The Effects of Surface Compliance on Knee Joint Loads During Level and Downhill Running

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Abstract

Running is known for providing numerous health benefits that untimely leads to decreased cardiac mortality. Despite these health benefits, running can cause overuse injuries in up to 79% of runners with a high incidence rate occurring in the knee[1,2]. Decline running in particular is associated with increased GRFs, resulting in increased knee contact forces. While decreased surface stiffness can result in increased leg stiffness in level running, it is difficult to determine how a softer surface would affect the knee joint in a decline setting. We predict a softer running surface would cause a reduction in knee joint loads during decline running because the runner's momentum would be directed more along the surface in decline running than level running. Therefore, we hypothesized that decreasing surface stiffness in decline running would result in a reduction in knee joint loads. The purpose of this study was to test the effect surface stiffness has on the knee joint loads in decline and level running in order to help prevent injury in runners.

14 healthy recreational runners provided written informed consent to university approved procedures. Knee joint patello-femoral compression and tibio-femoral compression

and shear forces were modelled from GRFs and kinematics during 10° decline and level running at a mean speed of 3.2 ms^{-1}. Participants ran with and without a 3 cm thick polyvinyl chloride (PVC) sponge shock absorbing mat (closed cell, Shore 00 65) places along the entire runway. Maximum forces were analyzed with 2-way ANOVA followed by Schefe post hoc tests, all $p < 0.05$.

Decline running on the softer surface lower the A/P GRF 56%, the patello-femoral compression, tibio-femoral compression, and shear forces 15%, 6%, 15% compared to the stiffer decline condition ($p < 0.05$). Surface stiffness had no effect on these forces in level running ($p > 0.18$). inclination angle and surface stiffness had no effect on maximum GRF. We predict the mat reduced knee forces in decline because the initial impact between the runner and the surface was more in line with the surface in decline, producing a larger displacement of the mat material in decline when compared to level.

We demonstrated that reducing surface stiffness can reduce knee loads during decline running at a 10° decline. Further, we found that the same decrease in surface stiffness had no effect on knee joint loads in level running. This interaction shows how there is a need for further analysis when investigating running biomechanics by investigating more than one setting and surface.

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By

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Chapter I. Introduction

Running as exercise enhances peoples' overall health and well-being and is one of the most popular forms of exercise with an estimated ten million people running regularly in the U.S. [3]. Health benefits associated with running are lower blood pressure, higher insulin sensitivity, increased cardiac output, and much more which ultimately leads to decreased cardiovascular mortality[4]. Although running has great health benefits, about 37-79% of runners experience some sort of running related injury every year [1,5,6]. DeHaven investigated running injuries even further and discovered 42% of the 93 recorded track and field runners injuries were at the knee, showing the deleterious impact running can affect ones knee joint [2]. Many of these injuries are due to high loading in the joints and other tissues and can be affected by many variables such as velocity, step length, and surface inclination[7–14]. People run up and down various inclines, loading their joints in different ways to help their body handle different stresses in the most efficient way possible[15].

The dissipation of the impact loads or the mechanical energy during the stance phase of running can be performed by body tissues such as muscles, tendons, cartilage and bones or by external devices that can be used to help cushion the load put on the body tissues. The external running surface can also change the joint load magnitude, whether it be the shoe worn, or the surface run on. Different shoes are made in different ways in order to specialize the types of loading different people put on their joints. Although it has been thoroughly researched, maximalist shoes have seemed to have minimal effects on how external loads are experienced in decline running when compared to level[1]. This was found in a study by Chan et al. where a

new shoe advertised to have the most support and significantly more cushioning than other running shoes on the market was tested against a conventional running shoe. Chan et al. found that despite having the characteristics advertised, the external loads experienced in the knee during running were very similar between the group using the maximalist running shoe and the control, demonstrating the minimal effect shoes can have on the distribution of external loads[1]. Although everyone has the same structures, they are all different in very subtle ways, loading their joints in different ways. While shoes may not be able to directly affect the magnitude of loads applied to the joints, perhaps the surface run on could[1,16]

Running surfaces can play a big role on runner's performance in more than just one way. For example, those who run on a very compliant track had their running speed reduced to 0.70 times the speed they could run on a hard surface due to the increased contact time in the stance phase of running[17]. This increased contact time could potentially allow runners to evenly disperse loads throughout the lower extremity tissues more than a hard surface. The hard surface emits the reaction force in a much more precise location due to the lack of compliance as the foot strikes the ground. This is especially true for those who strike heel to toe as forefoot runners have a 12% reduction tibiofemoral loading than those who run heel to toe[8]. This is even more relevant when focusing on the knee joint during running. This is because the knee joint has been shown to have the greatest role in energy absorption during running, meaning a more even distributed GRF could result in these forces being less impactful[12]. This finding also gives evidence as to why downhill running puts such a large load on the knee as running downhill greatly increases impact peaks in heel to toe runners[13].

