to calculate the net joint moments and reaction forces, and optimization techniques to estimate muscle forces we determined the axial compressive load and rate of loading at the ankle, knee, and hip. Effect sizes (ES) were used to categorize comparisons into small (ES>0.2), medium (ES>0.5) and large (ES>0.8) effects. Subjects consisted of 15 habitual FFS and 15 habitual RFS competitive runners. Each subject ran with their habitual strike pattern and then converted to the opposite strike pattern. Converting to a FFS pattern had the opposite effect of converting to a RFS pattern at the knee and hip. Converted FFS runners had decreased knee (ES=0.98) and hip (ES=1.23) contact forces while converted RFS runners had increased knee (ES=0.77) and hip (ES=0.71) contact forces. Habitual FFS runners had higher contact forces at the ankle (ES=0.51) and hip (ES=0.81) compared to habitual RFS runners. The habitual FFS runners also demonstrated a substantially larger compressive load at the ankle and knee during the first 40% of stance. This could be important when considering the high number of cyclical loads experienced by these runners.

Introduction

Running is associated with improved cardiovascular health but also increased risk of injury to the lower limbs. The incidence of lower extremity injury from running has been reported to range from 20% to 79% (Van Gent et al., 2007). Many factors have been identified for increased risk of running injury, ranging from innate factors such as age and gender to modifiable factors such as lifestyle and running mileage. An additional risk factor may be foot strike pattern. Foot strike pattern is defined by Cavanagh and Lafortune (1980) as the point of initial contact of the foot with the supporting surface. The foot is sectioned into thirds moving from most posterior to anterior: rearfoot, midfoot, and forefoot. For the purpose of our study, one group is composed of runners that practice a rearfoot strike pattern (RFS). The habitual midfoot and forefoot strike runners are combined into a second group, forefoot strike pattern (FFS), because of the similar kinetics and kinematics associated with both.

RFS is the dominant strike pattern practiced by runners (Hasegawa et al., 2007), but with a growing trend to run in minimalist shoes or even barefoot, FFS running may become more common. This trend has been encouraged by studies such as Lieberman et al. (2010) and popular books such as "Born to Run" by Christopher McDougall. At its core, this trend is based on the notion that humans evolved without footwear, and their continued use leads to increased injury risk. Many of the major shoe companies are following suit with the production of minimalist shoes such as the Nike Free, New Balance Minimus, and Vibram fivefingers.

Minimalist running is of interest because of its tendency to force runners to run with a FFS pattern. This minimalist approach does not provide the heel cushion of a typical running

shoe therefore, runners tend to run with a midfoot or forefoot strike pattern in order to distribute the impact force over a greater area and allow greater absorption of the impact by the plantar flexor muscles (de Wit et al., 2000; Lieberman et al. 2010; Divert et al., 2005). Cavanagh and Lafortune (1980) described the midfoot strike as a distribution of the pressure between the heel and metatarsal-phalangeal joint creating a center of pressure near the middle of the foot. De Wit et al. (2000) also observed an effect on the kinematics of the knee and ankle during barefoot running. The knee was more flexed and the ankle was more plantar flexed in preparation for foot contact. This kinematic change suggests the body is in a better position to absorb the impact.

The kinetics of RFS and FFS has been characterized but their association with injury is not well understood. RFS has a large vertical ground reaction force (GRF) impact peak (Cavanagh & Lafortune, 1980; Lieberman et al., 2010). FFS is characterized by an attenuated or absent impact peak, but an increased active peak (Laughton et al., 2003; Lieberman et al., 2010)). The rate of loading (ROL) for FFS and RFS is somewhat conflicting. Lieberman et al. (2010) found a seven fold decrease in ROL for barefoot FFS compared to barefoot RFS, whereas Laughton et al. (2003) found no difference for the ROL between RFS and converted FFS. Attempts have been made to associate vertical GRF and ROL with running injuries (Milner et al., 2006; Crossley et al., 1999). The findings are inconsistent and conflicting, making GRF an unreliable predictor of running related injuries (Zadpoor & Nikooyan, 2011).

Scott and Winter (1990) suggested that the ground reaction forces are not as important as the forces acting at the site of the injury. It is typically the internal loads that damage the biological tissues. While GRF's reach magnitudes of approximately 2.5 times body weight during running, the joint contact forces are estimated to reach 8-15 body

weights (Edwards et al., 2008; Sasimontonkul et al., 2007). Therefore, the use of the external GRF is only a surrogate measure of the internal load and will vastly underestimate the loads experienced by the body. For example, a plantar flexor moment of 0.30 body weights * meter (BWm) and moment arm of 0.05m for the triceps surae will produce and compressive force of 6 BWs at the ankle (Stief et al., 2008; Maganaris et al., 2000). The internal joint load is composed of the joint reaction forces (JRF) and the muscle forces acting across the joint (Sasimontonkul et al., 2007). Therefore, assuming a joint reaction force of 2.5 BWs and muscle forces of 6 BWs, the total compressive load at the ankle would easily exceed 8 BWs. The internal loading is difficult to measure without using invasive techniques, but there has been success using musculoskeletal modeling to estimate the loading (Sasimontonkul et al., 2007; Scott & Winter, 1990; Glitsch & Baumann, 1997; Edwards et al., 2008).

