

**Factors affecting lower extremity loading during running**

by

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## CHAPTER 1. GENERAL INTRODUCTION

### Introduction

On any given day, both in the United States and all over the world, someone will probably be running. All you really need is yourself and a pair of running shoes...and perhaps not even those. Yet despite the simplistic nature of running as an activity, many different influences are present. These influences include such things as the location of the run, running surface, intensity and duration of the run, footwear, injury history, and training goals, among others. The interaction of all these variables makes determining their effects very challenging. This makes determining the causes of running injuries as well as developing prevention strategies difficult. In addition, these variables are not always consistent; they can change from stride to stride. While increasing our knowledge base may not allow us to control running variables or prevent injury, it may at least help us to better understand what is happening when we do run.

The purpose of this line of research is to determine the effects of loading on the lower extremity during various conditions of running. These conditions may reduce the risk of injury or explain the stresses placed on the lower extremity. Repetitive impacts and the attenuation of these impacts are influenced by the internal and external running environment. When impacts exceed a certain threshold, they begin to have negative consequences. This threshold is determined by the ability of the body to attenuate the impacts, stresses, and strains experienced by tissues in the lower extremity. These loads can lead to injuries such as osteoarthritis, stress fractures, stress syndrome, joint deterioration or muscular microtears. Impacts are measured in a laboratory setting through the use of ground reaction forces or accelerations.

The purpose of the first study was to systematically alter the knee flexion at contact as participants ran off of a platform and document how this angle affected both the impact force and impact acceleration. Past research indicates a possible relationship between knee flexion angle at contact and the attenuation of impacts. Thomas and Derrick (2003) showed that increasing knee flexion angle was a strategy used by runners under conditions of uncertainty. It was speculated that this may have been employed to reduce the impact loading on the lower extremity, which in turn could help the body increase shock absorption. Denoth (1986) indicated that increasing knee flexion decreased impact forces at the leg.

Impacts from weight-bearing activities can also influence bone remodeling. Overloading of bone can lead to damage, and inadequate time for repair can result in overuse injury (Grimston and Zernicke 1993). Proper control of impacts may result in bone strains and strain rates that provide the mechanical stimulus needed for osteogenesis. For example, bone loading below the injury threshold can initiate an adaptive response that strengthens the bone, provided the load is above the threshold for remodeling. One study showed that male and female runners had a 40% higher bone density than non-runners (Lane et al 1986).

At this time we have a poor understanding of the factors that lead to the positive and the negative effects of impacts. One reason for this is there is not a clear understanding of the relationship between ground reaction forces and impact accelerations in the lower extremity during gait. Increases in both impact forces and impact accelerations during gait may be related to inefficient running form, poor training strategies, or inadequate footwear that may lead to injury. The degree of injury potential

may also be influenced by the mass being accelerated at footstrike due to the influences on impact forces and accelerations.

Knee flexion angle at contact might also affect loading on the lower extremity. The knee joint plays a critical role in altering the mechanical characteristics of the body during an impact. Fatigue, stride length, and surface irregularities can alter the knee flexion at contact and thus alter the response of the body to an impact. It may be found that increasing the knee flexion at contact will reduce the effective mass being accelerated, leading to decreased impact forces and therefore a lower injury potential. However, the influence of knee flexion on impact forces and accelerations it is not yet known. There is evidence that knee flexion at contact has a positive relationship with impact acceleration and a negative relationship with impact forces. Flexion of the knee can also increase shock attenuation (Lafortune, Lake and Hennig, 1996).

Knee flexion at contact could be altered during downhill running. The purpose of the second study was to investigate the relationships between knee flexion at contact and the joint contact forces while running off a platform. The platform was introduced to produce a pattern of running similar to downhill running. When running, changes occur in the lower extremity because of the variability in runners and the adaptability to various conditions. These changes may influence the contact forces in the knee joint. This may determine if increasing the amount of knee flexion at contact will change the knee contact forces and rates of loading.

Changes occur in the lower extremity as a result of variability in runners such as gait patterns and adaptability to various conditions such as footwear and running surface. These changes will likely influence the contact forces in the knee joint. It is not known if

there are any relationships between knee flexion at contact and joint contact forces or joint contact forces in the knee joint. Increasing knee angle may result in an increase in muscle activity in the lower extremity. This increased activity could result in an increased metabolic cost as well as increased compressive forces. These joint contact joint contact forces might play a role in stress fractures or other overuse injuries.

Two common strategies for reducing the impacts during gait are to increase the amount of midsole cushioning in the runner's footwear or to decrease the length of the runner's stride. The purpose of the third study was to determine how joint contact forces in the lower extremity are affected by modifications in midsole hardness and decreases in stride length. Lafortune et al. (1996) indicated that stiffness of the contact surface (such as running surface or midsole cushioning) may play a larger role in decreasing impact than changes in knee angle. They found that softer surfaces resulted in a decreased severity of impact as well as decreased leg stiffness. According to Hardin and Hamill (2002) midsole stiffness had no effect on peak tibial acceleration. Some research has reported that shoe cushioning does not aid in shock attenuation at the leg (Nigg et al., 1987, 1988, 1998) or reducing injury risk (Gardner et al., 1988; Benseal and Kaplan, 1986; Schwellnus et al., 1990). The effect of stride length on impact forces has a clearer role. Derrick et al., 1998) indicated that an increase in stride length decreased tibial acceleration, from 6.1 g during preferred stride length to 5.7 g with a 20% stride length reduction. However, it is not yet known what effect these strategies may have on joint contact forces.

## **Dissertation Organization**

This dissertation is organized as three papers to be submitted to scientific journals. This initial chapter includes the general introduction and a review of relevant literature on the topic of loads on the lower extremity. The next three chapters are the three manuscripts. The primary author for all three articles is Joshua M. Thomas, PhD candidate at Iowa State University and Assistant Professor of Human Performance and Wellness at Trinity International University. Timothy R. Derrick is a co-author on all three articles. Dr. Derrick is an Associate Professor in the Department of Kinesiology at Iowa State University. His role was to assist with research and preparation of manuscripts, as well as being the primary writer of the Matlab program. Dr. Jason Gillette, Assistant Professor of Kinesiology at Iowa State University, contributed to all three articles in manuscript preparation. W. Brent Edwards, PhD candidate in Kinesiology at Iowa State University, also contributed to this manuscript by assisting with Matlab programming. The first manuscript is found in Chapter 2 and is titled “Flexing the Knee While Running Off a Raised Platform.” The second manuscript comprises Chapter 3 entitled “Relationship Between Knee Flexion at Contact and Joint Contact Forces in the Knee.” The final manuscript makes up Chapter 4 with the title of “Joint Contact Knee Forces With Changes in Stride Length and Midsole Stiffness.” The articles are followed by Chapter 5 which includes general conclusions discussing the results of the research and debating future paths of research.

## **Literature Review**

During running, one of the most important considerations is the consequence of load on the body, especially the lower extremity. Repetitive impacts on the body that



occur with every footstrike can have negative effects on the structures of the leg.

Research has shown a possible link between high mileage, high intensity running and osteoarthritis in the hip (Marti 1989). However, there does not appear to be evidence linking running and osteoarthritis in the knee. (Sohn et al 1985; Cymet and Sinkov 2006). Maughan and Miller (1983) found that 28% of injuries in marathon runners were in the anterior knee. While these injuries were not identified as osteoarthritis, it may indicate the injury potential of loading on the knee.

Stress fractures are a common type of overuse injury in runners, with the most common injury site being the lower extremity (Jones, Harris, Vinh et al. 1989; Matheson, Clement, McKenzie et al. 1987; Brubaker, James 1974; Gudas 1980; Clement, Taunton, Smart et al. 1981; Ha, Hahn, Chung et al. 1991; Belkin 1980). Studies have shown a 21% incidence of stress fractures among track runners (Bennell, Malcolm, Thomas et al. 1996) and an 8-13% incidence of stress fractures among recreational runners (Brunet, Cook, Brinker et al. 1990). Within the lower extremity, tibial stress fractures are most common (Hulkko, Orava 1987; Matheson, Clement, McKenzie et al 1987; Brubaker, James 1974; Gudas 1980; Orava, Puranen, Ala-Ketola 1978; Devas 1958; Sullivan, Warren, Pavlov, et al. 1984; Taunton, Clement, Webber 1981). An incidence of 4.0% was found in military recruits during 12 weeks of training, with the highest injury rates correlating with the weeks of the highest training volume (Almeida 1999). A ten-year study of 6000 athletes at the University of Minnesota found an overall incidence of stress fractures of only 1% (Arendt et al. 2003). The incidence was higher in female athletes (1.9%) and distance runners (3.2%). The tibia was the bone with the highest stress injury rate (37%), although there were more injuries in the foot when considered as an anatomic region.

It is theorized that stress fractures may occur as a result of incomplete or inefficient osteogenic remodeling, in which the bone being resorbed is not given sufficient time to be replaced (Frost 1987; Markey 1987; Hershman, Maily 1990; Nattiv, Armsey 1997; Clement et al. 1981). Running is an activity that can produce this type of condition, especially among competitive or elite runners. Runners have a tendency to achieve a high volume of mileage, and run on average once a day, sometimes twice a day. This type of training may not allow the bone sufficient time to replace what has been resorbed.

McMahon, Valiant, and Frederick (1987) presented some important research observing the relationship between knee flexion angle and impacts. Their landmark study of “Groucho” running determined that runners who increased their knee flexion during stance decreased the transmission of impact through the body. The vertical force at midstance also decreased. Their results also showed that this increase in knee flexion was coupled with an increase in oxygen consumption by the runners.

Downhill running has been shown to influence knee flexion at contact. Runners decreased knee flexion at contact from approximately  $25^{\circ}$  during level running to approximately  $17^{\circ}$  when running downhill (Buczek and Cavanagh 1990). Downhill running also indicated a significantly greater amount of negative work done on the knee extensor muscles during stance. Downhill running also increases impact force peaks by 54% (Gottschall and Kram 2005).

Shorten and Winslow (1992) developed a method of spectral analysis for analyzing impact shock during running. Impact shock measured from accelerometers is often affected by noise from skin movement. Bone mounted accelerometers would not

have this skin artifact, but are not as practical to use due to their invasive nature. Shorten and Winslow used Fourier transformations to determine that the impact peak occurs between 12 – 20 Hz in the frequency domain. This allows for an easier interpretation of impact shock data. Lafortune, Lake, and Hennig (1995) also contributed to research on interpretation of accelerations and ground reaction forces. They developed a frequency transfer function to represent the relationship between axial acceleration from the tibia and ground reaction forces. The possibility of high tibial shock to predict stress injuries has also been noted in runners (Milner et al. 2006), as well as evidence that higher rates of loading may produce more injury than repeated loading at low rates of loading (Schaffler et al. 1989).

The influence of increased cushioning in shoes has also been studied (Markey 1987). This cushioning may come from midsole stiffness, cushioned shoe or orthotic inserts, or other methods. Some research indicates that cushioned inserts reduced tibial shock during gait by about 40%. (Voloshin and Wosk 1981), and that cushioned running shoes may decrease injury (James and Brubaker 1972). One study determined that running shoes had an increase of stress fracture occurrence after one month of use (Gardner, Dziados, Jones, et al. 1988). It could be assumed that these old shoes may have lost some of their cushioning ability. However, the same study indicated that shock absorbing insoles do not significantly reduce stress fractures. Other research supported these data (Schwellnus, Jordan, Noakes 1990; Bense, Kaplan 1986). Kersting and Bruggemann (2006) also found no difference in impact force with varying midsole cushioning, but speculated that it may have been due to adaptations of the individual runners in response to changing shoe conditions. Hardin et al (2004) also found that

lower extremity kinematics of runners adapted to passive changes such as changing midsole cushioning.

Some of the earliest research on joint contact forces during running was done by Burdett (1982). Because direct measurements are difficult in human subjects, Burdett used a model to estimate joint contact forces at the ankle. Anatomical measurements were taken from cadavers and the relevant data for the functional muscle groups of the ankle were input into the model. The model was designed to estimate joint forces during the stance phase. Kinematic and kinetic data for the model were collected from three subjects running at  $4.47 \text{ m}\cdot\text{s}^{-1}$  across a force platform. The results indicated ankle compressive forces of 8-13 body weights (BW) during midstance. Ankle shear forces in the anterior-posterior (AP) direction were 3.3 to 5.5 BW, and medial-lateral (ML) shear forces ranged from 0.5 to 0.8 BW. Peak muscle forces were highest in the plantar flexor group (gastrocnemius and soleus) at a range of 5.3 to 10 BW.

Scott and Winter (1990) developed a model to analyze joint contact forces during running and their connections to injury. They found peak vertical ground reaction forces of 2.1 BW at heel contact, and 2.7 BW occurring around midstance. The vastus medialis, vastus lateralis, and rectus femoris showed peak activity around 12% of stance. Ankle compressive forces ranged from 10.3 – 14.1 BW. Average peak compressive force was 11.2 BW, occurring just past midstance. Ankle AP shear forces ranged from -0.4 to -0.7 BW. The average peak bone-on-bone shear force was -0.7 BW just prior to midstance.

Sasimontongkul, Bay, and Pavol (2007) estimated contact forces at the distal tibia during stance in running. The influences of external forces such as ground reaction forces (GRF) and internal forces such as muscle forces were also accounted for. Compressive

forces from GRF averaged 1.97 BW and AP shear forces from GRF averaged 1.23 BW. The average peak tibia compressive force was 9.0 BW. The peak tibia AP shear force was 0.57 BW. The researchers noted that the peak tibia forces typically occurred around midstance, suggesting that midstance forces could play a more prominent role in the occurrence of stress fractures than forces at impact.

The aim of this research is to increase the knowledge of impact forces and impact accelerations during running, which would lead to a better understanding and interpretation of research involving gait and lower extremity loading. It is thought that increasing our knowledge of knee flexion angle and the effects on joint contact forces will lead to better understanding of factors influencing attenuation. This knowledge will hopefully show how impacts can have an effect on joint contact forces at the knee, which is a common injury site in the lower leg. This knowledge of joint contact forces will increase understanding of how lower leg injuries occur from repeated stress. Answering questions regarding midsole cushioning and stride length influences on knee forces will help us to understand whether or not these are effective strategies for injury prevention. This provides information for research projects designed to establish injury and osteogenic thresholds that will potentially improve bone health in humans.

## CHAPTER 2. FLEXING THE KNEE WHILE RUNNING OFF A RAISED PLATFORM

A paper to be submitted to *The British Journal of Sports Medicine*

Joshua M. Thomas, Timothy R. Derrick and Jason C. Gillette

### Abstract

**Objective:** The purpose of this study was to systematically alter the knee flexion at contact as runners land from running off of a platform and to document how this angle affects both the impact force and impact acceleration. The platform was used for its similarity to downhill running. **Participants:** Five male and five female participants volunteered to complete ten trials in each of four conditions. **Interventions:** Normal running (NR), normal running off a raised platform (NRP), flexed knee running off a raised platform (FRP), and extended knee running off a raised platform (ERP). **Main outcome measurements:** Kinematic data, ground reaction forces, leg and head acceleration data were collected. **Results:** Running off a raised platform increased peak impact forces (NR:  $1147 \pm 171\text{N}$ ; NRP:  $1826 \pm 352\text{N}$ ), peak leg accelerations (NR:  $7.6 \pm 5.3\text{g}'\text{s}$ ; NRP:  $13.9 \pm 5.8\text{g}'\text{s}$ ) and peak head accelerations (NR:  $1.4 \pm 0.4\text{g}'\text{s}$ ; NRP:  $4.0 \pm 1.4\text{g}'\text{s}$ ). Flexed knee running off the raised platform decreased peak impact forces to  $1672 \pm 463\text{N}$  but increased peak impact leg accelerations to  $16.3 \pm 5.7\text{g}'\text{s}$  compared to NRP. In contrast, extended knee running off the raised platform increased peak impact forces to  $2111 \pm 515\text{N}$  but peak leg accelerations did not change  $13.5 \pm 4.7\text{g}'\text{s}$  when compared to NRP. **Conclusions:** These results suggest that knee flexion decreased the mass that was being accelerated. This decreased mass was easier to accelerate and

resulted in greater peak values. When the knee angle changed, the peak impact force and the peak leg impact acceleration had a negative correlation (-0.85).