As the primary shock absorber of the body, the knee is especially prone to overuse injuries[14,18]. The knee has multiple forces placed on it, primarily from the joint reaction forces and muscle forces. The knee also experiences compression forces that are a summation of tibio-femoral and patello-femoral forces while running. The knee joint experiences a great magnitude of both tibio-femoral and patello-femoral forces while running. These forces have been found to be up to six times a person's body weight in patello-femoral forces and up to ten times ones' body weight for tibio-femoral joint forces[19–21]. These intense loads occur in0 the knee joint every step, making it beneficial to reduce these loads while running. Reducing stride length and increasing frequency has been shown to successfully reduce these loads in the knee joint during level running but with decline running having a greater impact on the body it is crucial to reduce these loads even more[11,21]. Because a softer surface has been shown to reduce external forces put on the leg during running, perhaps a softer surface could prove beneficial at reducing tibio-femoral and patello-femoral forces experienced when running downhill as decline running has been shown to have much higher forces associated with it.

Gottschall et al. found just how much impact downhill running can have on ones' body [11]. They found that while running downhill on an asphalt surface, vertical impact forces increased by 14% when compared to level running. Figure 1 shows that the initial contact peak in the stance phase of downhill running is much greater than the initial contact peak of the level running. Along with a higher peak, the latter half of the stance phase in the downhill running is significantly lower than initial peak. Due to these high impact forces associated with downhill running, the probability for a musculoskeletal injury increases significantly and is not recommended for those who are in a state of recovery from and injury[11].
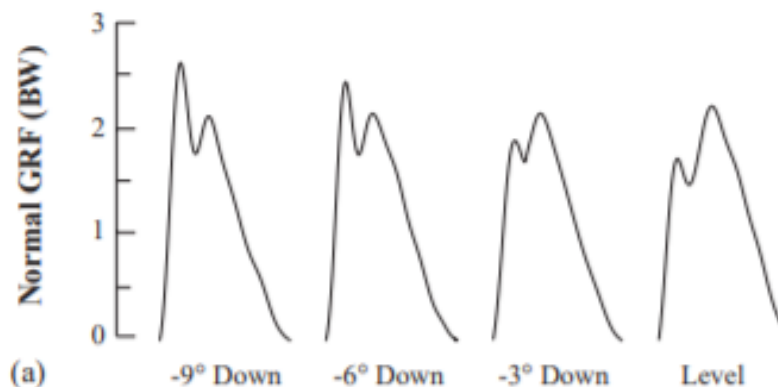
Figure 1. [11] shows the normal ground reaction force normalized by bodyweight

at -9, -6, and -3 degrees declined compared to the ground reaction force associated from level running

Joint loading is one of the many factors that are drastically affected by running downhill which takes a much greater toll on ones' body than level or uphill running[11]. Joint loading is affected not only by impact forces, but also muscle forces along with gravitational forces put on the runner while in motion. Compared to level running, downhill running requires more eccentric muscle work in an attempt to protect the joints in the leg from the increased impact forces while maintaining running velocity, putting a much greater load on the knee joint [10]. The leg muscles work so much eccentrically that they increase horizontal braking impulses in downhill running by 200% when compared to level running [11]. With downhill running affecting the body in so many ways, there are many questions regarding how one could reduce the intense impacts associated with it. With larger joint loads expected for downhill vs level running, we expect that a softer, more compliant running surface may reduce these loads and so potentially make downhill running less injurious.

Hypothesis

We hypothesized that a more compliant ground surface reduces loads experienced by the tibio-femoral and patello-femoral joints during downhill running due to the decreased impact forces that would occur in an altered declined run when compared to downhill running on harder surfaces. We expect an interaction effect such that the running surface in decline running will be larger than in level.

Purpose

The purpose of this study is to test the effect of compliant vs stiff running surfaces on the knee joint loads during level and decline running using padded rubber surfaces. The knee joint can be affected by many factors and can have the forces that act on it altered in a number of ways. Although many studies have investigated how to reduce these forces in many types of running, few investigate the direct relationship between different ground services in declined running. Because downhill running can take such a physical toll on the body, it is our goal to determine how ground stiffness could affect the forces at the knee joint in order to help reduce the injuries that are derived from the impacts of downhill running.

Delimitations

Delimitations of this study are the participants must be active runners who run a minimum of 10 miles per week and currently have no musculoskeletal injuries and have no surgeries performed on their lower extremity. If the participants have had any surgeries to a body part other than their lower extremity, it must have occurred over a year ago and they must exhibit no restrictions because of this surgery. There is only one predetermined ramp slope, one particular compliant surface, and one range of running speeds for this study.

