

Kinematics and kinetics of the lower limb in uphill and downhill running: A comparison of
forefoot strike and rearfoot strike runners

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Abstract

This study investigated the lower limb biomechanics during downhill and uphill running in habitual forefoot strike and habitual rearfoot strike runners. Fifteen habitual forefoot strike and fifteen habitual rearfoot strike recreational male runners ran at $3 \text{ m/s} \pm 5\%$ during level, uphill and downhill overground running on a ramp mounted at 6° and 9° . Results showed that hill running had similar impacts on joint angles in rearfoot strike and forefoot strike runners, causing a decrease in hip flexion at initial contact during downhill running, an increase in knee flexion angle at initial contact during uphill running and a decrease in peak hip flexion angle. In addition to differences in ankle joint angle due to landing pattern difference between rearfoot strike and forefoot strike runners, forefoot strike runners had a more flexed hip angle during downhill running. Forefoot strike runners had an absent impact peak in all running conditions, while the impact peaks only decreased during the uphill conditions in rearfoot strike runners. Active peaks decreased during the downhill conditions in forefoot strike runners while active loading rates increased during downhill conditions in rearfoot strike runners. Compared to the level condition, parallel braking peaks were larger during downhill conditions and parallel propulsive peaks were larger during uphill conditions. Peak hip flexion moment was significantly greater while peak knee flexion moment was significantly lower in both groups during the downhill 9° condition. Forefoot strike runners had larger peak plantar flexion moments and peak ankle power absorption compared to rearfoot strike runners during all conditions. Forefoot strike runners had decreased peak power absorption at the knee joint during downhill and level running conditions. Combined with previous biomechanics studies, our findings of no impact peak in forefoot strike runners suggests that this landing pattern may have potential in reducing overuse running injuries. Forefoot strike running reduces loading at the knee joint and can be used as an effective strategy to reduce stress at the knee joint experienced with rearfoot strike running.

Acknowledgements

The journey of my Masters has been one of the most remarkable experiences I have had the pleasure to accomplish, and as with most noteworthy accomplishments it was full of challenges and opportunities. At the beginning the task of choosing a research topic to invest the next two years was daunting. I am fortunate I was able to pursue an area I was passionate about, which made the long hours not even seem like work at times. I learned the importance of both managing my time with school, life and family, bridging theory and practice, but most importantly was the ability to delay gratification. The thesis is a long process, with constant work that sometimes goes unnoticed, so it is important to be able to celebrate the small accomplishments as they are all part of something bigger.

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Chapter 1

Introduction

Three primary types of landing patterns exist, namely rearfoot strike (RFS), midfoot strike (MFS), and forefoot strike (FFS). RFS running is the most common and has been observed in approximately 75% of runners, whereas approximately 1.5% of runners FFS (Hasegawa, Yamauchi, & Kraemer, 2007). Landing pattern and running biomechanics have been well documented as related risk factors in running injuries (van Gent et al., 2007) and a recent systematic review reported that approximately 25% of runners are injured at any given time and up to 40% of runners sustain a running related injury each year (Altman & Davis, 2012a). Running injuries are the most common sports related injury (Hreljac, 2004). Research has found that the primary difference between injured and non-injured runners was the magnitude of impact peak (Hreljac, Marshall, & Hume, 2000). Impact peaks are present in runners who RFS, but absent in FFS runners during level running (Lieberman et al., 2010). This impact peak may place the RFS runner at a higher risk of injury due to higher loading rate which sends a shockwave that can be measured in the tibia within a few milliseconds and the head 10ms later (Altman & Davis, 2012a; Cavanagh & LaFortune, 1980; Lieberman et al., 2010). To avoid running injury, a number of runners are attempting to switch running landing pattern (Rothschild, 2012). Preliminary findings suggest that switching landing pattern from a RFS to either a MFS or FFS can reduce stress at the knee joint (Vannatta & Kernozek, 2015).

FFS runners land with the ankle in a plantar flexed position and have greater knee flexion at initial contact compared to RFS runners who land with a dorsiflexed ankle (Lieberman et al., 2010; Shih, Lin, & Shiang, 2013; D. S. Williams, Green, & Wurzinger, 2012). FFS runners have a more compliant ankle joint compared to RFS runners, and this difference in ankle stiffness is thought to explain why an impact peak is present in RFS runners and absent in FFS runners

(Hamill, Gruber, & Derrick, 2014; Laughton, Davis, & Hamill, 2003). Current knowledge of FFS running biomechanics is limited to level running conditions (Daoud et al., 2012; Lieberman et al., 2010; Lieberman, 2012; Shih et al., 2013; D. S. Williams, McClay, & Manal, 2000; D. S. Williams et al., 2012). Uphill and downhill surfaces are commonly found in outdoor environments, and may impede running and present a risk factor for increased lower body injuries (Gehlsen, Stewart, Van Nelson, & Bratz, 1989). Running downhill increases impact peaks, so downhill running may increase the risk of overuse running injuries (Gottschall & Kram, 2005; Hreljac et al., 2000).

When compared with level running, research has found that running downhill at 9° increased impact peaks 54% and increased parallel braking force peaks 73%, while uphill running at 9° had an absent impact peak and 75% larger parallel propulsive force peaks (Gottschall & Kram, 2005). Downhill and uphill running at 6° also changed ground reaction force values, but to a lesser extent than the 9° conditions (Gottschall & Kram, 2005). However, uphill and downhill running have insignificant changes to peak joint moments when compared to level running (Buczek & Cavanagh, 1990; DeVita, Janshen, Rider, Solnik, & Hortobagyi, 2008; Telhan et al., 2010). The published studies did not mention the landing pattern of their participants and it seems these findings were from RFS runners. Since FFS running differs from RFS running during level conditions (Lieberman et al., 2010; Shih et al., 2013; D. S. Williams et al., 2012), only research in which FFS and RFS landing patterns are controlled during uphill and downhill running conditions can provide understanding of the impact of hill running on lower limb biomechanics.

A number of researchers have studied running biomechanics under uphill and downhill conditions, however most of these were done on a treadmill (Abe et al., 2011; Cai et al., 2010;

Gottschall & Kram, 2005; Hannon, Rasmussen, & Derosa, 1985; Ho et al., 2010; Padulo, Powell, Milia, & Ardigo, 2013; Telhan et al., 2010), with only a few studies which examined overground slope running (Buczek & Cavanagh, 1990; DeVita et al., 2008; Yokozawa, Fujii, & Ae, 2007). Although treadmill running has been validated for steady state running (Riley et al., 2008), evidence exists which suggests that treadmill and overground running are not identical (Elliott & Blanksby, 1976; Lavcanska, Taylor, & Schache, 2005; B. M. Nigg, Deboer, & Fisher, 1995; Riley et al., 2008; A. Schache et al., 2001; K. R. Williams, 1985). Results were mixed between researchers, but differences included decreased stride length, increased cadence, flatter foot at landing, and decreased peak values of GRF, joint moments, and joint powers associated with treadmill running (Elliott & Blanksby, 1976; B. M. Nigg et al., 1995; Riley et al., 2008). Therefore, limitations exist when generalizing findings from sloped treadmill running to sloped overground running.

To summarize the gaps in the literature, little research has been conducted on running with a FFS. Knowledge of overground hill running is limited, with the majority of research being conducted on an instrumented treadmill. Research is needed which examines both RFS and FFS runners when running on overground hill conditions. Selected kinematic and kinetic variables are defined and explained in Table 1. This will allow us to see if FFS runners still have distinct kinematic differences compared to RFS runners, if they have an absent impact transient, and if other differences in joint kinetics exist between the two groups. Current research between both groups is limited to level conditions, it is unknown if the known differences will remain when applied to hill conditions.

Variables

Independent Variables

The independent variables are landing pattern and slope conditions. The two different landing patterns that were used were rearfoot strike (RFS) and forefoot strike (FFS). Slope conditions used were downhill 9°, downhill 6°, level (0°), uphill 6°, and uphill 9°.

Dependent Variables

a) Kinematic variables

The kinematic parameters included hip, knee and ankle joint angles in the sagittal plane. A mean value at foot strike and peak angle for each joint angle were taken and compared between the conditions and groups.

b) Kinetic variables

Kinetic parameters included ground reaction forces (GRF) in the vertical and antero-posterior directions. From these variables, impact peak, active peak, braking force peak, propulsive force peak were defined and normalized to body weight (BW). Impact loading rate and active loading rate were calculated and normalized to body weight over time (BW/s). Braking impulse and propulsive impulse were integrated through the area under the braking and propulsive components of the parallel GRF, respectively, and were not normalized to body weight (N·s).

Through inverse dynamics, peak sagittal moment of force (Nm/kg), peak joint power absorption (W/kg) and peak joint power generation (W/kg) of the hip, knee and ankle joints were

calculated and normalized to body weight. Total joint power absorption and generation were integrated for the hip, knee and ankle joints and normalized to body weight (W/kg).

Table 1: Independent and dependent variables of interest. (BW = body weight)

Category	Description
<i>Independent Variables</i>	
Landing Pattern	Rearfoot strike (RFS)
	Forefoot strike (FFS)
	Downhill 9°
Slope Condition	Downhill 6°
	Level (0°)
	Uphill 6°
	Uphill 9°
<i>Dependent Variables</i>	
Kinematic parameters	Joint angle at foot strike (°)
	Peak joint angle (°)
Kinetic parameters	Impact force peak (BW)
	Active force peak (BW)
	Impact loading rate (BW/s)
	Active loading rate (BW/s)
	Braking force peak (BW)
	Propulsive force peak (BW)
	Braking impulse (N·s)
	Propulsive impulse (N·s)
	Peak moment of force (Nm/kg)
	Peak joint power absorption (W/kg)
	Peak joint power generation (W/kg)
	Total joint power absorption (W/kg)
	Total joint power generation (W/kg)

Purpose and Research Question

The purpose of this study is to examine the differences in kinematics and kinetics of the lower limb in RFS and FFS runners during level, uphill and downhill overground running conditions. The research question is to see if the differences in kinematics and differences

between RFS and FFS runners remain when applied to overground uphill and downhill conditions. To answer this question, the study will examine the kinematics and kinetics of the lower limb in habitual RFS and habitual FFS runners during level, uphill and downhill conditions.

Hypotheses

When comparing our test slope conditions ($\pm 6^\circ$ and $\pm 9^\circ$) to the level condition independent of foot strike we hypothesize the following:

- i. Knee flexion angle will decrease during the downhill condition and increase during the uphill condition compared to the level condition
- ii. Time to peak GRF, GRF, peak moments and peak power absorptions will increase during downhill conditions and decrease during the uphill conditions compared to the level condition

When comparing FFS and RFS runners at each slope condition we hypothesize the following:

- iii. FFS runners will have an absent impact transient and reduced time to peak GRF compared to RFS runners
- iv. FFS runners will have increased plantar flexion moments and ankle power absorption, and decreased power absorption at the hip and knee joints compared to RFS runners

Relevance

This study is significant because, to our knowledge, it is the first of its kind to measure the kinematics and kinetics of uphill and downhill running with consideration of landing patterns. The findings from the study can provide a better understanding of running

biomechanics, which is currently limited to mainly level running and uphill treadmill running. It will also add to understanding FFS running biomechanics for level, uphill and downhill running. The study findings will shed light on the prevention of running-related injuries, and provide information to runners and clinicians to help them choose appropriate landing strategies that will help reduce injury.

Limitations and Assumptions

Some of the limitations is that since we are not collecting data on a treadmill it will not be possible to have a steady running speed for all participants, let alone constant running speed between trials for the same participant. To control this as best as possible we used photocells to monitor speed and keep it within a 5% threshold for all participants. Midfoot strike runners were not selected for testing as their foot strike is highly variable and is characteristic of both a FFS and RFS landing pattern (Lieberman, 2012), therefore only the biomechanical differences between FFS and RFS runners were examined.

Chapter 2

Literature Review

Not every individual runs the same way, so to categorize the different ways individuals run, landing pattern at initial contact is used. Three primary foot strike patterns exist, namely, RFS, MFS, and FFS, with RFS being the most common seen in approximately 75% of runners and FFS being the least common, seen in approximately 1.5% of runners (Hasegawa et al., 2007). The type of landing pattern used by a runner is recognized as one of the major factors affecting the biomechanics of running (van Gent et al., 2007). The review of literature will examine an overview of running biomechanics, the effect of running landing pattern, and the effect of running surfaces. From this review, gaps and methodological flaws will be identified, justifying the need for new research. A list of definitions of the different landing patterns are provided below to elucidate the literature review.

Definitions of Landing Patterns

Rearfoot Strike (RFS): landing pattern characterized with the runner landing initially with the heel before the ball of their foot makes contact with the ground (Lieberman, 2012), foot strike angle (FSA) is greater than 8° dorsiflexion (Altman & Davis, 2012b) and strike index has the centre of pressure (COP) located in the posterior third of the foot (Cavanagh & Lafortune, 1980).

Midfoot Strike (MFS): landing pattern characterized with the runner landing simultaneously on the heel and ball of their foot (Lieberman, 2012), FSA is between 8° dorsiflexion and 1.6° plantar flexion (Altman & Davis, 2012b) and strike index has COP located in the middle third of the foot (Cavanagh & Lafortune, 1980).

Forefoot Strike (FFS): landing pattern characterized with the runner landing initially with the ball of their foot before the heel makes contact with the ground (Lieberman, 2012), FSA is greater than 1.6° plantar flexion (Altman & Davis, 2012b) and strike index has COP located in the anterior third of the foot (Cavanagh & LaFortune, 1980)

Running Gait Cycle

The running gait is illustrated in Figure 1, and it is comprised of two basic periods, stance and swing. Stance phase is approximately 40% of the gait cycle and begins when the foot contacts the ground and ends at toe off. Swing phase is approximately 60% of the gait cycle and begins at toe off and ends at initial contact (De Wit, De Clercq, & Aerts, 2000).

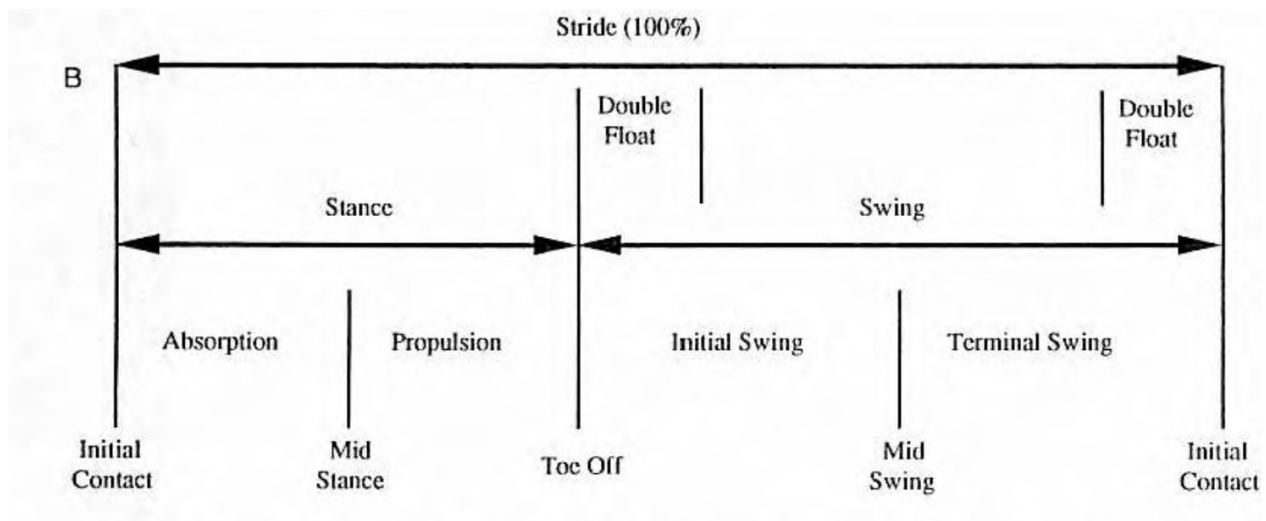


Figure 1: The running gait cycle including stance and swing phase (Ounpuu, 1994).

Stance phase is divided into 3 subphases: initial contact, midstance, and toe off. From initial contact to midstance, the lower extremity actively decelerates the forward-swinging leg and passively absorbs the shock of the ground reaction force. In midstance, the foot makes full contact with the ground and the body weight begins to shift towards the forefoot. From midstance to toe-off, the muscles of the lower extremity lengthen with concentric contraction of

the hip and knee extensors, preparing the foot for the propulsive push-off. Swing phase can be divided into 3 subphases: initial swing, midswing, & terminal swing. During initial swing and midswing, the foot advances forward in the air and in terminal swing the foot positions itself for foot strike and weight acceptance (Ounpuu, 1994).

Running gait is characterised by single-leg support and double-leg float periods. This is different from walking, which has one foot always in contact with the ground. The impact landing of one foot from an unsupported position when running at 4.5m/s results in the transmission of forces greater than 2.5 times the bodyweight throughout the lower limb (Cavanagh & LaFortune, 1980). The lower extremity must control and absorb these impact forces efficiently to avoid potential injuries.

The leg during the stance phase in running gait has been described as a spring-mass system in which the joints of the lower extremity lower the center of mass and absorb energy much like a spring compresses. Energy absorption is quickly followed by energy generation as the limb moves into extension, similar to the recoil of a spring, causing propulsion during the toe-off phase (Farley, Glasheen, & McMahon, 1993; Farley & Gonzalez, 1996). The longitudinal arch of the foot has been described as an “impact dampening structure” during the stance phase. With each foot strike, the lower limb endures significant impact force to the musculoskeletal structures. The impact at landing is created through collision of the shoe, foot, and lower leg mass. Landing pattern and cadence also affect the impact imposed on the lower extremity at landing (Heiderscheit, Chumanov, Michalski, Wille, & Ryan, 2011; Lieberman et al., 2010).

Overview of Running Biomechanics

The earliest running biomechanics was completed in 1927 when researchers examined sprint running (Furusawa, Hill, & Parkinson, 1927). Since then, research has also examined the

kinematics and kinetics of distance running, but the majority of these were done on RFS runners. The following overview will be exclusively based on RFS distance running.

Kinematics of RFS Running

Joint kinematics refers to spatial movement between body segments such as joint angular motion ($^{\circ}$). In running, the majority of motion is observed in the sagittal, allowing for the motion of flexion and extension. Some joint motion occurs in the frontal and transverse planes, but there is much less range of motion (ROM) in these planes compared to the sagittal plane. As velocity increases, ROM increases in all three planes (Mann & Hagy, 1980; Mann, 1981). The remainder of the kinematics section will report ROM values as average peak values (Ounpuu, 1994).

At initial contact the hip and knee flexion angles are 46° and 21° , respectively and the ankle is dorsiflexed 25° (Kulmala, Avela, Pasanen, & Parkkari, 2013). In the sagittal plane, total ROM of the hip, knee and ankle joints are 46° , 63° and 50° , respectively. In the frontal plane, the greatest ROM occurs at the hip joint with abduction and adduction occurring to produce a total ROM of 14° . In the transverse plane, total ROM for the hip and ankle joints are 16° and 14° , respectively (Mann & Hagy, 1980; Ounpuu, 1994).

Kinetics of RFS Running

Kinetics refers to the forces that cause motion. Ground reaction forces, joint moments, and joint powers are of the kinetics study interest. Joint kinetics is a study branch in biomechanics and its variables are calculated based on anthropometric measurements, kinematic positional data and ground reaction forces (GRF) through the inverse dynamics model. Inverse dynamics calculates kinetic variables indirectly from kinematics and inertial properties of moving bodies (Whittlesey & Robertson, 2004).

Ground reaction forces are the forces which are exerted by the ground during stance phase and are measured using a force platform. Figure 2 illustrates the vertical ground reaction force in RFS runners. The vertical GRF begins with an impact force peak (transient) in the first 20% of stance phase followed by an active peak at midstance (Lieberman et al., 2010). The vertical GRF range from 1.5 – 5x the magnitude of body weight, depending on the running speed (Lieberman et al., 2010). The parallel component of the GRF is initially negative after initial contact as a braking force is applied, reaching a peak of ~ -0.3 BW approximately a quarter into the stance phase before increasing and reaching zero at mid-stance. After mid-stance the parallel GRF becomes positive as a propulsive force is applied, reaching a peak of ~ 0.3 BW approximately three-quarters into stance phase before returning to zero prior to toe-off (Gottschall & Kram, 2005).

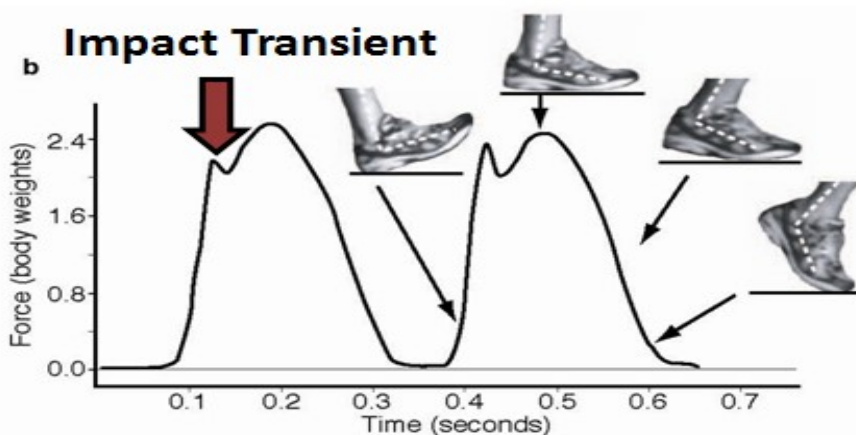


Figure 2: Vertical ground reaction force when running shod RFS at 3.5 m/s. (Lieberman et al., 2010)

Moments are defined as the product of a force and the distance to line of action of the force from the center of rotation. To apply this to the human body, the force is produced by muscles that are acting at a distance from the center of rotation of the joint. Moment values are presented as a net value; the net joint moment; which incorporates the effect of the agonist &

antagonist muscle groups and indicates the dominant muscle group. The effect of net joint moments is to cause a tendency for joint rotation. Figure 3 illustrates the joint moments in all three planes for the hip, knee and ankle joints during RFS running.

