

The Effects of Speed and Surface Compliance on Shock Attenuation Characteristics for  
Male and Female Runners

Janet S. Dufek<sup>1</sup>, John A. Mercer<sup>1</sup> & Janet R. Griffin<sup>2</sup>

<sup>1</sup>Sports Injury Research Center; Biomechanics Laboratory

University of Nevada, Las Vegas

Las Vegas, NV 89154-3034

<sup>2</sup>adidas Innovation and Research Test Laboratory

Portland, OR

Address for correspondence:

Janet S. Dufek

Department of Kinesiology and Nutrition Sciences

University of Nevada, Las Vegas

Las Vegas, NV 89154-3034

[jdufek@unlv.nevada.edu](mailto:jdufek@unlv.nevada.edu)

Word Count: 3845

Date Submitted: August 28, 2007

Revision Submitted: April 5, 2008

Second Revision Submitted: July 19, 2008

## Abstract

The purpose of the study was to examine the effects of running speed and surface compliance on shock attenuation (SA) characteristics for male and female runners. We were also interested in identifying possible kinematic explanations, specifically, kinematics of the lower extremity at foot-ground contact, for anticipated gender differences in SA. Fourteen volunteer recreational runners (7 male, 7 female) ran at preferred and slow speeds on an adjustable bed treadmill, which simulated soft, medium and hard surface conditions. Selected kinematic descriptors of lower extremity kinematics as well as leg and head peak impact acceleration values were obtained for 10 left leg contacts per subject-condition. Results identified significant SA values between genders across conditions and more specifically, across surfaces for females, with male runners demonstrating a similar trend. Regression modeling to predict SA by gender for surface conditions elicited unremarkable results, ranging from 30.9-59.9% explained variance. It appears that surface compliance does affect SA during running; however, the runner's ability to dissipate the shock wave may not be expressly explained by our definition of lower extremity kinematics at contact.

Keywords: Gender, Impact, Injury, Lower Extremity Function

## Introduction

Running is a common movement activity and overuse injury is not an uncommon result. James et al. (1978) determined that the most frequent cause of injury during running is a training error: too much, too fast or too soon. Gaining insight into the etiology of overuse running injury is challenging since this is a multifaceted problem (Hrlejac, 2004). For example, although magnitude of impact is likely related to causing overuse injury, the direction and rate of load application are likely also critical in the ramifications of this load application to the body.

An understanding of tolerance of the body to the shock wave generated between the foot and running surface at heel strike may be an important aspect relative to understanding running injury. Specifically, the shock wave generated at ground impact travels through the body and is mitigated by soft tissue. Ideally, this shock wave will be accommodated by the movement of the joints and soft tissue between the heel and the head. A measurement of these phenomena, termed “shock attenuation” (SA) has been previously presented in the literature (Derrick, 2004, Derrick & Mercer, 2004; Mercer et al., 2002).

Derrick (2004) surmised that knee joint angle at contact can influence the relationship between the leg and head acceleration values (SA) while Mercer et al. (2002) reported increased SA values for increases in running speed, stride length and stride rate. In a subsequent study, Mercer et al. (2005) sought to determine the effects of stride length on running velocity as well as SA characteristics. The researchers concluded that

lower extremity geometry may be a critical factor relative to characteristics of running impact since SA was influenced primarily by changes in stride length.

One practical method utilized by runners to attenuate this shock wave is via surface modification, e.g., concrete, asphalt, grass or even the shoe worn. In previous studies, it has been shown that shoes with lesser cushioning can result in greater knee joint flexion velocity (Frederick et al., 1983). Likewise, Hardin et al. (2004) reported knee and hip joint kinematic adaptations to surface modifications included increases in maximum angular velocity of the ankle, knee and hip with increases in surface stiffness. Extrapolating these results to the Mercer et al. (2005) conclusion that lower extremity geometry may be a factor influencing SA, it is reasonable to conclude that running surface may influence the mechanism of SA.

The majority of running literature has focused on the performance of male runners, which is contrary to a current focused emphasis on gender performance differences, due to the documented disparity in lower extremity non-contact anterior cruciate ligament injury rates for females (Bieze, 2004; Cowling & Steele, 2001; Ford et al., 2003; Ireland, 1999). Ferber et al. (2003) explored gender differences in running, focusing on lower extremity joint kinetics. They identified gender differences, primarily in hip and knee joint frontal and transverse plane energetics. These results suggest that there is a difference in the way in which males and females run. We questioned whether a concomitant difference exists in the dissipation of the shock wave developed at impact between males and females. Much of the contemporary gender injury research focuses on landing and cutting maneuvers (Hass et al., 2005; Hewitt, 2000), while running seems to have been overlooked.

The purpose of the study was to examine the effects of running speed and surface compliance on SA characteristics for male and female runners. Given the documented gender differences for impact activities including landing (Chappell et al., 2007; Nagano et al., 2007; Wikstrom et al., 2006) as well as frontal plane cutting movements (McLean et al., 2004; Wojtys et al., 2003), it is not unreasonable to anticipate that there would be a mechanistic difference between genders in the way the body accommodates the force of impact during running. Speed and surface perturbations could lend insight into factors that potentially differentiate between genders. We chose the primary dependent variable of SA as a discriminating performance variable, since it can theoretically represent the way the body accommodates the shock wave generated during running, which may be related to injury prevention. We hypothesized that there would be a significant gender difference in SA across running speeds and surface compliances. Our further interest lied in identifying possible kinematic explanations, specifically, geometry and/or velocity of the lower extremity, for anticipated gender differences.

## Methods

### *Subjects*

Seven female and seven male recreational runners who ran a minimum of 10 miles/week (Table 1) were recruited for this study. The subjects were all rearfoot strikers and had no obvious lower extremity misalignments as visually assessed or reported injuries. Each subject fit comfortably into provided standard laboratory shoes and no orthotic inserts were used by any subjects. All subjects were informed of the experimental procedures and signed a written informed consent form approved by the Institutional Review Board at the affiliated university prior to participation.