Chapter II. Review of Literature

This study will investigate the different knee joint loads that occur while running at a decline with a dampened and hard surface compliance. With there being a significant number of variables than can be modified, there are many areas of research that focus on each variable to observe its' independent effect it has on the loads that occur at the knee joint. Variables investigated in previous studies focused on factors such as step length, ground surfaces, foot strike pattern, declination angle, running footwear or even a combination of more than one of these variables. Finding which variable could influence the loads experienced by the knee during running is difficult as many studies can have confounding variables such as natural running styles which are difficult to control and hard to manipulate in studies. Some studies, however, focus solely on manipulating these innate patterns in people. Bowersock et al., Munro et al., and Wilson et al. are all studies that focused on manipulating participants natural running styles, while other studies such as Gottschall and Kram noted these innate styles, but did not attempt to manipulate them[8,11,22,23]. Many factors can influence the loads experienced by the knee during running, however; some of these variables can have a much greater effect than others. This review of literature is partitioned into sections: joint stiffness and surfaces, impact forces in running, step length in knee joint forces, uphill vs downhill running, and the summary.

Joint Stiffness and Surfaces

When investigating how surface compliance effects knee loading in downhill running, it is important to consider all the variables that influence the knee and the effects this can have

on joint loading. It is important to understand the stiffness of the leg and how different surfaces influence the loading of the knee joint when running. Joint stiffness is defined as the change in angular displacement along a specific joint in relation to the joint torque placed on it[24–26]. Humans are known for adjusting their leg stiffness for the specific compliance of the surface there are running on in order to achieve different stride frequencies at the same speed[27]. For example, when running on a more compliant surface, a runners leg becomes stiffer and compresses less when compared to running on a less compliant surface[28]. This is all done to minimize injury risk as the stiffness properties of the leg can influence the loading of the leg and how the joints handles the forces put on it[29]. To better understand how leg compliance works, Ferris et al. created a model the visualizes the spring mechanism observed in locomotion. Although simple, the model (Figure 2a) well demonstrates the shock absorbing, spring mechanism of the leg. This model shows how the leg starts off fully extended and then compresses as the load is put on the leg, and then returns to its original position as the load is transferred to the other leg. Ferris et al also uses a stick figure to demonstrate the spring-mass model for a compliant surface (Figure 2b). This model shows three moments throughout the stance phase of locomotion. The first stick figure shows the initial contact with the ground and the spring model below it simply shows the spring mechanism of the leg. The second figure shows the loading of the leg in the stance phase when the leg is loaded and then moves to the final contact with the ground where the leg returns to its original position of, ready to be loaded on the next step. Figure 2b also visualizes the grounds vertical displacement that occurs when loaded, meaning the leg would remain stiffer to accommodate for the compliance of the ground.
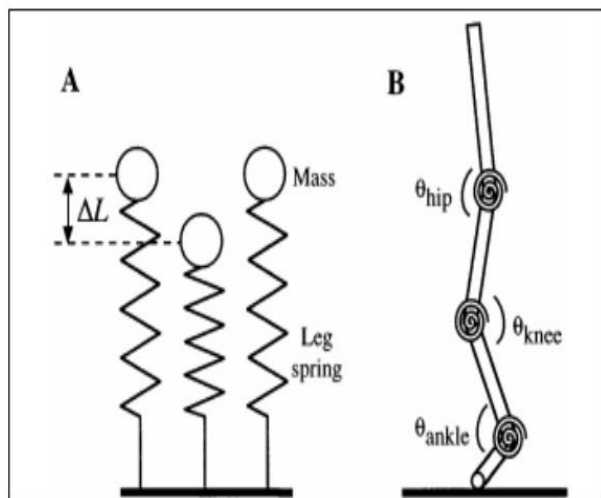
Figure 2a. Ferris et al (1998)[28]   Figure 2b. Ferris et al (1998)[28]
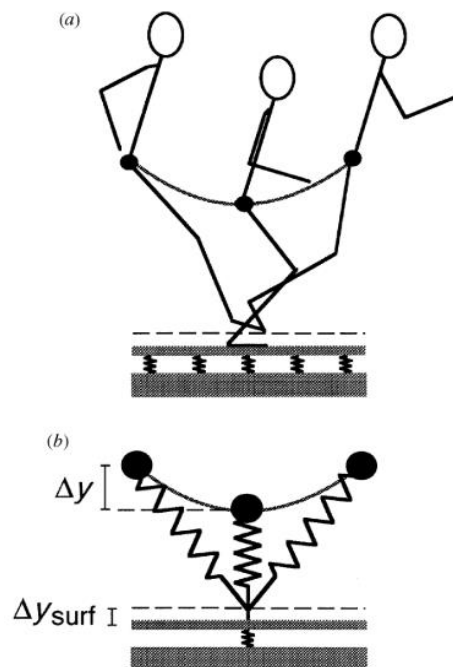
Both models simply visualize how the leg compresses during the ground contact phase of running and how the leg acts like a spring to maintain a level of homeostasis of stiffness between the leg joints and the ground. As the leg goes throughout the ground contact phase, there is vertical displacement of the hip as the leg is in contact with the ground. If these models were compared to a less compliant surface, there would be more vertical displacement in the spring model to accommodate for the lack of compliance in the grounds surface. This adjustment made by the leg joints occur naturally as the body instinctually maintains a consistent level of stiffness between the ground and the leg joints which could possibly help reduce injury risk throughout locomotion.