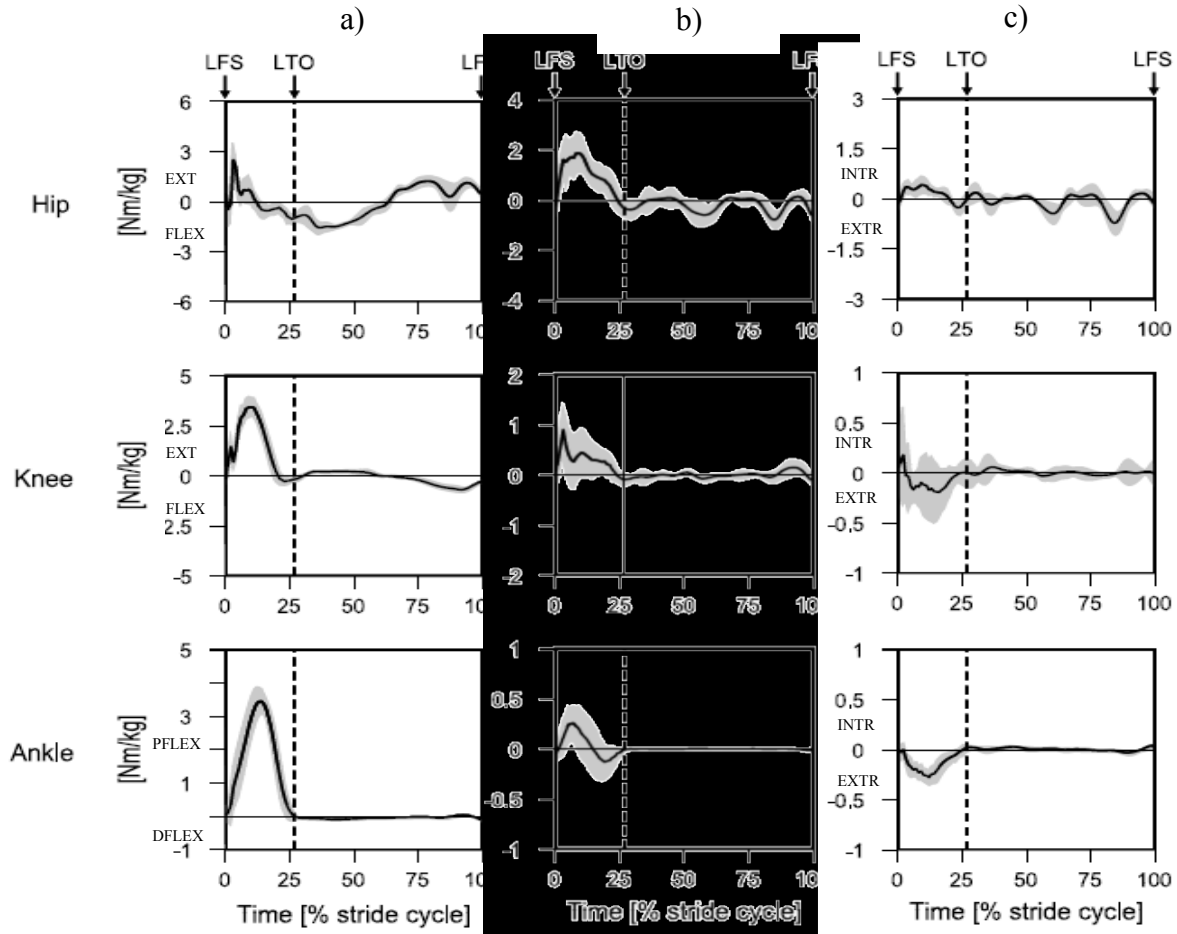


Figure 3: Joint moments in the hip, knee and ankle joints during a full gait cycle. a) Sagittal-plane torques, b) Frontal-plane torques, & c) Transverse-plane torques. Data represent the group means ($n=8$) (solid black line) $\pm$ one SD (gray shading). The dashed vertical line indicates the average time (% stride cycle) of toe-off. LFS: left foot-strike; LTO: left toe-off; EXT: extension; FLEX: flexion; PLEX: plantar flexion; DFLEX: dorsiflexion; ABD: abduction; ADD: adduction; INV: inversion; EV: eversion; EXTR: external rotation; INTR: internal rotation. (A. G. Schache et al., 2011)

During stance phase, joint moments of the lower extremity show a primarily extensor pattern with the relative timing of these extensor peaks occurring at different times for each joint (Winter, 1983). The hip peaks at 20%, the knee at 40% and the ankle near 60% stance phase

(Winter, 1983). The hip moment reverses from an extensor moment to a flexor moment before midstance to decelerate the backward rotating thigh's direction to push it forward into swing. After initial contact the knee flexes under the influence of weight bearing until midstance, and this flexion is stopped due to the peak knee extensor moment. Finally, the ankle joint develops a large plantar flexion moment shortly after midstance when the ankle is dorsiflexed with the foot flat on the ground and the leg rotates over it (Winter, 1983). During stance phase the peak extension moment of the hip and knee were 2.11 and $3.54 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$, respectively, and the ankle had a peak plantar flexion moment of $2.54 \text{ N}\cdot\text{m}\cdot\text{kg}^{-1}$ (Kulmala et al., 2013).

Joint powers are produced from the net joint moments determined through inverse dynamics. These moments represent the muscular response to moments applied to skeletal segments from the external forces which include ground and joint reaction forces, segmental weights and inertial torques (Alexander, 1991; Elftman, 1939; Roberts & Belliveau, 2005). It represents the rate of doing work and is described as power generation or power absorption. This is related to the type of muscle contraction, with a net power absorption occurring during an eccentric (lengthening under tension) muscular contraction while net power generation occurs during a concentric (shortening under tension) muscular contraction. A muscle will have a net power generation when the contraction produced is in the same direction as the pull. Figure 4 compares the power absorption and generation at the hip, knee and ankle during running, and illustrates the consistent reversal of power between these joints. During initial swing, the knee absorbs energy during flexion while the hip generates energy during flexion. These motions are both controlled by the rectus femoris muscle which crosses two joints, allowing to contract concentrically at the knee while the hip generates power during hip extension. This hip extension

is also controlled by a double joint muscle group; the hamstrings; that contract eccentrically at the knee and concentrically at the hip which allows efficient energy transfer between joints.

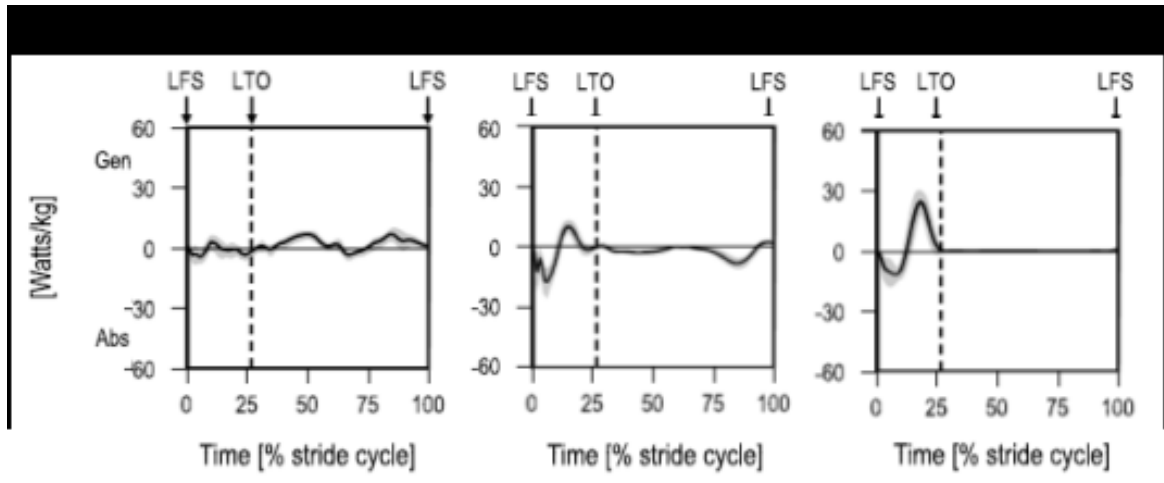


Figure 4: Net powers developed about the hip, knee, and ankle joints across the full gait cycle. Data represent the group ($n=8$) mean (solid black line) $\pm$ one SD (gray shading). The dashed vertical line indicates the average time (% gait cycle) of toe-off for each speed condition. LFS: left foot-strike; LTO: left toe-off; Abs: absorption; Gen: generation. (A. G. Schache et al., 2011)

When running at a speed of 3.50 m/s the hip, knee and ankle joints have a peak power absorptions of 2.15, 15.69 and 7.77 W/kg, respectively. Whereas peak power generation for the hip, knee and ankle joints when running at the same speed were 3.80, 7.72 and 16.09 W/kg, respectively (A. G. Schache et al., 2011). RFS runners absorb the majority of their power through the knee joint and generate most of their power through the ankle joint.

In the sagittal plane during absorption and propulsion phase of RFS, the hip continuously extends by concentric contraction of the hip extensors (semitendinosus, semimembranosus, biceps femoris & gluteus maximus), followed by the eccentric contraction of the hip flexors (rectus femoris, psoas major, iliacus & sartorius). In the initial swing phase, the hip flexes as a result of the concentric contraction of the hip flexors. In the terminal swing, the hip stops flexion and extends from the concentric contraction of the hip extensors. In the coronal plane, absorption is

followed by generation as the hip abductors concentrically contract to contribute to weight acceptance and to control the position of the pelvis. The hip continues to abduct in initial swing and adduct in terminal swing which is aided by the movement of the pelvis which is controlled the stance limb hip abductors (Ounpuu, 1994).

During absorption phase the knee flexes under the eccentric control of the knee extensors (rectus femoris, vastus lateralis, vastus intermedius & vastus medialis) to control the height of the body center of gravity. In the propulsion phase, the knee extensors concentrically contract. In initial swing, the rectus femoris eccentrically contracts which forms a small net knee extensor moment that controls excessive knee flexion. In terminal swing, inertia causes the knee to rapidly extend and is then slowed later in terminal swing through eccentric contraction of the knee flexors (semitendinosus, semimembranous, biceps femoris, gracilis & sartorius) (Ounpuu, 1994).

During the initial part of stance phase in RFS runners there is a small dorsiflexor moment which is followed by a net plantar moments for the absorption phase. During the propulsion phase, the ankle plantar flexors (soleus, medial & lateral gastrocnemius) contract concentrically to cause plantar flexion. This helps propel the stance limb into swing and double float. During initial swing, the ankle dorsiflexes from the concentric contraction of the ankle dorsiflexors (tibialis anterior) (Ounpuu, 1994).

Running Landing Pattern

Landing patterns vary amongst runners, but there are generally three landing patterns, namely, RFS, MFS, and FFS. A RFS landing pattern was observed in approximately 75% of runners, whereas a FFS landing pattern was observed in 1.4% of 415 elite runners (362 men and 53 women) at the 15 km point in a half marathon (21.1 km) race (Hasegawa et al., 2007). A MFS landing pattern generates a wide range of impact peaks which vary between what would be

observed in a RFS and other times resemble that of a FFS runners (Lieberman, 2012). For the purpose of this thesis only RFS and FFS runners will be observed. FFS running has only recently began to receive for research attention so only a few studies exist. Differences between RFS and FFS runners exist primarily at initial contact, but these initial differences cause changes in the biomechanics through the entire lower extremity.

Kinematic Aspect

The most noticeable differences between RFS and FFS runners is that RFS runners land with their ankle in a dorsiflexed position, whereas FFS runners land with their ankle in a plantar flexed position (Shih et al., 2013; D. S. Williams et al., 2012). Williams and colleagues (2012) compared 10 male and 10 female recreational runners who ran with a FFS and RFS landing pattern at $3.35 \pm 5\%$ m/s during level overground running. They reported that FFS group had a plantar flexion angle of $12.46 \pm 6.67^\circ$ at initial contact compared to the RFS group who had a dorsiflexion angle of $14.85 \pm 6.15^\circ$, however they did not find any significant differences at the hip or knee joint between the two groups (D. S. Williams et al., 2012). Shih and colleagues compared 12 male runners who were habitual RFS runners and had them run at 2.5 m/s on a treadmill with a RFS and FFS landing pattern. When their participants ran with a FFS they had a plantar flexion angle of $10.05 \pm 3.66^\circ$ at initial contact compared to a dorsiflexion angle of $10.80 \pm 3.78^\circ$ at initial contact when they ran with a RFS. They also reported that running with a FFS decreased hip flexion 6% and increased knee flexion by 66% at initial contact compared to when they ran with a RFS (Shih et al., 2013).

Neither group in the aforementioned studies used habitual FFS runners as their participants. Williams and colleagues included 12 MFS and 8 FFS runners in their FFS group and Shih et al. used 12 habitual RFS runners and asked them to run with a FFS (Shih et al., 2013;

D. S. Williams et al., 2012). Including MFS runners in the FFS group would decrease the average plantar flexion angle of the group since MFS runners land with a flatter foot from the ball of the foot and heel contacting the ground simultaneously (Lieberman, 2012). Also, ability and state of training is one of the characteristics that can effect biomechanical measurements, so if a group is not used to running with a FFS, they will not perform as well as habitual FFS runners (K. R. Williams, Snow, & Jones, 1989). Finally, both of these studies only examined sagittal plane kinematics so it is necessary to compare the kinematics in the frontal and transverse planes as well.

Kinetic Aspect

Several studies examined the kinetic aspects of RFS and FFS running. The study by Kulmala and colleagues examined 38 female team sport athletes (19 RFS and 19 FFS) as they ran overground at 4.0 m/s (Kulmala et al., 2013). This study classified their runners based on foot strike angle (FSA); RFS if $FSA > 8^\circ$ and FFS in $FSA < 8^\circ$. However, the original work by Altman and Davis (2012) classifies FFS runners with a $FSA < -1.6^\circ$, resulting in some of FFS runners in Kulmala and colleagues research to actually be MFS runners (Altman & Davis, 2012b; Kulmala et al., 2013). This would lead to some inaccuracies in their FFS findings (K. R. Williams et al., 1989).

When runners land with a FFS pattern they have an absent impact peak, illustrated in figure 5 (Lieberman et al., 2010). The study by Kulmala et al. found that FFS runners still had an impact peak but it was reduced by 36% compared to the RFS group, but this could be due to some of their FFS participants utilizing a MFS. They also found a reduction of 47% in vertical loading rate compared to their RFS group, whereas the study by Shih and colleagues found a 30% decrease (Kulmala et al., 2013; Shih et al., 2013).

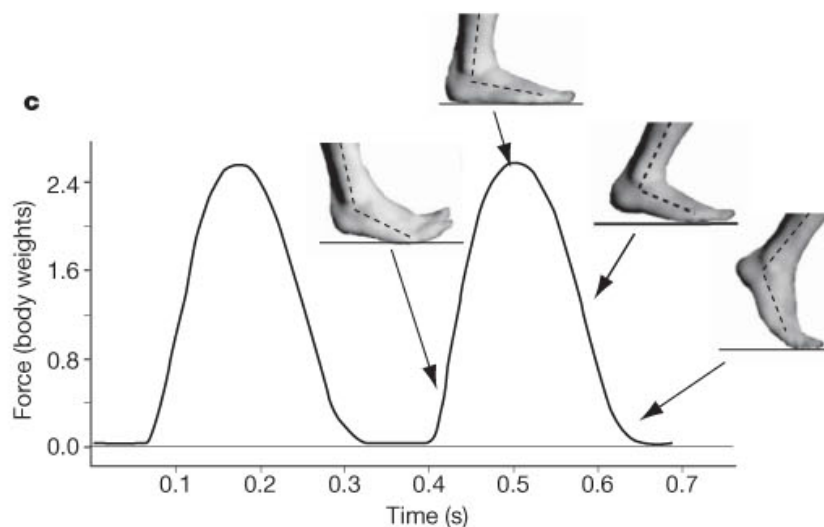


Figure 5: Vertical ground reaction forces when running barefoot with a FFS at 3.5 m/s (Lieberman et al, 2010).

The only study which measured joint moments between RFS and FFS runners found no significant differences between the two groups in peak hip flexion, extension or abduction moments. The FFS group had a 24% decrease in knee abduction moment compared to the RFS group, but no differences existed in knee extension moments. The FFS group had a 23% larger peak plantar flexion moment at the ankle compared to the RFS group (Kulmala et al., 2013).

The only study which measured joint powers between RFS and FFS runners found a 59% reduction in peak power absorption at the hip and a 54% reduction in peak power absorption at the knee in FFS runners compared to the RFS group. These reductions in hip and knee power absorptions was accompanied by a 68% increase in power absorption at the ankle joint. Total power absorption in the lower limb was reduced by 23% when running with a FFS compared to a RFS (D. S. Williams et al., 2012).

Running Surface Biomechanics

Running surfaces can cause changes to running biomechanics. Differences such as slope (uphill and downhill) and surface hardness (concrete, grass, and rubber) can affect the

biomechanics while running (Buczek & Cavanagh, 1990; Cai et al., 2010; Gottschall & Kram, 2005; Ho et al., 2010; Hong, Wang, Li, & Zhou, 2012). Research on running surfaces is extensive and the remainder of the review will focus on the effects of slope on running biomechanics.

A number of researchers have studied running biomechanics under uphill and downhill conditions, however the majority were examined while participants ran on an instrumented treadmill (Abe et al., 2011; Cai et al., 2010; Chumanov, Wall-Schefer, & Heiderscheit, 2008; Gottschall & Kram, 2005; Hannon et al., 1985; Hardin, Van den Bogert, & Hamill, 2004; Lussiana, Fabre, Hebert-Losier, & Mourot, 2013), with only a few researchers examining overground sloped running (Buczek & Cavanagh, 1990; DeVita et al., 2008; Yokozawa et al., 2007). Before overground and sloped treadmill running can be compared, a comparison on the two running surfaces during level running needs to be completed.

Overground versus treadmill running

Treadmills are often used during training, especially when the weather is not ideal for outdoor running. Treadmills are often used by clinicians and scientists to conduct running studies as they allow for a confined set up which is easy to control and reproduce conditions. Although treadmill running has been validated for steady state running (Riley et al., 2008), evidence exists that shows that treadmill and overground running are not identical (Elliott & Blanksby, 1976; Lavcanska et al., 2005; B. M. Nigg et al., 1995; Riley et al., 2008; A. Schache et al., 2001; K. R. Williams, 1985). Table 2 summarizes the main findings between overground and treadmill running and table 3 includes the participant and trial information in the aforementioned studies.

Distinct changes occur during treadmill running. When an individual runs on a treadmill they have reduced hip flexion, increased knee flexion and the foot lands in a flatter position,

Table 2: Summary of differences for overground and treadmill running studies. Findings are treadmill trials compared to overground trials.

Overground versus Treadmill Running		Reference
Temporospatial		
Stride Length	↗ 4.6% for level and 8.4% for uphill running at 6.40 m/s NC for level running between 3.33 to 4.80 m/s, and ↘ 3.1% in males, 10.2% in females for level running between 4.82 to 6.2 m/s ↘ 4.9% for level running at 3.9 m/s ↘ 4.0% for level running at 3.80 m/s	(Nelson, Dillman, Lagasse, & Bickett, 1972) (Elliott & Blanksby, 1976) (Schache et al., 2001) (Riley et al., 2008)
Stride Frequency	↘ 5.1% for level and 8.3% for uphill running at 6.40 m/s NC for level running between 3.33 to 4.80 m/s, and ↗ 3.5% in males, 10.9% in females for level running between 4.82 to 6.2 m/s	(Nelson et al., 1972) (Elliott & Blanksby, 1976) (Riley et al., 2008)
Stance Time	↗ 2.8% for level running at 3.80 m/s ↗ for level: 8% at 6.40 m/s; ↗ for uphill: 9.5% at 3.35 m/s, 11.7% at 4.88 m/s, 10.2% at 6.40 m/s; ↗ downhill: 6.3% at 4.88 m/s, 9.6% at 6.40 m/s NC for level running between 3.33 to 6.2 m/s	(Nelson et al., 1972) (Elliott & Blanksby, 1976) (Schache et al., 2001)
Swing Time	↗ 8.1% for level running at 3.9 m/s ↘ For uphill: 25.6% at 3.35 m/s, 11.6% at 4.88 m/s NC for level running between 3.33 to 4.80 m/s, and ↘ 7.3% in males, 18.5% in females for level running between 4.82 to 6.2 m/s ↘ 3.4% for level running at 3.9 m/s	(Nelson et al., 1972) (Elliott & Blanksby, 1976) (Schache et al., 2001)
Kinematics		
Joint Angles	↗ Lumbar extension 36.8% at IC, ↘ anterior pelvic tilt 13.7% at IC, ↗ max anterior pelvic tilt 24.1%, ↘ hip flexion 13.3% at IC, ↗ hip extension 50.8% at TO, ↗ max hip extension 15.2% ↘ Max knee flexion 6.0%, ↗ min knee flexion 22.3% Foot in flatter position at IC	(Schache et al., 2001) (Riley et al., 2008) (Nigg et al., 1995)
ROM	↗ Hip flexion-extension 2.6%	(Schache et al., 2001)
Kinetics		

GRF	↘ 5.3% for level running at 3.80 m/s	(Riley et al., 2008)
Moments	↘ Knee flexion moment 27.0%, ↘ knee varus moment 18.9%, ↗ ankle plantar flexion moment 16.6%	(Riley et al., 2008)
Power Absorption	↘ Hip power absorption 35.6%, ↘ knee power absorption 21.9%, ↗ ankle power absorption 29.8%	(Riley et al., 2008)

NC: no significant change; IC: initial contact; TO: toe off; ROM: range of motion; GRF: ground reaction force. All findings represent significant findings of $p < 0.05$

Table 3: Participant and study information for findings of overground versus treadmill running studies. Standard deviations are in parentheses when provided.