<Table 1 about here>

### *Instrumentation*

Subjects were instrumented with lower body reflective markers and tracked with a 12-camera (Vicon Instrument Corp., Oxford Electronics, Oxford, UK) motion capture system (120 Hz). Two lightweight uni-axial piezoelectric accelerometers (PCB Piezotronics Inc., model 352C68, 1082Hz (11 subjects); 1004Hz (3 subjects)) were attached firmly to the distal aspect of the left tibia using a firm elastic band and tape and to the frontal aspect of the forehead, respectively to measure segmental impact acceleration. The head accelerometer was firmly mounted to a plastic headpiece, which was then tightly secured to the participant's head using a ratcheting mechanism. The accelerometers were interfaced through a Type 9865B 8 channel amplifier to a data acquisition system using Bioware (Kistler Instrument Corp., Version 3.21) software. A Precor M9.3s treadmill with adjustable bed stiffness was used to alter the surface stiffness over six running conditions.

### *Procedures*

Prior to data collection, treadmill stiffness was characterized by static loading and measuring with a meter stick the deflection of the front edge of the treadmill. At each treadmill stiffness setting (marked I, II, and III) the treadmill was systematically loaded with weights and the deflection was measured. The weights were placed in the center of the treadmill slightly toward the front. This position was selected to estimate the location of a runner on the treadmill during the stance phase. To ensure the treadmill was at a zero location before each level was tested, the front of the bed of the treadmill was lifted and allowed to equilibrate to the starting location between load sessions. To ensure the

treadmill was adequately compressed, the final weight added was that of one of the experimenters before tracking the upward deflection as each weight was removed. A single trial was completed for each level and the levels were termed “soft”, “medium”, and “hard” based on this load-deformation testing. The result of this test, characterizing the three stiffness settings, is illustrated in Figure 1. The intent of this testing was not to provide a discrete stiffness value, but rather to determine the relationship among the three treadmill stiffness settings.

<Figure 1 about here>

The motion capture system was calibrated per manufacturer’s instructions. Sixteen 25 mm reflective markers were attached the lower extremity of each subject in accordance with the Vicon Plug-in-Gait model. Markers were placed bilaterally on the: anterior superior iliac spine, posterior superior iliac spine, lateral epicondyle of the knee, thigh (aligned with the greater trochanter and lateral epicondyle), lateral malleoli, lateral tibia (aligned with the lateral epicondyle and lateral malleolus markers), head of second metatarsal, and heel. These markers tracked the motion of the pelvis, thigh, leg, and foot of both lower extremities, however only the left lower extremity was of concern in the present analysis (given the constraint of single-limb accelerometry). Subjects were allowed to warm up at the medium level of treadmill stiffness at their preferred 20 minute run pace for four minutes (average preferred pace =  $2.66 \pm 0.36$  m/s). This established the preferred condition speed for each participant. The slow speed was set at 10% less than this preferred running speed. Data capture was initiated by manual keystroke, which sent a synchronizing square wave to both systems, and was obtained for 17 consecutive seconds during each running condition. The six conditions were counterbalanced such

that medium stiffness, preferred speed and medium stiffness, slow speed were the first for each subject at the two different speed settings. The soft stiffness settings for preferred and slow speed, respectively and hard stiffness were counter balanced for each subject (Table 2).

<Table 2 about here>

### *Data reduction*

Following visual observation of position-time histories and in accordance with previous laboratory procedures for locomotion activities, the kinematic marker data were low-pass filtered using a quintic spline with a mean standard error of 15 (Woltring, 1985). Foot contact was determined by examination of the vertical velocity and acceleration values of the ankle joint as well as the vertical position of the heel marker. Using the accelerometer-time histories, peak leg (LgPk) and peak head (HdPk) acceleration values during the impact phase were identified on a trial-by-trial basis. Ten consecutive left stance phase trials per subject-condition were evaluated resulting in 60 trials per subject across conditions. These data were used to compute shock attenuation (SA), using the following formula:

$$SA = [1 - (a_{\text{head}} / a_{\text{leg}})] \cdot 100, \text{ where}$$

$a_{\text{head}}$  = peak head impact acceleration and;

$a_{\text{leg}}$  = peak leg impact acceleration during the support phase of running.

### *Statistical analysis*

All statistical tests were conducted using Statistical Analysis Software (SAS, version 8.2; Cary, NC), with  $\alpha = 0.05$  adopted as the level of significance for all tests. A two-way (gender x condition) mixed model analysis of variance (ANOVA) was first



conducted for the dependent variable of SA. This was followed by two-factor (speed x surface) repeated measures ANOVAs, by gender, to examine anticipated gender differences across levels of running perturbations for SA, with *post hoc* comparisons when appropriate. Finally, stepwise multiple regression models ( $\alpha = 0.15$  for variable entry; overall model significance  $\alpha = 0.05$ ) were developed to predict SA and LgPk, by gender, from a set of selected lower extremity sagittal and frontal plane kinematic measures at contact (Appendix A). The specific set of independent variables identified for use in the regression analysis were selected following computation of a cross-correlation matrix of 45 originally identified kinematic variables and was guided by our interest in contact kinematics. The regression procedure was conducted specifically to address the potential relationship between lower extremity kinematics at contact and SA. In order to ascertain greater understanding of the SA measurement relative to lower extremity kinematics, we utilized the same stepwise multiple regression procedures to predict LgPk from the set of 13 contact kinematic variables.

## Results

Mean and standard deviation values for SA, by gender-condition, are given in Table 3. Results of the two-way repeated measures ANOVA (gender x condition) identified significant main effects for gender ( $F_{1,5} = 6.60$ ;  $p = 0.0246$ ) and condition ( $F_{1,5} = 2.70$ ;  $p = 0.0290$ ), with no significant interaction. This result confirmed our hypothesis that there would be a difference in SA between genders. Mean and standard deviation values for SA by gender collapsed across speed and surface are given in Table 4. The follow-up analyses of these data (speed x surface repeated measures ANOVAs by gender) revealed a significant surface effect for female runners ( $F_{1,6} = 11.43$ ;  $p = 0.0017$ )

only with no other significant main effects or interactions for either gender. *Post hoc* (least squared means) procedures identified significant ( $p < 0.05$ ) differences between soft and medium and medium and hard surface conditions for females. Given this result, data were collapsed across running speed for subsequent regression modeling. Due to the fact that there was a significant surface effect for females and given the trend of the male data followed that of the females, we did not collapse across surface for subsequent regression modeling.