Along with preventing injury, leg joint stiffness can also mediate the energy cost of running by influencing the amount the work the muscles have to exert[30,31]. The knee joint has a predominate role in the overall stiffness and energy usage of the leg throughout the stance phase[12]. Jin et al investigated the effects that velocity has on joint stiffness throughout the stance phase of running and found that the joint stiffness of the knee joint was lowest of the three joints in the stance phase of running. This finding demonstrates how the knee joint is responsible for more energy absorption than the hip or ankle, giving it a coordination role between the ankle and hip joints in the stance phase[12]. This finding suggests that the knee would be the joint that would show the greatest change in stiffness when comparing differences in leg stiffness between compliant and non-compliant surfaces.

Impact Forces in Running

Ground reaction forces (GRF) put on runners can vary from person to person depending on running velocity, foot strike pattern, stance time, previous injuries, incline during running, or surface run on. There are many ways one can manipulate these variables so that the forces put on the body put someone at a reduced risk of injury. Munro et al. developed normalized GRF data on runners in order to assess gait. A big factor in the GRF experienced by the body was the foot strike pattern of runner. Those who strike with their heel experience a greater impact force than those who strike with their toe[8,13,22]. Rooney et al. investigated contact and joint forces in rear foot strike and forefoot strike runners by converting half of the rear foot strike runners to forefoot strike runners as well as convert half of the forefoot strike runners to rear

foot strike runners. When doing this, Rooney et al observed decreased contact forces in the knee for the runners who were converted from rear foot strike to forefoot strike runners. The exact opposite was observed for the group who was converted from forefoot strike to rear foot strike runners [32]. Many studies have investigated this and have found similar results where a forefoot foot strike pattern can dissipate or even eliminate the impact peak of the vertical GRF when compared to the heel strike runners[8,9,13,23]. While the impact peak may be decreased, the cumulative of GRFs are still similar as forefoot runners have a slightly higher active peak than heel strike runners[13]. One factor that is clearly influenced by foot strike pattern is the muscle activation in lower extremity which can affect the torques and loads experienced at each of the joints in the leg as muscle forces make up a majority of patello-femoral and tibio-femoral forces that occur in the knee[13].

Step length on knee joint forces

Bowersock et al. investigated a similar running scenario as Munro et al. with manipulation of preferred step length and the tibiofemoral joint forces during running with consideration of foot strike patterns in the participants. By decreasing ones' step length by 10% at a given velocity resulted in 12% decreased peak tibio-femoral joint force as well as 14% lower tibio-femoral joint force impulse per step, and a 20% lower tibiofemoral joint force average

loading rate, regardless of the runners foot strike pattern[8]. TIbio-femoral joint contact force had the same effect observed when ones' step length was increased by 10% with tibio-femoral joint contact force increasing by 10%[8]. Wilson et al used a very similar protocol with manipulation of step length to determine its effect on patellofemoral joint stress. With a 10% decrease in stride length, patello-femoral joint stress decreased by 22% and 16.3% average

Fig 3 shows the average patellofemoral joint stress for each condition with standard error of the mean shaded

decrease in peak patellofemoral joint stress [23]. These findings coincide with previous studies that observed similar trends as step length decreased[8]. Figure 3 shows the patellofemoral joint stress for participants preferred step length, step length increased by 10%, and step length decreased by 10%. This graph clearly shows the relationship observed between step length and joint stress. With downhill running known to have greater joint loads than level or inclined, these findings suggest that decreasing step length could potentially decrease these forces seen in downhill running[11,13].

Different running velocities have been observed to affect joint loads and forces in the leg throughout locomotion in a number of ways. Lin and Hahn investigated running velocities effect on joint stiffness in the ankle, knee, and hip. The tested speeds for running ranged from 1.8 to 3.8 m/s and joint stiffness at the knee increased as the speed increased[7,12]. Munro et al. found that as running speed increases, the average vertical ground reaction force relative to

body weight increases as well[22]. Increased running speed has been observed to effect joint torques, joint work, and GRF with speeds increasing from 2.5 to 3.5 m/s and 3.5 to 4.5 m/s as well as cause an increase in stride length [33].This observation can explain why Munro et al. found an increased peak vertical GRF for increased running speeds in their studies. Although joint stiffness and GRF increase as running speed increases, knee joint impulse can decrease with increased running speeds. Petersen et al. assessed knee joint impulse in relation to increased running speeds and found a 27% reduction in knee joint cumulative impulse by increasing running speed from 8 to 12 km/h. As helpful as this finding is in reducing knee joint impulses, such a drastic increase in speed for a set distance is hard to achieve in many runners. This tactic would not be ideal for downhill running as runners work to lower themselves as projectiles as their center of mass is being lowered[15]. This would inhibit ones' ability to increase speed downhill since muscle work being down in downhill running is already accelerated to safely lower ones' center of mass and prevent injury[15].