### **Introduction**

The bones of astronauts, patients restricted to bed rest, military recruits and athletes participating in impact activities are affected by the degree of mechanical stimulation. Repeated impacts can have detrimental or beneficial effects on the body. Overloading of bone can lead to damage [1-3] and inadequate time for repair can result in bone fatigue or fracture.[4] However, impacts also result in bone strains and strain rates that provide the mechanical stimulus needed for osteogenesis.[5-6] At this time we have a poor understanding of the circumstances that lead to the cessation of biologically positive impacts and the initiation of biologically negative impacts.

Repetitive impacts are characteristic of cyclic activities such as running. These impacts and the attenuation (or reduction) of these impacts are influenced by the internal and external running environment. When impacts exceed a certain threshold, they begin to have negative consequences. This threshold is determined by the ability of the body to attenuate the impacts and the stresses and strains experienced by tissues in the lower extremity. Loading below the threshold can initiate an adaptive response that strengthens the bone.[7] If bone is not given adequate time to recover or if the loading exceeds the threshold, the risk of injury increases. These loads may then lead to injuries such as stress fractures, stress syndrome, joint deterioration or muscular microtears. The potential for injury may be affected by anatomical abnormalities, unusual kinematics, insufficient recovery time, excessively hard surfaces, footwear mismatched with the individual or training failures such as excessive mileage. A more complete understanding of the factors

that affect impacts and the measurement of impacts is necessary for further progress in establishing injury and osteogenic thresholds.

Kinematics such as the geometric alignment of the body segments at ground contact can influence the severity of the impact[8] or the body's ability to attenuate the impact.[9] Both the internal and external environment can influence this geometric alignment and thus affect the impact, the attenuation of the impact, and the biological tissues that are responsible for the attenuation. The knee joint plays a critical role in altering the mechanical characteristics of the body during an impact. Factors such as fatigue, stride length, intensity of light, surface irregularities, and even the length of grass under foot can alter the knee flexion at contact and thus alter the response of the body to an impact. After controlling for running speed, the conditions mentioned above increased the knee flexion at contact by an average of  $2.7^\circ$ . [8] Although this is a relatively small change, simulation studies suggest that increasing the knee flexion at contact can cause the impact force to decrease by as much as 68 N/degree.[10] An increase in knee flexion decreases the mass that is accelerated during the impact.[11] A smaller "effective" mass is easier to accelerate and results in an increase in the peak impact acceleration. Since the effective mass is altered, the increased peak leg accelerations that result from greater knee flexion do not necessarily correlate with increased peak impact forces.

There is evidence that knee flexion at contact has a positive relationship with the impact acceleration and a negative relationship with impact forces. This response was simulated using a mass-spring-damper model.[8] This model had a spring that separated a large, upper mass from a smaller, lower mass and a spring-damper system that separated the lower mass from the ground. The lower mass was designed to approximate



the effective mass of a runner. When 5% of body mass was shifted from the lower mass to the upper mass in the model the peak impact force decreased from 950 to 850 N. This also caused an increase in the peak acceleration of the lower mass (from 5.5 to 6.6 g).

The current research project was designed to test this effect in actual runners. Thus the purpose was to systematically alter the knee flexion at contact as participants ran off a platform and document how this angle affected both the impact force and impact acceleration. The platform was chosen to produce a condition of uncertainty, which has been shown to produce increased knee flexion angles.[8-9] The platform condition was also comparable to running downhill, which can increase impact forces and rates of loading.[12] This platform allowed us to exaggerate the loading environment. We hypothesized that increasing the knee flexion angle at contact would decrease peak impact forces and increase peak leg acceleration in accord with the mass-spring model. In addition, we hypothesized increased attenuation of the impact with increased knee flexion. Understanding the relationship between impact forces measured from a force platform and accelerations measured from leg mounted accelerometers is important because it may allow us to use accelerometers in the collection of continuous impact data during exercise and free living situations. In turn, this will lead to research designed to establish injury and osteogenic thresholds that will eventually improve bone health in humans.

### **Materials and methods**

Five male ( $71.6 \pm 7.3$  kg,  $1.80 \pm 0.05$  m,  $29.0 \pm 7.7$  years) and five female ( $65.6 \pm 8.3$  kg,  $1.64 \pm 0.06$  m,  $21.6 \pm 1.1$  years) participants were recruited for this study. All participants were recreational runners, but no requirements for mileage or pace were

mandatory. None of the participants had lower extremity abnormalities that would affect their ability to run and land off a raised platform.

### Protocol

Upon entering the laboratory, all participants read and signed an informed consent document approved by the Institutional Review Board at Iowa State University.

Retroreflective markers were placed on the right toe, heel, ankle, knee, hip, and shoulder.

Accelerometers were attached to the medial-distal aspect of the tibia and over the frontal bone of the forehead. Accelerometers were attached with double-sided tape on the skin, wrapped with an elastic band and covered by athletic tape to minimize skin movement.

Prior to data collection, participants were allowed to self-select their running speed.

Participants were then required to perform all trials within a range 5% above or below that speed. Average self-selected speed was  $2.7 \pm 0.4 \text{ m}\cdot\text{s}^{-1}$ . Participants completed ten trials for each of four conditions while running with a heel-toe running style. All 40 trials were completed in a single session. The first condition included running across a force platform with a self-selected knee angle or normal running (NR). The remaining three conditions involved running off a 22.5 cm raised runway onto the force platform. The raised platform conditions included normal platform running (NRP), flexed knee platform running (FRP) and extended knee platform running (ERP). During FRP, participants were asked to land with as much knee flexion as they felt comfortable with. During the ERP, participants were asked to land as upright as possible, with minimal knee flexion. The NR was always performed first, followed by NRP, and then the ERP and FRP conditions were balanced across the participants.

### Equipment

Immediate feedback of running speed was obtained using a microwave based radar gun (model Stalker ATS, Radar Sales) interfaced with a laptop computer. A more precise measurement was recorded during the post analysis and was calculated from the average horizontal velocity of the hip marker during the stance phase of the running cycle. Kinematic data were collected using a six-camera motion capture system (Peak Motus) and used to calculate knee flexion angles. Markers were digitized at a rate of 120 Hz and were low-pass digitally filtered at 16 Hz using a fourth-order Butterworth filter prior to the calculation of kinematics. A strain gage force platform (model OR6-7-2000, AMTI) was used to collect ground reaction forces and determine peak impact forces. Peak impact force was identified as the first peak of the bimodal vertical ground reaction force curves. Two quartz shear piezoelectric accelerometers (model 353B17, PCB Piezotronics, Inc.) were used to measure peak impact accelerations at the leg and head (see Derrick, Hamill & Caldwell for detailed procedures).[13] Both forces and accelerations were sampled at 3600 Hz.

### Statistical Analysis

Self-selected speed measured from the radar gun ( $2.7 \pm 0.4 \text{ m} \cdot \text{s}^{-1}$ ) was slightly lower than speed calculated from the kinematic system (Table 1.1). Since there were slight differences between the running speeds of the different conditions, the data were analyzed using a repeated measures ANCOVA with kinematically determined running speed as a covariate. As the independent variable, knee flexion was the single factor with four levels determined by the four conditions. A Tukey's post hoc analysis was used to identify differences between the conditions ( $\alpha = 0.05$ ) when appropriate. The key

dependent variables were peak impact force, peak leg impact acceleration (PL), peak head impact acceleration (PH), and impact attenuation. Attenuation was calculated in both the time domain and the frequency domain. In the time domain, attenuation was calculated as  $(1-PH/PL)*100$ . Spectral methods were used to estimate head and leg power spectrums, and frequency domain attenuation was calculated using a log function to look at changes to particular frequencies in the signal.[13-14]

## **Results**

### Normal Running vs. Normal Platform Running

Participants chose to land more extended when running off the platform compared to normal running (NR:  $17.3\pm 6.2^\circ$ , NRP:  $13.6\pm 4.5^\circ$ ,  $p = .001$ ). Contact velocity was measured from the vertical velocity of the heel marker and this also increased when running off the platform (NR:  $0.55\pm 0.2$  m·s<sup>-1</sup>, NRP:  $0.99\pm 0.3$  m·s<sup>-1</sup>,  $p < .001$ ). This combination of a more extended geometric alignment and greater contact velocity

introduced significantly more load to the body compared to normal running (Table 1.1).

Table 1.1 Mean and (sd) of selected kinetic and kinematic variables during four different running conditions. Means are calculated from all trials and all subjects. Significant differences were determined at  $\alpha = 0.05$ .

	Normal Running	Extended Running - Platform	Normal Running - Platform	Flexed Running - Platform
Knee Angle at Contact (deg)	17.3 bcd (6.2)	10.6 acd (3.7)	13.6 abd (4.5)	20.6 abc (5.2)
Impact Force (N)	1147 bcd (171)	2111 acd (515)	1826 ab (352)	1672 ab (463)
Head Acceleration (g's)	1.4 bcd (0.4)	5.1 acd (1.8)	4.0 abd (1.4)	2.3 abc (1.2)
Leg Acceleration (g's)	7.6 bcd (5.3)	13.5 ad (4.7)	13.9 ad (5.8)	16.3 abc (5.7)
Attenuation (%)	75.0 bcd (10.7)	59.3 acd (14.2)	68.1 abd (12.6)	84.2 abc (7.9)
Running Velocity (m/s)	3.05 bcd (0.4)	2.80 acd (0.3)	2.91 ab (0.3)	2.93 ab (0.3)
Heel Contact Velocity (m/s)	- 0.55 bcd (0.2)	- 1.07 ac (0.3)	- 0.99 ab (0.3)	- 1.13 a (0.4)
Maximum Knee Flexion Angle (deg)	48.8 bd (5.1)	46.6 acd (3.9)	50.1 bd (4.4)	64.2 abc (10.4)
Stance Time (s)	0.259 d (0.04)	0.257 d (0.04)	0.261 d (0.04)	0.299 abc (0.04)

a. Significantly different from NR

b. Significantly different from ERP

c. Significantly different from NRP

d. Significantly different from FRP

Impact force increased by 679 N (59%), peak leg acceleration increased by 6.4 g's (84%) and peak head acceleration increased by 2.6 g's (178%). Most of the differences in the vertical ground reaction forces occurred during the impact region of the curves (fig 1.1). Relative to normal running, more of the impact was transmitted from the leg to the head during platform running. The time domain analysis showed 75.0% attenuation during NR and only 68.1% during NRP. Fourier analysis indicated that the increase in the leg acceleration signal seen from NR to NRP was represented across the entire frequency spectrum (fig 1.2); however, the head acceleration signal had the greatest increases in the impact frequencies, which have been determined to be between 10-20 Hz.[13] The differences in attenuation between NR and NRP were mostly in the

low frequency region (<10 Hz). In the low frequency portion NR attenuation did appear to be greater than NRP attenuation. During the impact component (10 – 20 Hz) there was little difference between those two conditions in terms of attenuation. The decrease in attenuation during the platform running was likely due to a less compliant body configuration represented by greater knee extension at contact.

#### Altering Knee Angle off the Platform

All four conditions were significantly different from each other in the knee flexion at contact. All of the participants were able to voluntarily increase their knee flexion at contact and nine out of ten were able to increase knee extension at contact (Table 1.1). During normal running the participants flexed their knees an average of  $17.3^{\circ}$  as the foot made contact with the ground. When the participants ran off the platform they decreased knee flexion at contact to an average of  $13.6^{\circ}$ . Relative to the normal condition, it appears that the participants found it easier to increase knee flexion at contact ( $+6.9^{\circ}$ ) than it was to decrease knee flexion at contact ( $-3.0^{\circ}$ ) while running off the raised platform. Landing with a more extended knee could help participants to stop the downward movement of the body by shifting absorption responsibility from the muscles to other tissues. Maximum knee flexion remained fairly consistent between normal running and normal platform running ( $48.8 \pm 5.1^{\circ}$  vs.  $50.1 \pm 4.4^{\circ}$ ). When asked to increase knee flexion while running off the raised platform participants had a knee flexion at contact that averaged  $20.6 \pm 5.2^{\circ}$ . This had the effect of increasing maximum knee flexion to  $64.2 \pm 10.4^{\circ}$ . When asked to make contact with a more extended knee, the participants responded with an average knee flexion at contact of  $10.6 \pm 3.7^{\circ}$ . This decreased maximum knee flexion ( $46.6 \pm 3.9^{\circ}$ ), relative to the NRP condition. Thus maximum knee

flexion was also more affected by the flexed running ( $+14.1^\circ$ ) than it was by the extended running ( $-3.5^\circ$ ). This suggests there is a certain amount of knee flexion needed to produce an adequate impulse to propel the runner off the ground.

While running off the platform, the extended running landings produced the highest peak impact forces at  $2111 \pm 515$  N (fig 1.3). There was no significant difference in peak impact forces between normal platform running and flexed platform running. However, flexing the knee caused a significant drop (20.8%,  $p < .001$ ) in the impact forces compared to the extended knee. On the other hand, peak leg impact accelerations were significantly higher in flexed platform running ( $16.3 \pm 5.7$  g's) than either normal platform running ( $13.9 \pm 5.8$  g's,  $p = .034$ ) or extended platform running ( $13.5 \pm 4.7$  g's,  $p = .014$ ) (Table 1.1). Thus, as the knee became more extended at contact, peak impact forces increased and peak leg accelerations decreased. In the platform conditions, subjects reduced the peak impact force by  $-40.7 \pm 21.2$  N per degree of knee flexion at contact while increasing peak impact acceleration by  $.25 \pm .48$  g's per degree of knee flexion (fig 1.4).

#### Knee Flexion at Contact and Attenuation of the Impact

As expected, attenuation was influenced by the knee flexion at contact. All four conditions were significantly different from each other. Flexed platform running had the highest attenuation of the three platform conditions ( $84.2 \pm 7.9$  %) and extended platform running had the least ( $59.3 \pm 14.2$  %). Normal platform running fell between those two conditions ( $68.1 \pm 12.6$  %). The frequency domain analysis agreed with these results but also provided further insight (fig 1.2). There were no differences in the leg impact frequencies between the platform conditions; however there were differences in the head impact frequencies. Extended platform running had the greatest power at the impact

frequencies and flexed platform running had the least. This translates to greater attenuation in the impact frequencies when the knee is flexed. The transfer function in Figure 1.2 clearly shows this.

## **Discussion**

It was hypothesized that increasing the knee flexion at contact would decrease peak impact forces, increase peak leg acceleration and increase attenuation of the impact as it traveled from the leg to the head. As knee flexion at contact increased from more extended to more flexed running, there was a decrease in impact forces, which was significant between the ERP and FRP conditions ( $p < .001$ ). The results also support the increased leg acceleration and increased attenuation hypotheses. The runners were able to voluntarily increase and decrease the knee angle during contact and this altered the magnitude of the impact. In general, a more flexed knee at contact resulted in lower peak impact forces, greater peak leg acceleration and greater attenuation (Table 1.1). The more flexed knee flexion at contact created a geometric configuration of the lower extremity that decreased the effective mass.[8,15] This decreased mass was easier to accelerate and thus peak leg acceleration increased when the knee became more flexed at contact. A more flexed knee flexion at contact also attenuated the impact to a greater extent than the extended knee.

Modifying the knee flexion at contact is a common adaptation. Runners modified their knee angle in a range from 12.8 to 18.5 degrees at contact as a voluntary response to altered environmental conditions.[8] Therefore the intentionally modified knee angles in the present research (range 10.6 to 20.6 degrees) represented a practical range. These



were comparable to the knee flexion angles at contact in previous research on downhill running.[16]

With the introduction of the raised platform, the body falls a greater distance and the contact velocity increases (Table 1.1). This process is similar to downhill running. The peak leg accelerations in the current study were slightly higher than those found by Hardin and Hamill while running down a 12% grade on a treadmill (13.9 g's vs. 12.5 g's).[17] Peak head accelerations were also greater (4.0 g's vs. 2.4 g's) and attenuation was less (68.1% vs. 80.8%) than with downhill running. Denoth's research showed that increasing the knee flexion angle decreased the mass being accelerated and led to higher impact accelerations.[11] The present research supports these results. Although not estimated directly, the effective mass likely decreased as knee flexion increased at contact. This decreased mass leads to higher accelerations at the leg if all else remains equal. These results also support the results of Gerritsen et al.[10] Denoth and Gerritsen et al. both found that increasing knee flexion resulted in a decrease in impact forces.[10-11] In comparison, their simulation results predicted a reduction of 68 N per degree of knee flexion, while results of the current study suggest a lower change of approximately 40 N per degree of knee flexion.