<Tables 3 and 4 about here>

Regression results for prediction of SA from contact kinematic variables are presented in Table 5. Six independent SA prediction models were computed, by surface (3) and gender (2). Overall, results were unremarkable, resulting in an average percent explained variance (EV) across all models of only 40.3%, suggesting little relationship between contact kinematics and SA. More careful examination of these results indicated that, across conditions, EV was greater for males (44.8%) vs females (35.9%). In addition, the medium condition was least predicted across gender (33.4% EV) while the soft condition was best predicted (50.0% EV) with the hard condition predicted at 37.5% EV.

<Table 5 about here>

As a follow-up to this examination, and based upon previous research suggesting that leg impact acceleration magnitude may be an important phenomena relative to biological adaptation and shock wave transmission (Dufek et al., 2008; Mercer et al., 2005) we replicated the regression analysis in predicting LgPk from lower extremity geometry and/or velocity (kinematic variables at contact) by gender and surface. Results

of the LgPk predictions were unique from the SA predictions. First, overall EV was greater, averaging 48.0% EV across conditions and genders, in contrast to 40.3% EV for SA. Prediction models for LgPk were stronger for females across surfaces (54.3% EV) vs males (41.7% EV). Also, the medium surface condition was best predicted across gender for LgPk vs the soft condition for SA. Regression results for prediction of LgPk from contact kinematic variables are presented graphically in Figure 2.

<Figure 2 about here>

## Discussion

Results of this study support our hypothesis that differences exist between genders in the mechanisms employed to manage the impact generated at ground contact during running (Table 3). These differences appear more divergent when running surface is manipulated rather than running at a slower than preferred speed (Table 4). We were unable to find strong support for the previously suggested hypothesis that lower extremity kinematics at contact influences SA (Table 5).

In general, kinematics of the lower extremities at foot contact between genders was quite similar and followed the same direction of change with surface compliance with one notable exception. The frontal plane position of the knee joint at contact was different between genders, with females exhibiting 1.8 degrees of valgus at contact across conditions versus a value for males of 2.4 degrees of varus. An inspection of individual responses has lead to the observation that this parameter was quite variable between genders. For example, the range of female values was 14.3 degrees valgus to 4.7 degrees varus while the range for males was 1.3 degrees valgus to 8.7 degrees varus. While these

data show that some female runners did contact the ground with the knee joint in a varus position (n=2), most contacted in a valgus orientation, which is counter to the results observed for males. Relating back to injury propensity for females versus males (Bieze, 2004; Cowling & Steele, 2001; Ford et al., 2003; Ireland, 1999), data from the current study suggest that the notion of cutting and non-sagittal plane motion may not be the sole cause of the predominance of injuries to females. One of the causes may be more general in nature in that it may be related to frontal plane knee joint position during contact in running (and not only cutting and non-sagittal plane motion) and associated gender differences observed at the knee joint. Continuous valgus stress for female runners may predispose them to acute ACL injuries in other non-planar activities due to the chronic stress exposure experienced as a result of their frontal plane contact kinematics during running.

Specific composition of SA prediction models by gender-surface (Table 5) provides additional insight into potential gender differences. Accepting the limitations of small sample size, dominance of sagittal plane predictor variables and that the models are not exceptionally strong in prediction of SA from lower extremity kinematics at contact, primary contributing factors can be gleaned. In general, kinematics of the hip joint (position as well as linear and horizontal velocity) was more strongly related to SA characteristics for females. This is in direct contrast (with the exception of the soft condition) for males, who exhibited knee and ankle joint kinematics at contact as the only factors related to SA. Predictive dominance of knee and ankle joint kinematics for males may have led to the previously stated suggestion (Mercer et al., 2005) that SA may be strongly influenced by lower extremity geometry at contact since these data were

obtained from males only. Therefore, stride length (Mercer, et al., 2003) may be more closely related to SA for males than females. The hip joint dominance exhibited by females suggests that trunk inclination may be critical for females, extrapolating that the vertebral column may play a vital role in SA for this group.

Prediction of LgPk from contact kinematics was greater for females vs. males (Figure 2). As well, prediction models were generally stronger for LgPk vs SA. This result may suggest that lower extremity kinematics at contact during running may be more directly related to leg impact acceleration, and not dissipation of the impact shock wave which is characterized by SA. The mechanisms responsible for attenuating the impact of foot-ground contact may be totally independent of the orientation of the foot-leg at contact at the speeds tested in this experiment.

Interpretation of the SA parameter, a ratio of LgPk and HdPk is dependent on the response of both segment acceleration values (*i.e.*, tibia and head). As such, one must understand the directional change, or lack of change, of each variable. The results of the current study were similar to previously reported results of SA during running (Mercer, et al., 2002, 2003, 2005) which reported little change in HdPk with a greater change in LgPk across various experimental manipulations (Figure 3). The consistency of the HdPk value across surface conditions between genders allows one to interpret the observed increase in SA as being directly related to increases in LgPk between genders.

Furthermore, Figure 3 illustrates the gender difference in magnitude of LgPk for females across conditions as being greater than that of males for comparative surface conditions. The interpretation of greater SA as being related specifically to the observed increases in LgPk would be limited if an interaction response between LgPk and HdPk existed among

experimental conditions. However, such a response has yet to be reported in the literature.