The majority of the internal loading is made up of muscle forces crossing the joints. Kleindienst et al. (2007) found greater energy absorption at the ankle and metatarsal-phalangeal joints during FFS running compared to RFS. In contrast, the RFS running demonstrated greater energy absorption at the knee. EMG data on the triceps surae (electrodes were placed on the medial and lateral gastrocnemius, and the soleus) during barefoot running had increased muscle pre-activation in preparation for impact (Divert et al., 2005). Since barefoot running tends to force runners to run with FFS, it is expected that FFS runners with have similar pre-activation of the triceps surae indicating increased activity at the beginning of stance.

There has yet to be a study investigating how strike patterns affect the internal loading of the lower extremity. Therefore, the main purpose of this study was to use musculoskeletal modeling techniques to compare the internal joint loading and rates of

loading at the ankle, knee and hip of habitual RFS runners and habitual FFS runners. A secondary purpose was to determine if converted strike pattern runners are an adequate model for habitual runners when studying internal loading of the lower extremity joints. The FFS is expected to have greater internal joint loading at the ankle because of the increased activation of the triceps surae. However, the internal joint loading at the knee and hip are expected to be greater for the RFS pattern due to greater energy absorption. The loading rate at the joints is not expected to differ between groups because of the limited contribution of the external GRF.

Methods

Subjects

Fifteen forefoot and 15 rearfoot strike runners were recruited from a population of competitive runners. They were required to be free from injuries that could affect their mechanics during running. The study was approved by the Iowa State Institutional Review Board, and upon arrival all participants read and signed the informed consent.

Protocol

Participants were provided with the same brand and model of running shoes. Anthropometrics were measured and recorded for each participant. These included body mass, height, thigh length and circumference, calf length and circumference, foot length and breadth, and malleolus height and width.

The participants were asked to select a running speed that was comfortable and could be maintained for a long run but that was quicker than an easy recovery run. The participants then performed a five minute warm up on a treadmill at this speed. The same speed was then used for the over-ground running trials. Sixteen retro-reflective markers were placed on the

right leg, shoe, pelvis, and trunk: head of the fifth metatarsal, dorsi-foot, medial and lateral

malleoli, anterior distal and proximal calf, posterior calf, medial and lateral femoral

epicondyle, anterior and lateral thigh, both greater trochanter, sacrum, and both acromion

processes. An additional 5 markers were located on the heel counter of the shoe,

Position data were recorded from the markers during a standing trial and two over-

ground running conditions using an 8 camera motion capture (Vicon MX, Vicon, Centennial,

CO, USA), 200Hz. For the running trials, participants ran down a 30m runway and were

instructed to strike either of two adjacent force platforms (AMTI, Watertown, MA), 2000

Hz, with their right foot. They performed practice running trials to avoid targeting and allow

for natural gait. The participants were required to run within ± 5% of their selected speed.

This was monitored by measuring the horizontal velocity of the sacral marker. Ten good

trials were collected. Trials were discarded if the participant did not strike the force platforms

with their entire right foot, made any noticeable adjustments to strike the platform, or ran

outside their speed range.

For the first condition, participants were instructed to run naturally with no mention

of strike pattern. Ten acceptable trials were collected and the participant's strike pattern was

identified. The initial point of contact was calculated as an average of the center of pressure

under the foot over the first five data points of stance. This was reported as the heel strike

index (HSI), a percentage of the foot length from the heel. The participants were then

categorized into one of two groups, habitual RFS (HSI<33.3%) or habitual FFS

(HSI>33.3%). They were instructed on how to run with the opposite strike pattern, and were

allowed to practice until they felt comfortable with the new strike pattern. Then the

participants performed ten acceptable running trials using the converted/opposite strike pattern. These conditions were termed converted RFS and converted FFS.

Data Processing

Data analysis was performed in Matlab (7.9.0, R2009b) for the stance phase. Stance phase commenced when vertical GRF was greater than or equal to 10 N and ended when it fell below the 10 N threshold. Raw kinematic and kinetic data were filtered using a 4th order zero lag Butterworth filter. In order to reduce impact artifact in the joint moment curves, the kinematic and kinetic data were both filtered with a low-pass cutoff of 20 Hz (Bisseling & Hof, 2006). The Cardan joint rotational sequence used to calculate joint angles was flexion/extension, adduction/abduction and internal/external rotation. Anthropometric measurements were used to determine segment masses, center of mass and moments of inertia according to the equations of Vaughn et al. (1992). Inverse dynamics calculations were used in conjunction with rigid body assumptions to determine joint moments and joint reaction forces. Kinematic data was up sampled to 2000 Hz for the inverse dynamic calculation.