Distributing forces over an increased time interval reduces the peak force values. Both active tissue (muscle) and passive tissue (heel pad, cartilage, bone) can increase the time that the vertical center of mass velocity of a runner is brought to zero. Eccentric muscle contractions absorb the runner's energy but will result in increased muscle forces and increased metabolic cost. Passive tissue can also deform when loaded but likely to a lesser degree than muscle. Making ground contact with a more flexed knee likely shifts a

higher percentage of the energy absorption responsibility from passive to active tissue. Although it is beyond the scope of this study to examine each tissue's contribution toward absorbing the runner's energy, it is likely that the greater knee extensor muscle forces would occur during flexed knee running; this may lead to greater compressive loads in the knee joint.

Although the impacts in this study were large compared to normal running, they do not represent an activity that, by itself, is a risk to most runners. Running off a platform does not occur often enough to cause a repetitive motion injury and the magnitudes are not large enough to cause an acute injury. On the other hand, if you consider running off a platform to be similar to downhill running then increasing a single peak impact force by 59% when you run off the platform (Table 1.1, normal running vs. normal platform running) may be enough to increase the risk of injury. If this is coupled with an excessive extension error, the runner could increase peak impact force by 84% over normal running (Table 1.1, normal running vs. extended platform running). This increase in the external forces acting on the body may be especially significant if the tissues being stressed are already fatigued by prior cyclic loads.

The results of this experiment provide insight into the relationship between impact forces and accelerations. Accelerometers can provide a portable and inexpensive way to monitor impact loads in the body over extended periods of time, but it is critical that we understand the limitations. Accelerometers could be useful when assessing the potential for stress fractures or osteogenesis because of their portability. However, it may not be adequate to simply measure peak leg accelerations. Variations in effective mass and knee flexion at contact can alter the interpretation of peak acceleration values. The

inverse relationship between peak impact force and peak leg acceleration (fig 1.4) may suggest that the leg accelerometer is not a good measure of bone loading. On the other hand, the relationship becomes positive when the effective mass does not change. Furthermore, the inclusion of a second accelerometer and the calculation of attenuation could improve the prediction of bone loading. The results of this study suggest that high peak leg accelerations should be interpreted in conjunction with attenuation results. For instance, high impact accelerations with high attenuation would likely be a less severe loading environment than high accelerations with low attenuation.

These results indicated that running with an extended knee is a poor choice because of the increased loading on the leg. However, it is not yet known if muscle forces or joint contact forces will affect this position. It may be that increasing knee flexion may result in higher joint contact forces. More knowledge on how joint contact forces are affected by joint position may introduce additional considerations for running kinematics.

## **Conclusions**

Increased flexion of the knee at contact resulted in decreased peak impact force and increased peak impact acceleration. This suggests that the mass that is being accelerated decreased when the knee was flexed at contact. This increased the impact attenuation and may have decreased the potential for injury to the musculoskeletal system. A further refinement could look at joint contact forces and rates of loading in the lower extremity. These loads would be more directly related to injury potential and osteogenic stimulation and may better illuminate the effects of altered knee angle on bone loading.

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### Figure Captions

Figure 1.1. Ensemble curves of vertical ground reaction forces during the stance phase for running overground (NR) and running off a platform (NRP). Curves are averages of all participants and all trials. Stance is normalized to normal running.

Figure 1.2. Power spectral densities of head and leg accelerations in the frequency domain. Also included is attenuation measured as the transfer of impact between the leg and head in the frequency domain. Vertical lines between 10-20 Hz represent the impact phase.

Figure 1.3. Ensemble curves of vertical ground reaction forces during the stance phase for three conditions of knee flexion at contact while running off a raised platform. Curves are averages of all participants and all trials. Stance is normalized to normal running off the platform (NRP).

Figure 1.4. The relationship between impact forces and leg accelerations with respect to increasing knee flexion. Pearson  $r$  correlation was  $-0.85$ , indicating an inverse relationship between the variables.

Figure 1.1

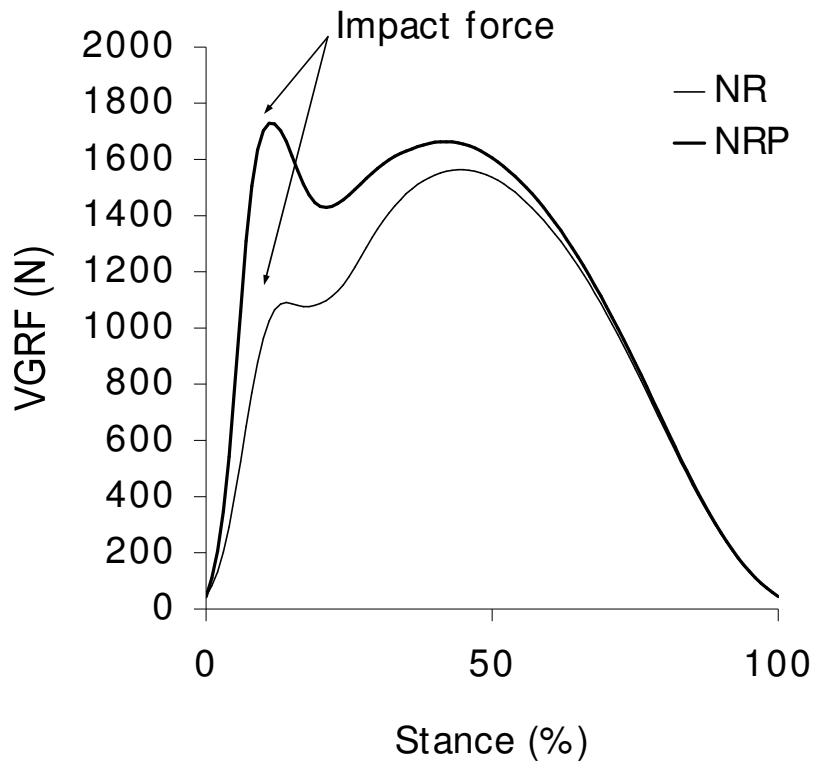


Figure 1.2

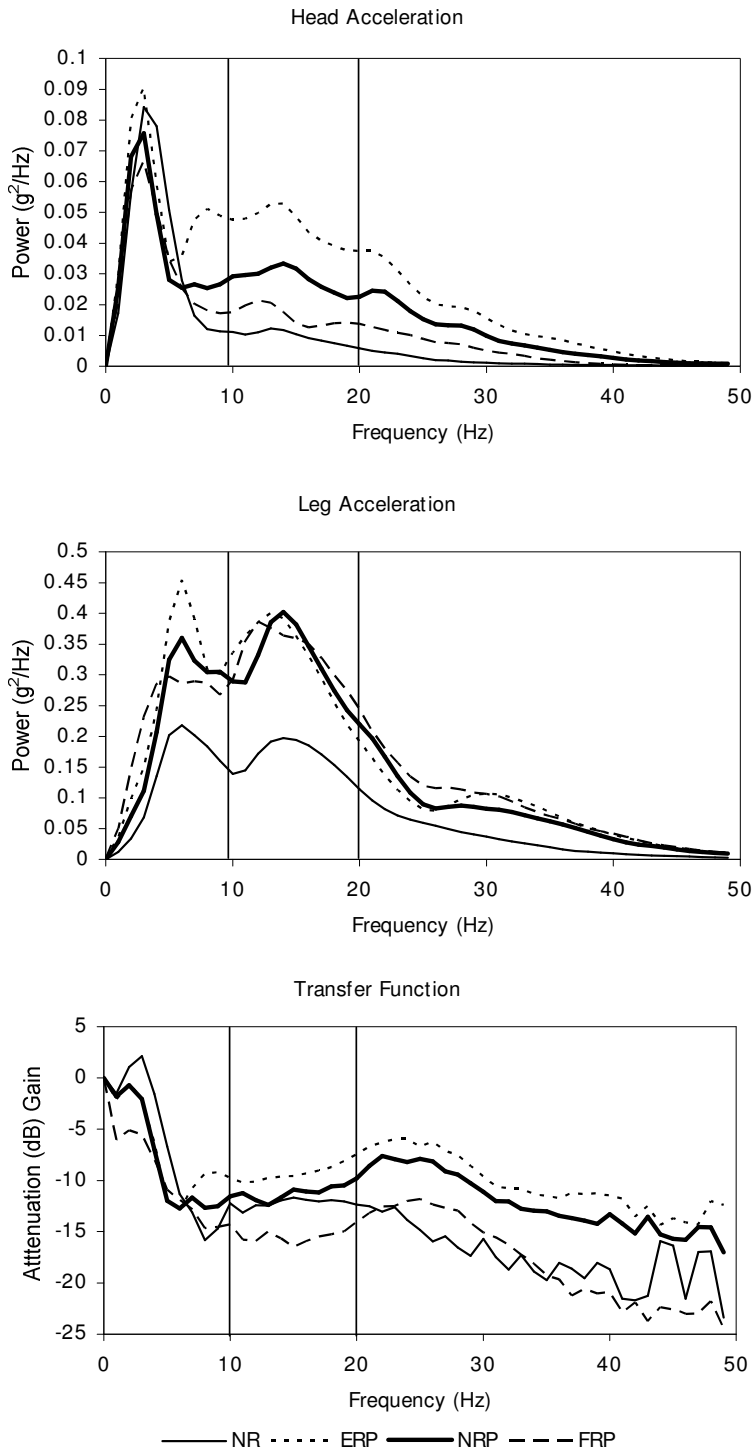


Figure 1.3

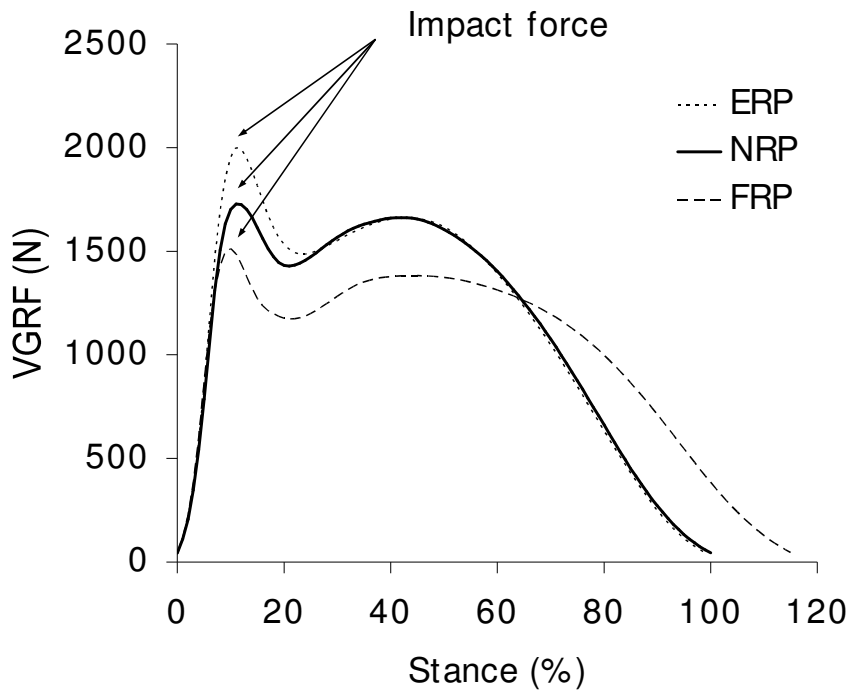
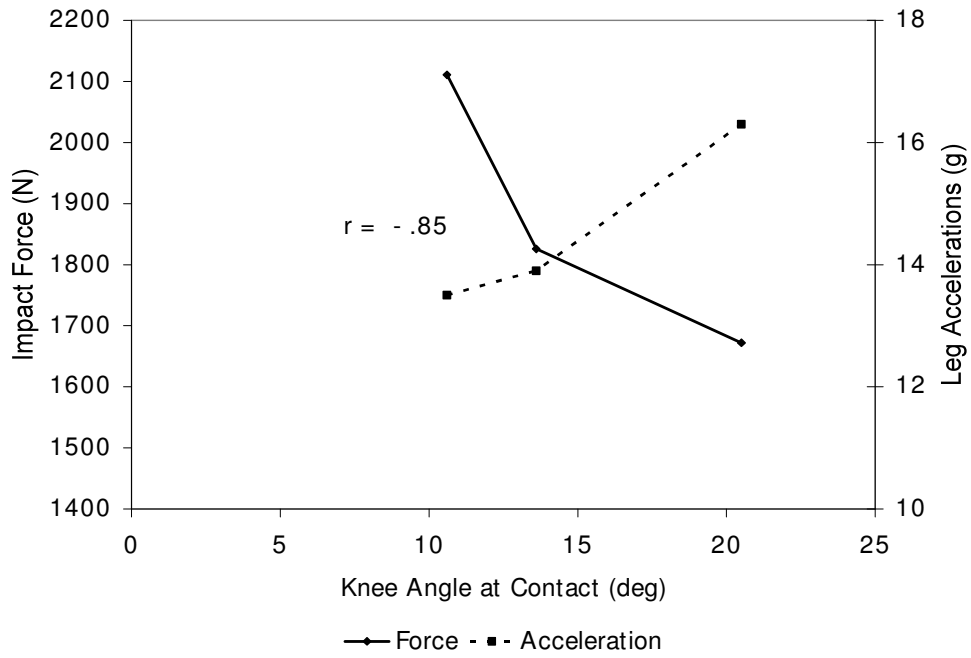


Figure 1.4





### **CHAPTER 3. RELATIONSHIP BETWEEN KNEE FLEXION AT CONTACT AND JOINT CONTACT FORCES IN THE KNEE**

A paper to be submitted to *The Journal of Applied Biomechanics*

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#### **Abstract**

The purpose of this study was to investigate the relationship between knee flexion at contact and joint contact forces at the knee. Knee flexion at contact is the degree of knee flexion present at initial footstrike. Joint reaction forces are determined through inverse dynamics. Ten runners completed ten trials in one normal running condition and three conditions involving a raised platform, which was used to exaggerate the loading environment. The three platform conditions were running with a preferred knee flexion at contact, running with a more flexed knee at contact, and running with a more extended knee at contact. Kinematic and kinetic data were captured and input into a two dimensional model of the lower extremity. From this model maximal dynamic muscle forces, moment arms, and muscle orientations were exported. Muscle forces were estimated using a minimized sum of muscle stress squared cost function. These variables were input into an optimization model in Matlab to calculate vertical and sagittal plane joint contact forces in the knee. Flexed knee platform running produced the highest compressive ( $14.1 \pm 1.5$  BW) and highest AP shear forces ( $3.4 \pm 0.8$  BW). Extended knee running produced the lowest muscle forces in the platform conditions (Compressive:  $13.0 \pm 1.7$  BW; AP Shear  $2.8 \pm 0.4$  BW).

## **Introduction**

Stress fractures are a common type of overuse injury in runners, and they occur most often in the lower extremity (Jones et al. 1989; Matheson et al. 1987; Brubaker & James 1974; Gudas 1980; Clement et al. 1981). Studies have shown an incidence of stress fractures of 21% among track runners (Bennell et al., 1996) and an incidence of 8-13% for recreational runners (Brunet et al., 1990), with tibial stress fractures the most common site of injury in the lower extremity (Hulkko & Orava, 1987; Matheson et al. 1987; Brubaker & James 1974; Gudas 1980; Orava et al. 1978; Devas 1958; Sullivan et al. 1984; Taunton et al., 1981).

Past research has investigated the relationship between kinematics of the lower extremity and the corresponding impact forces. High impact forces have been related to increased injury risk (Clement & Taunton, 1981; Milner et al. 2006). One strategy for reducing loads on the body when running is to increase knee flexion at ground contact. Research indicates that increases in knee flexion angle can decrease impact forces when running off a 22.5 cm platform (Thomas Unpublished manuscript). This may help the body increase shock absorption. McMahon, Valiant, and Frederick (1987) determined that increasing knee flexion during the stance phase of running decreased the transmission of impact through the body, as well as reducing peak vertical force at midstance. However, knee flexion may also increase muscle forces and therefore joint contact forces. The increase in muscle forces could increase the stiffness of the lower extremity. This increased stiffness may increase the compressive forces in the knee joint. Knee joint contact forces are comprised of joint reaction forces (due primarily to the ground reaction forces) and muscle forces (due to the contraction of muscles that cross

the knee joint). These joint contact forces are a more direct measure of the loads that lead to stress fractures than ground reaction forces or tibial accelerations.