Uphill vs Level vs Downhill running

Different levels of incline and decline can greatly affect the kinematics and kinetics of ones' lower limb when running. Figure 4 below shows level running in rear foot strike runners where there is the impact peak observed followed by the active peak in normal GRF (a) as well as the parallel GRF (b)[11]. The parallel GRF shows how the initial phase of running is the braking
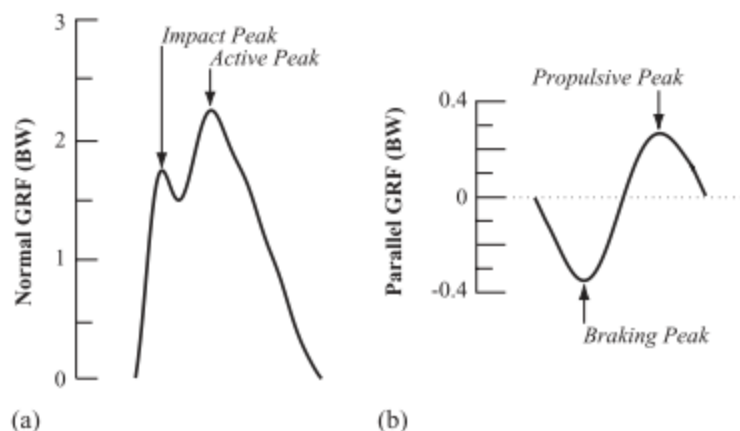
Figure 4 Gottschall and Kram

Phase where the runner works eccentrically to absorb the shock administer by the ground followed by the proposition phase where the runner pushes off to continue their momentum. As incline increases in running, there is a much lower braking peak as the runner must work more to keep their momentum[11]. The inverse effect is observed as the incline decreases, with a much greater braking peak and a minimal propulsive peak. The same type of effect is also observed in the normal GRF with a much higher impact peak that is greater than the active peak when running on a decline. The magnitudes of the normal impacts also make it clear how much the incline one is running at can make. Gottschall and Kram calculated an average normal impact force peak for 7 participants to be 767N ± 184N for a 6-degree incline while the 9-degree decline had normal impact force peaks of 1504N ± 362N[11]. The decline impact force peak is more than twice the inclined peak, and the inclined peak was only for the 6-degree condition as the 9-degree incline was unable to be calculated. While the impact peaks varied greatly, the active force peaks remained consistent regardless of the incline level. This shows that incline level really effects the

impact the body experiences in the initial portion of the stance phase. The table below shows

different conditions for each incline level and the forces associated with them.

Effects of downhill and uphill angle on running normal impact force peaks and loading rates

Condition (deg)	Normal impact force peak (N), $n = 10$	Normal impact force peak (N), $n = 7$	Ave impact loading rate (kN/s), $n = 10$	Ave impact loading rate (kN/s), $n = 7$	Max impact loading rate (kN/s), $n = 10$	Max impact loading rate (kN/s), $n = 7$	Normal active force peak (N) $n = 10$
−9 Down	1504±273[†]	1504±362[†]	47.7±9.3[*]	49.1±12.6[*]	94.6±29.8	83.8±24.8[*]	1341±177
−6 Down	1283±255[*]	1305±322[*]	44.8±8.9[*]	44.8±11.5[*]	88.2±28.5	79.2±23.2[*]	1325±205
−3 Down	1149±263[*]	1174±375[*]	41.3±9.4	42.6±10.8	78.0±25.9	73.0±12.1	1375±198
Level	974±193	988±252	34.9±6.2	36.3±6.7	71.2±20.8	66.1±13.0	1373±199
+3 Up	848±176	840±222	27.8±8.4	32.5±7.8	57.5±20.0	53.5±13.5	1384±168
+6 Up	—	767±184[*]	—	29.4±5.7[*]	—	48.9±15.8[*]	1362±174
+9 Up	—	—	—	—	—	—	1334±182

Values represent means for all 10 subjects±standard deviation and 7 subjects that initiated foot strike on their rear-foot. Conditions that differ from level running are denoted with an '[*]', $p<0.05$ or with an '[†]', $p<0.01$.

Table 1. Gottschall and Kram

Throughout the table there is the common trend of forces increasing as the incline decreased.

This argument holds true for all cases until the normal active force peaks. All other conditions

refer to the impact forces, and all show how an increased decline causes an increase in forces

put on the body. Telhan et al confirmed the findings of Gottschall et al by analyzing lower limb

joint kinematics during moderately sloped running[34]. While looking at lower limb joint

kinematics, Telhan et al. observed an increase in impact peak of the vertical GRF during the

declined running portion of their protocol, speculating these increased impact forces could lead

to musculoskeletal injury[34].