Reference	Participant #	Speed (m/s)	Age (years)	Height (m)	Mass (kg)
(Nelson et al., 1972)	16 males	3.35, 4.88 and 6.40 m/s Slopes: -5.71°, 0, +5.71°	19.6	1.75	70.3
(Elliott & Blanksby, 1976)	12 females	3.33 to 6.2 m/s	21.7	1.67	N/A
	12 males		26.9	1.76	N/A
(Nigg, Deboer, & Fisher, 1995)	N/A	3.0, 4.5, 5.0 and 6.0 m/s	N/A	N/A	N/A
(Schache et al., 2001)	9 males and 1 female	3.9 m/s	28.3 (4.7)	1.78 (0.05)	66.4 (11.2)
(Riley et al., 2008)	10 males and 10 females	3.80 m/s	25.2 (4.6)	1.75 (0.08)	66.4 (11.2)

ultimately reducing the stride length (B. M. Nigg et al., 1995; Riley et al., 2008; A. Schache et al., 2001). In order to maintain the same running speed, an individual is required to run with increased stride frequency. This causes a reduction in peak GRF which effects kinetic variables. Knee flexion and varus moments are reduced while ankle plantar flexion moments increase. Power absorption is reduced at the hip and knee joints, but increases at the ankle joint (Riley et al., 2008).

All of those differences were observed when comparing treadmill and overground running during level conditions, however many researchers have examined uphill and downhill running using treadmills (Abe et al., 2011; Cai et al., 2010; Chumanov et al., 2008; Gottschall & Kram, 2005; Hannon et al., 1985; Hardin et al., 2004; Lussiana et al., 2013). Since most commercial treadmills for runners only allow for level and uphill running, as special wedges need to be constructed to set a treadmill to a negative slope (Gottschall & Kram, 2005), it is important that downhill running is measured on overground conditions. If differences exist between treadmill and overground running during level conditions, it should not be assumed that the results that researchers are finding during sloped treadmill running are identical to overground sloped running.

Uphill and downhill running

Many runners frequently train on uphill and downhill slopes to improve their aerobic ability. This is because physiological variables such as oxygen consumption, heart rate, and blood lactate concentration are greater during uphill running compared to level running, which implies that the mechanical load on the lower limbs would be greater because of increased muscle activity (Gregor & Costill, 1973; Pivarnik & Sherman, 1990; Staab, Agnew, & Siconolfi, 1992). Few biomechanical studies have been completed on overground uphill and downhill

surfaces as the majority were done on instrumented treadmills. Researchers reported the slope in either degrees or percentage grade and the equation to convert is below:

$$\text{Degrees} = \tan^{-1}(\text{slope percent}/100)$$

$$\text{Slope Percent} = \tan(\text{degrees}) \times 100$$

The majority of researchers used treadmill angles between -10° to 10° (Abe et al., 2011; Cai et al., 2010; Chumanov et al., 2008; Gottschall & Kram, 2005; Hannon et al., 1985; Lussiana et al., 2013), with only one researcher using a declination angle of -12° (Hardin et al., 2004). The running studies done on a ramp were at -4.74° (Buczek & Cavanagh, 1990), $+5.2^{\circ}$ (Yokozawa et al., 2007), and -10° to $+10^{\circ}$ (DeVita et al., 2008). Since most of the research is focused only on such a limited range of slope, biomechanical knowledge of slope running is limited. Therefore it is necessary to look at slopes outside this range. There is an increase in popularity in obstacle races such as Spartan Race® and Tough Mudder® where participants ascend and descend down ski slopes with up to a 25% grade. It is important that future research examines the biomechanics of running at greater slopes. Table 4 summarizes the main findings of slope running trials and table 5 includes participant and trial information for the aforementioned studies.

Temporospatial variables are relatively consistent. During uphill running, stride length decreases and consequently causes an increase in stride frequency (Gottschall & Kram, 2005; Hannon et al., 1985; Padulo et al., 2013). During downhill running, stride frequency decreases which can be interpreted as an increase in stride length (Hannon et al., 1985; Padulo et al., 2013). The findings by De Vita and colleagues (2008) disagree with previous findings as they found a decrease in stride length during downhill running and an increase during uphill running, therefore the gap needs to be filled which compares temporospatial variables between level, downhill and uphill conditions.

Table 4: Summary of findings for sloped running trials. Findings are compared to level trials unless noted otherwise.

Reference	Slope & speed	Measurement & Findings
(Hannon et al., 1985) N=8	-5.71° (-10%) +5.71° (+10%) 4.47 m/s	↘ 4.1% stride frequency ↗ 4.3% stride frequency
(Buczek & Cavanagh, 1990) N=7	-4.74° (-8.3%) 4.5 m/s	↗ 9.1% Peak knee flexion angle, ↘ 31.1% knee flexion angle at FS, ↗ 56.3% peak power absorption at ankle, NC for peak extension moments
(Gottschall & Kram, 2005) N=10	-3° -6° -9° +3° +6° +9° 3.0 m/s	↗ 18% GRF ↗ 32% GRF, ↗ 23% average loading rate ↗ 54% GRF, NC GRF, ↘ 22% GRF, ↘ 23% average loading rate ↗ 4% stride frequency, GRF and average loading rate not reported
(DeVita et al., 2008) N=13	-10° +10° 3.35 m/s	↘ 5% stride length vs uphill, ↗ 2.5x GRF (25N/kg) vs uphill, ↗ 5% stride length vs downhill, ↘ 2.5x GRF (10N/kg) vs downhill,
(Telhan et al., 2010) N=19	-4° +4° 3.13 m/s	NC in joint moments, , ↗ 24.1% GRF, ↗ 30.1% knee power absorption, ↗ 900% hip power absorption (-100% BW vs 10% BW in level) NC in joint moments, ↗ 1400% hip power generation, (140% BW vs 10% BW in level)
(Padulo et al., 2013) N=18	+1.15° (+2%) +4.00° (+7%) 4.17 m/s	↘ 1.81% stride length, ↗ 1.85% stride frequency ↘ 4.30% stride length, ↗ 4.46% stride frequency, ↗ 12.06% GRF

Table 5: Participant and study information for findings of sloped running trials. Standard deviation is in parentheses when provided.

Reference	Participant #	Speed (m/s)	Age (years)	Height (m)	Mass (kg)
(Hannon et al., 1985)	8 males	4.47	22.3	1.77	62.2
(Buczek & Cavanagh, 1990)	7 males	4.5	26.9 (3.2)	1.80 (0.42)	73.4 (6.7)
(Gottschall & Kram, 2005)	10 (5 males & 5 females)	3	30.4 (5.1)	1.72 (0.06)	62.6 (7.6)
(DeVita et al., 2008)	13 (8 males & 5 females)	3.35	22.3 (2.8)	1.74 (0.10)	69.9 (13.2)
(Telhan et al., 2010)	10 males	3.13	26.6 (5.9)	1.80	74.2
	9 females	3.13	23.9 (2.5)	(0.52)	(8.1)
				1.66 (0.62)	56.2 (4.8)
(Padulo et al., 2013)	18 males	4.17	33.0 (8.5)	1.71 (0.04)	62.6 (5.2)

Knee flexion angles at initial contact and peak knee flexion angles were the only kinematic variable which had significant differences between level and hill running. During uphill running knee flexion angle at initial contact decreased by 31% while peak knee flexion increased by 9% (Buczek & Cavanagh, 1990). The same study found no differences at the ankle joint. Kinematic differences are currently limited to uphill running at the knee and ankle joint, therefore research needs to be done to expand the literature to include the hip joint, downhill trials and expand findings to include all three planes of movement.

Findings for peak GRF are relatively consistent. During downhill running on slopes $\geq 3^\circ$ peak GRF increases and decreases during uphill running on slopes $\geq 3^\circ$ (DeVita et al., 2008; Gottschall & Kram, 2005; Telhan et al., 2010). One study reported a 12% increase in peak GRF during 4° uphill running (Padulo et al., 2013). Average loading rates increased 23% when running on a -6° slope and decreased 23% when running on a $+6^\circ$ slope (Gottschall & Kram, 2005). No significant differences existed in peak joint moments when running on slopes between $\pm 5^\circ$ (Buczek & Cavanagh, 1990; Telhan et al., 2010), therefore research needs to extend into

larger slopes to see if differences arise. There is an increase in power absorption during downhill running at all three lower limb joints, however no single study reported on all three joints at once (Buczek & Cavanagh, 1990; Telhan et al., 2010). It is necessary to include joint power data for all three joints during uphill and downhill conditions.

Chapter 3

Methods

Participants

A total of 30 male recreational distance runners were recruited at races in the Ottawa area and through posters placed around the university and running clubs in the Ottawa area (Table 5). All prospective runners had to be between the ages of 18 – 35 years of age, run a minimum of 15 km/week, and be free from musculoskeletal injuries 6 months prior to participation. Only male subjects were recruited as men and women are not identical in running gait (Ferber, Davis, & Williams, 2003), and the aim of the study was not to analyze sex differences. Potential participants came for a preliminary screening session where we analyzed their landing pattern via strike index (Buczek & Cavanagh, 1990) and had them fill out a running history questionnaire (*Appendix C*). Participants were given a pair of Merrell Vapor Glove minimalist running shoes with reflective markers placed on the heel, medial and lateral malleoli, and between the 2nd and 3rd metatarsal heads. They were instructed to run overground at $3 \pm 5\%$ m/s and strike an imbedded force plate 4m away from the starting line.

The participants whose foot centre of pressure at foot strike was located in the anterior third of the foot were placed into the forefoot strike group, and the participants whose foot center of pressure was located in the posterior third of the foot were placed into the rearfoot strike group. Runners who landed with a midfoot strike landing pattern were excluded from participation as their foot strike is highly variable and characteristic of both a FFS and RFS landing pattern (Lieberman, 2012). Two groups, FFS and RFS group, with each 15 runners were formed. Table 6 provides the anthropometric data of the participants. Written informed consent following the guidelines of the Health Sciences and Science Research Ethics Board at the University of Ottawa was given by all participants prior to any data collection.

Table 6: Anthropometric data of the participants (mean $\pm$ SD)

	FFS Runners (n = 15)	RFS Runners (n = 15)	Total (n = 30)
Age (year)	26.1 $\pm$ 4.9	26.8 $\pm$ 4.5	26.5 $\pm$ 4.7
Height (m)	1.77 $\pm$.06	1.79 $\pm$.06	1.78 $\pm$ 0.05
Mass (kg)	74.9 $\pm$ 10.9	78.7 $\pm$ 9.1	76.8 $\pm$ 10.0

Materials and Instrumentations

Running Path and Ramp

The level running path was 10m long with two embedded force plates approximately half way in the running path. The ramp, was 3.7m long with a 4.9m long platform at the top portion of the ramp (Figure 6). The slope of the ramp was set to $\pm 6^\circ$ and $\pm 9^\circ$ which allowed comparison of results to Gottschall and Kram's study which was an intensive study in treadmill running with the same degrees (Gottschall & Kram, 2005). A force platform was embedded in the ramp, and reinforced structures were be added below the ramp under the force plate to minimize vibrations.

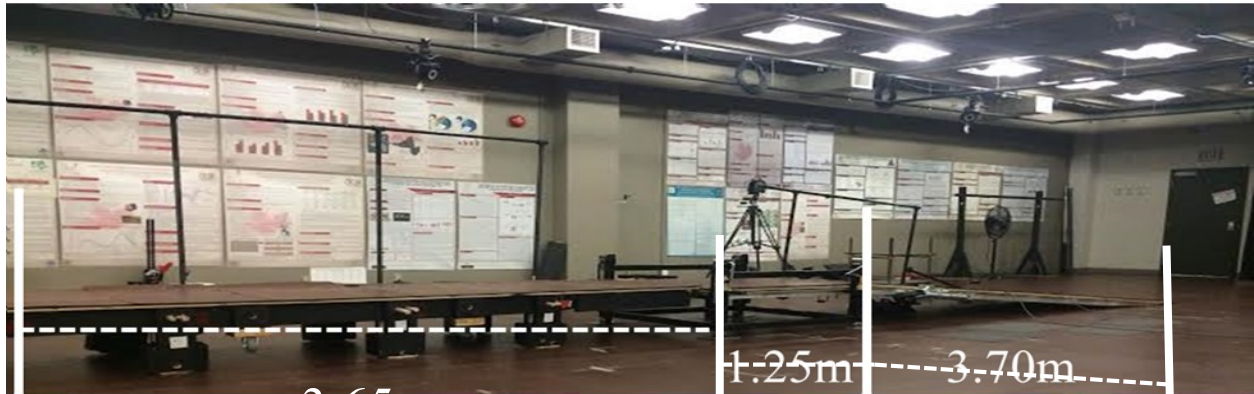


Figure 6: Image of ramp device.

Running Shoes

All participants ran in a standardized shoe, since allowing individuals to wear their own shoes would result in varying sole stiffness between participants, which would result in uncontrolled differences in kinematics and kinetics (B. M. Nigg, Baltich, Maurer, & Federolf, 2012; Wakeling, Pascual, & Nigg, 2002). The shoe used throughout testing was the Merrell Vapor Glove (Merrell, Michigan, USA), which was classified as a minimalist shoe and illustrated in figure 7. Subjects warmed up in the shoe prior to the first session and kept the shoes in between the two sessions to wear during their regular training to become accustomed to the shoes. We chose a minimalist shoe as our testing shoes since it had minimal cushioning in either the forefoot or heel of the shoe. A cushioned heel is designed to absorb some of the GRF experienced by running with a RFS (Lieberman et al., 2010; Willy & Davis, 2014), so to minimize the effects of the shoe on our results, we used a shoe with minimal cushioning in the heel.



Figure 7: Image of the Merrell Vapor Glove minimalist running shoe (<http://www.wiggle.co.uk/merrell-vapor-glove-shoes-aw13/>)

Motion Analysis System and Force Plate

Running movements were captured and analyzed using a 10 camera Vicon Motion Analysis System (MX-13, Oxford Metrics, Oxford, UK) recorded at 250Hz to obtain the kinematic data. Two force plates (model 9286AA, Kistler Instruments Corp, Winterthur, Switzerland) recorded at 1000Hz to collect GRF data was used for kinetic analysis of the lower extremity in running. One force platform was embedded in the level runway and one was embedded in the ramp. A system calibration of 10,000 frames was done prior to testing.

Protocol

Minimalist Shoe Testing

The right shoe of each Merrell Vapour Glove minimalist shoes were measured for the following variables: mass, stack height, and heel-to-toe drop (Table 7). Mass was measured using a precision balance with an accuracy of ± 2 g (Combiics 2, Sartprois, Mississauga, ON, Canada). For stack height and heel to toe drop a pair of skin for calipers with an accuracy of ± 2 mm (Harpenden Skinfold Caliper CE 0120, Baty International, West Sussex, UK). Stack height (mm) was measured at the centre of the heel and height at forefoot (mm) was measured at the

metatarsal heads. Heel to toe drop was the sub score value of the stack height (mm) subtracted by the height at the forefoot (mm).

Table 7: Results for the right shoe of the Minimalist Vapour Glove minimalist running shoes.

Size (US)	Mass (g)	Stack Height (mm)	Height at Forefoot (mm)	Heel to Toe Drop (mm)
8	148	5.4	6.2	-0.8
9	158	4.6	4.8	-0.2
9.5	158	4	4.6	-0.6
10	170	6	6.2	-0.2
10.5	174	5.6	5.8	-0.2
11	172	5.6	5.2	0.4
11.5	178	4.8	5.6	-0.8
Average	165 ± 11	5.1 ± 0.7	5.5 ± 0.6	-0.3 ± 0.4

Static and Dynamic Calibrations

Before beginning the dynamic calibration, the recording volume was masked to eliminate unwanted reflections which were picked up by the cameras. The dynamic calibration was done using a T-shape wand (240 mm) with three reflective markers. The T-shaped wand was waved around the area required to create a recording volume until 10000 frames were captured by the cameras of the three reflective markers.

Static calibration was performed using an L-shape frame (ErgoCal, 14 mm) placed at the centre of the capturing volume, which set the origins of the global coordinate system. Once calibrated, the global coordinate system defined the X, Y, and Z axis of the recording volume. The X-axis indicated the sagittal plane with movement from anterior-posterior direction, Y-axis indicated the frontal plane with movement in the medial and lateral direction, and the Z-axis indicated movement about the transverse plane.

Participant Preparation

During the initial assessment, a detailed description of the purpose of the study along with testing protocols and procedures were given to each participant. They were asked to come to the University of Ottawa Human Movement Biomechanics Laboratory, located at 200 Lees

Ave, for testing. The preliminary session lasted 0.5 hours and the testing session lasted 1.5 hours. Once in the lab, the participant was given a mini presentation to become accustomed to the testing protocol and procedures along with the testing environment. Written informed consent following the guidelines of the Health Sciences and Science Research Ethics Board at the University of Ottawa was given by all participants prior to starting data collection.

Participants were asked to remove any jewelry or any accessories to remove any false reflections could have been picked up by the cameras. They wore a tight spandex shorts and t-shirt (Under Armour), and Merrell Vapor Glove running shoes. Anthropometric measurements of height (mm), body mass (kg), leg length (mm), knee width (mm) and ankle width (mm) were recorded. A total of 45 reflective markers were placed at various anatomical positions according to the UOMAM marker set (*Appendix A & B*) (Lamontagne, Beaulieu, Varin, & Beaulieu, 2009). Anatomical landmarks for marker placement were located through palpation.

Landing Pattern Assessment

All recruited participants were asked to come to the University of Ottawa Human Movement Biomechanics Laboratory, located at 200 Lees Ave, for an initial assessment to classify their running strike pattern. Participants were given a pair of Merrell Vapor Glove minimalist running shoes with reflective markers placed on the heel, medial and lateral malleoli, and between the 2nd and 3rd metatarsal heads. They were instructed to run overground at $3 \pm 5\%$ m/s and strike an imbedded force plate 4m away from the starting line.

The participants whose foot centre of pressure at foot strike was located in the anterior third of the foot were placed into the forefoot strike group, and the participants whose foot center of pressure was located in the posterior third of the foot were placed into the rearfoot strike group (Cavanagh & LaFortune, 1980). Runners who landed with a midfoot strike landing pattern were

excluded from participation as their foot strike is highly variable and characteristic of both a FFS and RFS landing pattern (Lieberman, 2012).

Data Collection during Level and Hill Running

The main testing session lasted 1.5 hours and included the subject preparation, a warm-up, static trial, level trials, and uphill and downhill trials at 6° & 9°. We selected these angles to allow for comparison to a well cited study by Gottschall and Kram (2005) who did a study using the same speed on RFS participants while running on a treadmill. They only found significant findings during their slopes of 6° & 9° and not during their 3° condition, so we decided on only examining those conditions.

Warm-Up

Participants were given adequate warm-up time (approximately 5 to 15 minutes), until they felt comfortable in running on the ramp, before proceeding with data collection. During the warm-up, participants ran in the laboratory. Adequate rest period was given upon completing the warm-up. Prior to beginning the trials, three tests of leg dominance were used: step-up test, kicking leg, and balance-recovery test. The limb that is used in at least two of the tests was classified as the dominant leg.

Static Trial

Upon completing the warm-up phase, participants completed a static trial. The static trial was done by having the participants stand in the middle of the capture volume with their arms up in a T position and their thumbs pointing up. The participants were asked to stand still during the 5 second static trial.

Dynamic Trial

Following the static trial, participants completed the dynamic trials in a quasi-random order at a speed of $3.0 \text{ m/s} \pm 5\%$. Speed was monitored by using custom built photocells 3m apart (*Appendix D*). For the level trial, participants began the run 4m away from the force platform and struck it with their dominant foot. For uphill trials, participants began the run 4m away from the force platform and ran up the ramp before striking the force platform with their dominant foot. For downhill trials, participants began the run at the top of the running deck and continued running down the ramp before striking the force platform with their dominant foot. They were given as many practice runs as needed to get comfortable with striking the force plate naturally, without modifying their gait.

Participants were instructed to run, keeping the head up, and trying to run as the normally would. Ten trials of each running condition were collected. Only stance phase's data was collected since the ramp was not long enough to collect an entire gait cycle.

Only trials where the foot landed entirely on a force plate without altering their gait and the participant ran between 5% of 3 m/s was deemed as valid. A total of fifty dynamic trials were collected for each of the participants for the level and hill conditions. Participants were able to request a break at any time throughout the trials. The success of the trials was determined by the researcher.

Data Processing

Kinematic, Temporospatial and Kinetic Parameters

All trials were cropped and time-normalized to 100% stance phase and averaged through five trials for the dominant foot. Trials were labeled and processed using the Vicon Nexus Software (v1.8). Labeling was completed using the UOMAM v5_3 marker set to create the 3D

model. All gaps were filled for missing trajectories before the data will be filtered. Kinematic data was filtered using a Woltring 15 MSE filter and all analog data was filtered with a Butterworth 4th order, zero lag filter with a cut-off of 50Hz. Once all missing gaps were filled and filters are applied, a dynamic pipeline was used to process all the data which is then sent to Matlab (MathWorks, Natick, MA, USA) and Excel (Microsoft, Washington, USA) to retrieve biomechanical information from .c3d files and to examine the data.

Using the VICON Nexus software (v1.8), kinematic variables were normalized to 100% stance phase. Lower limb joint angles of the hip, knee and ankle were presented in the sagittal plane. Peak joint angle and joint angles at initial contact of the lower limb throughout stance phase were obtained. Initial contact and toe off were identified using force platform values. Initial contact was defined as the moment when the vertical GRF data first became greater than 20 N and toe off was defined after it fell below 20 N.

For each condition, we located impact force peaks, active force peaks, peak braking forces, and peak propulsive forces that are presented as times of body weight (BW). Active and impact loading rates (BW/s) were calculated from the vertical GRF and braking and propulsive impulses (N·s) were calculated from the parallel GRF. We determined the impact force peak by finding the highest value from foot strike to the positive slope became negative (within 25 ms after foot strike). Peak braking forces were determined by finding the minimum parallel GRF value and peak propulsive forces were determined by finding the maximum parallel GRF value. Impact loading rates were calculated by dividing the impact force peak by the time from foot strike to the impact force peak, whereas active loading rate equaled the active force peak divided by the time from foot strike to the active force peak. Parallel impulse data was obtained by

integrating all the negative (parallel braking) or positive (parallel propulsive) values of the parallel GRF during stance phase, yielding an average impulse during stance phase.