<Figure 3 about here>

Of note and in comparison to previous literature, runners in this study ran much more slowly averaging 2.66 m/s (Table 1), versus comparable experiments with male runners with velocity values of 3.50 m/s or greater (Derrick, 2004; Mercer et al., 2002, 2005). The slower performance speeds in the current study resulted in lesser SA values (77.4% average across all subject-conditions, Table 3), versus previously reported values of 90% and greater (Mercer et al., 2002). The less-demanding, preferred pace and slower run environments may not have stressed the biological system to the same degree as in previously reported experiments. It is not known if perhaps the body adopts a different strategy of attenuating greater leg impacts produced by faster running speeds, and if this possibility has influenced the current results. The current study focused on the influence of surface compliance on shock attenuation. Running speed was introduced as a second factor of interest to explore whether performance, in the form of impact acceleration, was different between genders across speed conditions. We did not observe any interaction effects between gender and speed on SA (Table 4). Examination of male and female runners at a greater variety of performance speeds would lend insight into the possibility of unique mechanisms for SA between genders at different levels of physiological output. A suggested limitation of uni-axial accelerometry as an explanation for the lack of a significant speed effect was dismissed owing to the findings of Derrick, et al. (2004) who reported that change in the leg orientation during stride length manipulations of up to

20% of preferred stride length had little influence on the orientation of the accelerometer at impact.

Challis and Pain (2008) have suggested that soft tissue motion (wobbling mass) may influence loads on the rigid body system during impact activities. One may speculate that the female subjects in the current study have a greater percentage of soft tissue (*i.e.*, adipose tissue) and therefore comparative results of SA between genders may be influenced by this factor. The current investigation was limited in that we did not identify potential wobbling mass effects, nor did we quantify percent body fat of the participants.

This study incorporated a traditional group statistical design in order to examine potential gender (*i.e.*, group) differences. However, viewing the data on an individual subject level suggests that there may exist unique SA mechanisms that are not necessarily gender-related, but may be more so based upon individual runner performance. To explore this hypothesis, we repeated the regression analysis on a per-subject basis and predicted SA for each individual data set. We observed that explained variance for females was between 0-68.6% (average = 26.8%) whereas the explained variance was between 0-89.2% for male subjects (average = 30.6%). Based upon this analysis, it seems that the ability to predict SA from the kinematic data sets defined in the current study is not related to gender. Rather, there may be some performance factors such as running experience or variability of performance within-runner that may more specifically define unique mechanisms of SA experienced by runners.

We sought to explore the potential relationship between lower extremity geometry and/or velocity at running contact as defined kinematically and the ability to attenuate

impact (SA) between genders. Results of this investigation identified significant differences between genders in SA characteristics between preferred and slow speeds among soft, medium and hard running surfaces (Table 3) which supported our hypothesis. Contrary to the literature reported for male runners (Mercer et al., 2002, 2005) there were no differences in SA between running speeds for males or females (Table 4) perhaps owing to the slower average running speeds elicited by subjects in the current study. A significant difference in SA between surfaces was identified for females, with male runners exhibiting a similar, although non-significant trend (Table 4). Our attempt to relate the mechanism of SA to lower extremity kinematics at contact using sagittal and frontal plane linear and angular kinematic measures did not elicit a strong outcome, but did suggest that trunk inclination as reflected by hip joint kinematics may have a greater effect on SA for females versus males. Mercer et al. (2003) suggested that stride length may be a primary factor influencing SA in their previous work using male runners. In the current study, knee and ankle joint parameters were stronger predictors of SA for males, which provides some support for the stride length-SA relationship previously suggested (for male runners). In order to more fully understand the possible relationship between SA and gender, additional research is warranted. A broader range of running speeds, faster speeds, various experiential levels of runners, addition of a low back accelerometer to more fully understand the transfer of the shock wave generated at impact and continued investigation into variability of running performance are all suggested as research topics which can contribute to this area of inquiry relative to running performance and injury prevention.



## Acknowledgments

The authors wish to express thanks to Angela Bestwick, David DeLion, Kaori Teramoto and Amanda Tritsch for their assistance during data collection.

## References

Bieze, J. (2004). ACL injury risk studies focus on knee flexion, valgus motion.

*Biomechanics*, 11(9), 9-10.

Challis, J.H. & Pain, M.T.G. (2008). Soft tissue motion influences skeletal loads during impacts. *Exercise and Sport Sciences Reviews*, 36(2), 71-75.

Chappell, J.D., Creighton, R.A., Giuliani, C., Yu, B. & Garrett, W.E. (2007). Kinematics and electromyography of landing preparation in vertical stop-jump. *American Journal of Sports Medicine*, 35(2), 235-241.

Cowling, E.J. & Steele, J.R. (2001). Is lower limb muscle synchrony during landing affected by gender? Implications for variations in ACL injury rates. *Journal of Electromyography and Kinesiology*, 11, 263-268.

Derrick, T.R. (2004). The effects of knee contact angle on impact forces and accelerations. *Medicine & Science in Sports & Exercise*, 36(5), 832-837.

Derrick, T.R., & Mercer, J.A. (2004). Ground/foot impacts: Measurement, attenuation, and consequences. *Medicine & Science in Sports & Exercise*, 36(5), 830-831.

Dufek, J.S., Mercer, J.A., Teramoto, K. Mangus, B.C. & Freedman, J.A. (2008). Impact attenuation and variability during running in females: A lifespan investigation. *Journal of Sport and Rehabilitation*, 17(3).

Ferber, R., McClay Davis, I. & Williams, D.S. III. (2003). Gender differences in lower extremity mechanics during running. *Clinical Biomechanics*, 18, 350-357.

Ford, K.R., Myer, G.D. & Hewitt, T.E. (2003). Valgus knee motion during landing in high school female and male basketball players. *Medicine & Science in Sports & Exercise*, 35(10), 1745-1750.

Frederick, E.C., Clarke, T.E., Larsen, J.L. & Cooper, L.B. (1983). The effects of shoe cushioning on the oxygen demands of running. In Nigg, B.M. and Kerr, B.A. (eds.), *Biomechanical Aspects of Sport Shoes and Playing Surfaces* (pp. 107-114). Calgary: University Printing Services.

Hardin, E.C., Van Den Bogert, A. & Hamill, J. (2004) Kinematic adaptations during running: Effects of footwear, surface, and duration. *Medicine and Science in Sports and Exercise*, 36(5), 838-844.