Wells et al. conducted a similar study where they analyzed the effect of downhill

running on female runners. Their study focused on the peak vertical impact force, starting at

level running, and decreasing the decline by 5% for each condition up to 20% decline[14]. They

observed the same trend as Gottschall and Kram, where the peak vertical force shifted from the active peak to the impact peak as the downhill grade increased[11,14]. Figure 5 below[14] shows side by side comparison of the vertical GRF throughout the stance phase as the decline grade became more extreme.
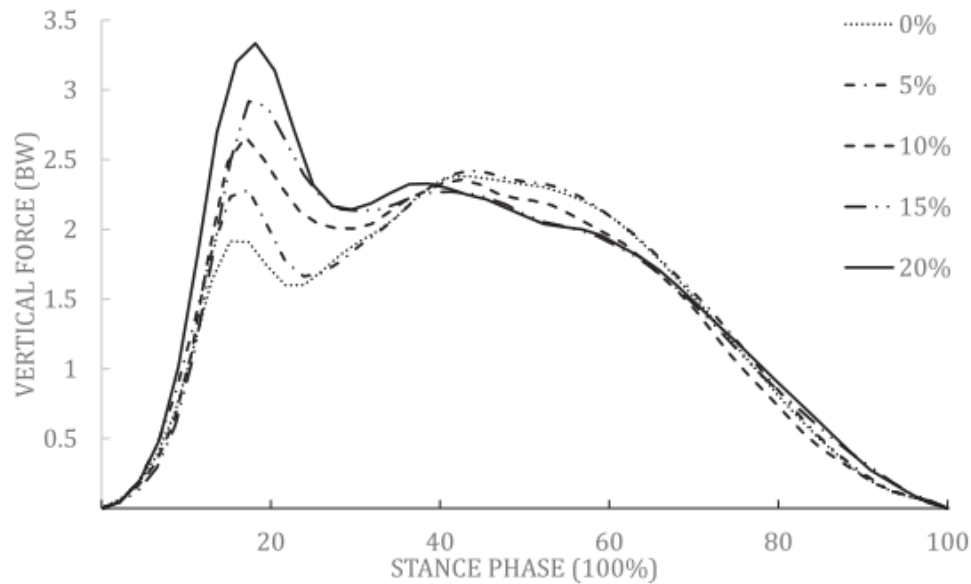
Figure 5. Wells et al

This figure really visualizes the shift seen in the impact peak, with GRF going from 2 times ones' body weight to almost 3.5 times ones' body weight[14]. There is a much smaller shift in the propulsive peak with the GRF only changing about 0.5 times ones' body weight, however; this is peak is not what causes musculoskeletal injury in runners. While Wells et al. found a significant difference in the GRF and loading, very little difference was observed in the braking forces. This lack of difference suggest that braking force is not main factor in declined running and has little effect on the downhill running, musculoskeletal injury relationship.

Hardin et al. conducted a two-part study that manipulated ground surface, incline level, and shoe midsole to observe the effect on joint stiffness, metabolic cost, impact loading, and joint kinematics over time. The experimental group that ran at a decline did so at a -12% grade for 30 minutes with no surface manipulations. The other experimental group ran at a level grade at a given surface stiffness for 6 minutes. They would then rest until their heart rate was less than 120 bpm and then repeated the protocol at a different surface stiffness[35]. Hardin et al. found that increased stiffness of the ground surface resulted in reduced stiffness in the lower extremity joints to maintain a uniform combined stiffness between the runner and surface, which agrees with previous studies that investigated joint stiffness[28,35,36]. They also found the runners had higher metabolic costs at the dampened surfaces, perhaps due to the work required to maintain a stiffer lower extremity[35]. In the second part of the study that investigated prolonged downhill running, Hardin et al. saw no kinematic adaptations at the knee to compensate for this increased GRF put on the body[35]. Although there was observed adaptions at the hip and ankle, the knee could still potentially receive a greater load as it did not vary between the level and declined surface.

Summary

Many factors can influence the loads experienced at the knee joint during running. Incline, step length, running velocity, foot strike pattern and surface compliance can all influence the knee joint loads in running. When looking at the factors that influence the knee joint loads and leg stiffness in Hardin et al. two-part experiment, it leads to the inference of

what would happen if these two scenarios were combined. With the idea that there is a

constant total stiffness between the runner and the surface[28,35,36] and downhill running putting

greater impact forces on the body[11,13,14,34,35], what if the declined surface was cushioned so that

the loads on the knee joint could be reduced. The knee is the target of more running related

injuries than any other lower extremity joint and receives a much greater load in downhill

running[2,11,13], so this study will investigate the loads experienced by the knee in downhill

running during hard surface running, and soft surface running.

Chapter III. Methods

This study was designed to determine if there is a difference in the loads experienced by the knee in declined and level running with two stiffness conditions and aimed to compare the effect of surface compliance as a function of surface interaction. We hypothesized that the loads experienced by the knee during declined running will be reduced when the surface stiffness is reduced due to the predicted reduced impact peak seen in running. This chapter describes the characteristics of the participants, the study procedures, the instruments used, and statistical analysis used.