Through the Matlab program, we extracted peak sagittal joint moments (N·m/kg), peak joint power absorption and generation (W/kg) for the hip, knee and ankle joints from the modelled data. Joint moments and power joint were normalized to body mass to allow for comparison between subjects and to previous studies. Total power absorption and generation were obtained by integrating all the negative or positive values of the joint powers during stance phase, yielding an average total power of each joint during the two phases of stance phase.

Statistical Analysis

All measurements were expressed as mean and standard deviation. A two-way ANOVA using SPSS 20 (SPSS Inc. Chicago, IL, USA) was used to examine the effects of slope and landing pattern on the vertical and parallel GRF variables, with slope and landing pattern used as separate factors. All significant slope-by-landing pattern interactions were reported. If a significant slope-by-landing pattern interaction was present, a Bonferroni *post hoc* test was conducted in order to determine where these differences occurred. Levene's test revealed that not all variables had homogeneity of variance, therefore, F ratios were considered significant at $p < .01$.

As main effect test of two-way ANOVA showed significant effects of slope and landing patterns on the dependent measures, Bonferroni *post hoc* test was run to examine our hypotheses (see **Introduction: Hypotheses**). To answer hypotheses i and ii, uphill and downhill conditions were tested against the level condition independent of foot strike pattern to determine the effect of slope on our dependent variables. To answer hypotheses iii and iv, FFS and RFS groups were

tested against each other at each slope condition to examine the effect of landing pattern on our dependent variables.

Chapter 4

Manuscript #1

Ground reaction forces in forefoot strike and rearfoot strike runners during overground downhill and uphill running

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Abstract

We investigated the normal and parallel ground reaction forces during downhill and uphill running in habitual forefoot strike and habitual rearfoot strike runners. Fifteen habitual forefoot strike and fifteen habitual rearfoot strike recreational male runners ran at $3 \text{ m/s} \pm 5\%$ during level, uphill and downhill overground running on a ramp mounted at 6° and 9° . Results showed that forefoot strike runners had an absent impact peak in all running conditions, while the impact peaks only decreased during the uphill conditions in rear foot strike runners. Active peaks decreased during the downhill conditions in forefoot strike runners while active loading rates increased during downhill conditions in rearfoot strike runners. Compared to the level condition, parallel braking peaks were larger during downhill conditions and parallel propulsive peaks were larger during uphill conditions. Combined with previous biomechanics studies, our findings of no impact peak in forefoot strike runners suggests that this landing pattern may have potential in reducing overuse running injuries. Forefoot strike running may be a strategy to use especially during downhill running, as research has shown that downhill running has the greatest risk of overuse running injury.

Keywords: Biomechanics; Slope; Incline; Decline;

Introduction

Ground reaction force (GRF) data is essential in biomechanics, giving us the physical interaction between two bodies. While running, a runner generates a force acting on the Earth from the runner's muscular efforts and body mass while the Earth exerts an equal but opposite reaction force. GRF data can be used to determine foot strike pattern, understand propulsion and braking, as well as to compute loading rate, joint moments and powers. By using GRF variables such as vertical force impact peak we are able to predict injury risk as this variable was significantly lower in the injury-free versus injured group of long distance runners (Hreljac et al., 2000).

Majority of publications on running biomechanics are directed to the study of rearfoot strike (RFS) runners. This type of landing pattern is characterized by GRF centre of pressure

being located in the posterior third of the foot at initial contact and a dorsiflexed ankle (Cavanagh & LaFortune, 1980). Another landing pattern seen in runners, forefoot strike landing, has its GRF centre of pressure located in the anterior third of the foot at initial contact, and is characterized by a plantarflexed ankle (Cavanagh & LaFortune, 1980). These different landing patterns affect GRF curves and may affect why differences in rates and locations of injuries vary between runners employing different landing patterns (Daoud et al., 2012).

When RFS runners run at a pace of 3 m/s, the vertical component of the GRF quickly rises and falls in the first 10% of stance phase, forming an impact peak of ~ 1.6 body weight (BW). The vertical component of the GRF continues to rise at a slower rate to a second peak of ~ 2.5 BW at mid-stance, termed the active peak, before returning to zero at toe-off (Gottschall & Kram, 2005). FFS runners have been shown to have an absent impact in the first 10% of stance phase (Lieberman et al., 2010), but have a larger active peak at midstance than RFS runners (Laughton et al., 2003). The parallel component of the GRF is initially negative after initial contact as a braking force is applied, reaching a peak of ~ -0.3 BW approximately a quarter into the stance phase before increasing and reaching zero at mid-stance. After mid-stance the parallel GRF becomes positive as a propulsive force is applied, reaching a peak of ~ 0.3 BW approximately three-quarters into stance phase before returning to zero prior to toe-off (Gottschall & Kram, 2005).

GRF during level running is well researched at various running speeds. The reality is that many runners run outdoors where they often encounter hills. So far GRF data for running uphill or downhill is generally limited to collection methods using instrumented treadmills (Gottschall & Kram, 2005; Padulo et al., 2013; Telhan et al., 2010), with only a few publications examining overground hill running (Buczek & Cavanagh, 1990; DeVita et al., 2008). Limited running

studies conducted on ramps are likely due to the lab space limitation or the difficulty of force platform setting. Gottschall and Kram did a thorough investigation of hill treadmill running on primarily RFS runners while running on $\pm 3^\circ$, $\pm 6^\circ$, $\pm 9^\circ$ at a constant speed of 3.0 m/s (Gottschall & Kram, 2005). During downhill 6° running, impact peak increased by 32% while average loading rate increased by 23%, whereas, during uphill 6° running, impact peak decreased by 22% and average loading rate decreased by 23%. Neither uphill nor downhill running affected normal active peaks. Parallel braking forces increased during downhill conditions and decreased during uphill conditions compared to the level condition, while parallel propulsive forces decreased during downhill conditions and increased during uphill conditions. Although this research advanced our knowledge of hill running, it was conducted on a treadmill, and research evidences show that treadmill running is similar but not identical to overground running during level conditions. Moreover the study was done in RFS. Therefore we must interpret these results with caution (Riley et al., 2008).

Several questions still remain unanswered which can only be answered by studying GRF during overground downhill and uphill running at several angles. For example, it has only been reported that FFS runners have an absent impact peak during level running, so will this impact peak remain absent across hill conditions? Do normal active peaks and loading rates change during downhill and uphill running? How do parallel GRF forces change during overground uphill and downhill running? Finally, are RFS and FFS runners similar in GRF during overground hill running?

The purposes of this study were first to quantify the GRF during overground downhill and uphill running in both habitual RFS and FFS runners, furthermore to examine the possible differences in GRF between different slope conditions, and finally to compare differences in the

measures between RFS and FFS at each running condition. To do so, we mounted a force platform to a custom built ramp at several angles. We tested 4 hypotheses: compared to level running in both RFS and FFS runners, (1) normal impact force peaks would be larger during downhill running and smaller during uphill running, (2) normal active force peaks would remain unchanged whereas active loading rates would be larger during downhill running and smaller during uphill running, (3) parallel braking force peaks would be greatest during the downhill 9° condition while parallel propulsive force peaks would be greatest during the uphill 9° condition. Finally, compared to RFS runners, (4) FFS runners will have an absent impact peak for all running conditions.

Methods

Participants

Participants were recruited at races in the Ottawa area and through posters placed around the university and running clubs in the Ottawa area. All prospective runners had to be between the ages of 18 – 35 years of age, run a minimum of 15 km/week, and be free from musculoskeletal injuries 6 months prior to participation. Potential participants came for a preliminary screening session where we analyzed their landing pattern via strike index (Buczek & Cavanagh, 1990). Participants were given a pair of Merrell Vapor Glove minimalist running shoes with reflective markers placed on the heel, medial and lateral malleoli, and between the 2nd and 3rd metatarsal heads. They were instructed to run overground at $3 \pm 5\%$ m/s and strike an imbedded force plate 4m away from the starting line.

The participants whose foot centre of pressure at foot strike was located in the anterior third of the foot were placed into the forefoot strike group, and the participants whose foot center of pressure was located in the posterior third of the foot were placed into the rearfoot strike

group. Runners who landed with a midfoot strike landing pattern were excluded from participation as their foot strike is highly variable and characteristic of both a FFS and RFS landing pattern (Lieberman, 2012). Two groups, FFS and RFS group, with each 15 runners were formed. Table 1 provides the anthropometric data of the participants. Written informed consent following the guidelines of the Health Sciences and Science Research Ethics Board at the University of Ottawa was given by all participants prior to any data collection.

Table 1: Anthropometric data of the participants (mean $\pm$ SD)

	FFS Runners (n = 15)	RFS Runners (n = 15)	Total (n = 30)
Age (year)	26.1 $\pm$ 4.9	26.8 $\pm$ 4.5	26.5 $\pm$ 4.7
Height (m)	1.77 $\pm$.06	1.79 $\pm$.06	1.78 $\pm$ 0.05
Mass (kg)	74.9 $\pm$ 10.9	78.7 $\pm$ 9.1	76.8 $\pm$ 10.0

Instrumentation

We constructed a 3m steel ramp with embedded force platform that could be raised to form slopes of 6° and 9° (10.5% and 15.8%). A 4.7m wooden platform was attached to the ramp to provide a running surface for participants at the top of the ramp. Wheel jacks were placed on the outside of the ramp beside the force platform and a car jack was placed under the ramp below the force platform to help reduce vibrations and bending of the ramp (Figure 1). The force platform (model 9286AA, Kistler Instruments Corp, Winterhur, Switzerland) was placed approximately 2/3 of the way up the ramp and approximately 4m away from the starting line during the level trial. Force platforms captured GRF at a frequency of 1000 Hz. A 10 camera Vicon Motion Analysis System (MX-13, Oxford Metrics, Oxford, UK) recorded movement at 250 Hz for motion analysis. All participant wore the same Merrell Vapor Glove minimalist running shoes as during the preliminary screening. Each shoe was approximately 165 $\pm$ 11 g and had 5 mm of cushioning and a heel-to-toe drop of 0 mm.

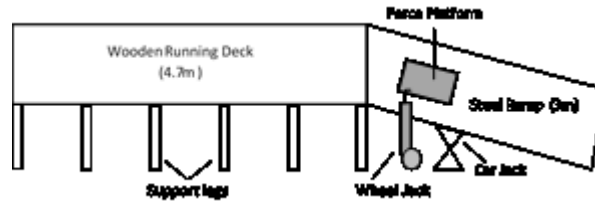


Figure 1: Ramp with embedded force platform mounted at 9°.

Data collection

Participants returned for the main testing session one week after their preliminary session. Participants were outfitted in 45 reflective markers according to the University of Ottawa Motion Model (Lamontagne et al., 2009) (modified Plug-in Gait Model). Testing included 5 different slope conditions (downhill 9°, downhill 6°, level, uphill 6° and uphill 9°) completed in a quasi-random order; both 6° and both 9° conditions were completed consecutively due to the lengthy process of changing the angle of the ramp. Running speed was set at $3 \pm 5\%$ m/s (5.6 min/km, 8.9 min/mi), which was monitored using custom built photocells. Participants started running approximately 4m away from the force plate at each condition.

Data Processing and Analysis

GRF data was cropped and time normalized to 100% stance phase, except for when calculating loading rates and parallel impulses. GRF data of stance phase was filtered using a fourth-order recursive, zero phase-shift, Butterworth low-pass filter with a cut-off frequency of 50 Hz and kinematic data was filtered using a Woltring 15 MSE filter (Woltring, 1986). Modelled data was further processed and analyzed in Matlab R2013a (MathWorks, Natick, MA, USA) to retrieve GRF information.

For each condition, we located impact force peaks, active force peaks peak braking forces, and peak propulsive forces (BW) that are presented as times of body weight (BW).

Active and impact loading rates (BW/s) were calculated from the vertical GRF and braking and propulsive impulses (N·s) were calculated from the parallel GRF. We determined the impact force peak by finding the highest value from foot strike to the positive slope became negative (within 25 ms after foot strike). Peak braking forces were determined by finding the minimum parallel GRF value and peak propulsive forces were determined by finding the maximum parallel GRF value. Impact loading rates were calculated by dividing the impact force peak by the time from foot strike to the impact force peak, whereas active loading rate equaled the active force peak divided by the time from foot strike to the active force peak. Parallel impulse data was obtained by integrating all the negative (parallel braking) or positive (parallel propulsive) values of the parallel GRF during stance phase, yielding an average impulse during stance phase.

To obtain lower limb kinematic measurements, Vicon Nexus Software (vers) using the UOMAM model was used to retrieve sagittal joint angles of the hip, knee and ankle joints for FFS and RFS groups during the 5 examined slope conditions. Measurements of initial contact and toe-off were defined using force platform values. Initial contact was defined as the moment when the vertical GRF data first became greater than 20 N and toe off was defined after it fell below 20 N.

A two-way ANOVA using SPSS 20 (SPSS Inc. Chicago, IL, USA) was used to examine the effects of slope and landing pattern on the vertical and parallel GRF variables, with slope and landing pattern used as separate factors. All significant slope-by-landing pattern interactions were reported. If a significant slope-by-landing pattern interaction was present, a Bonferroni *post hoc* test was conducted in order to determine where these differences occurred. Levene's test revealed that not all variables had homogeneity of variance, therefore, F ratios were considered significant at $p < .01$. In the tables, the data is presented as means and SD (standard deviation).

Results

GRF was studied in RFS and FFS runners as they ran at $3 \text{ m/s} \pm 5 \%$ during overground level, uphill and downhill running and measures were compared between the two groups. All results are presented as mean and standard deviation (SD). Overall, a significant landing effect ($p < .01$) was observed in all variables and a significant slope effect ($p < .01$) was observed in all variables. There was a significant slope-by-landing pattern interaction ($p < .01$) for 8 of the 11 variables. No significant slope-by-landing pattern interaction was found in the normal active peaks, and hip and knee angles at initial contact. *Post hoc* tests revealed that the normal impact peak and consequently the impact loading rate were significantly different ($p < .01$) between RFS and FFS runners during all slope conditions.

Level versus slope

Table 2 summarizes the vertical ground reaction force values for the FFS and RFS groups during running at 5 examined conditions. Active peaks were largest in the level condition for both groups. However, when running during the downhill conditions the FFS group showed significantly ($p < .01$) lower active peaks compared to the level trial (Figure 2a). Moreover the FFS group didn't show any significant differences in active loading rate between 5 running conditions. In the RFS group active loading rates were significantly higher for the downhill conditions compared to level running condition. Impact force peaks significantly ($p < .01$) decreased by 29.2% and 38.3% from level to uphill 6° and level to uphill 9° , respectively, in the RFS group but were not significantly different for downhill conditions. Impact loading rates followed similar trends as there were no significant differences for downhill conditions, but it decreased by 33% for both uphill conditions when compared to the level condition.

Table 2: Vertical ground reaction forces for the FFS and RFS groups during running at 5 examined conditions (means $\pm$ SD)

Condition (°)	Impact Force Peak (BW)	Active Force Peak (BW)	Impact Loading Rate (BW/s)	Active Loading Rate (BW/s)
FFS				
-9	-- [‡]	$2.29 \pm 0.22^{\dagger}$	-- [‡]	24.05 ± 5.92
-6	-- [‡]	$2.33 \pm 0.22^{\dagger}$	-- [‡]	21.31 ± 4.32
0	-- [‡]	2.51 ± 0.25	-- [‡]	21.50 ± 3.66
6	-- [‡]	2.42 ± 0.22	-- [‡]	21.73 ± 4.33
9	-- [‡]	2.34 ± 0.21	-- [‡]	21.19 ± 4.16
RFS				
-9	$1.71 \pm 0.30^*$	2.24 ± 0.17	$87.70 \pm 35.43^*$	$21.40 \pm 4.89^{\dagger}$
-6	$1.48 \pm 0.26^*$	$2.18 \pm 0.21^*$	$87.88 \pm 33.82^*$	$21.71 \pm 4.18^{\dagger}$
0	$1.54 \pm 0.32^*$	$2.37 \pm 0.19^*$	$102.90 \pm 35.07^*$	20.63 ± 3.28
6	$1.09 \pm 0.20^{*\dagger}$	$2.32 \pm 0.15^*$	$68.45 \pm 19.66^{*\dagger}$	19.88 ± 3.89
9	$0.95 \pm 0.27^{*\dagger}$	$2.28 \pm 0.14^*$	$68.44 \pm 32.30^{*\dagger}$	18.14 ± 4.50

The two-way ANOVA revealed statistically significant ($p < .01$) in all variables for effect of landing and statistically significant differences ($p < .01$) in all variables for effect of slope. There was a statistically significant ($p < .01$) slope-by-landing pattern interaction for all analyzed parameters. Post hoc comparisons for between group differences are indicated by * that represents significant differences between RFS and FFS at each slope condition. Significant differences within group are represented by [†] that represents significantly different from the level trial for that foot strike. [‡] indicates absent impact force peak and impact loading rate in FFS runners.

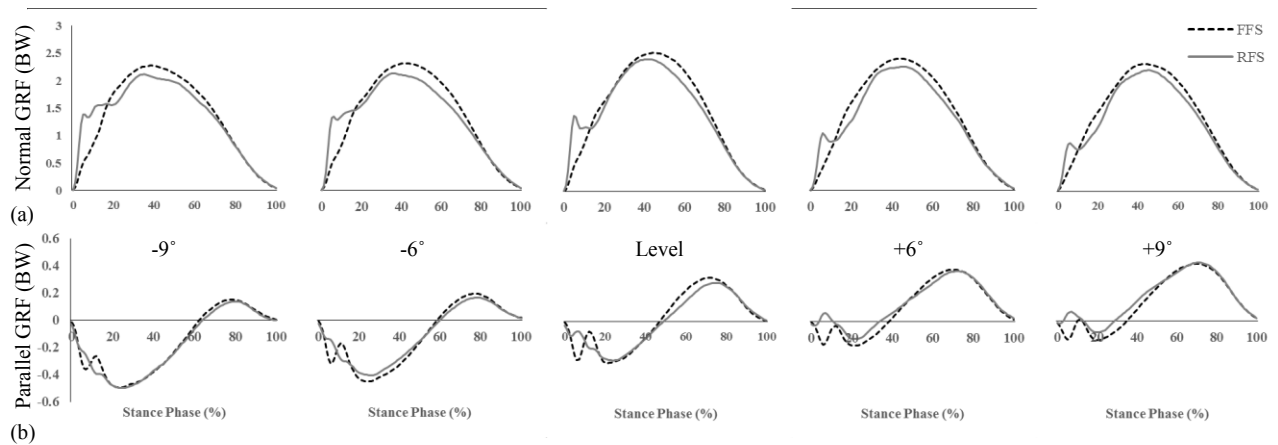


Figure 2: Normal (a) and parallel (b) ground reaction force normalized to percent of stance phase for FFS and RFS participants running at $3 \pm 5\%$ m/s.

Table 3 reports the parallel ground reaction forces that include peak braking force and propulsive forces for the FFS and RFS groups during the 5 examined slope conditions. Peak

braking force significantly increased for downhill conditions and significantly decreased for the uphill conditions, whereas peak propulsive forces significantly decreased for downhill conditions and significantly increased for uphill conditions (Figures 3a & 3b). Parallel impulses followed the same trend as parallel peak where braking impulses were larger during downhill conditions and propulsive impulses were larger during uphill conditions.

In general, hip flexion angles at initial contact increased as slope increased from the downhill 9° condition to the uphill 9° condition. Knee flexion angles only increased significantly for the uphill conditions. Ankle angles at initial contact did not follow the same trend for both groups. No significant changes occurred in the RFS group, however plantar flexion angle decreased as slope increased from the downhill 9° condition to the uphill 9° condition (Table 4).

Table 3: Parallel ground reaction forces for the FFS and RFS groups during the 5 examined slope conditions (means $\pm$ SD)

Condition (°)	Braking Peak (BW)	Propulsive Peak (BW)	Braking Impulse (N·s)	Propulsive Impulse (N·s)
FFS				
-9	-0.52 $\pm$ 0.08 [†]	0.15 $\pm$ 0.04 [†]	-36.08 $\pm$ 9.90 [†]	6.14 $\pm$ 2.22 [†]
-6	-0.47 $\pm$ 0.06 [†]	0.20 $\pm$ 0.05 [†]	-29.64 $\pm$ 8.94 [†]	8.60 $\pm$ 2.79 [†]
0	-0.34 $\pm$ 0.09	0.33 $\pm$ 0.06	-13.67 $\pm$ 4.57	17.80 $\pm$ 4.81
6	-0.21 $\pm$ 0.08 [†]	0.38 $\pm$ 0.06 [†]	-7.57 $\pm$ 4.17 [†]	24.19 $\pm$ 6.04 [†]
9	-0.19 $\pm$ 0.08 [†]	0.42 $\pm$ 0.08 [†]	-5.19 $\pm$ 3.11 [†]	29.63 $\pm$ 7.69 [†]
RFS				
-9	-0.53 $\pm$ 0.07 [†]	0.14 $\pm$ 0.05 [†]	-40.45 $\pm$ 7.28 ^{*†}	5.89 $\pm$ 2.58 [†]
-6	-0.43 $\pm$ 0.07 ^{*†}	0.17 $\pm$ 0.05 [†]	-33.47 $\pm$ 9.09 ^{*†}	9.13 $\pm$ 3.70 [†]
0	-0.31 $\pm$ 0.08	0.28 $\pm$ 0.04 [*]	-18.11 $\pm$ 6.70 [*]	18.19 $\pm$ 4.30
6	-0.14 $\pm$ 0.07 ^{*†}	0.37 $\pm$ 0.06 [†]	-5.38 $\pm$ 3.98 [†]	30.93 $\pm$ 7.09 ^{*†}
9	-0.11 $\pm$ 0.06 ^{*†}	0.43 $\pm$ 0.06 [†]	-2.90 $\pm$ 2.29 [†]	39.54 $\pm$ 6.87 ^{*†}

The two-way ANOVA revealed statistically significant ($p < .01$) in all variables for effect of landing and statistically significant differences ($p < .01$) in all variables for effect of slope. There was a statistically significant ($p < .01$) slope-by-landing pattern interaction for all analyzed parameters. Post hoc comparisons on for between group differences are indicated by * and represent differences between RFS and FFS at each slope condition, whereas within group differences are represented by [†] and represent parameters that are significantly different from the level trial for that foot strike.