Hass, C.J., Schick, E.A., Tillman, M.D., Chow, J.W., Brunt, D. & Caruagh, J.H. (2005). Knee biomechanics during landings: Comparison of pre- and postpubescent females. *Medicine and Science in Sports and Exercise*, 37(1), 100-107.

Hewitt, T. Neuromuscular and hormonal factors associated with knee injuries in female athletes: Strategies for intervention. (2000). *Sports Medicine*, 29:313-327.

Hreljac, A. (2004). Impact and overuse injuries in runners. *Medicine and Science in Sports and Exercise*, 36, 845-849.

Ireland, M.L. (1999). Anterior cruciate ligament injury in female athletes: epidemiology. *Journal of Athletic Training*, 34, 150-154.

James, S., Bates, B.T., & Osternig, L.R. (1978). Injuries to runners. *The American Orthopaedic Society for Sports Medicine*, 6(2), 40-50.

McLean, S.G., Lipfert, S.W. & Van Den Bogert, A. (2004). Effect of gender and defensive opponent on the biomechanics of sidestep cutting. *Medicine and Science in Sports and Exercise*, 36(6), 1008-1016.

Mercer, J.A., Vance, J., Hreljac, A., & Hamill, J. (2002). Relationship between shock attenuation and stride length during running at different velocities. *European Journal of Applied Physiology*, 87, 403-408.

Mercer, J.A., Bates, B.T., Dufek, J.S. & Hreljac, A. (2003). Characteristics of shock attenuation during fatigued running. *Journal of Sport Sciences*, 21(11), 911-919.

Mercer, J. A., Bezodis, N. E., Russell, M., Purdy A., DeLion, D. (2005). Kinetic consequences of constraining running behavior. *Journal of Sports Science and Medicine*, 4, 144-152.

Nagano, Y., Ida, H., Akai, M. & Fukubayashi, T. (2007). Gender differences in knee kinematics and muscle activity during single limb drop landing. *The Knee*, 14, 218-223.

Wikstrom, E.A., Tillman, M.D., Kline, K.J. & Borsa, P.A. (2006). Gender and limb differences in dynamic postural stability during landing. *Clinical Journal of Sports Medicine*, 16(4), 311-315.

Wojtys, E.M., Huston, L.J., Schock, H.J. & Boylan, J.P. (2003). Gender differences in muscular protection of the knee in torsion in size-matched athletes. *Journal of Bone and Joint Surgery*, 85-A(5), 782-789.

Woltring, H.J. (1985). On optimal smoothing and derivative estimation from noisy displacement data in biomechanics. *Human Movement Science*, 4, 229-245.

Table 1. Subject Descriptive Data (mean followed by standard deviation values).

---

	<b>Females (n = 7)</b>	<b>Males (n = 7)</b>
Age (yrs)	24.3 (3.4)	25.4 (4.6)
Height (cm)	169.6 (5.2)	175.4 (7.2)
Mass (kg)	67.0 (3.6)	80.6 (8.0)
Preferred Run Velocity (m/s)	2.72 (0.35)	2.87 (0.41)
Slow Run Velocity (m/s)	2.45 (0.30)	2.59 (0.37)

---

Table 2. Illustration of condition counterbalance procedure.

---

<b>Subject</b>	<b>Preferred Speed</b>			<b>Slow Speed</b>		
S1	Medium	Soft	Hard	Medium	Soft	Hard
S2	Medium	Hard	Soft	Medium	Hard	Soft
S3	Medium	Soft	Hard	Medium	Soft	Hard
S4	...	...	...	...	...	...

---

Table 3. Mean and standard deviation values for shock attenuation (percent) by gender-condition.

---

	<b>Slow Speed</b>			<b>Preferred Speed</b>			<b>Average</b>
	Soft	Med	Hard	Soft	Med	Hard	
Male	72.4	74.2	73.3	66.8	72.2	74.3	<b>71.7*</b>
	(9.0)	(6.5)	(10.5)	(13.4)	(8.0)	(10.1)	<b>(2.2)</b>
Female	81.9	83.8	85.0	81.0	83.5	87.1	83.7*
	(4.2)	(5.4)	(7.0)	(6.8)	(6.7)	(7.3)	<b>(2.0)</b>

---

Note: SA value of 100% indicates a total absorption/dissipation of the impact shock wave generated at contact as measured at the head;  
 \* indicates significant ( $p < 0.05$ ) differences between gender.

Table 4. Mean and standard deviation values for shock attenuation (percent) collapsed across surface and speed by gender.

---

	<b>Speed</b>			<b>Surface</b>			
	Slow	Preferred	Average	Soft	Med	Hard	Average
Male	<b>71.1</b>	73.3	<b>72.2</b>	69.6	73.1	73.8	72.1
	<b>(10.6)</b>	(8.5)	<b>(1.5)</b>	(11.3)	(7.1)	(9.9)	<b>(2.2)</b>
Female	<b>81.0</b>	83.6	<b>80.8</b>	81.4	83.6*	86.1**	83.7#
	<b>(7.0)</b>	(5.6)	<b>(0.3)</b>	(5.3)	(5.8)	(7.0)	<b>(2.3)</b>

---

Note: SA value of 100% indicates a total absorption/dissipation of the impact shock wave generated at contact as measured at the head;  
 # indicates significant ( $p < 0.05$ ) differences between **surface** conditions: \* **med > soft**, \*\* **hard > med**

Table 5. Shock Attenuation Prediction Models (Percent Explained Variance) by Gender-Surface

Surface (gender)	Model	Explained Variance (%)
<u>Soft</u>		
Female:	$(H\theta * -0.6) - (Hh\nu * 27.8) + (K\theta * 1.2) - (A\theta * 0.5) + (A\omega * 80.3) + 85.3$	40.1
Male:	$(H\omega * 102.7) + (K\omega * 40.5) - (Kv\nu * 36.2) + (A\omega * 90.4) + (Av\nu * 42.3) + 73.7$	59.9
<u>Medium</u>		
Female:	$(Hv\nu * 20.4) - (Hh\nu * 33.6) + (K\theta * 0.8) + (K\omega * 30.1) - (A\theta * 0.9) + 83.7$	30.9
Male:	$(Kv\nu - 32.8) + (Av\nu * 62.6) + 80.4$	36.0
<u>Hard</u>		
Female:	$(H\theta * -0.64) + (K\theta * 0.73) - (Kh\nu * 13.35) + 97.4$	32.0
Male:	$(K\theta * -0.85) + (A\omega * 84.2) + 84.3$	38.5

Note: See Appendix A for variable abbreviations.