It has been shown that the knee angle can affect the knee compressive forces. During leg press and squat exercises, tibiofemoral compressive force increased by approximately 2000 N as knee flexion increased 20° to 80° (Zheng et al., 1998). However, that study also showed a 2000 N decrease in tibiofemoral compressive force with the same increase in knee angle during a knee extension exercise. The authors suggested that a dominant factor in compressive force may be the activation of the muscles. A flexed knee is also subject to shear forces due to quadriceps tension (DeMorat et al., 2004). Increasing knee flexion at contact from 10.6° to 20.6° decreased impact force by over 400N (Thomas Study 1). Additionally, peak leg acceleration increased almost 3 g's while attenuation increased from 59.3% to 84.2% during that same increase in knee flexion.

Scott and Winter (1990) were among the first researchers to emphasize the importance of looking beyond ground reaction forces when assessing the importance of skeletal loading to injury. They developed a model to analyze joint contact forces during running and their connections to injury. They found peak vertical ground reaction forces of 2.1 body weights (BW) during the impact, and 2.7 BW occurring around midstance. The vastus medialis, vastus lateralis, and rectus femoris showed peak activity around 12% of stance. However, ankle compressive forces ranged from 10.3 – 14.1 BW and occurred just after midstance. Ankle shear forces ranged from -0.4 to -0.7 BW and occurred just prior to midstance.

A similar study estimated contact forces at the distal tibia during stance in running (Sasimontokul et al., 2007). The influences of external forces and internal forces were also accounted for. Compressive forces from ground reaction forces (GRF) averaged 1.97 BW and AP shear forces 1.23 BW. The average peak tibia compressive force was 9.0 BW. The peak tibia AP shear force was 0.57 BW. The researchers noted that the peak forces typically occurred around midstance, suggesting that these midstance forces could play a more prominent role in the occurrence of stress fractures than ground reaction forces at contact. However, neither Scott and Winter or Sasimontokul et al. investigated rates of loading, which may be important in the incidence of stress fractures (Milner et al. 2006).

The purpose of this study was to investigate the relationships between knee flexion at contact and the joint contact forces while running off a platform. The platform will exaggerate the loading environment hopefully making it easier for the subjects to alter their knee flexion at contact. In addition, the platform will be similar to downhill running. Previous research by Thomas and Derrick (Study 1) indicated that runners increased their knee flexion at contact which resulted in decreased impact forces. It was speculated that this increased knee flexion may result in increased muscle activity due to the changes in the preferred running pattern. Our hypothesis is that as knee flexion at contact increases, muscle forces will increase along with compressive and AP shear forces at the knee.

## **Methods**

### Subjects

Recreational runners (five male ( $71.6 \pm 7.3$  kg,  $1.80 \pm 0.05$  m,  $29.0 \pm 7.7$  years) and five female ( $65.6 \pm 8.3$  kg,  $1.64 \pm 0.06$  m,  $21.6 \pm 1.1$  years) were recruited for this study. No standards for mileage or pace were required to participate. None of the subjects had lower extremity abnormalities that affected their ability to run and land off a raised platform. Informed consent was obtained for all subjects according to the Iowa State University Institutional Review Board.

### Protocol

Retroreflective markers were placed on the right toe, heel, ankle, knee, hip, and shoulder. Prior to data collection, subjects were allowed to self-select their running speed (mean  $2.7 \pm 0.4$  m·s<sup>-1</sup>). Subjects were then required to perform all trials within a  $\pm 5\%$  range. Subjects completed ten trials for each of four conditions in a single session. The four conditions were normal running (NR), normal running off a 22.5 cm platform (NRP), running with a flexed knee off a platform (FRP), and running with an extended knee of a platform (ERP). Subjects ran 10 times in each condition.

### Instrumentation

Running speed was monitored for variances within  $\pm 5\%$  of preferred via immediate feedback obtained using a microwave based radar gun (model Stalker ATS, Radar Sales) interfaced with a laptop computer. Running speed was measured more precisely during post analysis and was calculated from the average horizontal velocity of the hip marker during the stance phase of the running cycle. Kinematic data were collected using a six-camera motion capture system (Peak Motus). Markers were

digitized at a rate of 120 Hz and were low-pass digitally filtered at 16 Hz using a fourth-order Butterworth filter prior to the calculation of the joint angles. A strain gage force platform (model OR6-7-2000, AMTI) was used to collect ground reaction forces at 3600 Hz. Peak impact force was identified as the first peak of the bimodal vertical ground reaction force curves.

### Knee Model

Sagittal plane lower extremity joint moments were calculated during the stance phase of each trial. Kinematic ensemble averages for each subject and condition were imported into a scaled SIMM 4.0 musculoskeletal model (MusculoGraphics, Inc., Santa Rosa, CA) consisting of 34 lower extremity muscles. Maximal muscle forces were adjusted by physiological cross-sectional area, active length-tension, passive length-tension, and force-velocity relationships. Maximal dynamic muscle forces, sagittal plane moment arms and muscle orientations were exported from SIMM for each 1% of the stance phase. Actual muscle forces were estimated using static optimization techniques. The maximal dynamic muscle force values were used to calculate bounds for the muscle forces during the optimization. In addition, nonphysiological changes in muscle forces were prevented using activation dynamics (Pierrynowski & Morrison, 1985a, 1985b). The equality constraint ensured that the sum of the muscle forces multiplied by their moment arms equaled the sagittal plane hip, knee and ankle joint moments calculated with the inverse dynamics:

$$\mathbf{M}_{j=1}^3 = \sum_{m=1}^{34} F_m \cdot x_m$$

where  $M$  represents the three sagittal plane joint moments,  $F_m$  represents the force for each muscle and  $x$  represents the moment arm for each muscle. There are many sets of muscle forces that satisfy this equality constraint and therefore the optimization used a cost function to select the best solution. The cost function to be minimized was the sum of the squared muscle stresses. This cost function has been shown to provide a reasonable relationship between muscle force and EMG (Glitsch & Baumann, 1997).

Finally, the muscle forces were added to the joint reaction force to determine the joint contact force. Sagittal plane compressive and anterior/posterior shear forces were calculated at the proximal tibia by reorienting the forces to be perpendicular and parallel to the tibial plateaus respectively. Rates of loading were calculated using the first central difference method and then peak values were identified.

#### Statistical Analysis

Self-selected speed measured from the radar gun ( $2.7 \pm 0.4 \text{ m}\cdot\text{s}^{-1}$ ) was slightly different than that calculated from the kinematic system (NR:  $3.05 \pm 0.4 \text{ m}\cdot\text{s}^{-1}$ , NRP:  $2.91 \pm 0.3 \text{ m}\cdot\text{s}^{-1}$ , FRP:  $2.93 \pm 0.3 \text{ m}\cdot\text{s}^{-1}$ , and ERP:  $2.80 \pm 0.3 \text{ m}\cdot\text{s}^{-1}$ ). Thus, data were analyzed using a repeated measures ANCOVA using running speed as a covariate. There was a single factor with knee flexion at contact and four levels to account for the four conditions. A Tukey's post hoc analysis was used to identify differences between the conditions ( $\alpha = 0.05$ ) when appropriate. The key dependent variables were knee flexion at contact, peak joint moments, peak muscle forces, peak compressive and AP shear forces, and peak compressive and AP shear rates of loading.

## Results

There were significant differences in knee flexion at contact between all conditions (NR:  $17.3 \pm 6.2^\circ$ ; NRP:  $13.6 \pm 4.5^\circ$ ; FRP:  $20.6 \pm 5.2^\circ$ ; ERP:  $10.6 \pm 3.7^\circ$ ; Table 2.1). Participants chose to land more extended when running off the platform compared to normal running. All of the participants were able to voluntarily increase their knee flexion at contact and nine out of ten were able to increase knee extension at contact. Relative to the normal condition, it appears that the participants found it easier to increase knee flexion at contact ( $+6.9^\circ$ ) than it was to decrease knee flexion at contact ( $-3.0^\circ$ ) while running off the raised platform (Figure 2.1). The ensemble curves show that the magnitude of knee flexion in FPR was much different than the other three conditions.

Statistical results of muscle force changes are reported in Table 2.2. Muscle forces produced in the hamstrings, quadriceps, and gastrocnemius are presented as a function of stance phase (Figure 2.2). Within the muscle groups, the highest muscle force value in the four conditions was found in the quadriceps muscle group (7.6 – 10.7 BW). The quadriceps reached a peak around 40 – 60% of stance; the gastrocnemius peaked slightly later, at around 60%. The hamstrings produced lower forces (1.3 – 0.73 BW), and were only active during the first 20% of stance.

Within the running conditions, increasing knee flexion at contact (FRP) produced the highest muscle force values compared to the other three conditions in the hamstrings (1.3 BW) and quadriceps (10.7 BW). The quadriceps force was significantly higher in FRP than the other three conditions. NRP was also significantly higher than NR and ERP. The FRP hamstring force was significantly higher than ERP and NRP, but not NR. NR produced significantly higher force than ERP as well. Increasing knee flexion at contact



produced the lowest muscle force in the gastrocnemius (4.0 BW) compared to the peak muscle force recorded during NRP (4.6 BW), which was significantly different from the other three conditions. Flexed knee running also peaked earlier than the other conditions in the hamstrings, while peaking later in the quadriceps. The gastrocnemius force was fairly uniform across all conditions.

The normal running condition produced the lowest compressive forces. Of the three platform conditions, peak compressive forces were lowest during extended platform running (Table 2.1). This value was over 1 BW lower than flexed platform running ( $13.0 \pm 1.7$  BW to  $14.1 \pm 1.5$  BW). Peak AP shear forces were also lowest in the extended platform running ( $-2.8 \pm 0.4$  BW/s), and significantly lower than FRP ( $-3.4 \pm 0.8$  BW/s). Negative AP shear forces indicate a posteriorly directed force. Ensemble curves of the compressive forces show that there appears to be little difference between the three platform conditions during the first 30% of stance (Figure 2.3). The compressive force was lowest in normal running from approximately 5% to 60% of stance. The AP shear force was also lowest in normal running from 20 – 50% of stance. In the later stages of stance (40 – 100%), flexed knee running has the highest compressive and AP shear forces. The flexed knee running produced the highest peak compressive and AP forces, although these peaks occurred later than the other conditions. The FRP compressive force occurred around 55 – 60% compared to 45 – 50% for the other three conditions. The FRP AP shear peak occurred around 60% compared to 50 – 55% in the other three conditions.

None of the rates of loading were significantly different. The highest compressive rate of loading was in the extended platform condition ( $370.6 \pm 119.1$  BW/s). There was little difference between NRP and FRP. Normal running was the lowest rate of loading

(208.8±49.4 BW/s). The AP shear rates of loading were highest in FRP (-88.3±34.0 BW/s), and lowest during NRP (-79.5±27.3 BW/s). The overall lowest rate was in the normal running condition (NR: -56.7±16.6 BW/s).

## **Discussion**

Running with a flexed knee appears to have a significant effect on joint contact forces during running. Flexing the knee at contact produced the highest muscle forces in the hamstrings and the quadriceps muscle groups, although the hamstring force peak (FRP:1.1 BW) was not much higher than the other peaks (NR:0.74 BW, NRP:0.57 BW, ERP:0.49 BW). Quadriceps force while flexing the knee at contact was almost 2 BW higher than the normal platform running condition.

Most previous research has looked only at the ankle, which prevents the comparison of muscle forces in the quadriceps and hamstrings. The gastrocnemius force in the present study was higher than Sasimontonkul et al. (2007). Although the value was not reported, a figure examination shows the gastrocnemius force to be no higher than 1 BW, which is much lower than the present values near 4 BW. The current values were closer to Burdett (1982), which had values ranging from 5.3 BW to 10.0 BW, although Burdett included both the gastrocnemius and soleus together as a plantar flexor muscle group.

Compressive forces in the current study were comparable to Burdett (1982) and Scott and Winter (1990), and slightly higher than Sasimontonkul et al. (2007). The current study found compressive forces ranging from 12.5 – 14.1 BW, while the previous research results were 8.0 – 13.0 BW (Burdett 1982), 10.3 – 14.1 BW (Scott and Winter 1990), and 9.0 BW (Sasimontonkul et al. 2007).

Anterior-posterior shear forces in the current study were comparable to Burdett (1982) but much higher than Sasimontongkul et al. (2007) and Scott and Winter (1990). Current results were in the range of -2.9 BW for normal running, similar to 3.3 – 5.5 BW in Burdett (1982). Sasimontongkul et al. (2007) found AP Shear forces of 0.57 BW, and Scott and Winter (1990) ranged from -0.4 to -0.7 BW. Those two studies had average running speeds ranging from 3.5 to 5.1  $\text{m}\cdot\text{s}^{-1}$ , which was higher than the current research which ranged from 2.8 to 3.05  $\text{m}\cdot\text{s}^{-1}$ . Running speed may have been a factor in the difference in magnitudes.

Overall, normal running produced the lowest compressive forces. This was likely a result of not having a platform drop. In the platform conditions, flexed knee running produced the highest peak compressive and peak AP shear forces. Extended knee running produced the lowest peak compressive and AP shear forces of the three platform conditions. It has been shown that increased knee flexion while running can reduce ground reaction forces. It has been suggested that this increased flexion may also result in an increased metabolic cost because of the increased muscle activity (Denoth 1986). The current research supports this theory as we see that increasing knee flexion resulted in increased muscular activity.

The results showed that knee compressive forces and quadriceps muscle forces were higher in the FRP than the other three conditions. The quadriceps force was 19.2% greater than the next highest condition (NRP), while compressive force was about 2.2% greater. Meyer and Haut (2005) suggest a flexed knee position may increase the risk for ACL injury due to compressive forces. They suggested that compressive loading could cause an anterior shift of the tibia relative to the femur due to the angle of the tibial

plateau, and that this shift could cause ACL injury. It is unlikely that ACL injury would occur in normal distance running, but it is a consideration for activities involving running while changing direction such as soccer or basketball, or landing from a platform.

Due to the lack of a platform drop, it may have been expected that the NR condition would produce the lowest muscle force. The drop would require the muscle to be more active to slow the runner's vertical center of mass. However, the NR condition only produced the lowest force in the quadriceps group (Figure 2.1). In the gastrocnemius the lowest peak muscle force was in the FRP condition, while in the hamstring it was ERP.

Glitsch and Baumann (1997) indicated that the gastrocnemius, rectus femoris, sartorius, tensor fascia latae and the hamstrings bear the greatest load during stance phase in walking and forefoot running. The current results showed the highest load in the quadriceps (7.6 – 10.3 BW), followed by the gastrocnemius (3.8 – 4.5 BW). The force at the hamstrings was small (0.49 – 1.1 BW). The hamstring force may have been low in our study due to the optimization protocols, which tend to minimize co-contraction in muscles (Gottlieb 2000). The use of a platform likely required a more dominant quadriceps action to arrest the downward motion of the center of mass (opposing knee flexion). Additionally, the subjects running range was limited by the length of wire attachments, which had been used to attach accelerometers for earlier research (Thomas Study 1). This short runway may have caused subjects to reduce hamstring activity during the propulsive stride off the force platform. The biarticular nature of the hamstrings may have also been penalized in the model in favor of the gluteals and quadriceps muscles.

There were a few limitations to this study. The model used was only two dimensional, incorporating sagittal plane movements which were used in the optimization. Although medial-lateral forces were not accounted for, most movement in running occurs primarily in the sagittal plane. Ensemble curves were also used as an average of the individual trials in each condition. This may have reduced the effects on rate variables such as rate of loading because of the differences in timing. Peaks occurring at different times will reduce the averages in the ensemble curves.

Although flexing the knee more during contact decreases the ground reaction impact force, it does not result in lower joint contact forces at the knee joint because of the increased muscle forces necessary to maintain the flexed posture. Since attenuation of forces is increased with knee flexion, the joint contact forces at the hip may be reduced. Subjects likely extended their knee more during the platform running rather than normal running due to the effect of the platform. The extra vertical height gave subjects more time to extend the knee as they reached for the force platform. This could result in higher injury potential because a straighter leg tends to increase impact forces.