FFS versus RFS

FFS runners had an absent impact transient and as a result did not have an impact loading rate available for any of the conditions, whereas both were present during all conditions in RFS runners (Figure 2a).

Active force peaks were significantly larger in FFS runners compared to RFS runners in all conditions except the downhill 9° condition (Figure 2a). Although active loading rate was generally larger in FFS for most conditions, it never reached a significant difference (Table 2).

Table 4: Sagittal joint angles of the hip, knee and ankle at foot strike for FFS and RFS groups during the 5 examined slope conditions (means $\pm$ SD)

Condition (°)	Hip Flexion Angle (°)	Knee Flexion Angle (°)	Ankle Angle (°)
FFS			
-9	30.8 $\pm$ 5.4 [†]	11.2 $\pm$ 6.0	-27.9 $\pm$ 6.3 [†]
-6	32.0 $\pm$ 5.7 [†]	11.1 $\pm$ 5.4	-25.1 $\pm$ 5.8 [†]
0	38.9 $\pm$ 5.5	13.9 $\pm$ 5.6	-16.1 $\pm$ 5.5
6	42.3 $\pm$ 5.3	21.2 $\pm$ 6.0 [†]	-10.7 $\pm$ 6.1 [†]
9	46.4 $\pm$ 4.4 [†]	24.5 $\pm$ 5.3 [†]	-7.2 $\pm$ 5.3 [†]
RFS			
-9	27.1 $\pm$ 7.5 ^{*†}	11.5 $\pm$ 4.9	4.4 $\pm$ 3.6 [*]
-6	27.5 $\pm$ 6.5 ^{*†}	10.9 $\pm$ 4.6	4.5 $\pm$ 3.7 [*]
0	34.7 $\pm$ 6.4 [*]	13.3 $\pm$ 5.4	6.3 $\pm$ 1.7 [*]
6	39.8 $\pm$ 7.8 [†]	18.5 $\pm$ 5.3 [†]	6.7 $\pm$ 1.9 [*]
9	43.8 $\pm$ 8.1 [†]	23.1 $\pm$ 5.7 [†]	8.0 $\pm$ 2.2 [*]

The two-way ANOVA revealed statistically significant ($p < .01$) in all variables for effect of landing and statistically significant differences ($p < .01$) in all variables for effect of slope. There was a statistically significant ($p < .01$) slope-by-landing pattern interaction for ankle angle. Post hoc comparisons on for between groups differences are indicated by * and represent differences between RFS and FFS at each slope condition, whereas within group differences are represented by [†] that represents significantly different from the level trial for that foot strike. Note negative ankle angle represents plantar flexion and positive ankle angle represents dorsiflexion.

Peak braking force peaks were significantly ($p < .01$) smaller in RFS runners for the downhill 6° condition and both uphill conditions compared to FFS runners (Table 3). Propulsive peaks were only significantly different in the level condition where FFS runners had an 18%

larger peak propulsive force compared to RFS runners. Braking impulses were significantly larger in RFS runners in the level and both downhill conditions compared to the FFS group, while propulsive impulses were larger for the RFS group during both uphill conditions.

FFS runners had significantly less hip flexion at initial contact than RFS runners for both downhill conditions and the level condition. No significant difference existed between both groups in knee flexion angle at initial contact. Ankle angles were significantly different in all conditions between both groups. RFS runners landed with the ankle in a dorsiflexed position, whereas FFS runners landed with the ankle in a plantar flexed position (Table 4).

Discussion

The study was to examine the GRF in habitual RFS and FFS runners as they ran overground on level and hill surfaces. Furthermore, we wanted to discover the differences in GRF force between different hill conditions in RFS and FFS respectively, and finally to compare if there are differences in GRF between RFS and FFS runners at each condition. The findings of this study showed that FFS runners had an absent impact transient during all running condition, which resulted in them not having an impact loading rate. These differences are most likely linked to foot placement at initial contact as FFS runners maintained a plantar flexed angle throughout, whereas RFS runners used a dorsiflexed landing position for all conditions. However, the differences in foot strikes did result in differences in peak active forces, as FFS runners had larger active peaks than RFS runners in all conditions, agreeing with the previous findings which showed greater peak vertical GRF in FFS runners (Laughton et al., 2003). Braking peaks and impulses decreased as slope increased, while propulsion peaks and impulses increased as slope increased in both groups, following similar trends to those done at identical slopes while running on a treadmill (Gottschall & Kram, 2005).

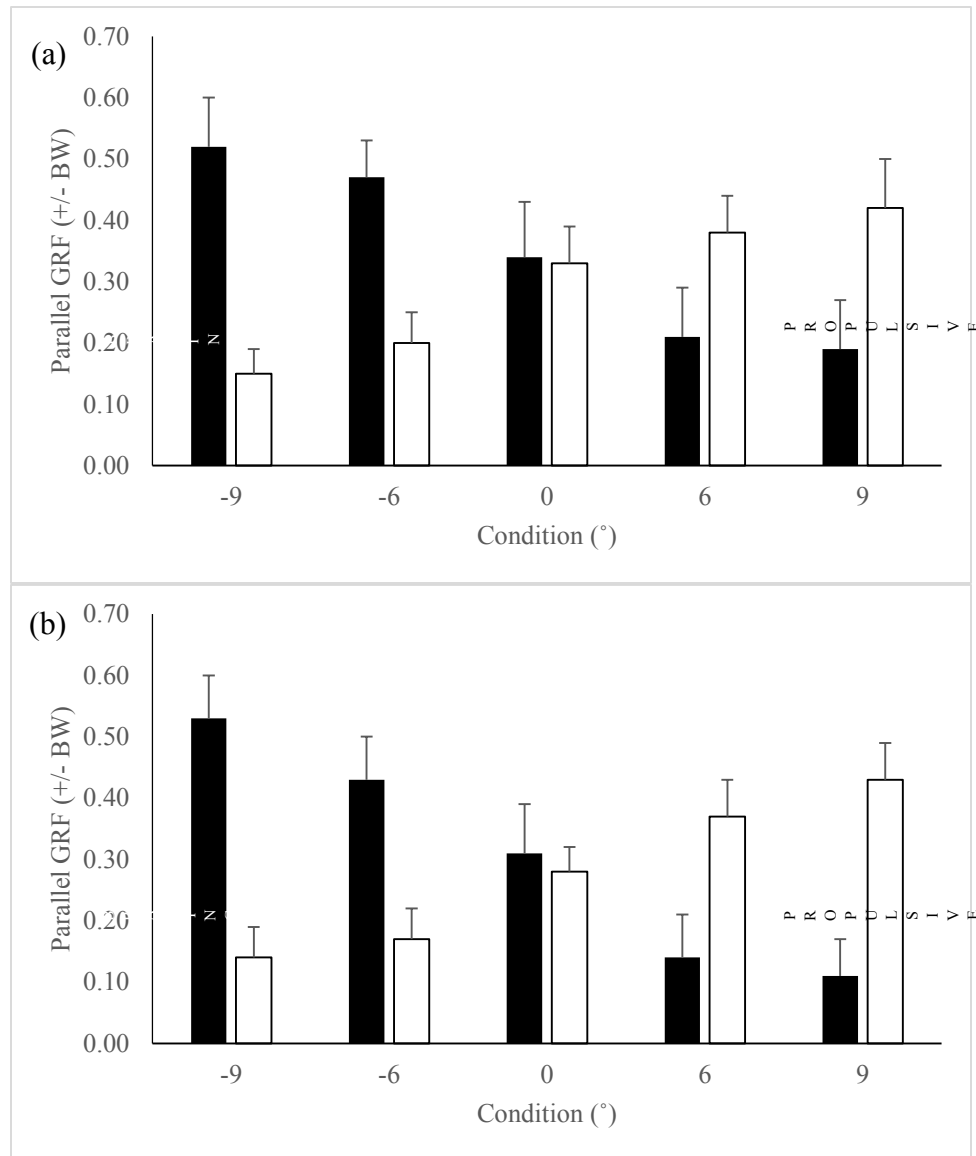


Figure 3: Parallel ground reaction forces (mean $\pm$ SD) in (a) FFS runners and (b) RFS runners. All conditions are significantly different ($p < .01$) from level condition. Parallel braking force is indicated by black bars and propulsive force is indicated by white bars. Note that mean parallel braking force peaks are presented as the absolute value.

Level versus slope

Since it was predicted that normal impact force peaks would be larger during downhill running and smaller during uphill running, we expected that impact loading rate would follow a similar trend. This hypothesis was based on the previous work of Gottschall & Kram in which

they reported that impact force peaks and impact loading rates increased in RFS runners as slope changed from uphill to downhill conditions (Gottschall & Kram, 2005). Our results deny our first hypothesis, normal impact force peaks would be larger during downhill running and smaller during uphill running, as impact force peaks and impact loading rates were only significantly ($p < .01$) lower during the uphill conditions and not during the downhill conditions in the RFS runners.

Since Gottschall and Kram did not report normalized GRF data we cannot directly compare our results with theirs, however, by normalizing their values based on their average participant mass we can get an approximate value. In the level condition the RFS group of our study showed 1.54 BW impact force peak. The calculated impact force peak of their result was 1.61 BW. They did not report GRF values for +9° condition, in the +6° condition they had an impact force peak of 1.25 BW compared to our 1.09 BW. The findings of impact force peak from both of us were significantly different from our level condition. RFS runners in this study had insignificantly lower normal impact peaks of 1.48 BW and 1.71 BW for the downhill 6° and 9° condition, respectively, compared to the calculated values of 2.13 BW and 2.45 BW for their downhill 6° and 9° condition, respectively (Gottschall & Kram, 2005).

Several possible reasons for the large differences between the results of our study and Gottschall & Kram's findings (2005) during the downhill conditions. First, they averaged 20 consecutive steps of downhill running on the treadmill whereas we only observed one step per trial. The second possible reason is the difference in running surfaces. Our participants ran overground on a ramp whereas their participants ran on an instrumented treadmill. Although treadmill running is similar to overground running it is not identical (Riley et al., 2008). No one

has compared impact peaks between overground and treadmill running so this may help explain our results.

We hypothesized that normal active force peaks would remain unchanged in both groups whereas active loading rates would be larger during downhill running and smaller during uphill running. Gottschall and Kram found no significant differences in normal active force peaks between level and sloped conditions (Gottschall & Kram, 2005). The work of De Vita and colleagues (2008) showed decreased stance times in downhill conditions compared to uphill conditions, from which it can be translated to decreased time to peak loading which would increase active loading rate (DeVita et al., 2008). Our results partially support our second hypothesis as active force peaks were significantly different for FFS runners during downhill conditions while active loading rates were only significantly different in RFS runners during downhill conditions.

Active force peaks were largest during the level condition for both RFS and FFS runners. Our findings confirm previous research which also showed insignificant changes in active force peaks in RFS runners (Dick & Cavanagh, 1987; Gottschall & Kram, 2005). Active peaks were only significantly ($p < .01$) lower during the downhill conditions in the FFS group, of which they were less than 10% different from the level condition. Average normal force for a complete gait cycle during hill running must equal product of mass, gravity and cosine of the angle. The cosine of 9° is .988, making average normal force almost identical to level running. The normal active force peaks in level running are related to the average normal force and inversely related to duty factor (Gottschall & Kram, 2005). Duty factor is the fraction of the stance period that one foot spends in contact with the ground and may be different between RFS and FFS runners

(Gottschall & Kram, 2005), which may explain why differences exist in active force peaks during downhill running in FFS runners and not in RFS runners.

Active loading rates represent the active peak which occurs at midstance divided by the time from initial contact to this peak value. These loading rates were generally insignificant between slopes, however, active loading rates did increase during downhill running only in the RFS group (Table 2). This increase could be a result of decreased stance time since peak active force was lower during the downhill conditions compared to the level and still caused greater active loading rates. Stance time was shown to decrease from uphill to downhill conditions (DeVita et al., 2008), it can partially explain why active loading rates are larger in downhill conditions and lower in uphill conditions. Changes in stance time between the conditions may not be large enough to cause large changes to active loading rate to cause significant changes.

It was hypothesized that braking peaks would decrease as slope increased, whereas propulsion peaks would increase in this study. Gottschall and Kram as well as other researchers thought that parallel braking forces corresponded to metabolic costs of running because as metabolic cost increases, concentric muscle contraction, i.e., propulsive forces must increase (Gottschall & Kram, 2005; Minetti, Ardigo, & Saibene, 1994). Our findings that the largest peak braking forces during the downhill 9° condition and the largest peak propulsion forces during the uphill 9° condition confirmed our third hypothesis.

In both RFS and FFS runners, peak braking forces increased as the conditions changed from the uphill 9° condition to the downhill 9°, whereas the opposite was true for the peak propulsion forces (Figures 3a & 3b). The same trend was found for braking and propulsive impulses (Table 3). To show constant velocity during level running, braking and propulsive must be equal and opposite in magnitude. This pattern was not evident in either the downhill or uphill

conditions. During downhill 9° running, the braking impulse increased an average of 144% between RFS and FFS runners, whereas the propulsion impulse decreased an average of 66.6% between both groups. This asymmetry existed during uphill 9° running as well, the braking impulse decreased by an average of 73% between both groups and the propulsion impulse increased by an average of 92%. It appears that runners do not symmetrically modify both parallel braking and propulsive impulses to sustain a constant running speed. Our photocell timers were 3 m apart and confirmed that all runners ran at a speed of 3 m/s between the 2 photocells, so it seems that runners must have utilized a different strategy to maintain constant speed which cannot be explained with only the GRF data. Previous research showed that RFS runners were able to modify both parallel impulses to sustain speed while treadmill running only during the +9° but not during the -9° condition (Gottschall & Kram, 2005).

Parallel braking forces correspond to aspects of the metabolic cost for running, where braking forces represent eccentric muscles contractions and propulsive forces represent concentric muscle contractions (Abbott, Bigland, & Ritchie, 1952; Margaria, 1976; Minetti et al., 1994). Uphill running has a large metabolic cost since concentric muscle contractions are more metabolically taxing than eccentric muscle contractions (Abbott et al., 1952; Minetti et al., 1994) and our findings show the largest peak propulsive forces during the uphill 9° condition. During downhill running, muscles contract eccentrically for most of stance phase causing the less taxing negative work (Gabaldon, Nelson, & Roberts, 2004), and largest peak braking forces occurred during the downhill 9° condition. Therefore, the changes we found in parallel GRF data can explain some of the metabolic costs associated with hill running.

FFS versus RFS

The results of this study support our fourth hypothesis, FFS runners did not have an impact transient and consequently no impact loading rate during any of the tested conditions, whereas both were present in all conditions in the RFS runners (Figure 2a). This lack of impact transient is characterized by a plantar flexed ankle at initial contact in all conditions. FFS runners maintained a plantar flexed ankle during all condition, but as the uphill angle increased there was a decrease in the plantar flexion angle.

To our knowledge it is the first study to show that FFS runners do not have an impact peak during uphill or downhill running similar to level running. This strengthens the argument made by Nigg in which he mentioned FFS runners do not need any cushioning from shoes to dampen the shock at impact because they do not generate any impact peak in the first place, justifying the use of minimalist running shoes for this group (B. Nigg, 2001). Several reasons have been used to explain variations in impact peak between RFS and FFS runners. The first is due to the angle of the ankle at foot strike. With FFS runners, the foot is initially plantarflexed and then experiences controlled dorsiflexion at a compliant ankle. However, in RFS runners, the foot is dorsiflexed and the ankle is stiff at foot strike. This causes a greater effective mass (M_{eff}) in RFS runners. A study by Lieberman and colleagues measured M_{eff} of $1.7 \pm 0.4\%$ body mass in FFS and $6.8 \pm 3.0\%$ in RFS running (Lieberman et al., 2010). These values generally represent the percentage mass of the foot and lower leg, respectively. The other reason explaining an absent impact peak is compliance. RFS runners land with less knee flexion and stiffer knee and ankle joints than FFS runners. FFS runners have greater knee flexion at impact and their ankle dorsiflexes during stance phase, which allows the lower extremity to soften forces more effectively. This is why toe running, a landing pattern in which the toes contact the ground first

but the heel never touches, can result in an impact peak because the runners' ankle is relatively stiff at impact (Lieberman, 2012).

Previous research on normal impact peak data found that it was the primary biomechanical variable which separated the injured versus non-injured runner (Hreljac et al., 2000). The injured group had normal impact peaks approximately 13% larger than the injury free group. Based on this finding it has been suggested that downhill running has the largest risk for musculoskeletal injury and uphill running would decrease this risk (Gottschall & Kram, 2005). This lack of impact transient in FFS runners during all conditions gives merit to the possibility of runners adapting a FFS landing pattern as a way of preventing overuse running injuries.

Patellofemoral pain (PFP) is an overuse injury has been linked to the impact transient. A small case study done by Cheung and Davis (2011) had 3 RFS runners with PFP adopt a gait retraining program and after 3 months of the program vertical impact peaks were reduced and patellofemoral pain and associated functional limitations were improved (Cheung & Davis, 2011). Another study found that altering strike pattern to a FFS resulted in consistent reductions in patellofemoral joint stress independent of changes in step length, again warranting the use of FFS landing patterns to be utilized as a treatment for PFP (Vannatta & Kernozek, 2015). FFS running would be a potential solution for runners who experience overuse injuries, specifically PFP.

Active peaks were larger in FFS runners for all conditions except the downhill 9° condition, and this is consistent with previous findings done during level running between RFS and FFS runners (Laughton et al., 2003; D. S. Williams et al., 2000). They linked the increase in peak active force to greater tibial shock in FFS runners. FFS runners had lower ankle stiffness

but greater knee stiffness compared to RFS runners, which suggested that the knee may be a stronger regulator of leg stiffness than the ankle (Hamill et al., 2014; Laughton et al., 2003).

FFS runners had larger peak braking forces for the downhill 6° condition and both uphill conditions and a larger propulsive peak during the level condition compared to the RFS group (Table 3). The increase in peak braking force can be explained by the eccentric loading of the triceps surae group after landing with a plantar flexed ankle and have an increase in power absorption at the ankle joint compared to RFS runners (D. S. Williams et al., 2012). The dependence on this muscle group explains why many individuals who attempt running with a FFS landing pattern experience soreness in the same muscle group as this soreness has been associated with the eccentric activity of these muscles in the absorption phase (Newham, Mills, Quigley, & Edwards, 1983).

A couple limitations exist with our research. Due to the placement of the force platform on the ramp, not all subjects had a step on the ramp before making contact with the force platform, causing some participants to have an extended float period from the running deck before making the first step during downhill running directly onto the force platform. A possible solution for future studies would be to place the force platform further down the ramp to allow participants to have a step down the ramp before making contact with the force plate. The other possible limitation with our study is we used a ramp that was 3 m in length, which may have had some bending as runners ran over it. Although we tried to minimize the bending that occurred by adding supports under the force platform and ramp, it may not have eliminated it completely. This bending in the steel ramp could have reduced impact force peaks which may explain why our impact force peaks during downhill running were lower than those previously reported.

Conclusion

In summary, RFS and FFS runners presented different ground reaction force patterns during hill running. Hill running did not significantly affect impact force peaks in RFS runners, however impact peak tended to decrease as slope increased. Normal active force peaks only changed in FFS runners during downhill conditions while active loading rates only changed in RFS runners during downhill conditions. As slope changed from a steep downhill to a steep uphill, parallel braking forces decreased and parallel propulsive peaks increased. The most significant finding that is revealed for first time was that FFS runners had an absent impact peak in hill running conditions as observed that in level running. This finding supports the suggestion from research which used FFS running as an intervention strategy for runners with a history of overuse running injuries, such as patellofemoral pain syndrome, since many overuse running injuries have been linked to impact peaks.

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Chapter 5

Manuscript #2

Lower limb kinematics and kinetics of overground downhill and uphill running in forefoot strike and rearfoot strike runners

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Abstract

We investigated the lower limb kinetics during downhill and uphill running in habitual forefoot strike and habitual rearfoot strike runners. Fifteen habitual forefoot strike and fifteen habitual rearfoot strike recreational male runners ran at $3 \text{ m/s} \pm 5\%$ during level, uphill and downhill overground running on a ramp mounted at 6° and 9° . In addition to the known differences in ankle joint angles at initial contact between the two groups, results showed that forefoot strike runners had a more flexed hip angle during downhill running. Peak hip flexion moment was significantly greater while peak knee flexion moment was significantly lower in both groups during the downhill 9° condition. Forefoot strike runners had larger peak plantar flexion moments and peak ankle power absorption compared to rearfoot strike runners during all conditions. Forefoot strike runners had decreased peak power absorption at the knee joint during downhill and level running conditions. Therefore, forefoot strike running reduces loading at the knee joint and could be used as an effective strategy to reduce stress at the knee joint experienced with rearfoot strike running.

Keywords: Biomechanics; Slope; Incline; Decline;

Introduction

Distance runners can be classified as either forefoot strike (FFS), midfoot strike, or rearfoot strike (RFS) runners depending on which part of their foot makes initial contact with the ground and on the location of the centre of pressure (COP) with respect to the foot. A RFS landing pattern can be defined as the heel of the foot touching the ground first and the COP located in the posterior third of the foot. A midfoot strike landing pattern can be defined as the heel and ball of the foot touching the ground simultaneously and the COP located in the middle third of the foot. Finally, a FFS landing pattern can be defined as the ball of the foot initially touching the ground and the COP located in the anterior third of the foot (Cavanagh & Lafortune, 1980).