Participants

Fourteen participants (eight males and six females), ages 18 – 26 years old were recruited from the city of Greenville, North Carolina. Students of East Carolina University (ECU) and Greenville residents were included in the criterion for participants. Subjects were recruited with flyers, word of mouth, and website announcements. The participants were recreational runners who runs between 10 and 50 miles a week, are healthy and mobile, and are habitually heel strike runners to maintain the consistent ground reaction force pattern. Participants must have no current musculoskeletal injuries or any condition that causes pain of the lower limb, any surgeries that affected the knee joint, nervous system diseases, cardiovascular disease, or any other disease that would affect their ability to perform running tasks. All participants will read and sign an informed consent from prior to participation and all procedures were approved by the ECU Institutional Review Board (Appendix B and C).

Instruments and Software

All data collections were performed on a 15m ramp with an in-ground imbedded force platform (AMTI Model LG-6, Newton, MA) with a sampling rate of 960Hz and a gain of 4000. Rubber mats (Consolidated Plastics Ultra Sponge Mat, Stow, OH) were used to create a damped surface stiffness for the second condition. Participants running pace were timed using an infrared timing gate (Brower Timing Systems, Model IRD-T175, Salt Lake City, UT) so that a verified, consistent speed can be achieved. When timing gates where not working, posterior hip marker velocity was filtered and analyzed using Qualisys software to ensure the average velocity was within an acceptable range. Twelve Qualisys Oqus 300 motion capture cameras were used for motion capture, with two of the cameras collecting video (Qualisys Medical AB, Gothenburg, Sweeden) at 200Hz. Motion capture and force data was processed with Qualisys Track Manager (QTM) Software (Innovation Systems Inc, Columbiaville, MI) and then analyzed using Visual 3-D (V3D) program (C-Motion Inc, Rockville, MD). Quick Basic (IBM Basmark QuickBasic, Cleveland, Ohio) was used to process kinetic and kinematic data. Specific muscle and joint measurements including tibio-femoral and patello-femoral forces in the lower limb will be analyzed to compare the results obtained. Participants height and weight was taken prior to each data collection using a Seca 703 digital scale (Seca GMBN & C. Kg, Hamburg, Germany). Prior to each data collection, all equipment was calibrated and maintained according to the factory protocols.

Procedures

Running Conditions

Four running conditions consisting of a stiff surface and a softer surface as well as a level and decline run were used. Condition one (C1) was the controlled stiff floor with no extra padding at a decline of 10°. Condition two (C2) was the damped surface with two PVC sponge (shock absorbing, 100% closed cell, Shore00 65 mat) to create the soft running surface (length 914.4cm x width 91.4cm x height 3.1cm) at a decent of 10°. The rubber pieces were long enough to cover the entire ramp as well as the first few feet of the level surface at the bottom of the ramp. One mats specifications will be obtained using a manufacturer derived durometer value of Shore 00 65 $\pm$ 10. Condition three (C3) consisted of the controlled stiff floor with no extra padding on a level running setting and condition four (C4) had the damped surface of two PVC sponge mats at the level condition to create the softer surface.

Data Collection Phase

Prior to the participants arrival, the force plate was located in QTM program using an L frame to mark the coordinates in the system and 4 placement markers were used to define the area of the platform. The testing area and motion capture location were calibrated with a 750.1mm T-wand with an accepted calibration trial consisting of less than or equal to a 2mm average residual error per camera capture. 15 reflective markers and 3 reflective marker plates (for thigh, shank, and foot) were set out and ready to be used on the participant. The infrared timing gates were set up 2m apart (1m away from either side of the force plate) and verified it was working correctly.

Participants were asked to come prepared in spandex shorts and a tight-fitting shirt or asked to change upon arrival, have height and weight recorded, and self-selected running shoes on. Before beginning the data collection, informed consent was obtained upon their arrival as well a short questionnaire that provided more background on their running habits. Following the informed consent, the procedures were explained to the participants and any questions they had were answered. Athletic wraps were placed around the hips, right thigh, and right shank for marker and maker plate placement. Markers were placed using anatomical bony landmarks near segment endpoints; right and left superior iliac spine, posterior iliac spine, iliac crest, greater trochanters, and lateral and medial femoral condyles and malleoli. Markers for the first and fifth metatarsal heads were place on the outside of the participants running shoe and secured with tape. The thigh and shank plates were placed on top of the athletic wraps put on the thigh and shank segments and secured with tape while the shoe plate was placed on top of the participant's running shoe and also secured with tape. A calibration trial with all tracking markers was recorded as the participant stood on the force plate for five seconds, followed by a second standing calibration trial which was performed without the calibration markers (right and left iliac crest, greater trochanters, lateral and femoral condyles and malleoli, and the first and fifth metatarsal heads).

Following the standing trials, participants were given time to practice running down the ramp. Participants ran down the 15m ramp at a 10° decline, initially with no padding (highest stiffness) and again with two mat layers (softest stiffness). Velocity was determined using the infrared timing gates which were placed 1m from the center of the force platform so that participants could practice running at 3.35m/s $\pm$ 5% (7:00 min/mile). Five successful running

trials were recorded for both stiffness conditions with a successful trial being defined as

participants running at the required speed, ran the full length of the track, had their right foot

strike the force plate in a normal stride, with all markers being visible. Following the ramp

condition, the participants were asked to perform five successful trials in the level condition in

both the stiff and soft condition. Running condition order of both running surface and decline

were randomly presented to the participants to avoid order effects. Once all trials are collected,

all markers were removed, participants were thanked for their participation and reminded that

they could access their data obtained at any date if they desired to do so.