Joint angles were used to drive a musculoskeletal model that included 43 lower extremity muscles and subject specific bone and muscle scaling. The model used muscle points and maximum isometric forces described in Delp et al. (1990). Moment arms, muscle orientations, and velocity and length adjusted maximal dynamic muscle forces were calculated for each 1% of the stance phase of each trial. A static optimization was used to select a set of muscle forces that minimized the sum of stresses cubed for each frame of data (Glitisch & Baumann, 1997). The solutions were selected so that the muscle forces multiplied by the moment arms equaled the joint moments calculated experimentally. The

joint moments used in this optimization were the sagittal and frontal axes at the hip and ankle, and the sagittal plane axis at the knee. An upper bound for the muscle force solution was the maximal dynamic muscle forces estimated with the musculoskeletal model and the lower bound was zero. The optimized muscle forces were then filtered with a low-pass cutoff of 20 Hz to prevent non-physiological changes as the static optimization procedure moved from one frame to the next. Once the muscle forces were estimated, they were vector summed with the joint reaction forces and placed in the proximal segment coordinate system to obtain the joint contact forces.

Statistical Analysis

The variables of interest for this study were the compressive contact force and the rate of loading at the ankle, knee and hip. The peak loading was determined for each trial and then averaged within the FFS and RFS conditions for each subject. There were four conditions: habitual RFS, habitual FFS, converted RFS and converted FFS. Effect sizes (ES) were calculated to categorize comparisons between conditions into small (ES>0.2), medium (ES>0.5), and large (ES>0.8) effects (Cohen, 1992). A p-value of less than an alpha level of 0.05 was considered statistically significant for the habitual to converted comparison.

Two independent measures t-tests were performed, one to detect differences between habitual RFS runners and converted RFS runners, and the second to detect differences between habitual FFS runners and converted FFS runners. If there were no statistical difference from the prior tests, a repeated measures t-test was performed to detect differences between foot strike patterns. If there were significant differences in the original two t-tests, independent measures t-tests were used to compare habitual RFS to habitual FFS. The habitual FFS and RFS groups were matched for several of the subject characteristics: gender,

age, body mass, height, mileage, and speed. Independent measures t-tests were performed on each of the characteristics to determine if there was a significant difference between groups. Statistical analyses were performed in IBM SPSS Statistics 19 (IBM Corp., Somers, NY).

Results

The running velocities of the FFS and RFS groups were 4.36 (0.23) and 4.25 (0.26) m/s respectively (Table 1). The groups were not significantly different in age, height, body mass, mileage or running speed. The HSI for converted FFS condition was greater than the habitual FFS condition, 69.1 (6.1) % compared to 63.3 (7.1) %, but was not significantly different ($p > 0.05$). The habitual and converted RFS conditions had similar HSI at 22.9 (2.0) % and 20.7 (4.4) % ($p > 0.05$).

Joint Moments

Hip, knee and ankle sagittal plane joint moments as well as hip and ankle frontal plane joint moments were used in the optimization (Figure 1). The sagittal plane moments were prominently extensor. The notable exception was a hip flexor moment during the final 40% of stance and an ankle dorsiflexor moment at beginning of stance for the RFS condition. There was an abductor moment at the hip during the first 75% of stance and a predominant inverter moment and the ankle. The major differences between the habitual RFS and habitual FFS joint moments were a much earlier activation of the plantar flexor and a bimodal hip flexor moment in FFS condition. The bimodal hip flexor moment was considered an artifact due to the inability of the kinematic signal processing to adequately smooth the signal without attenuating the acceleration peaks to non-physiological levels (Bisseling and Hof, 2006; van den Bogert and de Koning, 1996).

Muscle Forces

Optimized muscle activations generally agreed with the joint moments (Figure 2). The major extensor muscles (gluteus maximus, vasti and triceps surae) had the greatest activations. The hamstrings showed activation and there was a small activation of the tibialis anterior during the first 20% of stance for the RFS as the foot was lowered to the ground by an eccentric contraction of this muscle. The tibialis anterior muscle also illustrates a limitation of the optimization process as the lower bound of this muscle was consistently violated with a small negative value (-.02 BW) at approximately 60% of stance.