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## Tables

Table 2.1. Mean and (sd) of compressive and shear forces and rates of loading during four knee flexion conditions during running.

	Normal Running	Extended Running - Platform	Normal Running - Platform	Flexed Running - Platform
Knee angle at contact (deg)	17.3 bcd (6.2)	10.6 acd (3.7)	13.6 abd (4.5)	20.6 bcd (5.2)
Peak compressive forces (BW)	-12.5 (1.7)	-13.0 d (1.7)	-13.8 (1.8)	-14.1 b (1.5)
Peak AP shear forces (BW)	-2.9 (0.5)	-2.8 d (0.4)	-3.0 (0.4)	-3.4 b (0.8)
Peak compressive rates (BW/s)	-208.8 (49.4)	-370.6 (119.1)	-351.9 (95.0)	-351.4 (118.6)
Peak AP shear rates (BW/s)	-56.7 (16.6)	-83.9 (51.7)	-79.5 (27.3)	-88.3 (34.0)

- a. Significantly different from NR
- b. Significantly different from ERP
- c. Significantly different from NRP
- d. Significantly different from FRP

Table 2.2. Mean and (sd) of muscle forces during four knee flexion conditions during running.

	Normal Running	Extended Running - Platform	Normal Running - Platform	Flexed Running - Platform
Quadriceps Force (BW)	7.6 cd (2.1)	8.0 cd (1.8)	8.7 abd (2.1)	10.7 abc (3.7)
Hamstring Force (BW)	0.98 bd (0.57)	0.73 ad (0.50)	0.82 d (0.59)	1.3 bc (0.76)
Gastrocnemius Force (BW)	4.3 c (0.65)	4.3 c (0.41)	4.6 abd (0.54)	4.0 c (0.70)

- a. Significantly different from NR  
b. Significantly different from ERP  
c. Significantly different from NRP  
d. Significantly different from FRP

## Figures

Figure 2.1. Ensemble knee joint angles during stance phase of running with four knee flexion conditions. Positive slope indicates the knee is flexing while negative slope indicates the knee is extending.

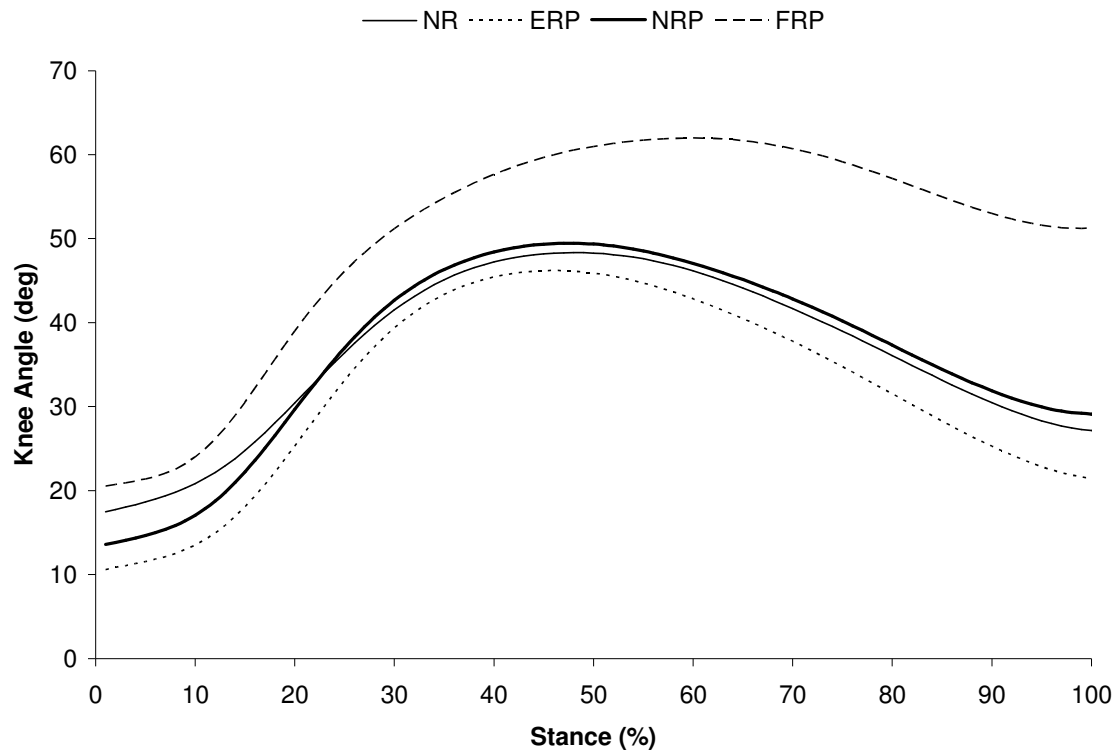
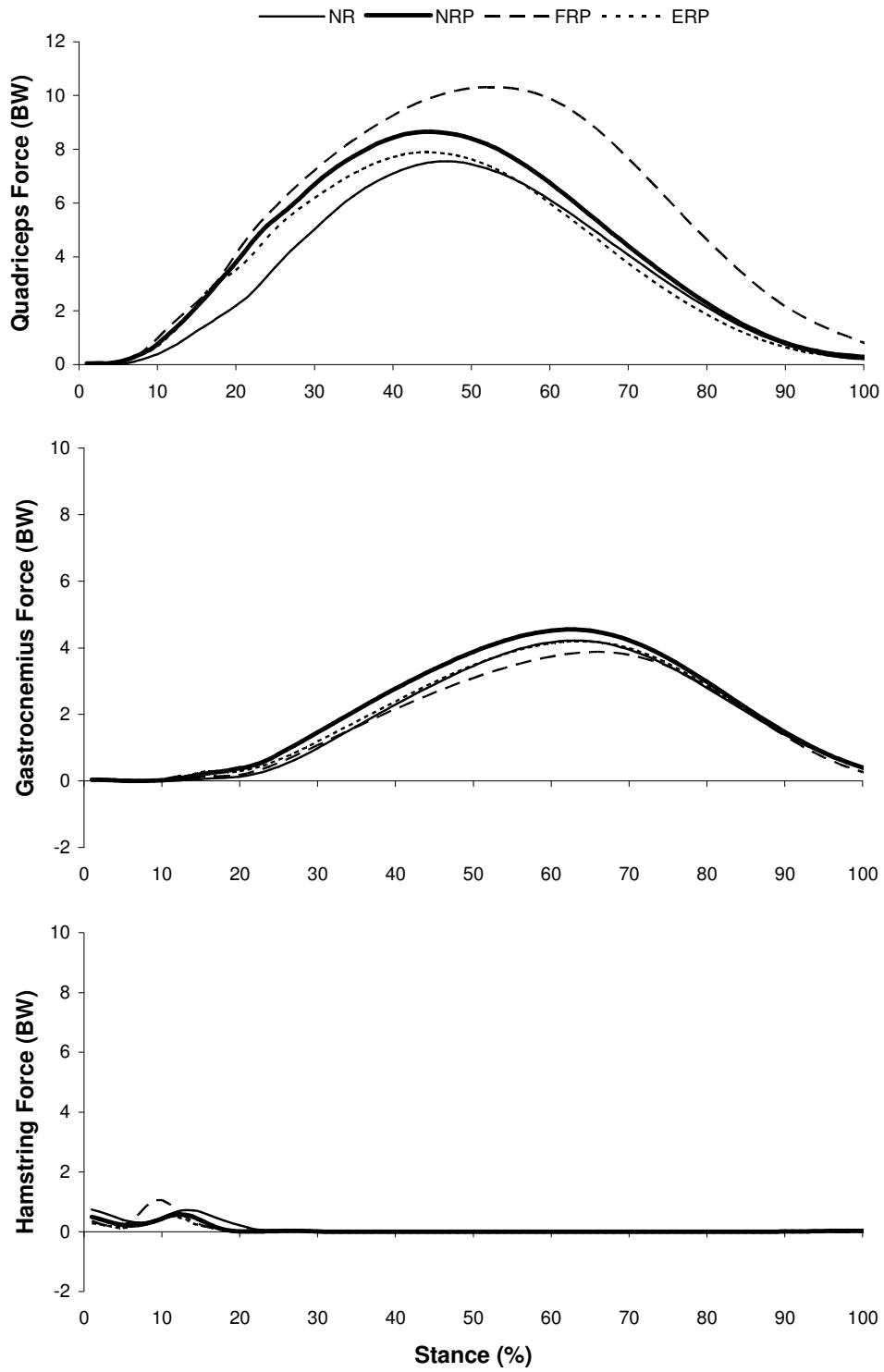




Figure 2.2. Muscle force activity during stance phase in running. Four different conditions of knee flexion are represented. Curves are ensemble representations.





## CHAPTER 4. KNEE JOINT CONTACT FORCES WITH CHANGES IN STRIDE LENGTH AND MIDSOLE STIFFNESS

A paper to be submitted to *Medicine and Science in Sports and Exercise*

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### Abstract

The purpose of this study was to determine how lower extremity joint loading is affected by altering the stride length and the shoe midsole hardness. Ten subjects running at an average of  $4.43 \text{ m}\cdot\text{s}^{-1}$  completed 10 trials in each of three conditions: normal running, running with a stride length shortened by 10% of normal, and running with a decrease in midsole stiffness (from 13.7 g's to 10.9 g's). Reaction forces calculated from inverse dynamics were summed with muscle forces estimated from static optimization to obtain joint contact forces. Peak compressive, anterior-posterior (AP) shear, and medial-lateral (ML) shear forces as well as rates of loading were examined at the hip, knee, and ankle. Reducing the stride length resulted in statistically significant reductions in peak ML shear (from 0.83 to 0.65 body weights (BW)) and compressive forces (14.3 to 13.6 BW) as well as peak AP shear rates (81.9 to 75.6 BW/s) at the ankle. Increased midsole cushioning also reduced peak ML shear forces (0.83 to 0.75 BW) at the ankle but to a lesser extent. At the knee, the reduced stride length significantly reduced the peak AP shear force (1.85 to 1.75 BW) and the peak compressive force (14.9 to 13.8 BW) while the shoe cushioning had no significant effects. Neither treatment produced any significant effects at the hip joint. Overall the 10% reduction in stride length seemed to have a greater affect than the increased shoe midsole cushioning.

Key Words: Gait, Compressive Force, Shear Force, Cushioning

## Introduction

Repeated impacts from activities such as running can lead to overuse injuries in the lower extremity. The incidence of stress fractures is 21% among track runners (**Error! Reference source not found.**) and an 8-13% for recreational runners (6). An incidence of 4.0% was found in military recruits during 12 weeks of training, with the highest injury rates correlating with the weeks of the highest training volume (1). A ten-year study of 6000 athletes at the University of Minnesota found an overall incidence of stress fractures of 1% but the incidence was higher in female athletes (1.9%) and distance runners (3.2%) (2). The tibia was the bone with the highest injury rate (37%), although there were more injuries in the foot when considered as an anatomic region.

While there are numerous possible causes of overuse injuries such as stress fractures, less is known about how to actually prevent them. Clement and Taunton (10) indicated that high impact forces are related to increased injury risk. Loading of the lower extremity may also cause overuse injury prior to actual fracture in the bone. For example, runners training for a marathon had a 28% rate of injury in the anterior knee (22). The rate of loading due to impact may also play a role in injury occurrence. Research has indicated that high rates of loading may produce more injury than repeated loading at a low rate of loading (28).

Direct measurements on internal structures of the lower extremity are difficult in human subjects. Burdett (7) used a musculoskeletal model of the lower leg to estimate joint contact forces at the ankle. Anatomical measurements were taken from cadavers, and the relevant data for the functional muscle groups of the ankle were input into the model. These data included muscle insertion points, line of pull, and physiological cross-

sectional area. The model was used to estimate joint forces during the stance phase of running. Kinematic and kinetic data for the model were collected from three subjects running at an average speed of  $4.47 \text{ m}\cdot\text{s}^{-1}$  across a force platform. The results indicated compressive forces of 8-13 body weights (BW) during midstance. AP shear forces were 3.3 to 5.5 BW, and ML shear force ranged from 0.5 to 0.8 BW. Peak muscle forces were highest in the plantar flexor group (gastrocnemius and soleus) at a range of 5.3 to 10 BW.

Scott and Winter (30) estimated peak vertical ground reaction forces of 2.1 BW at heel contact. They also estimated ankle compressive forces ranging from 10.3 – 14.1 BW, with an average peak compressive force of 11.2 BW. Ankle AP shear forces in their model ranged from -0.4 to -0.7 BW. Sasimontokul, Bay, and Pavol (27) estimated contact forces at the distal tibia during stance in running. Subjects ran at speeds ranging from 3.5 – 4.0 m/s. Compressive joint reaction forces averaged 1.97 BW and AP shear joint reaction forces averaged 1.23 BW. The average peak compressive joint contact force at the tibia was 9.0 BW. The peak AP shear joint contact force was 0.57 BW.

Two possible ways of reducing these joint contact forces are by attenuating the impacts at the foot through the use of a softer midsole in footwear or by reducing the impacts by shortening the stride length of the runners.

The effects of increasing midsole cushioning on impact loads are unclear. Some research indicates that cushioned inserts reduced tibial shock during gait by about 40% (32), and that cushioned running shoes may decrease injury (19). Lafortune et al (21) indicated that stiffness of the contact surface (such as running surface or midsole cushioning) may play a larger role in decreasing impact than changes in knee angle. They found that softer surfaces resulted in a decreased severity of impact as well as decreased

leg stiffness. Milgrom et al (22) suggested that footwear could be used to dampen the bending moments that may lead to stress fractures. One study determined that shock absorbing insoles do not significantly reduce stress fractures (14).

Other research gives evidence that midsole cushioning does not have an affect on lower extremity impacts (5,29) or attenuation (24,25,26). According to Hardin and Hamill (17) midsole stiffness had no effect on peak tibial acceleration. Kersting and Bruggemann (20) also found no difference in impact force with varying midsole cushioning, but speculated that it may have been due to adaptations of the individual runners in response to changing shoe conditions. Hardin et al (18) also found that lower extremity kinematics of runners adapted to passive changes such as changing midsole cushioning.

Research concerning the effect of stride length on impact is less debated. Derrick et al (13) indicated that an increase in stride length decreased tibial acceleration, from 6.1 g during preferred stride length to 5.7 g with a 20% stride length reduction. Other research has supported this, showing that a decrease in stride length resulted in decreased impacts (16).

The purpose of this study was to determine how joint loading is affected by altering the stride length and the shoe midsole hardness. We define joint loading by the peak contact forces and peak absolute rates of loading at the hip, knee and ankle joints. We hypothesized that reducing stride length would result in decreased joint loading and that increasing midsole cushioning would have lesser or no measurable effects on these parameters.

## **Methods**

### Subjects

Ten men ( $69.2 \pm 6.5$  kg,  $1.78 \pm 0.05$  m,  $22.2 \pm 3.2$  years) were recruited for this study. Nine of the subjects were current or former college cross-country runners. The average weekly mileage was  $63 \pm 20$  miles. Subjects were limited to those who could wear a size 9 or 11 shoe. None of the subjects had lower extremity abnormalities that would affect their ability to run.

### Protocol

Upon entering the laboratory, all subjects read and signed an Informed Consent Document approved by the Human Subjects Research Office at Iowa State University. Height, total body mass, thigh length, mid-thigh circumference, calf length, calf circumference, foot length, malleolus height, malleolus width, and foot breadth were measured on each subject in order to build a model for inverse dynamics analysis (31). All subjects wore the Adidas 1.1 running shoe with adjustable cushioning. Each shoe contained a small microprocessor, a sensor, and a motorized cable system in the heel of the shoe. Through the use of a magnetic “key” the cable length can be lengthened or shortened to create a softer or firmer cushion as desired. Retroreflective markers were placed on the anterior dorsifoot, heel, medial and lateral malleoli, anterior calf, medial and lateral knee, anterior thigh, right and left greater trochanter, right and left ASIS, and L5/S1 region of the lower back. Prior to data collection, subjects were allowed to self-select their running speed. Subjects were asked to run down a runway at a comfortable training velocity towards a force platform imbedded in the floor, landing on the force platform with the right foot without targeting the force platform. Trials were accepted if

the speed was  $\pm 5\%$  of their preferred running speed. Average self-selected speed was  $4.43 \pm 0.48 \text{ m}\cdot\text{s}^{-1}$ . Subjects completed ten trials for each of three conditions. The three conditions were a normal stride running condition (NS), a condition with a stride length reduced by 10% (RSL) and a condition with a cushioned stride via a soft midsole (CS). The shorter stride lengths were achieved through the use of floor markings. All 30 trials were completed in a single session.