A RFS landing pattern is the prevailing landing style used by runners. When researchers examined runners at the 15 km point of an elite level half marathon race approximately 75% of the 283 runners were categorized as RFS, less than 1.5% were categorized as FFS and the remaining were midfoot strike runners (Hasegawa et al., 2007). Although such a small percentage of the running community runs with a FFS, a recent subculture has emerged among recreational and competitive runners to adopt a FFS running style. This subculture is enhanced by several factors including Dr. Lieberman's research on barefoot running (which use midfoot and FFS landing patterns), Christopher McDougall's *Born to Run*, as well as claims of hypothesized performance advantages and decreased prevalence of lower limb injuries (Lieberman et al., 2010). A recent retrospective study showed that injury rates may be lower among FFS runners when compared to RFS runners (Daoud et al., 2012). Given that running has the highest injury rate among all sports, and between 19 – 79% of runners experiencing a running injury annually, runners have been altering landing patterns as a way to mitigate risk of injury (Hreljac, 2004). Switching from a RFS to a FFS landing pattern was used as a way to reduce patellofemoral joint pain and has been suggested as a way to treat runners with this injury (Vannatta & Kernozek, 2015). Some differences between RFS and FFS running must exist which explain the differences in injury rates.

FFS running is characterized by landing with the ankle in a plantar flexed position compared to a dorsiflexed ankle in RFS runners. When observing the vertical ground reaction force (GRF) pattern, RFS runners have an impact peak that occurs 25 ms after initial contact whereas this impact peak is absent in FFS runners. A study which evaluated lower extremity overuse injury potential found that a low impact force peak was the primary biomechanical variable which distinguished the injury-free from the injured group of runners (Hreljac et al.,

2000). FFS running also has greater plantar flexion moments and power absorption at the ankle joint, whereas larger peak knee flexion moments and power absorption at the knee were observed in habitual RFS runners (Hamill et al., 2014; Kulmala et al., 2013; Stearne, Alderson, Green, Donnelly, & Rubenson, 2014; D. S. Williams et al., 2000). These differences in impact peaks, joint moment and power at the knee joint may explain why FFS runners may possess less of an injury risk compared to RFS running.

Most of our knowledge on FFS running biomechanics is limited to level overground or treadmill conditions, with little to no research being done on uphill or downhill running. Runners encounter hills on a regular basis when running outdoors. Downhill running at 9° increased impact force peaks by 54% compared to the level condition in RFS runners, suggesting that downhill running may have greater risk of injury compared to level or uphill running (Gottschall & Kram, 2005; Hreljac et al., 2000). Uphill and downhill running has been shown to not effect sagittal joint moments between slopes of $\pm 5^\circ$, however disagreements in the literature exist for joint powers (Buczek & Cavanagh, 1990; Telhan et al., 2010). One study found no change in knee power absorption between level and downhill 4.7° running, but did find a 56% increase in peak power absorption at the ankle joint (Buczek & Cavanagh, 1990). Whereas the other study found a 30% increase in knee power absorption and a large increase in hip power absorption during downhill 4° running and an increase in hip power generation during uphill 4° running (Telhan et al., 2010). Future research is needed to expand our understanding of lower extremity kinetics for both RFS and FFS runners when running at even greater slopes.

Although previous studies which examined joint kinetics provided valuable knowledge concerning running biomechanics (Buczek & Cavanagh, 1990; Hamill et al., 2014; Kulmala et al., 2013; Lieberman et al., 2010; Stearne et al., 2014; Telhan et al., 2010; D. S. Williams et al.,

2000), none have provided a comprehensive analysis of both habitual FFS and habitual RFS runners during level, uphill and downhill running conditions. Many studies which compared running biomechanics between RFS and FFS runners had RFS runners change to a FFS landing pattern or incorrectly classified some midfoot strike runners as FFS (Kulmala et al., 2013; D. S. Williams et al., 2000; D. S. Williams et al., 2012). It is quixotic to believe that RFS runners switching to a FFS for a single testing session would have indistinguishable biomechanical differences from habitual FFS runners. To the authors' knowledge hill running biomechanics is limited to data collected on RFS runners, and no extensive study exists which includes kinetic information from FFS for all lower extremity joints. Published work has demonstrated the biomechanical differences existed between RFS and FFS runners during level running, so research needs to establish if these differences continue to exist during downhill and uphill running conditions as well. One study unintentionally reported some GRF data on midfoot strike and FFS runners, however this was due to the RFS runners adapting to a more anterior landing pattern on the steeper uphill slopes of 9° (Gottschall & Kram, 2005). It is surprising that no study has been completed which comprehensively examined the kinetics of lower limb during sloped running as they can all play a role in injury susceptibility, muscle fatigue, overall limb mechanics and finally running performance.

The purpose of this study is to examine the differences in kinematics and kinetics of the lower limb in RFS and FFS runners during overground level, uphill and downhill running conditions. More specifically, we want to examine biomechanical differences of the lower limb between level and hill running independent of foot strike, and also compare biomechanical measures of the lower limb between RFS and FFS runners at each slope condition. When comparing hill conditions to the level condition, we hypothesize that (1) peak knee flexion angle

will decrease during downhill conditions and increase during uphill running and (2) peak joint moments and peak power absorptions of the lower limb will increase during downhill conditions and decrease during uphill conditions. When comparing RFS and FFS runners at each slope, we hypothesize that (3) FFS runners will have increased plantar flexion moments and power absorption at the ankle, and decreased power absorption at the hip and knee joints compared to RFS runners. This study will add our understanding to FFS running biomechanics which is limited mainly to level running conditions because to our knowledge it is the first study of the kinematics and kinetics of lower limb of uphill and downhill running with consideration of landing patterns. The findings from this study would provide scientific recommendation to runners and shed light on the prevention of running-related injuries.

Methods

Participants

Participants were recruited at races in the Ottawa area and through posters placed around the university and running clubs in the Ottawa area. All prospective runners had to be between the ages of 18 – 35 years of age, run a minimum of 15 km/week, and be free from musculoskeletal injuries 6 months prior to participation. Potential participants came lab for a preliminary screening session where we analyzed their landing pattern via strike index (Buczek & Cavanagh, 1990). Participants were given a pair of Merrell Vapor Glove minimalist running shoes with reflective markers placed on the heel, medial and lateral malleoli, and between the 2nd and 3rd metatarsal heads. They were instructed to run overground at $3 \pm 5\%$ m/s and strike an imbedded force plate 4m away from the starting line.

The participants whose foot centre of pressure at foot strike was located in the anterior third of the foot were placed into the forefoot strike group, and the participants whose foot center

of pressure that was located in the posterior third of the foot were placed into the rearfoot strike group. Runners who landed with a midfoot strike landing pattern were excluded from participation as their foot strike is highly variable and characteristic of both a FFS and RFS landing pattern (Lieberman, 2012). Two groups, FFS and RFS group, with each 15 runners were formed. Table 1 provides the anthropometric data of the participants. Written informed consent following the guidelines of the Health Sciences and Science Research Ethics Board at the University of Ottawa was given by all participants prior to any data collection.

Table 1: Anthropometric data of the participants (mean $\pm$ SD)

	FFS Runners (n = 15)	RFS Runners (n = 15)	Total (n = 30)
Age (year)	26.1 $\pm$ 4.9	26.8 $\pm$ 4.5	26.5 $\pm$ 4.7
Height (m)	1.77 $\pm$.06	1.79 $\pm$.06	1.78 $\pm$ 0.05
Mass (kg)	74.9 $\pm$ 10.9	78.7 $\pm$ 9.1	76.8 $\pm$ 10.0

Instrumentation

We constructed a 3m steel ramp with embedded force platform that could be raised to form slopes of 6° and 9° (10.5% and 15.8%). A 4.7m wooden platform was attached to the ramp to provide a running surface for participants at the top of the ramp. Wheel jacks were placed on the outside of the ramp beside the force platform and a car jack was placed under the ramp below the force platform to help reduce vibrations and bending of the ramp (Figure 1). The force platform (model 9286AA, Kistler Instruments Corp, Winterthur, Switzerland) was placed approximately 2/3 of the way up the ramp and approximately 4m away from the starting line during the level trial. Force platforms captured GRF at a frequency of 1000 Hz. A 10 camera Vicon Motion Analysis System (MX-13, Oxford Metrics, Oxford, UK) recorded movement at 250 Hz for motion analysis. All participant wore the same Merrell Vapor Glove minimalist

running shoes as during the preliminary screening. Each shoe was approximately 165 ± 11 g and had 5 mm of cushioning and a heel-to-toe drop of 0 mm.

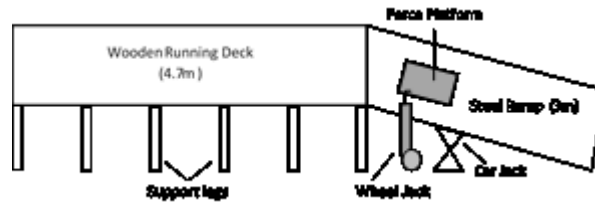


Figure 8: Ramp with embedded force platform mounted at 9°.

Data collection

Participants returned for the main testing session one week after their preliminary session. Participants were outfitted in 45 reflective markers according to the University of Ottawa Motion Model (Lamontagne et al., 2009) (modified Plug-in Gait Model). Testing included 5 different slope conditions (downhill 9°, downhill 6°, level, uphill 6° and uphill 9°) completed in a quasi-random order; both 6° and both 9° conditions were completed consecutively due to the lengthy process of changing the angle of the ramp. Running speed was set at $3 \pm 5\%$ m/s (5.6 min/km, 8.9 min/mi), which was monitored using custom built photocells. Participants started running approximately 4m away from the force plate at each condition.

Data Processing and Analysis

Kinematic and kinetic data was cropped and time normalized to 100% stance phase. GRF data of stance phase was filtered using a fourth-order recursive, zero phase-shift, Butterworth low-pass filter with a cut-off frequency of 50 Hz and kinematic data was filtered using a Woltring 15 MSE filter (Woltring, 1986). Modelled data was further processed and analyzed in Matlab R2013a (MathWorks, Natick, MA, USA) to retrieve biomechanical information.

For each condition, sagittal peak joint angles (°) and the angles (°) at foot initial contact of ground, peak sagittal joint moments (N·m/kg), and peak joint power absorption and generation (W/kg) for the hip, knee and ankle joints were determined. Total power absorption and generation were obtained by integrating all the negative or positive values of the joint powers during stance phase, yielding an average total power of each joint during the two phases of stance phase.

To obtain lower limb kinematic measurements, Vicon Nexus Software (version 1.8) using the UOMAM model was used to retrieve sagittal joint angles of the hip, knee and ankle joints for FFS and RFS groups during the 5 examined slope conditions. Measurements of initial contact and toe-off were defined using force platform values. Initial contact was defined as the moment when the vertical GRF data first became greater than 20 N and toe off was defined after it fell below 20 N.

A two-way ANOVA using SPSS 20 (SPSS Inc. Chicago, IL, USA) was used to examine the effects of slope and landing pattern on all measured variables, with slope and landing pattern used as separate factors. All significant slope-by-landing pattern interactions were reported. If a significant slope-by-landing pattern interaction was present, a Bonferroni *post hoc* test was conducted in order to determine where these differences occurred. Levene's test revealed that not all variables had homogeneity of variance, therefore, F ratios were considered significant at $p < .01$. In the tables, the data is presented as means and SD (standard deviation).

Results

The study investigated joint angle, joint moment and joint power in RFS and FFS runners as they ran at $3 \text{ m/s} \pm 5 \%$ during overground level, uphill and downhill running and compared the measures between RFS and FFS. Overall, a significant landing effect ($p < .01$) was observed

in angles of hip, knee and ankle joint, joint moment of knee and ankle joint moment and power of hip, knee and ankle, but not in peak knee extension moments, peak hip flexion moments and peak hip extension moments. A significant slope effect ($p < .01$) was observed in joint angles, joint moment and power of the hip, knee and ankle, except in peak hip extension angle. There was a significant slope-by-landing pattern interaction ($p < .01$) in the ankle angle at initial contact, peak plantar flexion angle, peak hip flexion moment and all joint power measurements. *Post hoc* tests results are presented as following.

Level versus slope

At initial contact, hip flexion angle significantly decreased during downhill trials and increased during uphill conditions compared to the level trial for both RFS and FFS runners. At the knee joint, knee flexion angle significantly increased during uphill trials for both groups of runners. The ankle joint did not follow a consistent pattern during different hill conditions in both groups. Plantar flexion angle significantly decreased as slope increased in FFS runners, however, dorsiflexion angles in RFS runners did not show any significant change as slope increase (Table 2).

Peak hip flexion angle significantly decreased during both downhill conditions and increased only for the uphill 9° condition in both groups of runners. No significant differences existed in peak hip extension angle and peak knee flexion angles. Peak ankle dorsiflexion angle was significantly lower during the downhill 9° condition and greater during the uphill 9° condition in both RFS and FFS runners. Peak ankle plantar flexion angle did not follow the same trend like dorsiflexion as running in slope conditions in both groups. It decreased during downhill conditions in the RFS group, but no differences existed in the FFS group (Table 3).

Table 2: Sagittal joint angles of the hip, knee and ankle at initial contact for FFS and RFS groups during the 5 examined slope conditions (means $\pm$ SD)

Condition (°)	Hip Flexion Angle (°)	Knee Flexion Angle (°)	Ankle Angle (°)
FFS			
-9	30.8 $\pm$ 5.4 [†]	11.2 $\pm$ 6.0	-27.9 $\pm$ 6.3 [†]
-6	32.0 $\pm$ 5.7 [†]	11.1 $\pm$ 5.4	-25.1 $\pm$ 5.8 [†]
0	38.9 $\pm$ 5.5	13.9 $\pm$ 5.6	-16.1 $\pm$ 5.5
6	42.3 $\pm$ 5.3	21.2 $\pm$ 6.0 [†]	-10.7 $\pm$ 6.1 [†]
9	46.4 $\pm$ 4.4 [†]	24.5 $\pm$ 5.3 [†]	-7.2 $\pm$ 5.3 [†]
RFS			
-9	27.1 $\pm$ 7.5 ^{*†}	11.5 $\pm$ 4.9	4.4 $\pm$ 3.6 [*]
-6	27.5 $\pm$ 6.5 ^{*†}	10.9 $\pm$ 4.6	4.5 $\pm$ 3.7 [*]
0	34.7 $\pm$ 6.4 [*]	13.3 $\pm$ 5.4	6.3 $\pm$ 1.7 [*]
6	39.8 $\pm$ 7.8 [†]	18.5 $\pm$ 5.3 [†]	6.7 $\pm$ 1.9 [*]
9	43.8 $\pm$ 8.1 [†]	23.1 $\pm$ 5.7 [†]	8.0 $\pm$ 2.2 [*]

The two-way ANOVA revealed statistically significant ($p < .01$) in all variables for effect of landing and statistically significant differences ($p < .01$) in all variables for effect of slope. There was a statistically significant ($p < .01$) slope-by-landing pattern interaction for ankle angle. Post hoc comparisons on for between group differences are indicated by * and represent differences between RFS and FFS at each slope condition, whereas within group differences are represented by [†] and represent parameters that are significantly different from the level trial for that foot strike. Note negative ankle angle represents plantar flexion and positive ankle angle represents dorsiflexion.

Comparing level running, downhill 9° running significantly effects hip joint moment in both FFS and RFS runners, which significantly increased peak hip flexion moment and significantly decreased peak hip extension moment was showed during the downhill conditions in the FFS group. No differences existed between hill conditions in peak knee extension moments in both groups, but peak knee flexion moment significantly decreased during the downhill 9° condition. Peak ankle plantar flexion moments were significantly lower in the downhill conditions in the FFS group, whereas peak ankle dorsiflexion moments were only larger during the downhill condition in the RFS group (Table 4).

Table 3: Peak sagittal joint angles of the hip, knee and ankle for FFS and RFS groups during the 5 examined slope conditions (means $\pm$ SD)

Condition (°)	Peak Hip Flexion Angle (°)	Peak Hip Extension Angle (°)	Peak Knee Flexion Angle (°)	Peak Ankle Dorsiflexion Angle (°)	Peak Ankle Plantar flexion Angle (°)
FFS					
-9	30.8 $\pm$ 5.4 [†]	6.8 $\pm$ 7.0	37.6 $\pm$ 6.8	15.1 $\pm$ 4.0 [†]	27.9 $\pm$ 6.3
-6	32.4 $\pm$ 5.5 [†]	7.0 $\pm$ 6.9	38.4 $\pm$ 5.8	15.9 $\pm$ 3.7	25.3 $\pm$ 5.6
0	38.9 $\pm$ 5.4	7.7 $\pm$ 6.7	39.3 $\pm$ 4.4	18.6 $\pm$ 3.8	27.5 $\pm$ 6.3
6	42.3 $\pm$ 5.3	5.6 $\pm$ 5.8	39.5 $\pm$ 5.2	21.2 $\pm$ 4.0	25.9 $\pm$ 5.7
9	46.4 $\pm$ 4.4 [†]	4.6 $\pm$ 5.3	42.0 $\pm$ 4.7	22.6 $\pm$ 4.0 [†]	25.1 $\pm$ 5.7
RFS					
-9	29.5 $\pm$ 7.4 [†]	10.3 $\pm$ 7.4	43.0 $\pm$ 4.5*	20.5 $\pm$ 3.8 ^{*†}	12.7 $\pm$ 5.9 ^{*†}
-6	29.4 $\pm$ 7.2 ^{*†}	12.2 $\pm$ 6.5	40.7 $\pm$ 6.2	20.0 $\pm$ 3.7 ^{*†}	14.8 $\pm$ 6.8 ^{*†}
0	35.9 $\pm$ 5.8*	14.2 $\pm$ 7.3	43.4 $\pm$ 5.2*	23.7 $\pm$ 4.1*	20.8 $\pm$ 5.4*
6	39.9 $\pm$ 7.8	12.3 $\pm$ 6.5	42.2 $\pm$ 4.6	27.1 $\pm$ 3.2 ^{*†}	21.8 $\pm$ 5.7*
9	43.8 $\pm$ 8.0 [†]	12.4 $\pm$ 7.1	44.2 $\pm$ 5.0	29.4 $\pm$ 3.0 ^{*†}	22.3 $\pm$ 6.4*

The two-way ANOVA revealed statistically significant ($p < .01$) in all variables for effect of landing and statistically significant differences ($p < .01$) in all variables except hip extension for effect of slope. There was a statistically significant ($p < .01$) slope-by-landing pattern interaction for ankle plantar flexion angle. Post hoc comparisons on for between subject differences are indicated by * and represent differences between RFS and FFS at each slope condition, whereas within differences are represented by [†] and represent parameters that are significantly different from the level trial for that foot strike.

Total lower extremity power absorption generally decreased as slope increased, but only reached significance in the FFS group for the downhill and uphill 9° conditions (Figure 2a). Total lower extremity power generation generally increased as slope increased, but only reached significance in the downhill 9° condition for both groups and both uphill conditions for the RFS group (Figure 2b). Peak hip power absorption generally decreased as slope increased, however it was only significantly different in the uphill conditions for FFS runners and the downhill 9° condition for RFS runners. Peak knee power absorption was significantly greater in the downhill 9° condition and uphill conditions, but only in the RFS group. No differences existed between hill conditions in either group for peak ankle power absorption (Figure 3a). Peak hip power

generation was greater in both groups during the uphill 9° condition but decreased during the downhill conditions only in the FFS group. Peak knee power generation was significantly lower in the downhill 9° condition for both groups but no differences occurred during uphill running. Finally, peak ankle power generation was significantly lower during the downhill conditions but only in the FFS group (Figure 3b).

Table 4: Peak sagittal joint moments of the hip, knee and ankle for FFS and RFS groups running at the 5 examined slope conditions (means $\pm$ SD)

Condition (°)	Peak Hip Flexion Moment (N·m/kg)	Peak Hip Extension Moment (N·m/kg)	Peak Knee Flexion Moment (N·m/kg)	Peak Knee Extension Moment (N·m/kg)	Peak Ankle Plantar flexion Moment (N·m/kg)	Peak Ankle Dorsiflexion Moment (N·m/kg)
FFS						
-9	1.05 $\pm$.38 [†]	1.59 $\pm$.62 [†]	.43 $\pm$.14 [†]	2.01 $\pm$.57	2.33 $\pm$.31 [†]	.023 $\pm$.03
-6	.88 $\pm$.38	1.87 $\pm$.34 [†]	.52 $\pm$.18	2.06 $\pm$.59	2.44 $\pm$.32 [†]	.001 $\pm$.03
0	.70 $\pm$.32	2.36 $\pm$.43	.64 $\pm$.24	1.82 $\pm$.44	2.99 $\pm$.30	.001 $\pm$.03
6	.45 $\pm$.18 [†]	2.07 $\pm$.50	.61 $\pm$.21	1.36 $\pm$.44	3.10 $\pm$.35	.003 $\pm$.03
9	.52 $\pm$.27	2.10 $\pm$.35	.59 $\pm$.18	1.50 $\pm$.45	2.89 $\pm$.28	.003 $\pm$.03
RFS						
-9	.91 $\pm$.37 ^{*†}	1.70 $\pm$.78	.28 $\pm$.21 ^{*†}	1.97 $\pm$ 1.19	1.50 $\pm$.90 [*]	.53 $\pm$.34 ^{*†}
-6	.89 $\pm$.28	1.70 $\pm$.65	.33 $\pm$.16 [*]	1.96 $\pm$ 1.10	1.62 $\pm$.89 [*]	.45 $\pm$.25 ^{*†}
0	.68 $\pm$.18	2.09 $\pm$.93	.45 $\pm$.17 [*]	1.66 $\pm$ 1.21	1.68 $\pm$.16 [*]	.24 $\pm$.16 [*]
6	.63 $\pm$.34 [*]	2.00 $\pm$.76	.47 $\pm$.18 [*]	1.42 $\pm$.91	1.87 $\pm$ 1.10 [*]	.26 $\pm$.20 [*]
9	.59 $\pm$.23	2.03 $\pm$.82	.50 $\pm$.19	1.32 $\pm$.86	1.94 $\pm$ 1.15 [*]	.20 $\pm$.22 [*]

The two-way ANOVA revealed statistically significant ($p < .01$) in all variables except hip flexion and extension moments and knee extension moments for effect of landing. There were statistically significant differences ($p < .01$) in all variables for effect of slope. There was no statistically significant ($p < .01$) slope-by-landing pattern interaction for any sagittal joint moments. Post hoc comparisons on for between subject differences are indicated by * and represent differences between RFS and FFS at each slope condition, whereas within differences are represented by [†] and represent parameters that are significantly different from the level trial for that foot strike.