Habitual vs. Converted Contact Forces

The peak contact forces at the hip, knee, and ankle differed slightly between habitual and converted conditions (Table 2). The peak axial contact forces ranged from -8.67 to -15.09 BWs. When a habitual RFS runner converted to FFS, the axial contact force increased at the ankle but decreased at the knee. The exact opposite occurred for habitual FFS runners converting to RFS; the axial contact force increased at the knee and but decreased at the ankle. The converted FFS condition was lower than the habitual FFS at the ankle (p=0.326, ES=0.36), knee (p=0.011, ES=0.98), and hip (p=0.008, ES=1.23) for peak axial contact forces. The converted RFS condition was lower than the habitual RFS conditions at the ankle (p=0.411, ES=0.27) and higher at the knee (p=0.026, ES=0.77) and hip (p=0.415, ES=0.25).

The patterns of joint contact loading for habitual versus converted for FFS and RFS were similar (Figure 3). The peak ROL was lower for the converted FFS compared to the habitual FFS at the ankle (p=0.193, ES=0.52), knee (p=0.101, ES=0.58), and hip (p=0.499, ES=0.29). The converted RFS was lower than the habitual RFS at the ankle (p=0.388, ES=0.36) and hip (p=0.185, ES=0.71) but higher at the knee (p=0.065, ES=0.41).

Habitual Rearfoot vs. Habitual Forefoot Contact Forces

Since there were statistically significant differences between habitual and converted conditions, we did not collapse the habitual and converted conditions for FFS and RFS. Habitual FFS was greater than the habitual RFS for the peak axial contact forces at the ankle (p=0.176, ES=0.51), knee (p=0.415, ES=0.30), and hip (p=0.041, ES=0.81). Habitual FFS and habitual RFS patterns of loading differed at the ankle; the FFS condition had larger forces for the first 40% of stance. The knee and hip were more similar in pattern. The habitual FFS had higher peak ROL than the habitual RFS at each: ankle (p=0.314, ES=0.37), knee (p=0.101, ES=0.63), and hip (p=0.383, ES=0.33).

Discussion

There has been little research comparing the internal loading of FFS and RFS runners. With trends toward minimalist running, there is an increasing need for an understanding of the differences in the potential for injury. The purpose of this study was to compare the two styles and the resulting internal joint loading at the ankle, knee, and hip. We were also interested in the ability of runners to convert from one strike pattern to another. The competitive running population was used for this study because of the increased proportion of FFS runners (Hasegawa et al., 2007).

It was hypothesized that the FFS runners would have higher axial contact forces at the ankle and the RFS runners would have higher axial contact forces at knee and hip. The habitual FFS condition was higher at the ankle, but contrary to our prediction the habitual FFS was also higher at the knee and hip. Our results lent support to the hypothesis that there would be no difference in rate of loading between habitual FFS and habitual RFS. Our method of filtering the ground reaction forces (low-pass 20 Hz) may have influenced the rate

of loading variables. This low of a cutoff will attenuate the RFS vertical impact force and the FFS braking impact force.

The internal joint contact forces are difficult to measure without invasive techniques such as instrumented prosthesis. There has been success using analytic methods to estimate internal joint loading (Sasimontonkul et al., 2007; Edwards et al., 2008; Glitsch & Braumann, 1997; Scott & Winter, 1990; Burdett, 1982). Internal forces at the ankle, knee, and hip have been consistently reported to range between 8-15 BW for axial compressive forces. Our values fall in the upper end of the range for the ankle and knee. This is may be due to the relatively faster running speed of the participants. Previous studies typically had speeds below 4 m/s, and our average was 4.3 m/s for all the subjects. It was estimated that increased running velocity results in increased joint contact forces (Edwards et al., 2010). The larger contact forces are also supported by Glitsch and Baumann (1997). They found a joint contact force near 18 body weights for their one FFS runner (5m/s).

Although the peak differences between habitual FFS and habitual RFS runners were slight, the repetition involved with distance running must be taken into consideration. The average mileage of the participants was above 9.2 miles per day, which would correspond with cyclical loading of 4750 cycles per day at a running speed of 4.5 m/s (Edwards et al., 2010). Even though the differences in the peak contact forces of the habitual runners were less than 0.80BW (Table 2) there were other differences that might suggest increased loading with the FFS. The habitual FFS runners experienced substantially greater compressive loading at the ankle during the first 40% of stance and somewhat greater loading at the knee during the same time period (Figure 4). This increased load placed over 33,250 cycles per week may become a factor with structural fatigue of the biological tissues.

The results for GRF correspond well with previous literature concerning FFS versus RFS (Figure 5) (Cavanagh & Lafortune, 1980; Lieberman et al., 2010; Laughton et al., 2002). Habitual and converted FFS runners demonstrated vertical GRF with an attenuated impact peak and an increased active peak. The AP GRF also had a bimodal breaking force occurring just before 10% and around 20% of stance. Both RFS conditions provided a typical vertical GRF with a steep impact peak followed by the active peak.