### Data Analysis

The Adidas 1.1 shoes were tested for cushioning differences using an Exeter Impact Tester to verify the degree of cushioning differences between midsole conditions. Motion capture data were collected with a Peak Motus 3D optical capture system (Vicon Peak, Centennial, CO) with a sampling frequency of 120 Hz. During dynamic trials, force platform data were collected concurrently with a sampling frequency of 1200 Hz using a strain gage force platform (model OR6-7-2000, AMTI). Marker coordinates were low-pass digitally filtered at 16 Hz using a fourth-order Butterworth filter prior to the calculation of the lower extremity kinematics. Running velocity was determined through use of the L5/S1 marker captured by the Peak Motus motion capture software and was calculated from the average horizontal velocity of the marker during the stance phase of the running cycle. The synchronized raw motion capture and force platform data were then exported to Matlab for data processing.

Dynamic motion capture data were referenced to a static trial for the calculation of dynamic joint centers and local (segmental) coordinate systems for the foot, calf, thigh, and pelvis. Three-dimensional Cardan joint angles were calculated using a flexion/extension, abduction/adduction, internal/external rotation sequence.



Three-dimensional joint moments and reaction forces were calculated using standard inverse dynamics. Joint moments and reaction forces were calculated in the global coordinate system and then rotated into their respective segment coordinate systems. Force and moment signs indicate loads acting on the proximal end of the distal segment of the joint. Negative forces indicate compression, medial, and posterior forces along their respective axes.

The stance phase joint angles for each trial were interpolated to a percentage of stance (1% increments) and imported into a scaled SIMM 4.0 musculoskeletal model (MusculoGraphics, Inc., Santa Rosa, CA). The SIMM model was used to obtain maximum dynamic muscle force, muscle moment arms, and muscle orientations for 44 lower extremity muscles.

Actual muscle forces were estimated using static optimization techniques. The cost function ( $u$ ) to be minimized was the sum of squared muscle stresses (11,12)

$$u = \sum_{i=1}^{44} (f_i / PCSA_i)^2$$

where  $f_i$  is the force generated by the  $i^{th}$  muscle, and  $PCSA_i$  is the physiological cross-sectional area of the  $i^{th}$  muscle. The optimization was constrained so that the resulting hip and sagittal plane knee and ankle moments equaled those from inverse dynamics.

Three dimensional joint contact forces were calculated as the sum of reaction forces and muscle forces crossing the joint. Rates of loading were calculated using the first central difference method. Peak joint contact forces and rates of loading at the hip, knee, and ankle were calculated and averaged across subjects. Group ensemble curves were generated for joint contact forces and rates of loading.

Outliers were determined using boxplots and those trials were removed from the data before statistical analysis was done. Data were analyzed as part of a multiple group experiment using a univariate ANOVA with a single level and three factors. A Tukey's post hoc analysis was used to identify differences between the conditions ( $\alpha = 0.05$ ) when appropriate. Footstrike pattern was not controlled and post hoc analysis indicated that four of the ten runners were forefoot strikers according to the footstrike index.

## **Results**

### Verification of Cushioning Data

Impact testing of the Adidas 1.1 shoes showed peak values of 10.9 g's for the cushioned setting. The alternate setting, which was used for the normal stride condition and the reduced stride length condition, tested at 13.7 g's. This provided verification that there were in fact differences in the cushioning properties between conditions. A compilation of 142 commercially available running shoes tested in the same laboratory showed a range of 7.2 to 13.5 g's using the same protocol (3).

### Ground Reaction Forces

Ensemble vertical ground reaction force curves are displayed in Figure 3.1. There is a noted similarity between normal running and the increased cushioning condition, while the reduced stride length VGRF appears to be lower through the first 50-60% of stance. The maximum values of the ensemble force peaks of NS and CS were similar, with values of 2.96 BW and 2.93 BW respectively. The maximum value of the RSL ensemble curve was 2.81 BW.

### Forces at the ankle

Peak joint contact forces at the ankle are presented in Table 3.1. AP shear forces during the RSL condition were significantly lower than the CS condition (2.34 to 2.73 BW). Peak AP shear forces occurred at approximately 50 – 60% of stance (Figure 3.1). The compressive forces were significantly lower (13.6 BW) in RSL compared to the other two conditions (NS: 14.3 BW;  $p < .001$ , CS: 14.3 BW;  $p=.024$ ). Peak compressive force occurred around midstance (55%). The ML Shear forces were highest in normal running (0.83 BW) and significantly different in all three conditions (RSL: 0.65 BW, CS: 0.75 BW). They reached a negative (medially directed) peak in early stance (~15%), with a positive (lateral) peak around midstance (50 – 60%). At the ankle, both compressive force and AP shear force peak around 50 – 60% of stance. The ML shear force has a positive peak near the same point, but its highest peak is a negative force around 15 – 20% of stance. The AP shear force also has a slight posteriorly directed force in late stance. During most of the stance phase the shorter stride condition appears to have slightly lower magnitudes than the other two conditions.

### Forces at the knee

Peak forces at the knee joint are presented in Table 3.2. AP shear forces were significantly lower (-1.75 BW) during the RSL condition compared to the other two conditions (NS: -1.85 BW,  $p < .001$ ; CS: -1.84 BW,  $p=.007$ ), peaking near 60% of stance (Figure 3.2). The compressive forces were also significantly lower (-13.8 BW to -14.8 BW) during the RSL condition. Compressive forces in the knee did not reach their peak value until approximately 50% of stance phase. There were no significant differences among the ML shear forces. Compressive and AP shear forces both peaked near

midstance, although at the knee the AP shear force is posterior directed rather than anterior directed as in the ankle. The AP shear force does show an early anterior component in the first 20% of stance. The ML shear force again peaks around 15% but remains negative throughout stance in contrast to the ankle. Again the shorter stride condition appears to reach lower magnitudes, although this is not quite as strong in the AP shear forces.

### Forces at the hip

Peak hip forces are presented in Table 3.3. AP shear forces were significantly lower (-1.49 BW) during RSL compared to the CS condition (-1.69 BW;  $p=.031$ ). As in the other joints, AP shear forces peaked near 60% of stance phase. (Figure 3.3). There were no significant differences in compressive forces. Hip compressive forces peaked around 18% of stance, with a slightly lower second peak near 40%. The ML shear force in the RSL was significantly lower (5.42 BW) than CS (6.31 BW;  $p=.005$ ), reaching their peak near 40% of stance. At the hip, the force patterns are more variable than at the ankle and knee. The AP shear force alternates between the anterior and posterior directions, before finally reaching a peak in the posterior direction near 60% of stance. The compressive force also deviates from the nice smooth pattern seen in the ankle and knee. At the hip the force does remain negative, but there are two peaks, the highest near 15% of stance and the other around 40% of stance. The magnitude is also about 4 BW lower than in the ankle and knee. At the hip the ML shear force also has a directional change, from negative at the knee to positive at the hip. The force curve is also bimodal, with an early peak near 15% stance and the highest peak around 40% of stance. Reducing stride length decreased the magnitudes, with a greater effect in ML shear force.

### Rates of loading

There were no significant differences in peak absolute rates of loading at the knee or hip (Table 3.1, 3.3). At the ankle, the average peak absolute AP shear rate of loading during RSL was significantly lower than NS and CS running (Table 3.2). However, there were some significant interactions when accounting for footstrike pattern. Ankle ML shear rate of loading had a significant interaction ( $p=0.017$ ; Figure 3.5). At the knee, all three axes indicated either a significant interaction or at least approached a significant interaction (Figure 3.6). The absolute rate of loading of AP shear ( $p = 0.062$ ) and compressive force ( $p = 0.051$ ) approached significance, while ML shear was in fact significant ( $p = 0.049$ ). AP shear absolute rate of loading decreased from RSL to CS in rearfoot strikers, but increased in forefoot strikers.

### **Discussion**

The purpose of this study was to determine how joint loading is affected by altering the stride length and the shoe midsole hardness. We defined joint loading by the peak contact forces and the peak absolute rates of loading at the hip, knee and ankle joints. We hypothesized that reducing stride length will result in decreased joint contact forces. We hypothesized that reducing stride length would result in decreased joint loading and that increasing midsole cushioning would have lesser or no measurable effects on these parameters.

The vertical ground reaction force results were in line with previous research trends. Research has produced VGRF values of 1.32 BW (9), 1.97 BW (27), approximately 3 BW (30), approximately 2 BW (15). There was little difference between normal running and the increased cushioning, which agrees with previous results

(5,14,17,24,25,26,29). The decrease in ground reaction forces with a shorter stride length also agrees with previous research. Hamill et al (16) had runners increase from a preferred stride frequency to a 20% higher stride frequency. Because running velocity was kept constant, this increased stride frequency would equate to a reduced stride length. With this change, peak impact accelerations at the leg were reduced. Active peaks were also reduced, although not as much. Derrick et al (13) found that tibial acceleration decreased from 6.1 g to 5.9 g to 5.7 g as subjects reduced their stride length from preferred to -10% to -20%. Additionally, reducing stride length also reduced the joint moments at the knee and the ankle.

Reducing stride length does appear to reduce joint contact forces, supporting our hypothesis. The results showed that reducing stride length decreased ML shear forces in the ankle and decreased AP shear forces at the knee. The ankle also saw a decrease in rate of loading of the AP shear component. Finally, reducing stride length resulted in decreased compressive forces at the ankle and at the knee. The compressive forces at the ankle (NS:14.3 BW, CS:14.3 BW, RSL:13.6 BW) were comparable to the results of Burdett (7) (8 – 13 BW), Scott and Winter (30)(10.3 – 14.1 BW) and Sasimontokul (27)(9 BW).

Our hypothesis on the effect of cushioning was supported at the hip and the knee. At these two joints there was little effect of midsole cushioning in relation to joint contact forces. At the ankle, there was no significant change in compressive force or AP shear force, but increasing cushioning did result in a decrease in ML shear force.

Another question which may be answered with these results is whether reducing stride length or increasing cushioning is a better strategy for decreasing joint contact

forces. The results strongly support that reducing stride length is a better strategy.

Reducing stride length resulted in several significantly lower conditions of joint contact forces when compared to the cushion condition. Reduced stride length produced lower AP shear forces at the hip, knee, and the ankle. It also produced lower ML shear forces at the ankle and the hip. Compressive forces at the ankle and knee were significantly lower in the shorter stride condition than the cushion condition. There was also a decrease in the AP shear rate of loading at the ankle. However, while reducing stride length may reduce the potential for injury, it does not come without cost. Changes from preferred stride length may result in increased oxygen consumption (8)

Observing the patterns of the ensemble curves allows us to better understand the timing of the loading on the lower extremity. At the ankle, both compressive force and AP shear force peak around 50 – 60% of stance. The ML shear force has a positive peak near the same point, but its highest peak is a negative force around 15 – 20% of stance. The shear force also has a slight posterior force in late stance. The reduced stride length condition reached lower peak magnitudes than the other two conditions.

At the knee, similar patterns emerged. Compressive and AP shear force peaked near midstance, although at the knee the AP shear force is posterior directed rather than anterior directed as in the ankle. The AP shear force does show an early anterior component in the first 20% of stance. The ML shear force again peaks around 15% but remains negative throughout stance in contrast to the ankle. Again the reduced stride length condition reached lower peak magnitudes.

At the hip, the AP shear force alternates between the anterior and posterior directions, before finally reaching a peak in the posterior direction near 60% of stance.

The compressive force also deviates from the pattern seen in the ankle and knee. At the hip the force does remain negative, but there are two peaks, the highest near 15% of stance and the other around 40% of stance. The magnitude is also about 4 BW lower than in the ankle and knee. At the hip the ML shear force also has a directional change, from negative at the knee to positive at the hip. The force curve is also bimodal, with an early peak near 15% stance and the highest peak around 40% of stance. Reducing stride length decreased the magnitudes, with a greater effect in ML shear force.

The patterns of the joint contact forces indicated that in most cases peak or near peak forces occurred near 40 – 60% of stance. This may indicate that injury potential during running may not simply be due to impact forces and impact accelerations at contact. The high joint contact forces at midstance also will play a role, and may in fact have more of a contribution to injury than ground reaction forces, as has been suggested previously (27,30).

During the data collection it was noted that some of the runners appeared to be more forefoot/midfoot strikers than rearfoot strikers. This presented a potential problem, as the adjustable cushioning in the Adidas 1.1 shoes was present only in the heel. Therefore, it would be possible that the cushioning would have no effect or a reduced effect if the runners were not striking with that part of the shoe. A footstrike analysis from the force platform center of pressure data confirmed that four of the ten subjects were forefoot or midfoot strikers. Statistical analysis of the interactions indicates that footstrike pattern may affect the rates of loading at the knee and also the ML shear rate of loading at the ankle. With ML shear rate of loading at the ankle, reduced strides appear to have no significant effect. However, it does appear that knowledge of the footstrike index



may be required to determine changes in normal or cushion. In ML shear rate of loading at the knee, being a rearfoot striker makes the cushion condition comparable to short stride, but being a forefoot striker has no effect. Although the interaction during AP shear rate of loading at the knee was not significant, during the cushion condition the forefoot strikers produced the highest rates of loading in the three conditions. The rearfoot strikers produced the lowest rates of loading in the cushion condition. It does not appear that being a forefoot striker had any effects on changes in joint contact forces or rates of loading as a function of stride length or cushioning.

### **Conclusions**

Reducing stride length while running appears to be a productive strategy for decreasing joint contact forces in the lower extremity. Conversely, increasing midsole cushioning does not appear to reduce joint contact forces in the lower extremity, except ML shear forces at the hip. Footstrike pattern does appear to cause interactions with midsole cushioning in rates of loading. Therefore it may be that the lack of cushioning differences is due to runners landing on the forefoot as opposed to the midfoot. Further study into the relationship between forefoot and rearfoot strikers and joint contact forces would be a helpful area of research.

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## Tables

Table 3.1. Mean and (sd) of AP shear, ML shear and compressive forces and rates of loading at the ankle.

	NS	RSL	CS
Peak AP shear forces (BW)	2.64 (0.9)	2.34 (0.9)	2.73 (1.0)
Peak ML shear forces (BW)	0.83 (0.2)	<b>0.65</b> <b>(0.2)</b>	<b>0.75</b> <b>(0.2)</b>
Peak compressive forces (BW)	14.3 (1.3)	<b>13.6</b> <b>(1.0)</b>	14.3 (1.3)
Peak AP shear rates (BW/s)	81.9 (14.0)	<b>75.6</b> <b>(12.9)</b>	83.6 (14.7)
Peak ML shear rates (BW/s)	51.3 (14.4)	45.7 (15.3)	50.8 (17.3)
Peak compressive rates (BW/s)	274.8 (57.9)	263.6 (49.0)	269.5 (47.9)

**Significant difference from NS**

*Significant difference between RSL and CS*

Table 3.2. Mean and (sd) of AP shear, ML shear and compressive forces and rates of loading at the knee.

	NS	RSL	CS
Peak AP shear forces (BW)	-1.85 (0.2)	<b>-1.75</b> <b>(0.2)</b>	-1.84 (0.2)
Peak ML shear forces (BW)	-1.19 (0.3)	-1.08 (0.3)	-1.18 (0.3)
Peak compressive forces (BW)	-14.9 (1.4)	<b>-13.8</b> <b>(1.3)</b>	-14.8 (1.4)
Peak AP shear rates (BW/s)	121.0 (34.5)	122.8 (39.5)	123.6 (45.4)
Peak ML shear rates (BW/s)	111.8 (51.8)	101.9 (50.0)	112.3 (54.2)
Peak compressive rates (BW/s)	500.7 (162.0)	465.7 (147.0)	500.4 (188.1)

**Significant difference from NS**

*Significant difference between RSL and CS*

Table 3.3. Mean and (sd) of AP shear, ML shear and compressive forces and rates of loading at the hip.