FFS versus RFS

At initial contact, FFS runners had significantly increased hip flexion compared to RFS runners during the level and both downhill conditions. No differences existed in knee flexion

angles. At the ankle, FFS runners maintained a plantar flexed ankle for all conditions whereas RFS runners maintained a dorsiflexed angle for all conditions (Table 2).

For peak sagittal joint angles, the greatest differences between the two groups occurred at the ankle joint. RFS runners had greater peak dorsiflexion angles for all conditions while FFS runners had increased peak plantar flexion angles. For peak knee flexion, differences only occurred during the downhill 9° and level condition, with RFS runners having greater peak knee flexion during those two conditions. No differences existed in peak hip extension but in peak hip flexion FFS runner had significantly greater values compared to RFS runners during the level and downhill 6° condition (Table 3).

FFS runners had significantly greater peak hip flexion moments during the downhill 9° condition while the RFS group had significantly larger hip flexion moments during the uphill 6° condition. No differences existed in hip extension moments. FFS runners had significantly larger peak knee flexion moments in all conditions except the uphill 9° condition, but no differences existed in peak knee extension moments. FFS runners had significantly larger peak ankle plantar flexion moments in all conditions, whereas RFS runners had significantly larger peak ankle dorsiflexion moments in all conditions (Table 4).

Total lower extremity power absorption was greater in FFS runners during all slope conditions (Figure 2a), while total lower extremity power generation was greater in FFS runners in all the conditions except the downhill 9° condition (Figure 2b). Peak hip power absorption was only significantly different between the two groups during the uphill 6° condition, while RFS runners had greater peak knee power absorption in the level and both downhill conditions. Peak ankle power absorption was much greater in FFS runners during all slope conditions (Figure 3a). FFS runners had greater peak hip power generation in all conditions except the uphill 6°

condition. No differences existed in peak knee power generation between the two groups (Figure 3b).

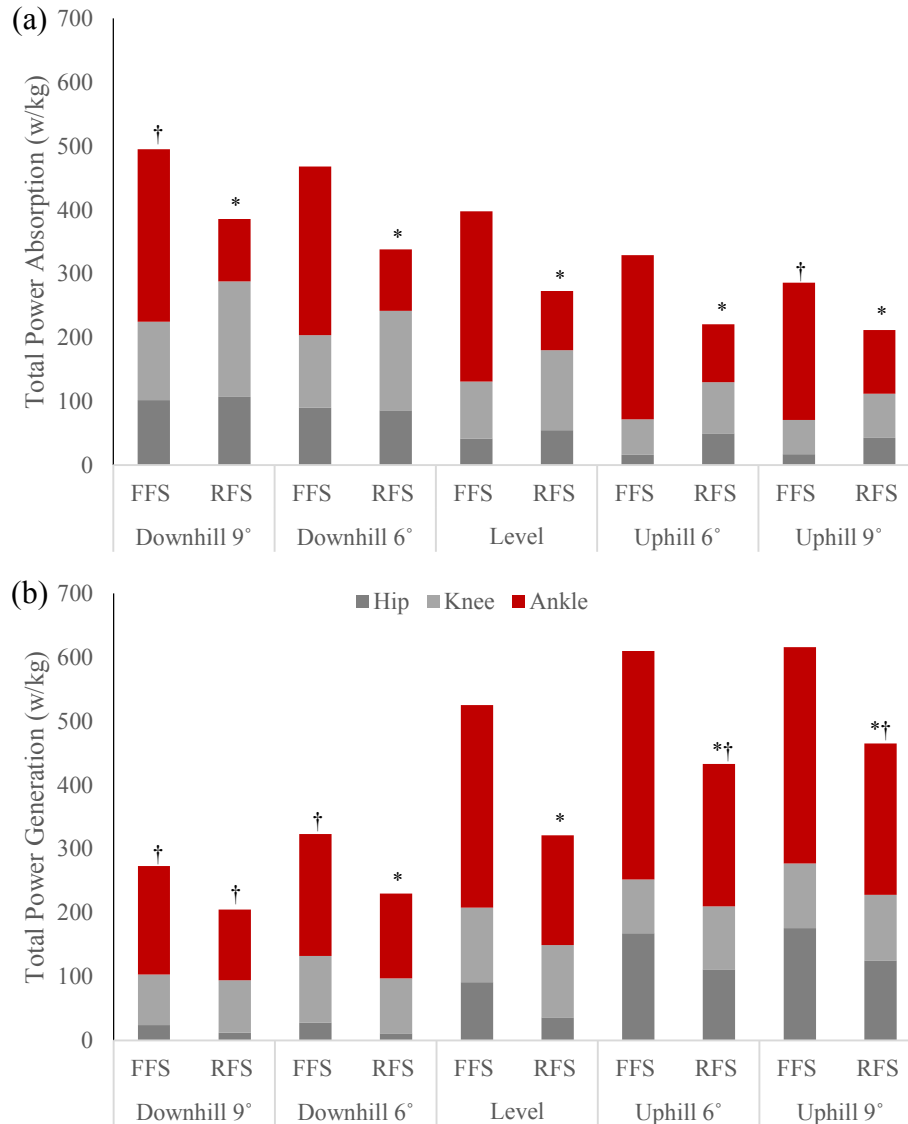


Figure 9: Total power (W/kg) (a) absorption and (b) generation for RFS and FFS runners during downhill and uphill running at 3 m/s. The two-way ANOVA revealed statistically significant ($p < .01$) in both variables for effect of landing and for effect of slope. There was a statistically significant ($p < .01$) slope-by-landing pattern interaction. Post hoc comparisons on for between subject differences are indicated by * and represent differences between RFS and FFS at each slope condition, whereas within differences are represented by † and represent parameters that are significantly different from the level trial for that foot strike.

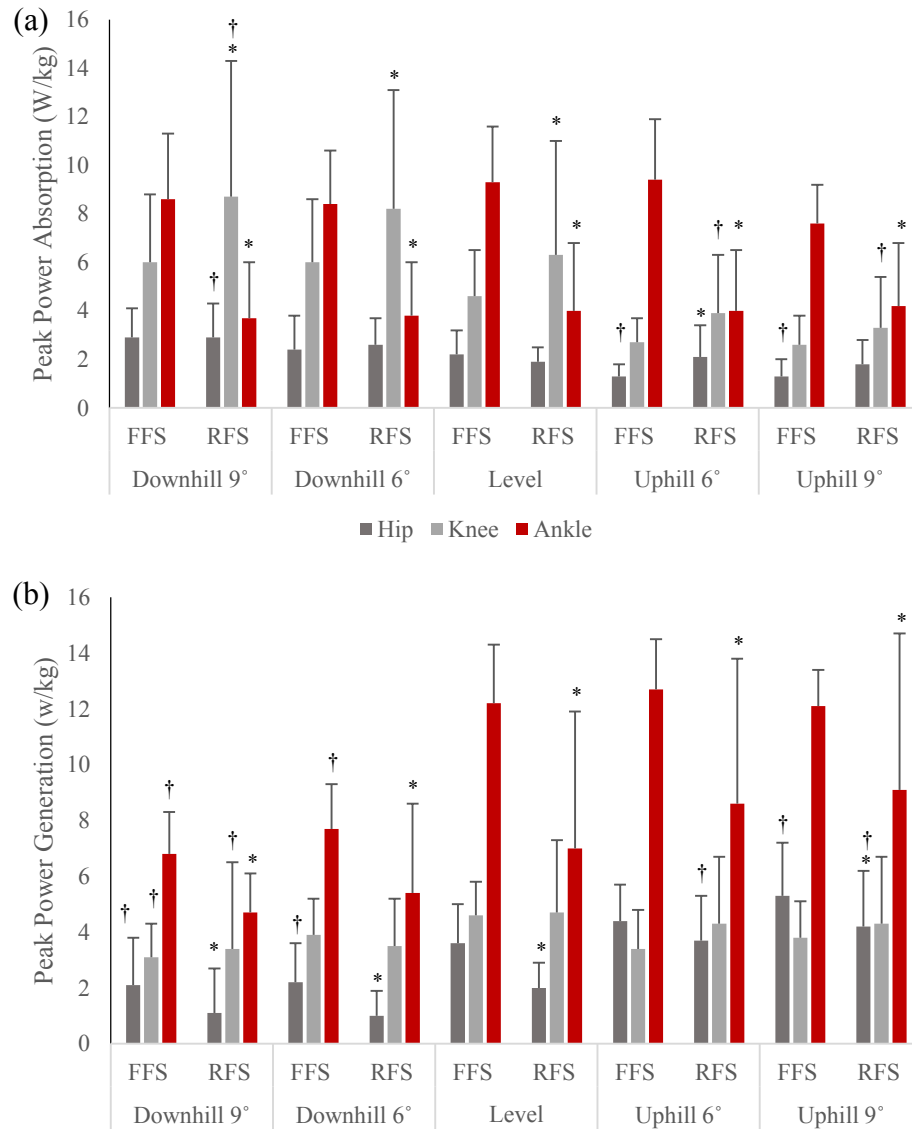


Figure 3: Peak joint power (W/kg) (a) absorption and (b) generation for RFS and FFS runners during downhill and uphill running at 3 m/s. The two-way ANOVA revealed statistically significant ($p < .01$) in both variables for effect of landing and for effect of slope. There was a statistically significant ($p < .01$) slope-by-landing pattern interaction. Post hoc comparisons on for between subject differences are indicated by * and represent differences between RFS and FFS at each slope condition, whereas within differences are represented by † and represent parameters that are significantly different from the level trial for that foot strike.

Discussion

The study was to examine the mechanics of the lower extremity in habitual RFS and FFS as they ran overground on level and hill surfaces. Furthermore, we wanted to investigate the

differences in kinetics through different hill conditions in RFS and FFS respectively, and to compare if differences in kinetics exist between RFS and FFS runners at each condition. The findings of this study showed that RFS and FFS runners demonstrated similar mechanical changes during downhill and uphill conditions. Specifically, hip and knee flexion angles at initial contact increased, peak hip flexion and peak ankle dorsiflexion angles increased, total power generation increased and total power absorption decreased as conditions changed from downhill 9° to uphill 9°, respectively. These findings are similar to those reported on RFS runners during sloped running (Buczek & Cavanagh, 1990; Telhan et al., 2010).

Level versus slope

It was predicted that peak knee flexion angle would be lower during downhill conditions and greater during the uphill conditions. This hypothesis was inferred from previous temperospatial research which found that stride length decreased during uphill running and increased during downhill running (Gottschall & Kram, 2005; Hannon et al., 1985; Padulo et al., 2013), so it was predicted that the shorter stride experienced with uphill running would result in greater peak flexion at the knee. Our findings deny our first hypothesis since peak knee flexion was not significantly ($p < .01$) during the uphill or downhill conditions in either the RFS or FFS group.

We were unable to measure stride length in our study since the length of our ramp and placement of the force platform did not allow all participants to have a complete gait cycle entirely on the ramp. However, if our runners followed similar stride length modifications during hill running it must be due to other kinematic variables which cannot be explained by the peak knee flexion angle alone. At initial contact, hip flexion angle increased as slope increased from the downhill 9° to the uphill 9° condition, while knee flexion angle significantly increased during

the uphill conditions (Table 2). Although hip flexion increases, knee flexion also increased during uphill running which would ultimately shorten a runner's stride. This could help explain why stride length has been reported to be longer during downhill conditions and shorter during uphill conditions (Gottschall & Kram, 2005; Hannon et al., 1985; Padulo et al., 2013).

We hypothesized that peak moments and peak power absorptions of the lower limb would increase during downhill conditions and decrease during uphill conditions in both RFS and FFS groups. Previous research has not found any significant changes in joint moments between level and hill running in RFS runners, but they used slopes of between $\pm 5^\circ$ (Buczek & Cavanagh, 1990; Telhan et al., 2010). We anticipated that as slopes became greater that differences in peak moments would emerge. Several studies have found that power absorption increased with a decrease in slope and power generation increased with an increase in slope (Buczek & Cavanagh, 1990; Telhan et al., 2010), so it was thought that changes would occur at all lower extremity joints during hill running. Our findings did not support our second hypothesis as there were no consistent changes in peak sagittal moments or in peak joint powers.

Resultant joint moments have been shown to be possible indirect indicators of joint loading (Andriacchi, 1994). During stance phase, joint moments of the lower extremity show a primarily extensor pattern with the relative timing of these extensor peaks occurring at different times for each joint (Winter, 1983). The hip peaks at 20% stance, the knee at 40% and the ankle near 60% stance (Winter, 1983). The hip moment reverses from an extensor moment to a flexor moment before midstance to decelerate the backward rotating thigh's direction to push it forward into swing. After initial contact the knee flexes under the influence of weight bearing until midstance, and this flexion is stopped due to the peak knee extensor moment. Finally, the ankle

joint develops a large plantar flexion moment shortly after midstance when the ankle is dorsiflexed with the foot flat on the ground and the leg rotates over it (Winter, 1983).

Joint moments were only significantly different during downhill running conditions (Table 4). Peak hip flexion moment was significantly greater whereas peak knee flexion moment was significantly lower during the downhill 9° condition in both RFS and FFS runners. Peak hip extension moment and peak ankle plantar flexion moment were significantly lower in both downhill conditions compared to the level condition in FFS runners. Both downhill conditions resulted in a significant increase in peak ankle dorsiflexion moments in RFS runners.

Joint moments are influenced by the moment arm, muscle contraction force, and joint angles. The differences we found during downhill running can be linked to changes in moment arm and muscle contraction forces. A study by Cai and colleagues (2010) which examined muscle activity during uphill and downhill running found that downhill running resulted in a decrease in muscle activity in the biceps femoris and rectus femoris during downhill running, but increased muscle activity in the gastrocnemius, and caused no differences in the tibialis anterior (Cai et al., 2010). Since our study found a decrease in peak hip flexion angle during the downhill 9° condition and their results found a decrease in muscle activity, the increase in peak hip flexion moment must be due to an increase in moment arm. Our results showed no differences in peak knee joint angles, so like with the peak hip flexion moment, the change in peak knee flexion moment during the downhill 9° condition must be due to a change in the moment arm. Differences in peak moments at the ankle joint should also be explained by differences in moment arm.

Joint moments can be used to predict muscle joint loading since $F_{\text{muscle}} = M / (r \sin \theta)$, where N is the net moment of force at the joint, r is the distance from the insertion point of the

muscle to the joint centre, and θ is the angle of the muscle's line of action and the position vector between the joint centre and muscle's insertion point (Robertson, Gordon & Whittlesey, Saunders, 2004). Since there is an increase in some joint moments during downhill running, it should be associated to increase in muscle joint loading.

Joint powers are produced from the net joint moments determined through inverse dynamics. These moments represent the muscular response to moments applied to skeletal segments from the external forces which include ground and joint reaction forces, segmental weights and inertial torques (Alexander, 1991; Elftman, 1939; Roberts & Belliveau, 2005; Winter, 1983). Peak joint power absorption was greatest during the downhill conditions as the muscles contract eccentrically for most of the stance phase (Abbott et al., 1952). Peak joint power generation was smallest during downhill conditions and largest during uphill conditions, as was to be expected, due to the increased metabolic cost of uphill running (Margaria, 1976; Minetti et al., 1994). Although total power absorption decreased with increases in slope, only the downhill 9° and uphill 9° conditions were significantly different from the level condition in FFS runners, with not slope differences occurring in RFS runners. Total power generation was significantly lower among both groups during the downhill 9° condition, but the uphill conditions were only significantly different in the RFS group. The ankle joint generated the most power among all lower extremity joints in all test conditions for both RFS and FFS runners (Figure 3b). Work by Roberts & Belliveau (2005) found that the mechanical power necessary for uphill running was achieved through an increase in power generation at the hip joint. We did find a significant increase in peak power generation in the hip joint for both RFS and FFS runners during uphill 9° compared to the level condition.

FFS versus RFS

In the published related studies FFS running biomechanics was only examined during level running. The research findings from this study provide information about lower limb biomechanics of FFS runners during uphill and downhill running. The significant differences in joint angles of the lower limb at initial contact between RFS and FFS runners are presented in hip and ankle joints, but not the knee joint. Moreover, differences in hip angle between groups existed in downhill and level running conditions, but not uphill conditions. While ankle joint angles showed significant differences in all running conditions between two groups. As was expected, at initial contact RFS runners maintained a dorsiflexed ankle and FFS runners maintained a plantar flexed ankle during all conditions. These ankle touchdown angles are in agreement with previous research (Lieberman et al., 2010; Shih et al., 2013; D. S. Williams et al., 2012). However, discrepancies exist in the literature with hip and knee angles at initial contact. Several researchers did not find any differences at these two joints, whereas Shih and colleagues found the hip to be 6% less flexed and the knee 66% more flexed when running in level condition with a FFS versus a RFS (Kulmala et al., 2013; Shih et al., 2013; D. S. Williams et al., 2012). Another study which only examined the ankle and knee joints found that knee angles did not show significant change between shod RFS and barefoot FFS running (Lieberman et al., 2010). Our results showed that the hip was more flexed in FFS runners than RFS runners during level and downhill running conditions and that no differences existed at the knee joint. A reason as to why our results resemble the work of Lieberman et al. is that like our participants, they had habitual RFS and FFS runners, whereas the work by the other groups used habitually RFS runners and instructed them to run with a FFS pattern (Lieberman et al., 2010; Shih et al., 2013; D. S. Williams et al., 2012). The study by Kulmala and colleagues classified some midfoot

strike runners as FFS runners, so caution must be used when applying their findings to those of habitual FFS runners (Kulmala et al., 2013).

Research has shown that RFS runners who were instructed to run with FFS pattern had similar patterns to habitual FFS runners, however they only examined the knee and ankle joints (D. S. Williams et al., 2000). Although RFS runners can mimic a FFS running pattern when instructed to, it is quixotic to believe that they would be identical to habitual FFS runners. The paper of Williams et al (2012) caused the majority of research on FFS runners to mimic their method instructing RFS runners to run with a FFS pattern (Shih et al., 2013; D. S. Williams et al., 2012). Ability and state of training has been shown to be characteristics which effect biomechanical measurements, so if a group is not used to running with a FFS pattern, they will be able to replicate it, but not be indistinguishable from habitual FFS runners (K. R. Williams et al., 1989). Although habitual FFS runners make up a very small portion of the running community (Hasegawa et al., 2007), we cannot compromise the quality of future research by studying participants who do not represent this cohort of runners.

FFS runners had larger peak plantar flexion moments and smaller peak dorsiflexion moments at the ankle joint compared to RFS runners during all slope conditions (Table 4). As previously mentioned, peak plantar flexion moment occurs around 60% stance phase (Winter, 1983), but peak dorsiflexion moment occurs at different points for FFS and RFS runners. In RFS runners, peak dorsiflexion moment occurs around 10% stance phase, but in FFS runners it happens just prior to toe-off, with FFS runners displaying a plantar flexion moment for most of stance phase. These differences in ankle moments cannot explained by the range of motion (ROM) occurring in the ankle joint between both groups. During the level condition, FFS runners have approximately 46° of ankle ROM, whereas RFS runners had 44.5° of ankle ROM.

However, differences between peak dorsiflexion angles and peak plantar flexion angles differ between both groups. FFS runners have larger peak plantar flexion angles, while RFS runners have larger peak dorsiflexion angles.

Although significant differences did not exist between RFS and FFS runners in peak hip and knee extension moments, the variability in RFS runners was much greater. In table 4 we can see that the SD of the RFS group is almost double that of the FFS group for all conditions. This translated into having vary different power absorption and generation values at the hip and knee joint within the RFS group. Some of our RFS runners used more of a hip strategy for power absorption and generation, whereas other RFS runners utilized more of a knee strategy. Although we did not separate these within group differences, future research should examine possible explanations and implications to using one strategy over the other.

RFS runners absorbed most of their power through the knee joint compared to the ankle joint in FFS runners for uphill and level running. These findings are consistent with what has previously been reported on research performed on FFS runners (Lieberman et al., 2010; D. S. Williams et al., 2000; D. S. Williams et al., 2012). Knee flexion angle cannot explain this phenomenon entirely as it is not significantly different at initial contact nor is peak flexion angle consistently different between both FFS and RFS runners. FFS and RFS runners had significantly different ankle angles at initial contact; RFS runners landed with a dorsiflexed ankle, whereas FFS runners landed with a plantar flexed ankle. This causes a difference in how the forces are transmitted through the body. FFS runners have the forces transmitted through the midfoot bones and muscles rather than through the calcaneous, talus and tibia directly, as in RFS runners. This is evident when examining peak power absorption during the downhill conditions since peak ankle power absorption in FFS runners is almost identical to peak knee power

absorption in RFS runners (Figure 3a). The increased plantar flexed ankle in FFS runners results in greater shortening of the triceps surae muscle group, which may require greater muscle activation due to the compromised length-tension relationship (Winegard, Hicks, & Vandervoort, 1997). There is a transition of eccentric to concentric contraction that occurs at midstance, so this may cause the triceps surae muscle group to be further stressed at midstance (Horita, Komi, Nicol, & Kyrolainen, 1999).

In FFS runners, total power generation distribution is 17.3% hip, 22.3% knee, and 60.4% ankle during level running and 28.6% hip, 16.4% knee and 55.0% ankle during uphill 9° running, whereas for RFS runners the distribution was 11.2% hip, 35.2% knee, and 53.6% ankle during level running and 26.8% hip, 22.2% knee, and 51.0% ankle for uphill 9° running. Uphill running experiences an increased role by the hip flexors to help propel the runner, but the triceps surae muscle group is still the primary muscle group responsible for generating mechanical power.

Both the hip flexors and gastrocnemius are two-joint muscles which can transfer mechanical power from one joint to another (Bobbert, Huijing, & Schenau, 1986; Bobbert & Schenau, 1988). Explanations as to why majority of power absorption and generation occur at the ankle joint may be due to their anatomical structure which could be suited to producing force economically. Plantar flexors have relatively short fascicles, and for a given force out, short-fibred muscles are metabolically less costly than long-fibred muscles such as the hip flexors because a lower volume of muscle must be active. These muscles also undergo the stretch-shorten cycle that may reduce the energy cost of running by allowing for elastic energy storage and recovery through the Achilles tendon.