There has only been one study looking at the ability of a converted strike pattern to represent a habitual strike pattern. Williams et al. (2000) proposed a mechanical link between segments that allowed habitual RFS runners to convert to a FFS and adequately represent habitual FFS runners. Our data provide some support that converted runners can represent the habitual population for both FFS and RFS but caution should be used when making these assumptions. The pattern of joint contact loading, moments, and muscle forces were similar between the habitual and converted conditions for each style of running. However, there were some statistically significant differences in the peak contact forces. The magnitude of the peak forces at the knee and hip joints were decreased in the converted FFS condition compared to habitual FFS. It appears that habitual FFS runners were able to convert to a RFS according to the principles proposed by Williams et al. (2000).

The internal joint moments also provided evidence that a converted runner can imitate a habitual runner. The patterns for the internal joint moments at the ankle, knee, and hip were similar for both converted and habitual conditions. However, the magnitudes of the peak moments for the converted condition are consistently decreased except for the converted RFS at the knee about the ML axis. Similarly, William's et al. (2000) found a significant decrease

in plantar flexion moment for converted FFS runners. These differences are minimal and may disappear with practice.

In conclusion, habitual FFS runners have greater peak axial contact forces than RFS runners, but the differences are less than 0.80 BW. These differences, in addition to the earlier onset and greater magnitude of the axial compressive loading at the ankle and knee throughout the first 40% of stance, may place FFS runners at a higher risk of injuries such as stress fractures and stress reactions. This may become especially significant over the weeks and months of high mileage training in elite distance runners.

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Tables

Table 1: Mean (SD) subject characteristics for habitual FFS and RFS groups. No significant difference between groups (p-value > 0.05).

	Habitual RFS	Habitual FFS
Body Mass (kg)	65.7 (7.6)	63.7 (8.7)
Height (m)	1.78 (0.07)	1.78 (0.08)
Age (yr)	20.6 (1.6)	22.0 (2.3)
Mileage (mi/wk)	59.9 (17.8)	63.7 (24.6)
Speed (m/s)	4.25 (0.26)	4.37 (0.23)

Table 2: Mean (SD) of the internal axial contact forces and rate of loading (ROL) at the ankle, knee, and hip during FFS and RFS running. * = significantly different between converted and habitual RFS; † = significantly different between habitual and converted FFS ($p<0.05$, EF>0.5).

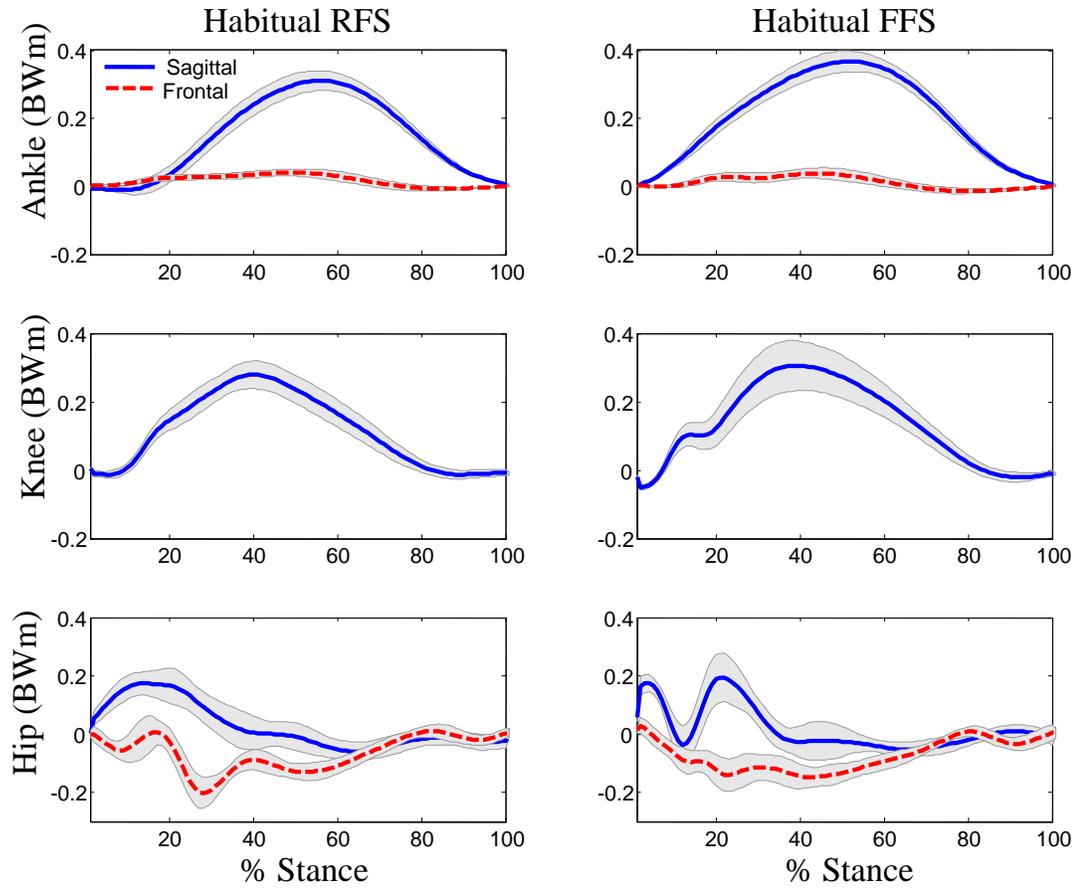|	Rearfoot Strike Pattern		Forefoot Strike Pattern	
Contact Forces (BW)	Habitual	Converted	Habitual	Converted
Ankle	-14.35 (1.41)	-13.97 (1.10)	-15.09 (1.49)	-14.56 (1.42)
Knee *†	-12.75 (1.43)	-13.85 (1.12)	-13.18 (1.44)	-11.77 (1.40)
Hip †	-8.76 (1.17)	-9.06 (0.74)	-9.52 (0.68)	-8.67 (0.91)
ROL (BW/s)				
Ankle	-268.3 (28.4)	-258.1 (35.0)	-278.8 (27.5)	-264.4 (31.4)
Knee	-242.0 (35.0)	-267.0 (36.3)	-268.8 (50.1)	-239.9 (43.1)
Hip	-299.4 (70.5)	-270.2 (44.0)	-319.7 (53.5)	-304.3 (68.8)