## Captions for Figures

Figure 1. Force-Deflection Profiles for Three Treadmill Settings. Compression and release responses of the progressively loaded treadmill bed for each of three selectable surface stiffness settings. Level 1 represents the hard surface, Level 2 represents the medium surface, and Level 3 represents the soft surface. It can be observed that the stiffness response among the three levels was not linear, with Level 3 (hard) being approximately 2.5 times Level 2 (medium).

Figure 2. LgPk Predicted from Lower Extremity Kinematics. Multiple regression results illustrating the strongest prediction of LegPk (71,9% EV) was elicited from females running on a medium stiffness surface. The medium surface also resulted in the strongest prediction model for males (51.1% EV).

Figure 3. HdPk and LgPk vs SA by Gender-Surface. Individual running trials (n = 140 per graph) displaying the relatively unchanging HdPk and variable LgPk producing each SA value. Figure 3a: Females-Soft Surface; Figure 3b: Males-Soft Surface; Figure 3c: Females-Medium Surface; Figure 3d: Males-Medium Surface; Figure 3e: Females-Hard Surface; Figure 3f: Males-Hard Surface.

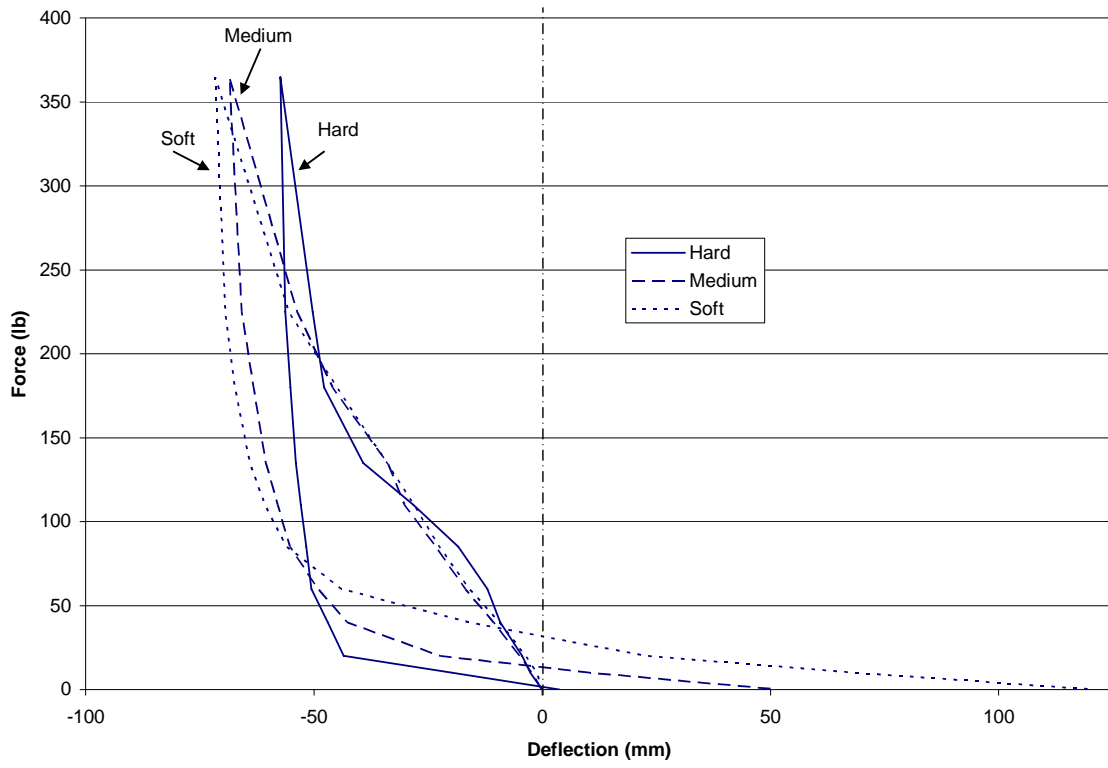


Figure 1. Force-Deflection Profiles for Three Treadmill Settings

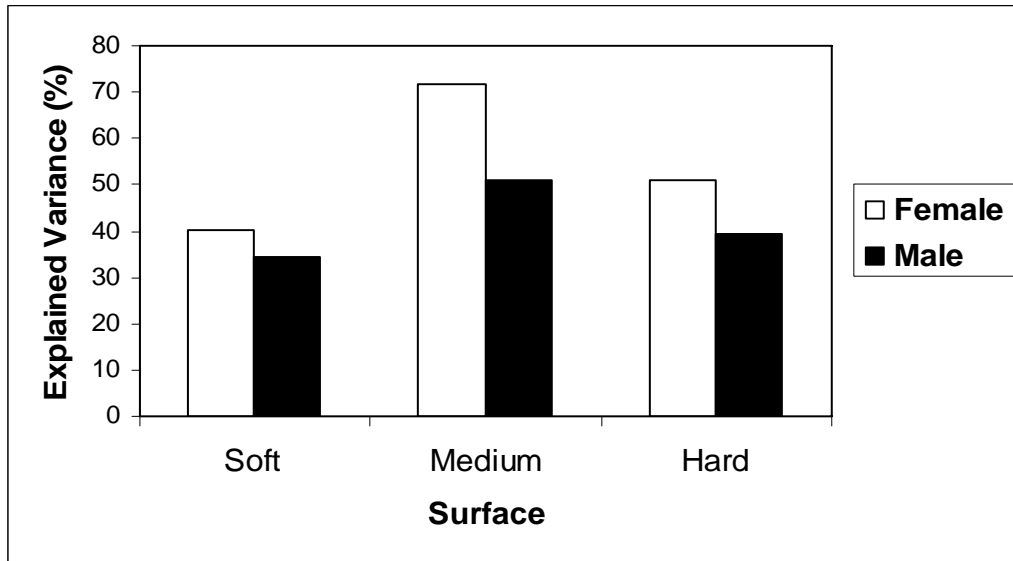
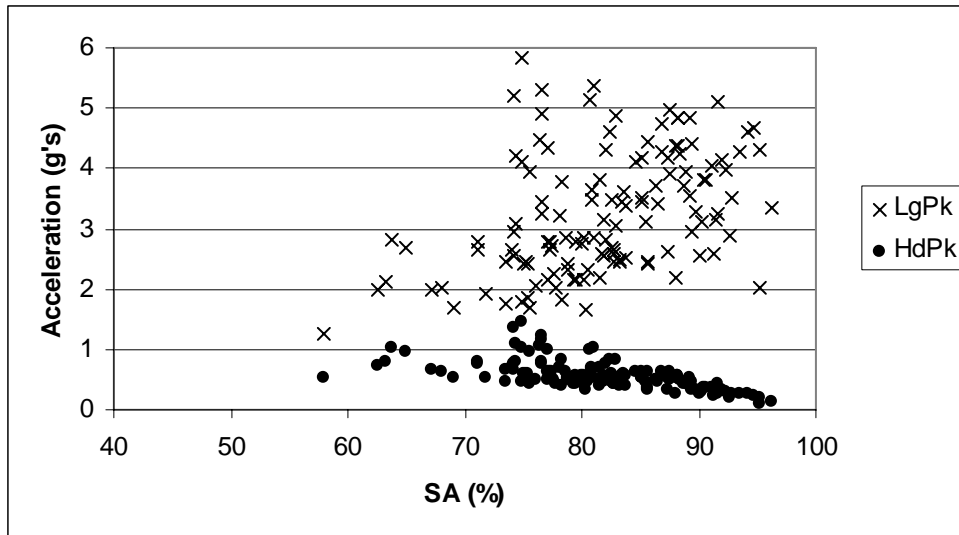


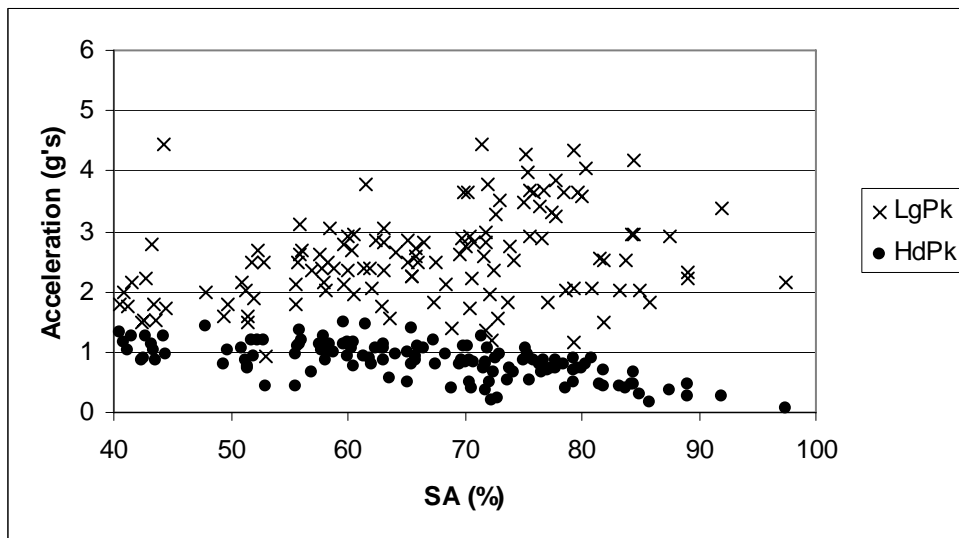
Figure 2. **LgPk Predicted from Lower Extremity Kinematics**