	NS	RSL	CS
Peak AP shear forces (BW)	-1.56 (0.4)	-1.49 (0.4)	-1.69 (0.3)
Peak ML shear forces (BW)	6.00 (1.7)	5.42 (1.4)	6.31 (1.6)
Peak compressive forces (BW)	-11.2 (1.7)	-10.9 (1.7)	-11.3 (1.9)
Peak AP shear rates (BW/s)	204.8 (110.1)	200.5 (107.0)	208.9 (114.3)
Peak ML shear rates (BW/s)	509.7 (252.9)	485.0 (239.0)	537.3 (278.1)
Peak compressive rates (BW/s)	825.8 (131.8)	853.3 (162.0)	822.5 (160.8)

**Significant difference from NS**

*Significant difference between RSL and CS*

## Figures

Figure 3.1. Vertical ground reaction forces (VGRF) as a function of stance phase. Note the lack of an impact peak, which may be a function of the ensemble averages.

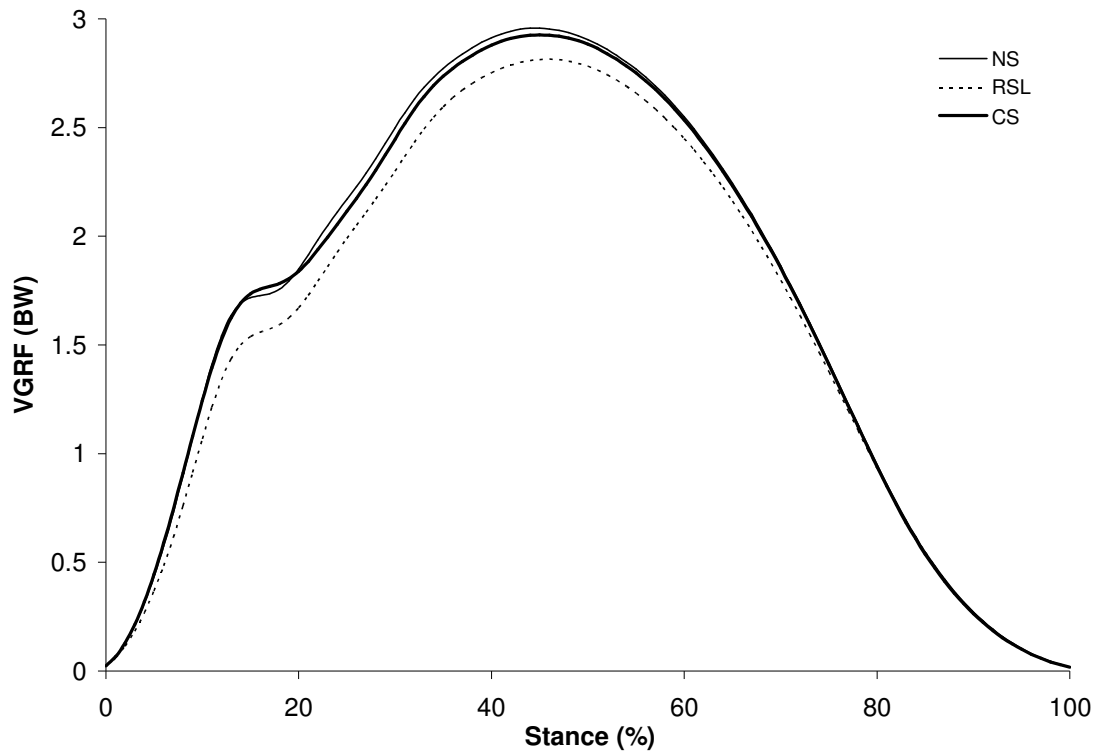


Figure 3.2. Ensemble curves of joint contact forces at the ankle joint. Curves are normalized to stance. AP shear adjusted to be parallel to the tibial plateaus. Negative forces indicate posteriorly directed AP shear forces and medially directed ML shear forces.

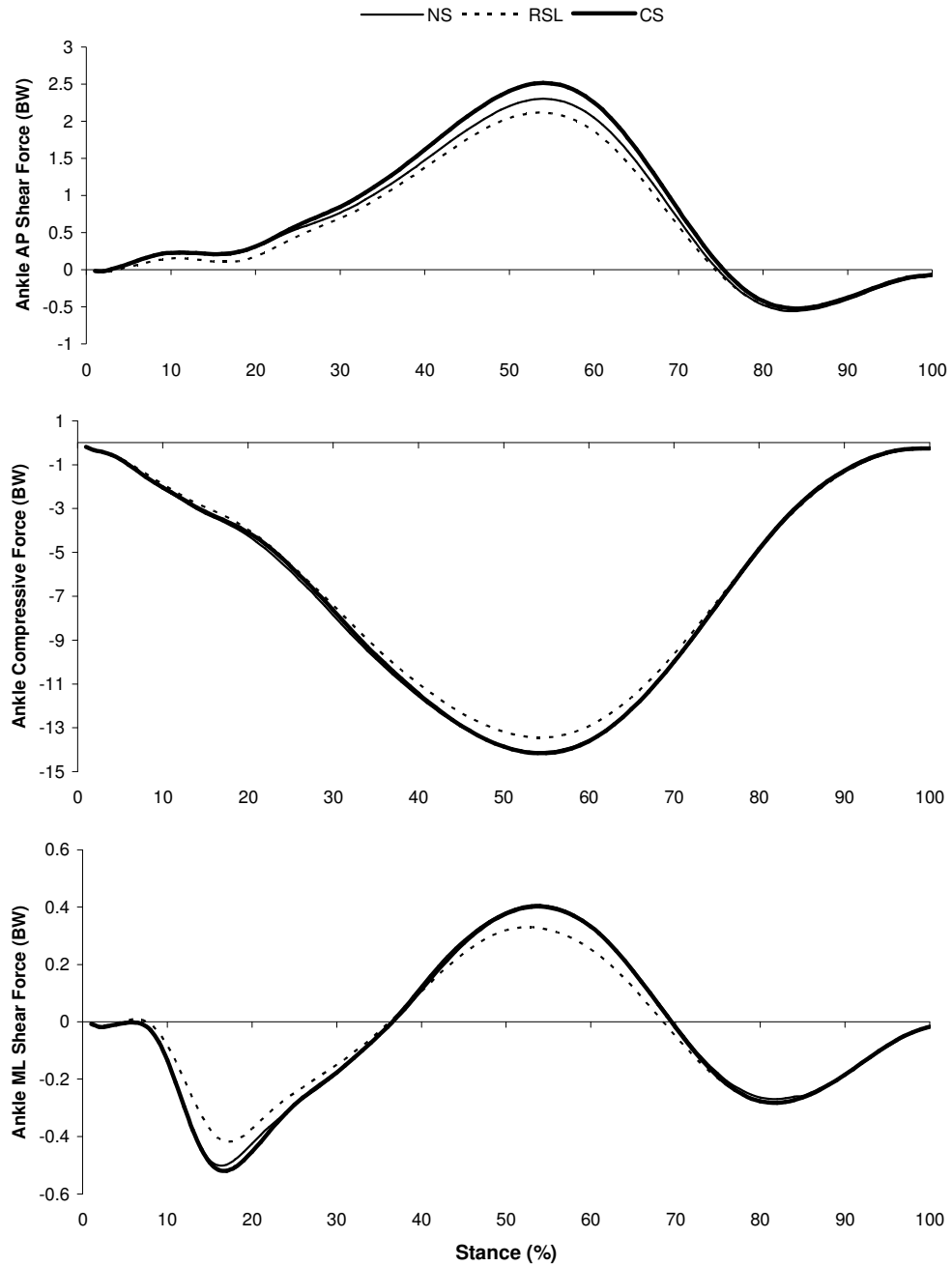


Figure 3.3. Ensemble curves of joint contact forces at the knee joint, normalized to stance. Curves are normalized to stance. AP shear adjusted to be parallel to the tibial plateaus. Negative forces indicate posteriorly directed AP shear forces and medially directed ML shear forces.

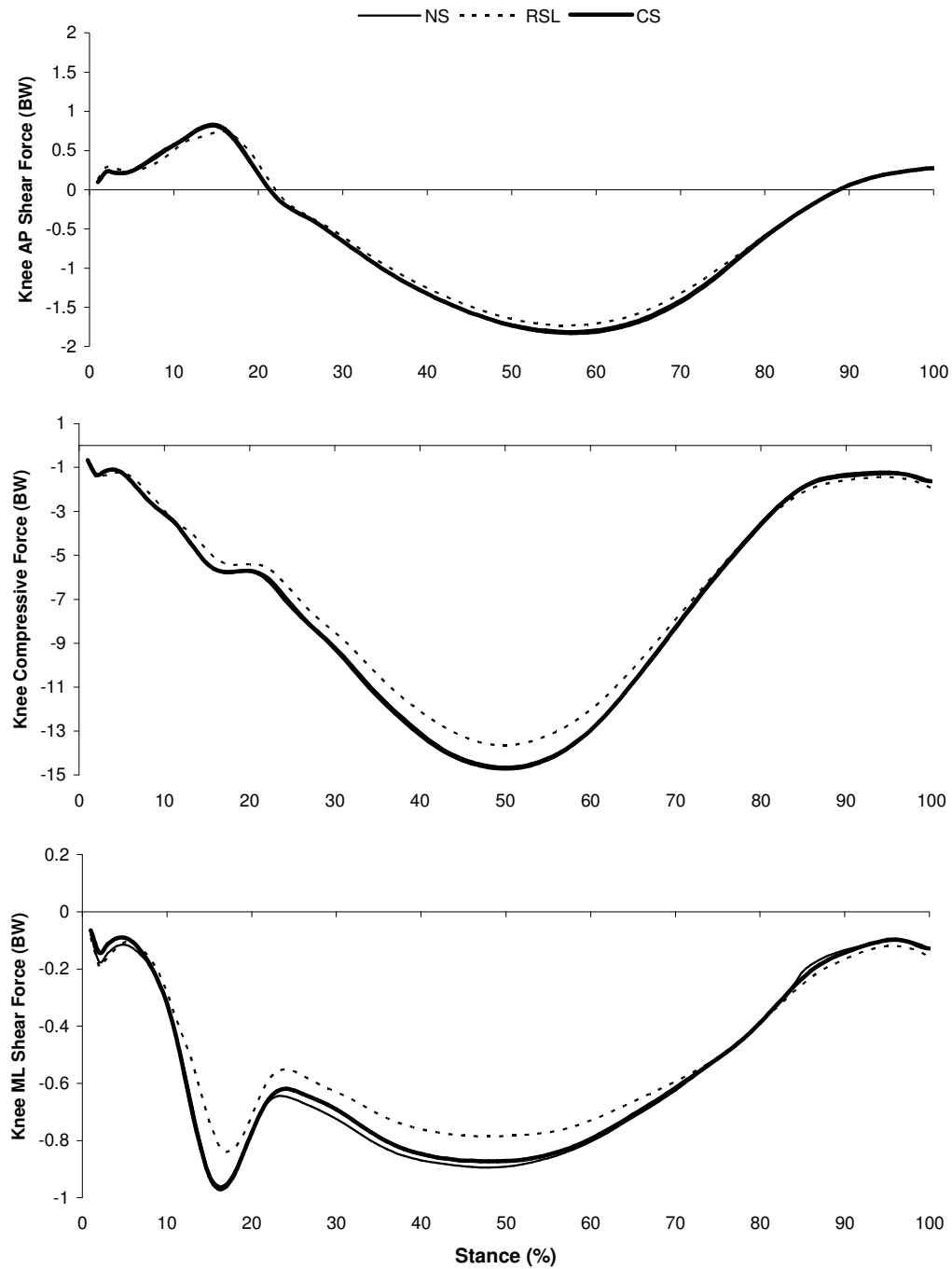


Figure 3.4. Ensemble curves of joint contact forces at the hip joint, normalized to stance. Curves are normalized to stance. AP shear adjusted to be parallel to the tibial plateaus. Negative forces indicate posteriorly directed AP shear forces and medially directed ML shear forces.

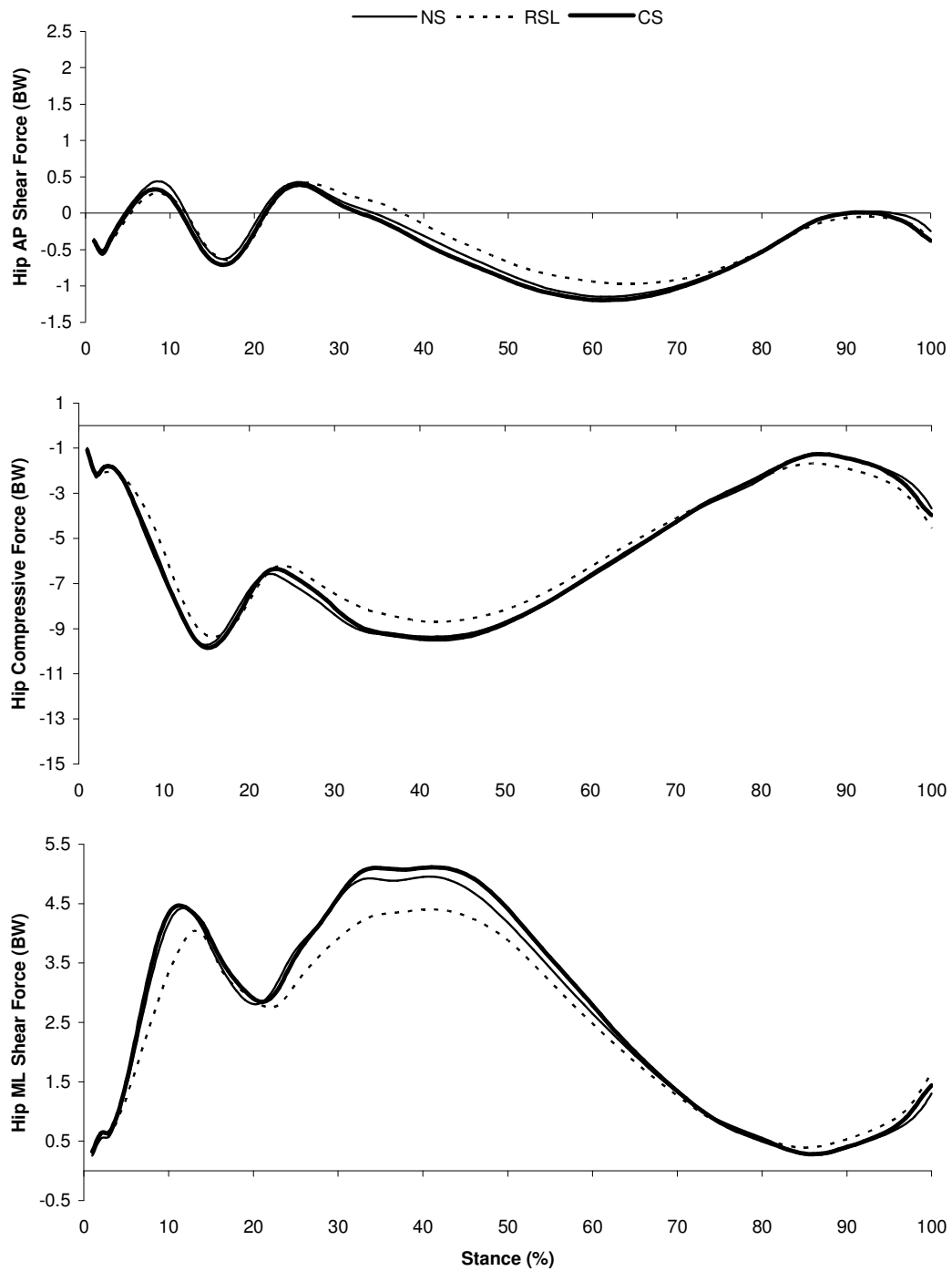




Figure 3.5. Interactions between footstrike and running conditions in ankle ML shear rate of loading.