Figures

Figure 1: The net internal joint moments used for muscle optimization in the habitual RFS and habitual FFS conditions and their SD bands. Moments were about the ML and AP axis for the hip and ankle, and the ML axis for the knee. Positive values indicate extension, plantar flexion, inversion and adduction.
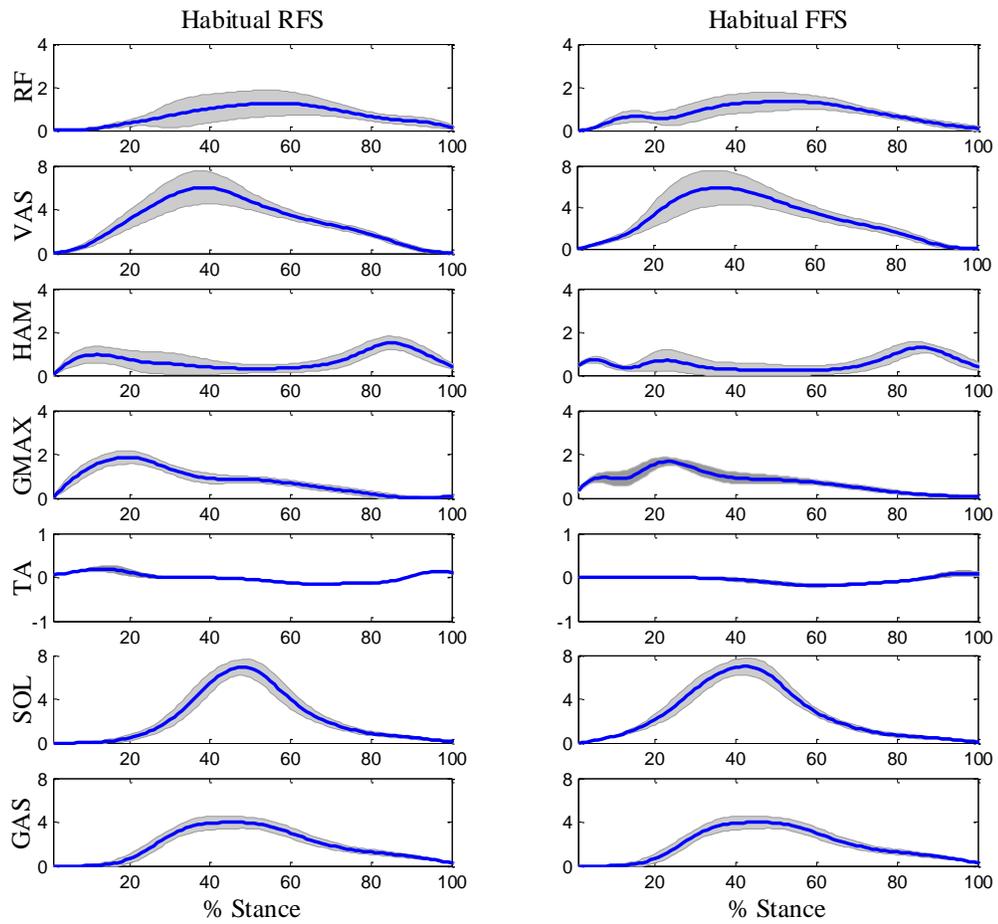
Figure 2: The ensemble curves for the estimated muscle forces (BWs) for habitual RFS and habitual FFS conditions and there SD band. (GAS = medial & lateral gastrocnemius; SOL = soleus; TA = tibialis anterior; GMAX = gluteus maximus; HAM = semitendinosus, semimembranosus & biceps femoris; RF = rectus femoris). Note the difference in scaling.
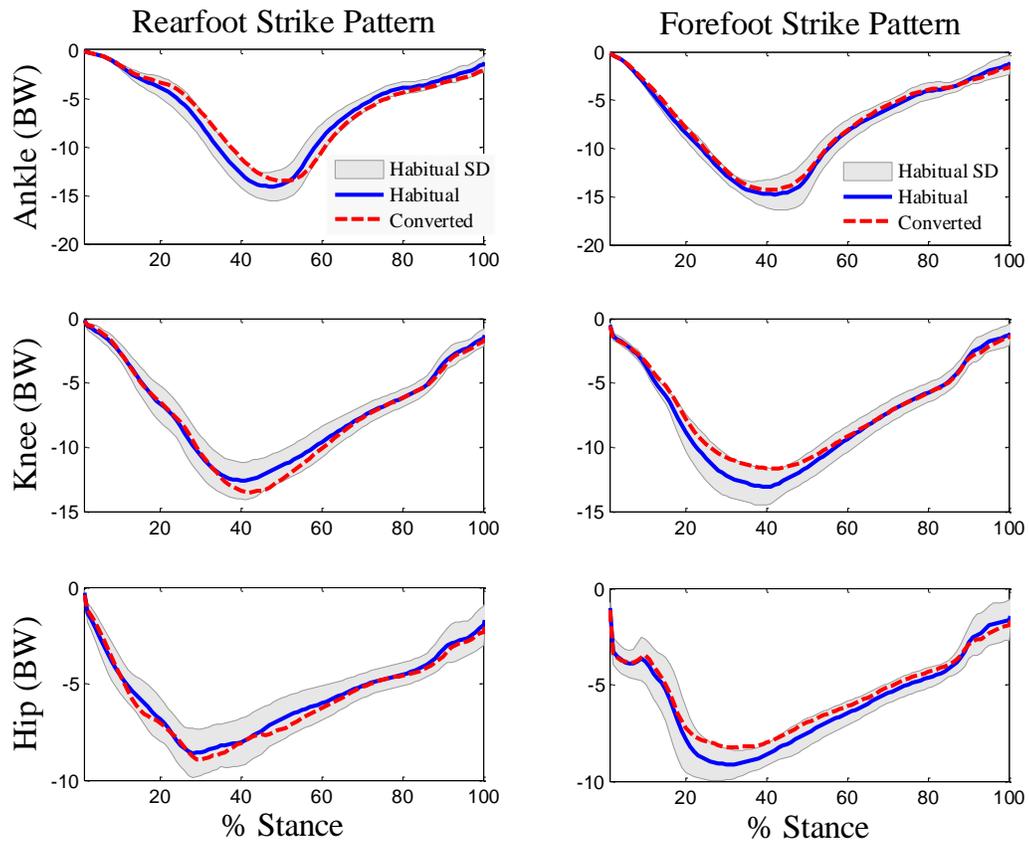
Figure 3: Comparison of axial contact forces at ankle, knee, and hip between habitual and converted strike patterns. Negative values represent compression. Note the difference in scaling between joints.
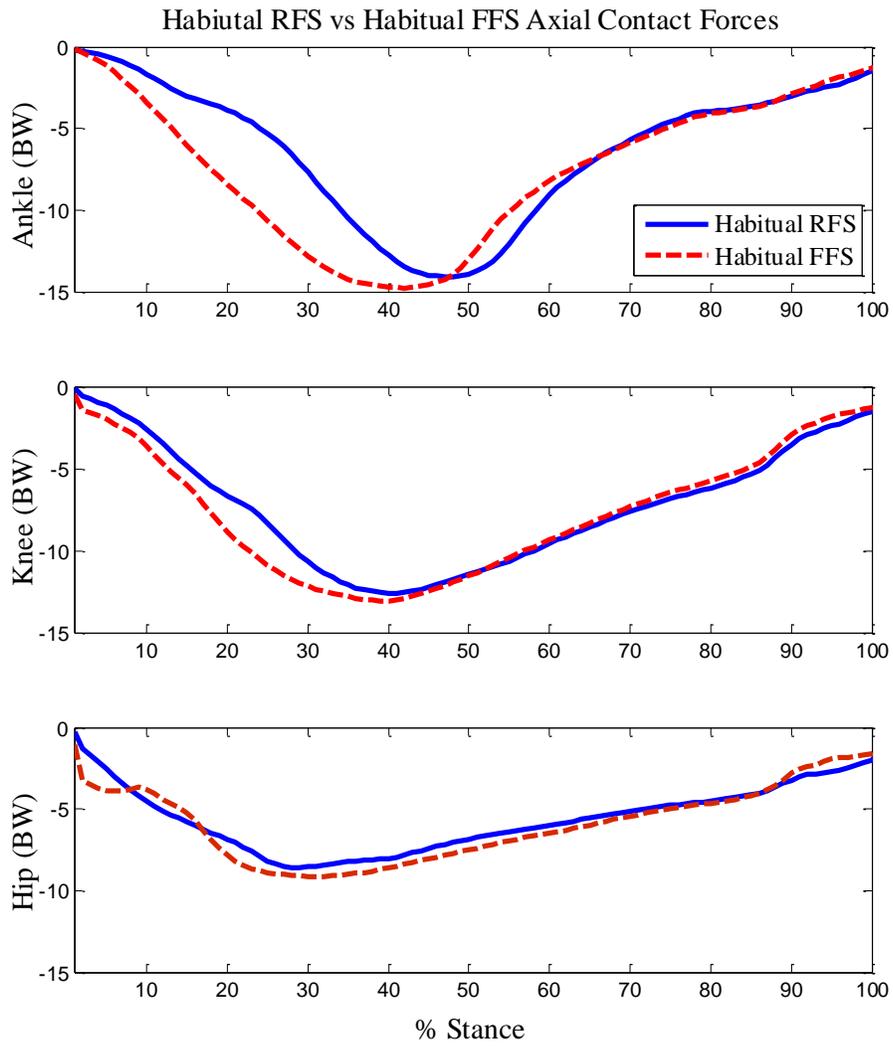