Figure 3 (a-f). **HdPk and LgPk vs SA by Gender-Surface:**

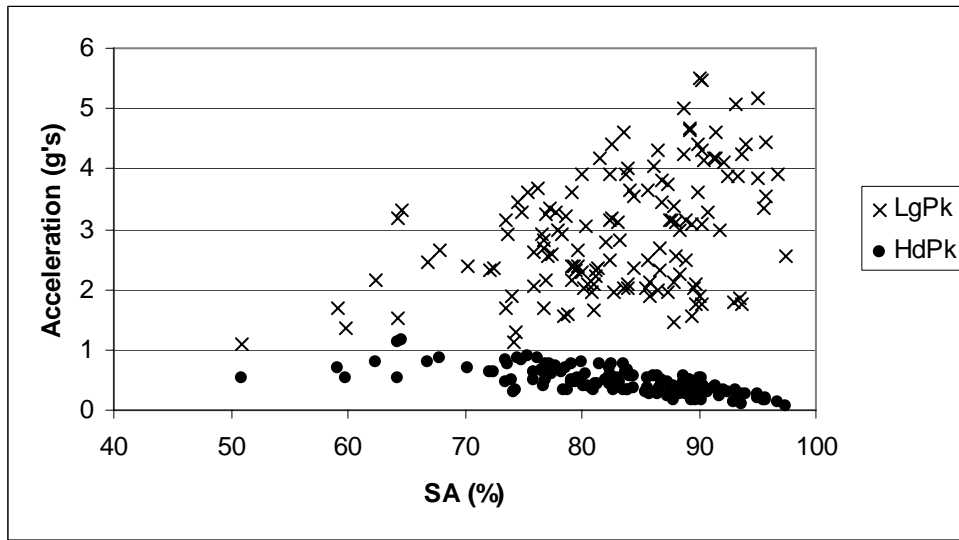
3a. **Females: Soft Surface**



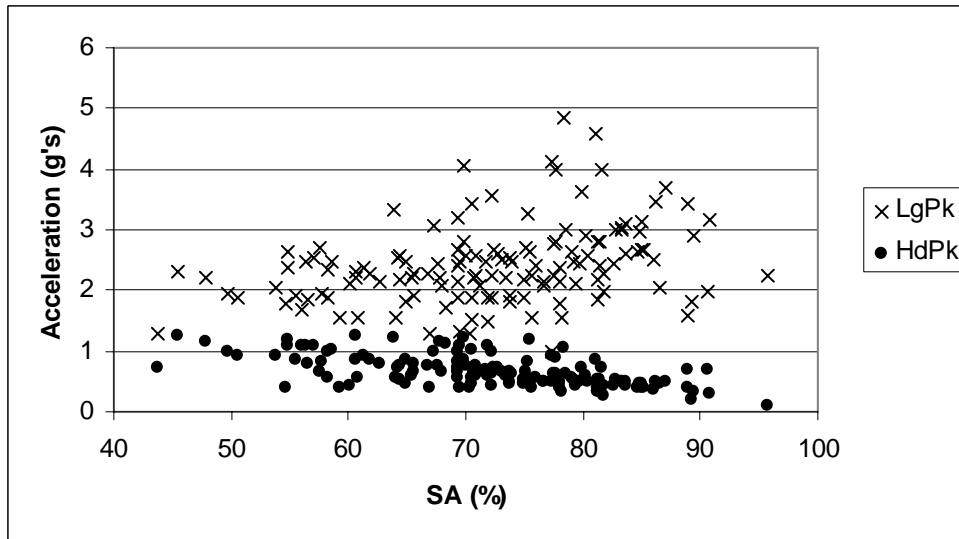
3b. **Males: Soft Surface**



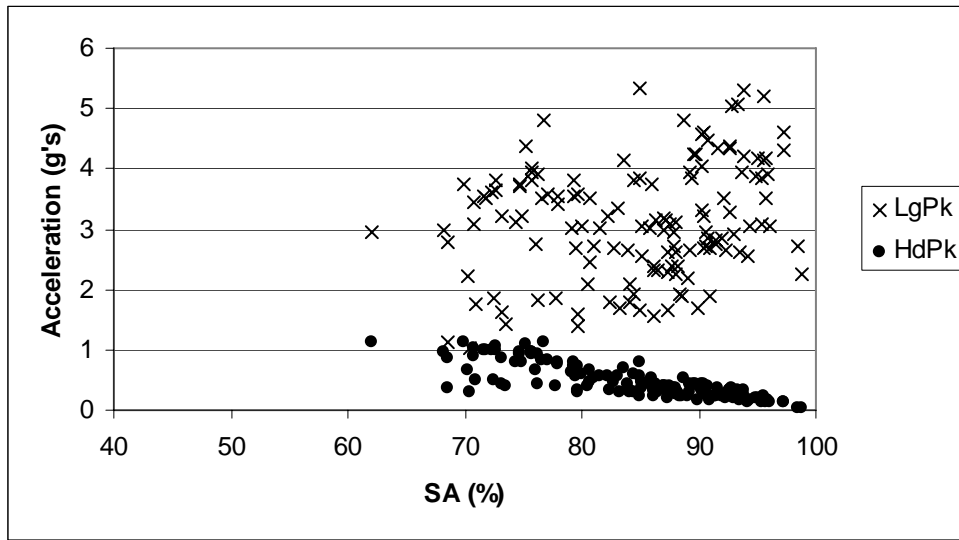
3c. Females: Medium Surface



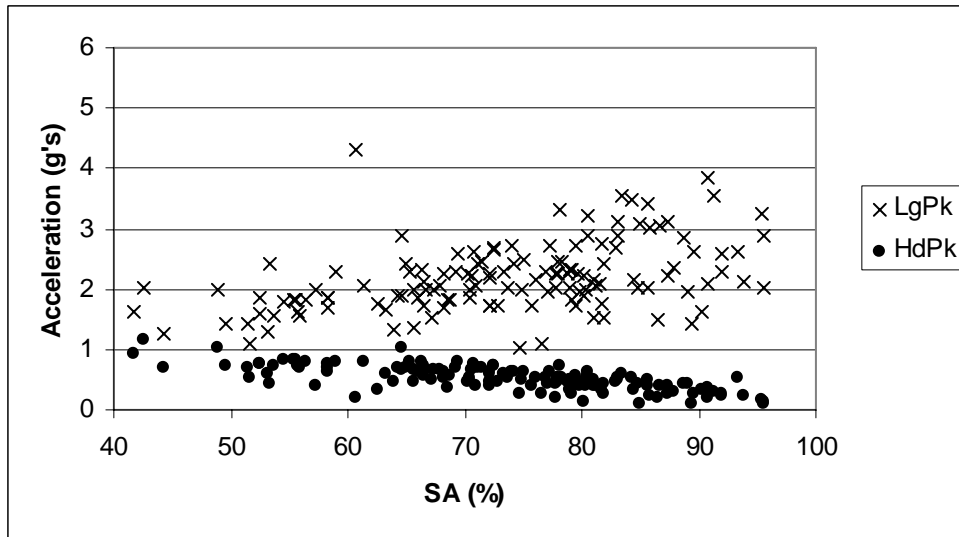
3d. Males: Medium Surface



3e. Females: Hard Surface



3f. Males: Hard Surface



Appendix A. Independent variables (and abbreviations) used for regression analyses.

---

<u>Independent Variable</u> (n = 13)	<u>Abbreviation</u>
<b>Hip Joint</b>	
Angular position at contact	H $\theta$
Angular velocity at contact	H $\omega$
Vertical velocity at contact	H $v_v$
Horizontal velocity at contact	H $v_h$
<b>Knee Joint</b>	
Angular position at contact	K $\theta$
Angular velocity at contact	K $\omega$
Vertical velocity at contact	K $v_v$
Horizontal velocity at contact	K $v_h$
Frontal plane position at contact	K $\theta_y$
<b>Ankle Joint</b>	
Angular position at contact	A $\theta$
Angular velocity at contact	A $\omega$
Vertical velocity at contact	A $v_v$
Horizontal velocity at contact	A $v_h$

---