FFS runners absorb and generate the majority of their power with the ankle joint through the eccentric and concentric contractions of the triceps surae muscle group. This supports

previous findings of increased eccentric work of the gastrocnemius and soleus muscles during FFS running (Olin & Gutierrez, 2013). This dependence on this muscle group explains why many individuals who attempt running with a FFS landing pattern experience soreness in the same muscle group. This soreness has been suggested to be associated with the eccentric activity of these muscles in the absorption phase (Newham et al., 1983). However, it is because of this large eccentric contraction that allows FFS runners to generate large concentric contractions through the triceps surae muscle group. The gastrocnemius muscle can perform a greater amount of positive power if it has been stretched immediately before it is allowed to shorten (Cavagna, Saibene, & Margaria, 1965).

Total power generation and absorption were greater among FFS runners in all conditions except the downhill 9° condition compared to RFS runners (Figure 2). This disagrees with the work done by Williams and colleagues which found that a FFS landing pattern was associated with significantly ($p = .00$) lower total joint power absorption compared to the RFS landing pattern. It should be noted that their study asked RFS runners to use a FFS landing pattern, so it was a novel task for all participants. Another problem with their findings is that they just summed the peak powers as their total power absorption (D. S. Williams et al., 2012). These results are puzzling as they suggest that FFS running may be more metabolically costly since total power absorption and generation are larger than in RFS, regardless of the condition. We did not measure running economy, however a study which measured running economy in RFS and FFS runners wearing minimalist running shoes found no significant difference in cost of transport ($\text{mL O}_2 \cdot \text{kg}^{-1} \cdot \text{m}^{-1}$) (Perl, Daoud, & Lieberman, 2012). More research is needed to investigate the implications of running economy in both types of runners for both uphill and downhill running conditions.

In a recent study, RFS runners had approximately twice the rate of repetitive stress injuries than runners who habitually used a FFS (Daoud et al., 2012). Of all the running injuries runners sustain, patellofemoral pain (PFP) is among the most prominent and can be defined as anterior knee pain or retro-patellar pain in the absence of other specific pathology (Crossley, Bennell, Green, & McConnell, 2001). One of the risk factors for PFP can be linked to impact peak associated with RFS running. FFS running is characterized by an absent impact peak during level running. Since FFS running has an absent impact transient and reduces the power absorption through the knee joint, it may prove to be an effective strategy for managing PFP. Current research has found that altering one's strike pattern from a RFS to a FFS pattern resulted in consistent reductions in patellofemoral joint stress (Vannatta & Kernozek, 2015). Due to the increased load placed on the triceps surae group, it is recommended that the transition from RFS to FFS running be done in a graduated manner.

Conclusion

In summary, hill running has similar impacts on joint angles in RFS and FFS runners, causing a decrease in hip flexion at initial contact during downhill running, an increase in knee flexion angle at initial contact during uphill running and a decrease in peak hip flexion angle. In addition to the most observable difference in ankle joint angle due to landing pattern difference between RFS and FFS runners, FFS runners had a more flexed hip angle during downhill running. Peak hip flexion moment was significantly greater while peak knee flexion moment was significantly lower in both groups during the downhill 9° condition. FFS runners had larger peak plantar flexion moments and peak ankle power absorption compared to RFS runners during all conditions. FFS runners had decreased peak power absorption at the knee joint during downhill and level running conditions. Therefore, FFS running reduces loading at the knee joint during

level and downhill conditions and can be used as an effective strategy to reduce stress at the knee joint experienced with RFS running. Since uphill running has been reported to have the lowest risk of overuse injuries due to the decreased impact peak, runners do not need to switch their landing pattern as no significant differences in knee joint loading occur between FFS and RFS runners.

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Chapter 6

General Discussion and Conclusion

We set out to examine the differences in kinematics and kinetics of the lower limb in RFS and FFS runners during overground level, uphill and downhill conditions. Furthermore, we wanted to investigate the differences of lower limb biomechanics between hill conditions in RFS and FFS respectively, and finally to compare if differences in the measures existed between RFS and FFS runners at each condition. The most remarkable differences existed between RFS and FFS runners which are likely linked to foot placement at initial contact as FFS runners maintained a plantar flexed ankle, whereas RFS runners maintained a dorsiflexed ankle for all conditions.

Level versus slope

We compared the uphill and downhill conditions to the level condition independent of foot strike and predicted that (i) peak knee flexion angle would decrease during the downhill conditions and increase during the uphill conditions, and (ii) the active loading rate, active GRF peak, peak moments and peak power absorptions would increase during the downhill conditions and decrease during the uphill conditions. Our findings deny our first hypothesis as peak knee flexion did not significantly change in either RFS or FFS runners in any hill condition. Our findings partially support our second hypothesis as active loading rates only changed in RFS runners during the downhill conditions and active GRF peaks only changed in FFS runners during the downhill conditions. Peak hip flexion moment significantly increased and peak knee flexion moment significantly decreased during the downhill 9° condition in both RFS and FFS

runners. Peak power absorptions did not change consistently among all three joints during hill running.

When examining level versus slope conditions, several kinematic alterations occurred. It was inferred that peak knee flexion would increase during uphill conditions since previous temperospatial research found that stride length decreased during uphill running, which we predicted was the result of increased knee flexion (Gottschall & Kram, 2005; Hannon et al., 1985; Padulo et al., 2013). Although we were unable to measure stride length in our study since the length of our ramp did not allow for a complete gait cycle, if our runners followed similar stride length modifications during hill running than other kinematic adaptations must exist than just changes at the knee joint. At initial contact, hip and knee flexion angle increased during uphill running which would ultimately shorten a runner's stride. This could help explain why stride length has been found to be shorter during uphill conditions and longer during downhill conditions (Gottschall & Kram, 2005; Hannon et al., 1985; Padulo et al., 2013).

When examining level versus slope conditions, several kinetic alterations occurred. Although previous research has found that peak sagittal moments did not change during hill running between the angles of $\pm 5^\circ$ (Buczek & Cavanagh, 1990; Telhan et al., 2010), we predicted that as the slopes became greater that significant differences would arise. During stance phase in level running, the lower extremity joint moments show a primarily extensor pattern (Winter, 1983), however our findings did not show any consistent changes in extension moments in either the RFS or FFS group. Our findings showed a significant increase in peak hip flexion moment and a significant decrease in peak knee flexion moment during the downhill 9° condition in both RFS and FFS runners. Previous research has shown that downhill running

decreases the muscle activity in the quadriceps and hamstring group (Cai et al., 2010), therefore these changes must be explained due to differences in moment arm.

FFS versus RFS

We compared FFS and RFS runners at each slope condition and predicted that (iii) FFS runners would have an absent impact peak and reduced active loading rate compared to RFS runners, and (iv) FFS runners would have increased ankle plantar flexion moments, increased peak ankle power absorption, and decreased peak power absorption at the hip and knee joints compared to RFS runners. Our findings agree with the first part of our third hypothesis as FFS runners had an absent impact peak in all slope conditions, however active loading rates were not significantly different between RFS and FFS runners during any condition. Our results partially support our fourth hypothesis as peak ankle plantar flexion moments and peak ankle power absorption were larger in FFS runners compared to RFS runners during all slope conditions. However, the FFS group had a lower peak knee power absorption in the level and downhill conditions and peak hip power absorption was only lower in the uphill 6° condition.

When examining FFS versus RFS runners, several kinetic alterations occurred. Impact peaks were absent in FFS runners during all slope conditions, whereas this distinct peak occurred in the first 25 ms after initial contact in all slope conditions for RFS runners. This lack of impact peak is thought to be the result of FFS runners landing with a compliant plantar flexed ankle at initial contact, whereas RFS have a very stiff ankle at initial contact (Hamill et al., 2014; Laughton et al., 2013). Previous research on impact peaks found that it was the primary biomechanical difference among injured versus non-injured groups, as the non-injured groups had significantly lower impact peaks (Hreljac et al., 2000). Since FFS runners do not have an impact peak during any slope condition, adaptation of this type of landing pattern may be

warranted as a strategy to prevent overuse running injuries. Patellofemoral pain (PFP) is an overuse injury which has been linked to the presence of an impact peak. A small case study that used a gait retraining program found a decrease in impact peaks and PFP in 3 habitual RFS runners after a 3 month intervention (Cheung & Davis, 2011). Another study found that altering strike pattern to a FFS resulted in consistent reductions in patellofemoral joint stress (Vannatta & Kernozek, 2015). Running with a FFS could be used as a potential solution for runners who experience overuse injuries, specifically PFP.

The main differences in peak sagittal moments between RFS and FFS runners occurred at the ankle joint. FFS runners had larger peak plantar flexion moments and smaller peak dorsiflexion moments compared to RFS runners during all slope conditions. Peak plantar flexion moments occur around 60% of stance phase (Winter, 1983), however the timing of the peak dorsiflexion moment occurs at different times for RFS and FFS runners. In RFS runners, peak dorsiflexion moment occurs around 10% stance phase, but in FFS runners it occurs much later, just prior to toe off, with FFS runners displaying a plantar flexion moment for the majority of stance phase. These differences may be explained by peak sagittal joint angles as FFS runners have greater peak plantar flexion angles and RFS runners have greater peak dorsiflexion angles.

RFS runners absorbed the majority of their power through the knee joint compared to the ankle joint in FFS runners. These findings are in agreement with previous research performed on FFS runners (Lieberman et al., 2010; D. S. Williams et al., 2000; D. S. Williams et al., 2012). How each group positions their foot at initial contact causes differences in how forces are transmitted through the body. The plantar flexed ankle in FFS runners causes these forces to be transmitted initially through the midfoot bones and muscles rather than the calcaneus, talus and tibia directly, as is the case when landing with a dorsiflexed ankle during RFS running.

FFS runners absorb and generate the majority of their power through the ankle joint through the activity of the triceps surae muscle group. This may be due to the anatomical structure of this muscle group. Plantar flexors have relatively short fascicles, and for a given force output, short fibred muscles are metabolically less costly than long-fibred muscles such as the hip flexors since a lower volume of the muscle must be active. The eccentric loading that occurs in the triceps surae group in FFS runners utilizes the stretch-shortening cycle which may reduce the energy cost of running by allowing for elastic energy storage and recovery through the Achilles tendon. Research has shown that the gastrocnemius muscle can perform a greater amount of power generation if it has been stretched immediately before it is allowed to shorten (Cavagna et al., 1965).

Landing with a RFS and FFS result in biomechanical differences not only during level running but also during hill running, particularly downhill running which are attributed to the different landing strategies. FFS and RFS presented different joint kinematics during level and hill running, showing significant differences at the ankle joint, however differences did not exist at the knee joint as was anticipated. Both RFS and FFS running resulted in significant increases in hip flexion moment and a significant decrease in knee flexion moment during the downhill 9° condition. FFS runners absorb the majority of their power through the ankle joint whereas RFS runners absorb the majority of their power through the knee joint.

The most interesting finding was that FFS runners had an absent impact peak in all conditions. This finding suggests that running with a FFS may have potential to decrease the risk of developing an overuse running injury and help reduce knee pain which is experienced with PFP. FFS runners had larger peak plantar flexion moments and peak ankle power absorption during all conditions compared to RFS runners which can be attributed to the increased muscle

activity of the triceps surae muscle group. Peak power absorption at the knee joint was lower in FFS runners during level and downhill running conditions, which provides further evidence that using a FFS decreases loading at the knee joint which can help with reducing pain experienced at the knee with PFP. These findings warrant the use of FFS running as a potential strategy to help reduce overuse running injuries, specifically PFP. However, those choosing to switch from a RFS to a FFS should do so in a gradual manner as FFS running increases the loading in the triceps surae muscle group as to avoid injury to the muscle group.

Chapter 7

Limitations and Future Directions

Several limitations exist with our research. Due to the placement of the force platform on the ramp, not all subjects had a step on the ramp before making contact with the force platform. This caused some participants to have an extended float period from the running deck before making the first step during downhill running directly onto the force platform. A possible solution to use with future research would be to place the force platform in the middle of the ramp to allow participants to have a step on the ramp with their non-dominant foot before contacting the force platform with their dominant foot.

Another possible limitation which existed in our research was that we used a 3m ramp, and although we did our best to reinforce the structure, some bending may have occurred. This bending in the steel ramp could have diminished some of the force at impact, as the ramp may not have been completely rigid. Future research will be required to validate the use of the ramp and establish the amount of bending which occurs during uphill and downhill running. If this research provides evidence that a significant degree of bending occurs, new supports will need to be implemented to reduce this as much as possible.

Finally, this research was only conducted on male participants and may not be representative of the biomechanics experienced by female runners (Ferber et al., 2003). Previous research has shown that men and women are not identical with their running gait. Women have greater hip adduction and knee abduction throughout most of stance phase and absorbed more power at the hip joint compared to men. Women also demonstrated greater hip internal rotation than men (Ferber et al., 2003). However, since the majority of these differences occur at the hip joint, these differences may be attributed to the anatomical structure differences between men

and women. Women have a larger hip width to femoral length ratio which leads to greater hip adduction, resulting in an increased Q-angle in females. This increased angulation of the femur contributes a greater static genu valgus and an increase in lateral patellar contact forces (Horton & Hall, 1989; Mizuno et al., 2001; Simoneau, Hoenig, Lepley, & Papanek, 1998; Woodland & Francis, 1992). The next phase of this research will follow the same protocol but on female participants. Following data collection a thorough sex comparison will be conducted which for the first time will include a sex comparison with respect to landing pattern and slope.

Chapter 8

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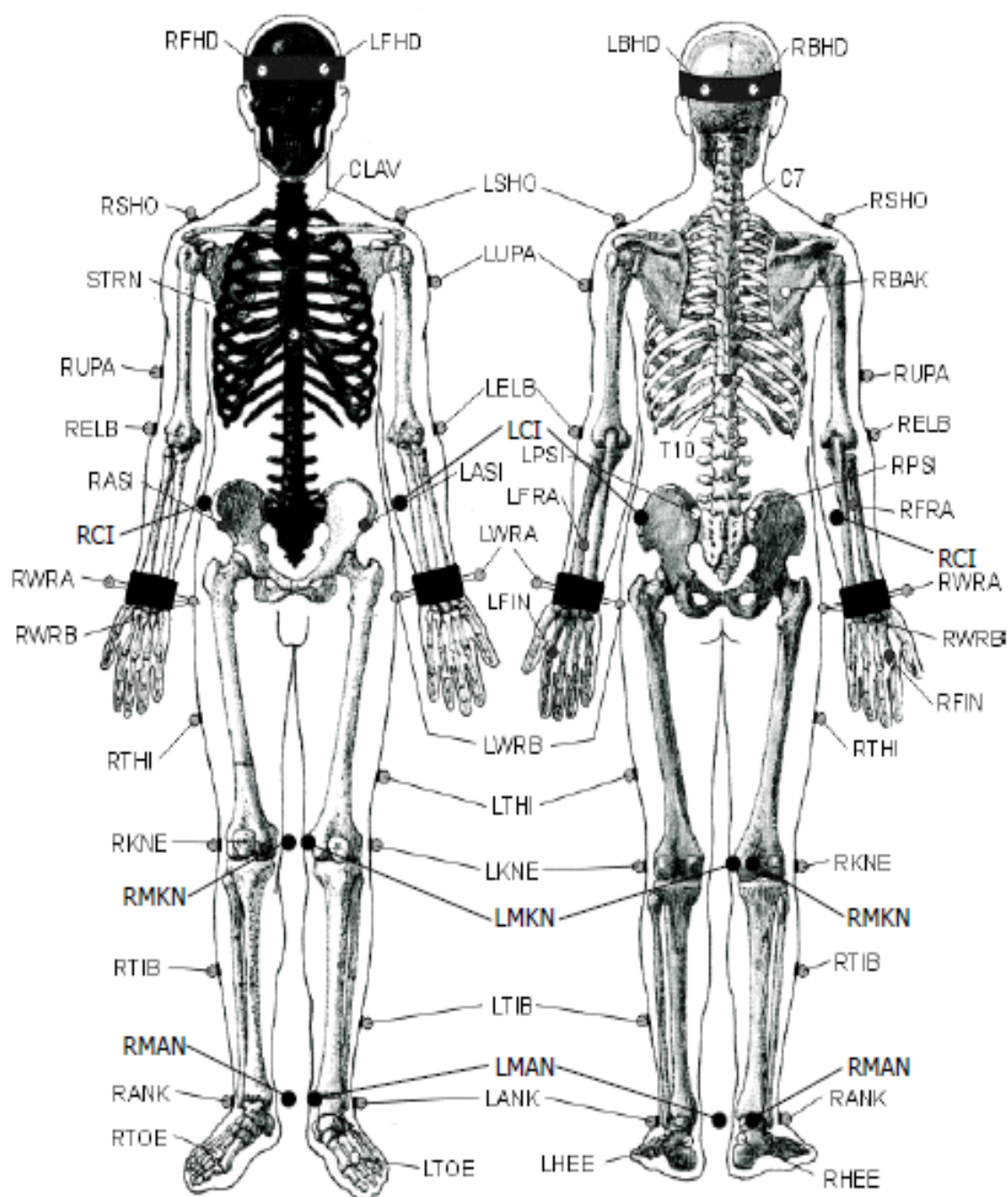
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Chapter 9

Appendix A

University of Ottawa Motion Analysis Model (OUMAM)

Appendix B

University of Ottawa Motion Analysis Model (UOMAM) Markerset

Head	
LFHD & RFHD	Left & Right Temple
LBHD & RBHD	Left & Right Back of Head
Torso	
C7	7 th Cervical Vertebrae
T10	10 th Thoracic Vertebrae
CLAV	Jugular Notch
STRN	Xiphoid Process
RBAK	Middle of Right Scapula
Arms	
LSHO & RSHO	Left & Right Acromio-Clavicular Joint
LUPA & RUPA	Left & Right Upper Arm
LELB & RELB	Left & Right Lateral Epicondyle
LFRA & RFRA	Left & Right Forearm
LWRA & RWRA	Left & Right Wrist Bar Thumb Side
LWRB & RWRB	Left & Right Wrist Bar Pinkie Side
LFIN & RFIN	Left & Right Dorsum of the Hand Head of 2 nd Metacarpal
Pelvis	
LASI & RASI	Left & Right Anterior Superior Iliac Crest
LPSI & RPSI	Left & Right Posterior Superior Iliac Crest
LIC & RIC	Left & Right Iliac Crest
Legs	
LTHI & RTHI	Left & Right Lateral Thigh
LMKN & RMKN	Left & Right Medial Epicondyle of the Knee
LKNE & RKNE	Left & Right Lateral Epicondyle of the Knee
LTIB & RTIB	Left & Right Lateral Shank
Feet	
LANK & RANK	Left & Right Lateral Malleolus
LMAN & RMAN	Left & Right Medial Malleolus
LTOE & RTOE	Left & Right 2 nd Metatarsal Head of Foot
LHEE & RHEE	Left & Right Posterior Calcaneus

Appendix C

Running Questionnaire

Participant ID: _____

Kinematics and kinetics of the lower limb in uphill and downhill running: A comparison of forefoot strike and rearfoot strike runners

Participant Questionnaire

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Participant Information

Name: _____ Date: _____

Address: _____

Phone (Day) _____ (Evening) _____

Date of Birth: _____ Sex: _____

Running Information

How long have you been running? _____

On average, how many times a week do you run? _____

How far is your regular run (distance)? _____

How many kilometers do you run a week? _____

Please break down your weekly kilometers (i.e. how many kilometers to long runs, tempo, hills, speed work, etc.)

Why are you interested in running? (Check all that apply)

- ☐ Fitness
- ☐ Weight Loss
- ☐ Recreation
- ☐ Stress Relief
- ☐ Social
- ☐ Recreational Racing
- ☐ Competitive Racing
- ☐ Other: _____

How many surface(s) do you run on? (Check all that apply and rank in frequency)

- ☐ Sidewalk/Asphalt _____
- ☐ Grass _____
- ☐ Trails _____
- ☐ Gravel _____
- ☐ Treadmill _____
- ☐ Other: _____

What kind of running surface/s you prefer?

Running Injuries

Have you ever had a running injury?

- ☐ Yes
- ☐ No

If yes, which of the following injuries have you had? (Check all that apply)

- ☐ Stress fracture (specify location) _____
- ☐ Muscle strains (specify location) _____
- ☐ Shin splints
- ☐ Plantar fasciitis
- ☐ Achilles tendinitis
- ☐ Iliotibial band syndrome
- ☐ Patellofemoral pain syndrome
- ☐ Other: _____

Please provide detail about above selection(s) (approximate date of injury, single occurrence vs. reoccurring, did you still run with injury, recovery time)

Are you currently injured?

- ☐ Yes
- ☐ No

When was the last time you had a running injury? _____

Do you ever experience pain while you run?

- ☐ Yes
- ☐ No

If yes, where do you experience the pain? _____

Have you ever passed out during or after exercise?

- ☐ Yes
- ☐ No

Have you ever been dizzy during or after exercise?

- ☐ Yes
- ☐ No

Have you ever had chest pain during or after exercise?

- ☐ Yes
- ☐ No

Has your physician ever suggested that you should not exercise for any reason?

- ☐ Yes (please explain) _____
- ☐ No

Foot and Footwear

Which of the following best describes your arch type?

- ☐ High arch
- ☐ Normal arch
- ☐ Low arch

Which of the following best describes your typical landing pattern when you run?

- ☐ Rear foot strike (landing heel first)
- ☐ Midfoot strike (landing with a flat foot)
- ☐ Forefoot strike (landing toes first)

What shoe do you normally run in? (Brand and model) _____

Do you feel your running shoes allow you to run pain free?

- ☐ Yes
- ☐ No

Are you happy with the current running shoes that you use?

- ☐ Yes
- ☐ No

How long have you had your current running shoes? _____

Do you use the same running shoes for each run?

- ☐ Yes
- ☐ No (please explain) _____

How often do you change your running shoes? _____

What makes you change your running shoes? _____

Appendix D

Photocell Timer