Figure 4: Comparison of the axial loading at the ankle, knee, and hip between habitual RFS and habitual FFS. Negative values represent compression.
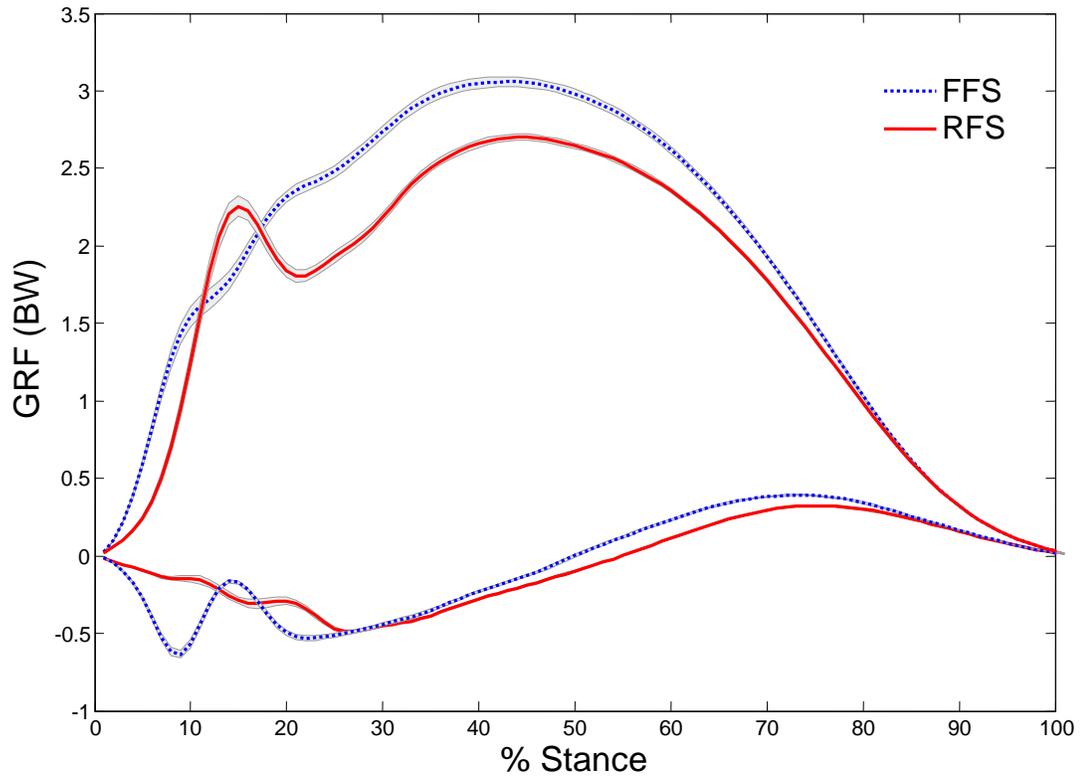
Figure 5: Ensemble curves for vertical and AP GRF during FFS and RFS running. The solid line represents RFS and the dotted represents FFS. Positive values represent forces in the vertical and anterior direction. (note: I need to add a legend on the graph and change it so that it is obvious what is RFS and FFS)