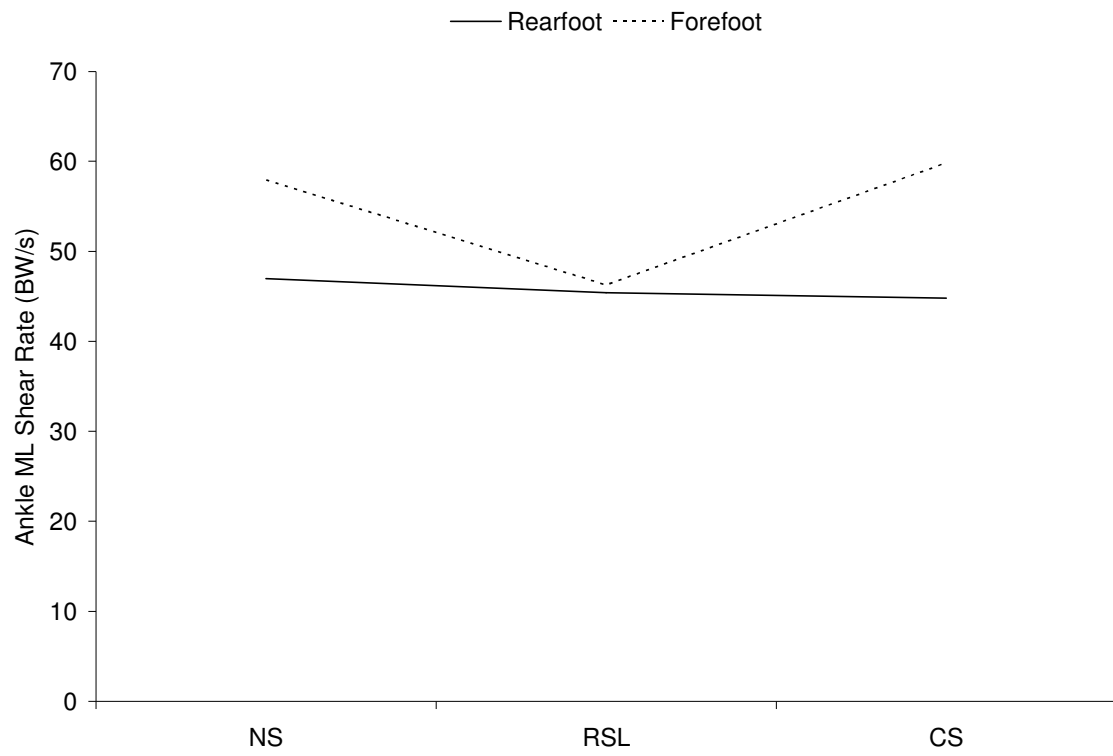
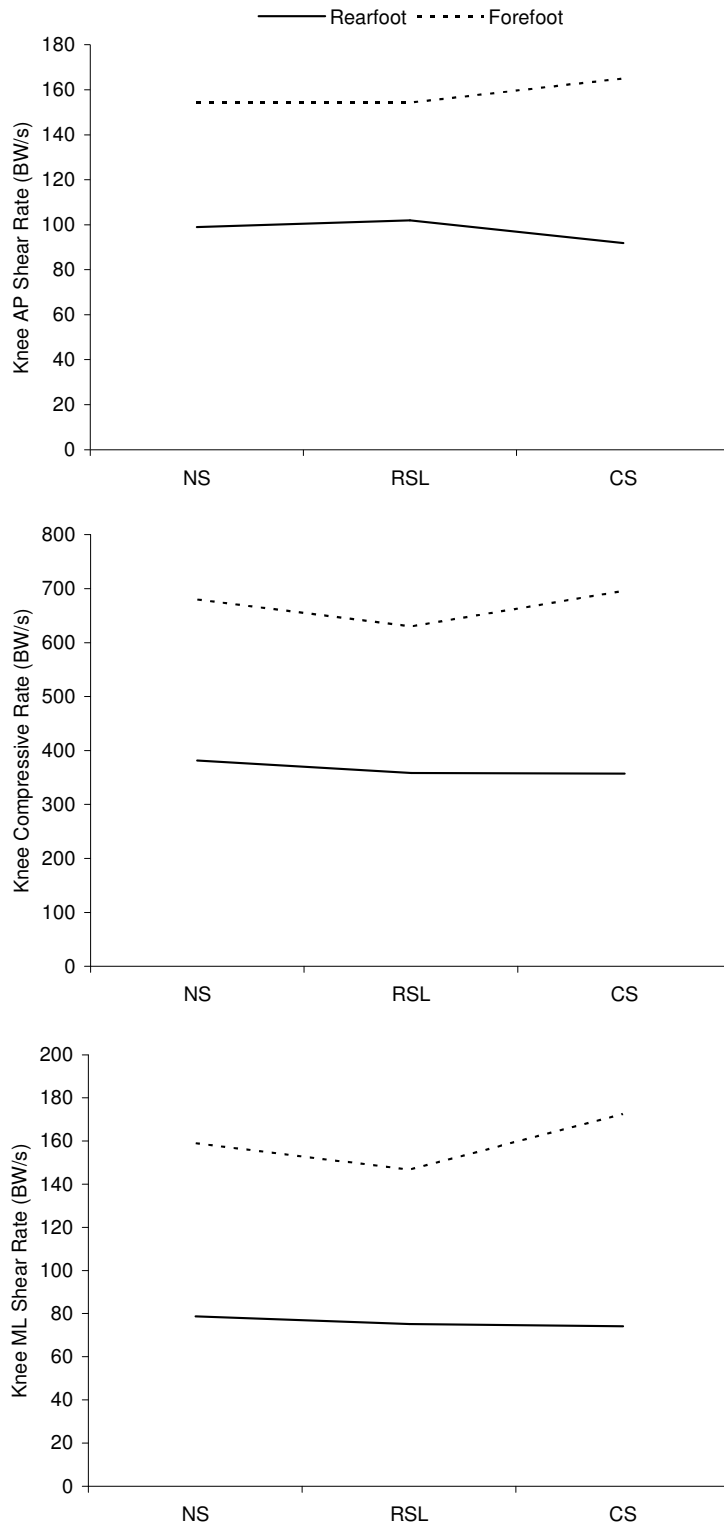


Figure 3.6. Interactions between footstrike and running conditions at the knee for compressive, AP shear, and ML shear rates of loading.



## CHAPTER 5. GENERAL CONCLUSIONS

### General Discussion

Running is a seemingly simple activity that in reality has many complex interactions when it comes to causes of injury. The popularity of running as a form of exercise, coupled with a fairly high rate of injury incidence among runners, makes understanding the complex interaction of injury risk an important topic. It is especially important that we understand the nature of impact loads on the lower extremity due to the possible links between high loads on the body and overuse injuries in the lower extremity. Injuries such as osteoarthritis, tendonitis, patellofemoral pain, iliotibial band syndrome, and stress fractures are all common running injuries that are likely related to loading or rates of loading.

The purpose of this line of research was to increase our understanding of loading on the lower extremity during running. There were three goals we hoped to achieve: The first goal was to determine how changes in knee flexion angle at contact would influence both tibial acceleration and ground reaction forces. The second goal was to determine the effect of changes in knee flexion angle at contact on joint contact forces in the knee joint. Finally, we hoped to determine the effectiveness of reducing stride length and increasing midsole cushioning as strategies for decreasing joint contact forces in the lower extremity.

The hypotheses of the first research project were that increasing the knee flexion angle at contact would decrease peak impact forces, increase peak leg acceleration and increase attenuation of the impact as it traveled from the leg to the head. These hypotheses were supported. As knee flexion angle increased at contact during running,

there was a concurrent decrease in impact forces. The runners were able to voluntarily increase and decrease the knee angle during contact and this altered the magnitude of the impact. The results also support the increased leg acceleration and increased attenuation hypotheses. In general, a more flexed knee at contact resulted in lower peak impact forces, greater peak leg acceleration and greater attenuation. The more flexed knee flexion at contact created a geometric configuration of the lower extremity that decreased the effective mass. This decreased mass was easier to accelerate and thus peak leg acceleration increased when the knee became more flexed at contact. A more flexed knee flexion at contact also attenuated the impact to a greater extent than the extended knee.

The present research supports the results of Denoth (1986) and Gerritsen et al. (1995). Denoth's research showed that increasing the knee flexion angle decreased the mass being accelerated and led to higher impact accelerations. Although not estimated directly, the effective mass likely decreased as knee flexion increased at contact. If no other changes occur, this decreased mass would lead to higher accelerations at the leg. Gerritsen et al. predicted a reduction of 68 N per degree of knee flexion via simulation, while results of the current study suggest a lower change of approximately 40 N per degree of knee flexion.

Distributing forces over an increased time interval reduces the peak force values. Both active and passive tissue can increase the time that the vertical center of mass velocity of a runner is brought to zero. Eccentric muscle contractions absorb the runner's energy but will result in increased muscle forces and increased metabolic cost. Contact with a more flexed knee shifts a higher percentage of the energy absorption responsibility from passive to active tissue. With the introduction of the raised platform, contact

velocity is increased as the body falls a greater distance. This process is similar to downhill running, supported by our peak leg accelerations which were comparable, though slightly higher, than those found by Hardin and Hamill (2002).

Although the impacts with the platform were large compared to normal running, they do not represent an activity that, by itself, is a risk to most runners. Running off a platform does not occur often enough to cause a repetitive motion injury and the magnitudes are not large enough to cause an acute injury. On the other hand, if you consider running off a platform to be similar to downhill running then increasing a single peak impact force may be enough to increase the risk of injury. Increases in the external forces acting on the body may be especially significant if the tissues being stressed are already fatigued.

The results of this experiment provide insight into the relationship between impact forces and accelerations. Our results suggest that high peak leg accelerations should be interpreted in conjunction with attenuation results. Accelerometers can provide a portable and inexpensive way to monitor impact loads in the body over extended periods of time, but it is critical that we understand the limitations. It may not be adequate to simply measure peak leg accelerations, as variations in effective mass and knee flexion at contact can alter the interpretation of peak acceleration values. The positive relationship between peak impact force and knee flexion at contact may suggest that the leg accelerometer is not a good measure of bone loading. On the other hand, the relationship is positive between leg acceleration and peak impact force when the effective mass does not change. The inclusion of a second accelerometer and the calculation of attenuation could improve the prediction of bone loading.

The results of the first research project allowed us to better understand how knee flexion angle can change accelerations and forces during impact. The results indicated that increasing knee flexion at contact decreased impact forces while increasing tibial acceleration. Knee flexion also increased attenuation of impact. This seems to imply that increasing knee flexion would be a viable strategy for decreasing the loads on the lower extremity that occur during running. However, it is possible that this could increase muscle activity and therefore metabolic cost. The relationship between increased tibial acceleration and decreased ground reaction force with increased knee flexion also lends support to the concept of effective mass. The increased knee flexion would decrease the effective mass being accelerated.

The hypothesis of the second research project is that increasing knee flexion angle at contact would increase joint contact forces and muscle forces in the lower extremity. This hypothesis was supported by our results in the quadriceps and hamstring groups. In the gastrocnemius, muscle forces decreased with increasing knee flexion. The flexed knee produced the highest muscle forces in the hamstrings and the quadriceps muscle groups. The flexed knee quadriceps force was almost 2 BW higher than the next highest condition, normal platform running. There was a delayed peak in the flexed platform running quadriceps force which was interesting. The increased knee flexion from normal running may result in a longer stance time to stop the vertical excursion of the center of mass. This longer stance time may result in the delayed peak. A lower center of mass may also require greater muscle force to raise it back to normal height, which could be produced from quadriceps activity during knee extension. The delayed quadriceps activity also likely accounts for the delay in compressive and AP shear force in flexed

knee running. Another interesting result is that extending the knee decreases quadriceps force below NRP, and to almost the same level as normal running.

Overall, normal running produced the lowest compressive forces. This was probably a result of not having a platform drop. In the platform conditions, flexed knee running produced the highest peak compressive and peak AP shear forces. Extended knee running produced the lowest peak compressive and AP shear forces of the three platform conditions. It has been suggested that increased knee flexion may result in an increased metabolic cost despite the decrease in ground reaction forces (Denoth 1986). The current research supports this theory as we see that increasing knee flexion results in increased muscular activity.

The second study showed the highest muscle forces in the quadriceps, followed by the gastrocnemius and the hamstrings. This was different than the results of Glitsch and Baumann (1997). The hamstring force may have been low in our study due to its role as a knee flexor; the use of a platform likely required a more dominant quadriceps action to arrest the downward motion of the center of mass.

Although flexing the knee more during contact decreases the ground reaction impact force, it does not result in lower joint contact forces at the knee joint because of the increased muscle forces necessary to maintain the flexed posture. Subjects likely extended their knee more during the platform running rather than normally due to the effect of the platform. The extra vertical height gave subjects more time to extend the knee as they reached for the force platform.

The results of the second study supported the theory that increasing knee flexion results in an increase in muscular activity. During the flexed knee conditions, there was

an increase in knee compressive forces, with the quadriceps having the highest muscle force. Increased knee flexion also increased compressive forces and AP shear forces during stance compared to the extended knee.

Our hypothesis of the final study was that reducing stride length will result in a decrease in muscle and joint contact forces. We also hypothesized that increasing midsole cushioning would have no effect on joint contact or muscle forces. Both hypotheses were supported by our results. The results showed that reducing stride length decreased ML shear forces in the ankle and decreased AP shear forces at the knee. The ankle also saw a decrease in rate of loading of the AP shear component. Finally, reducing stride length resulted in decreased compressive forces at the ankle and at the knee. These results were comparable to past research (Burdett 1982; Scott and Winter, 1990; Sasimontongkul et al., 2007). Our hypothesis about the effect of cushioning on joint contact forces was not supported. Although most forces did not change as a result of decreased midsole cushioning, it did result in a decrease in ML shear force at the ankle.

The research results from the third study also indicated that reducing stride length is a better strategy for reducing joint contact lower extremity forces than a cushioned midsole. Decreased stride length produced lower AP shear forces at the hip, knee, and the ankle than increased midsole cushioning. It also produced lower ML shear forces at the ankle and the hip. Compressive forces at the ankle and knee were significantly lower in the shorter stride condition than the cushion condition. There was also a decrease in the AP shear rate of loading at the ankle.

The patterns of the joint contact forces indicated that in most cases peak or near peak forces occurred near 40 – 60% of stance. This may indicate that injury potential



during running may not simply be due to impact forces and impact accelerations at contact. The high joint contact forces at midstance also will play a role, and may in fact have more of a contribution to injury.

During the data collection it was noted that some of the runners appeared to be more forefoot/midfoot strikers than rearfoot strikers. However, it does appear that knowledge of the footstrike index may be required to determine changes in normal or cushion. In ML shear rate of loading at the knee, being a rearfoot striker makes the cushion condition comparable to short stride, but being a forefoot striker has no effect. Although the interaction during AP shear rate of loading at the knee was not significant, during the cushion condition the forefoot strikers produced the highest rates of loading in the three conditions. The rearfoot strikers produced the lowest rates of loading in the cushion condition. It does not appear that being a forefoot striker had any effects on other joint contact forces or rates of loading as a function of stride length or cushioning.

The final study revealed that reducing stride length resulted in a decrease in compressive forces at the ankle and knee joints. The only effect of increasing midsole cushioning compared to normal running was to decrease ML shear force at the ankle. There was no effect of increasing midsole cushioning on compressive or AP shear forces. The results also seemed to indicate that reducing stride length is a better strategy than increased midsole cushioning for reducing joint contact forces. The decreased stride length resulted in significantly lower joint contact forces than the cushion condition in multiple variables.

## **Recommendations for Future Research**

In a practical sense, this new information may not be directly helpful to runners. Our research has indicated that increasing knee flexion during running will reduce ground reaction forces but may also increase joint contact compressive and AP shear forces. The overall impact of this has not been determined. The weight-bearing nature of running means that it is inevitable that the lower extremity will incur some form of loading. It remains to be determined if one of these variables has a greater impact on injury potential or human performance. Runners may have to determine whether they would prefer a decrease in impact forces or a decrease in metabolic cost. Runners who have gait patterns that are very stiff or straight-legged may want to increase their knee flexion in order to reduce the impact forces on their legs. Runners who have a tendency to overstride may also want to reduce stride length to decrease joint contact forces. The knowledge of midsole cushioning may prevent runners from buying running shoes that shoe companies claim have high levels of cushioning, but which are often accompanied by a high price tag.

Potential future research includes further study of the differences between forefoot strikers and rearfoot strikers in joint contact forces. It is also possible that new information on joint or bone geometry, or muscular activity at the lower leg could lead us to improve our three dimensional model used for estimating joint contact forces. Another area of interest would be to determine how barefoot running might affect joint contact forces as opposed to running in shoes.

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Perseverance and spirit have done wonders in all ages. ~George Washington

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