Data Processing

Walking kinematics data were collected using QTM, and then processed in V3D. A

subject specific linked rigid-segment model of the pelvis and right lower limb were created in

V3D using the static calibration file for each subject. Location of the individual reflective

markers within the global coordinate system as well as the local coordinate system was

determined, as well as the virtual joint centers, and each segments center of mass using

calibration recordings. Ankle and knee joint centers were determined calculated as 50% of the

distance between the medial and lateral malleoli calibration markers and medial and lateral

femoral epicondyle calibration markers. A similar method was used for the hip with the joint

center being determined by calculating 25% of the distance between the right and left greater

trochanters calibration markers. The segments longitudinal axis was creating a line from the

distal to proximal virtual joint centers of each segment, using these data above and each

segments center of mass was determined using the subject height and weight and anthropometrics.

V3D was used to calculate joint reaction forces and joint torques using linear and angular Newtonian equations of motion. Ground reaction force, center of pressure, segmental anthropometrics, kinematic position and acceleration data was used for the calculations to calculate joint toques and joint reaction forces. This approach used inverse dynamics beginning with the segment where the known ground reaction forces come from (foot), moving proximally to the shank, and then the thigh, using the prior segment to calculate the next. V3D always uses the right-hand rule to determine the direction of the calculated torque.

Biomechanical Knee Model

The biomechanical knee model used calculated the tibio-femoral and patello-femoral joint forces during the stance phase during level and decline running in both the padded and unpadded conditions. Force produced by the gastrocnemius, hamstrings, and quadriceps muscles as well as tibio-femoral and patello-femoral contact forces were calculated by inputting the joint forces, movement kinematics and moments of inverse dynamics. Figure 6 by DeVita et al. (2001) and Messier et al. (2011) shown below demonstrates the compressive and anterior-posterior shear forces within the tibio-femoral area of the knee joint. Inverse dynamics was used to determine joint reaction forces, moments, and kinematics to calculate forces produced by the quadriceps, hamstrings, and gastrocnemius as well as lateral support provided by the

lateral collateral ligament. The horizontal and vertical reaction forces as well as the summation

of the previous forces allow for an estimate of the tibio-femoral force[26,37,38].
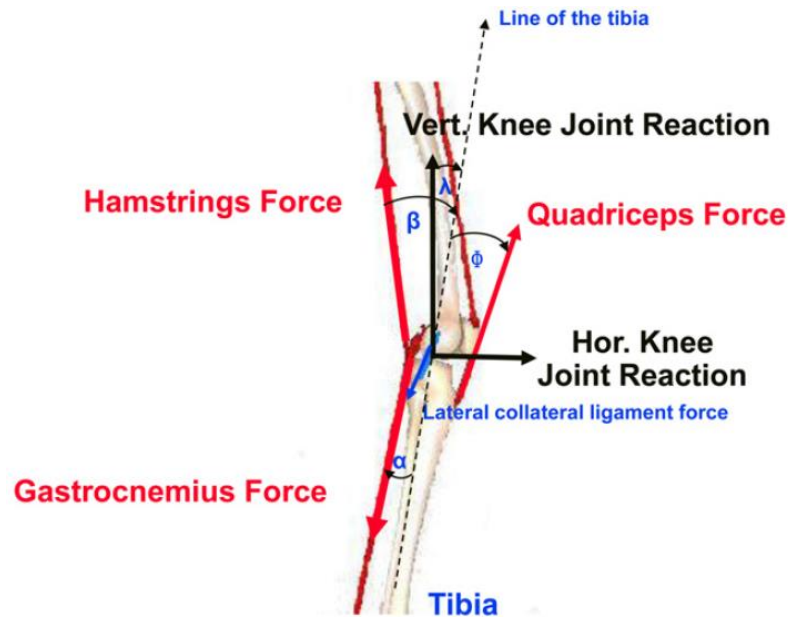
Figure 6. Messier et al. (2011)[37]

Patello-Femoral Joint Force

Patello-femoral joint is primarily to determine the contact force between the femur and

the patella which is primarily generated by the quadricep muscles along the femur. This focuses

primarily on the rectus femoris and vasti muscles, the patello-femoral joint reaction force

(PFJR), and the tension of the ligamentum patellae. Position data was used to calculate the

knee joint angle and was represented by the angle β assuming that PFJR is normal to contact

surface. Both tension values are assumed to be equal without account of friction on articular

cartilage so the angle β was divided in half, split by the PFJR vector which represented as the

angle value α. Trigonometry was used to calculate the forces and the resulting force

represented the net patello-femoral joint force. These calculations for patello-femoral joint

force were based on methods used by Ahmed et al. (1987) (Figure 7) and were supported by

Willson et al. (2015) in previous literature, assuming the knee joint was healthy[21,39].
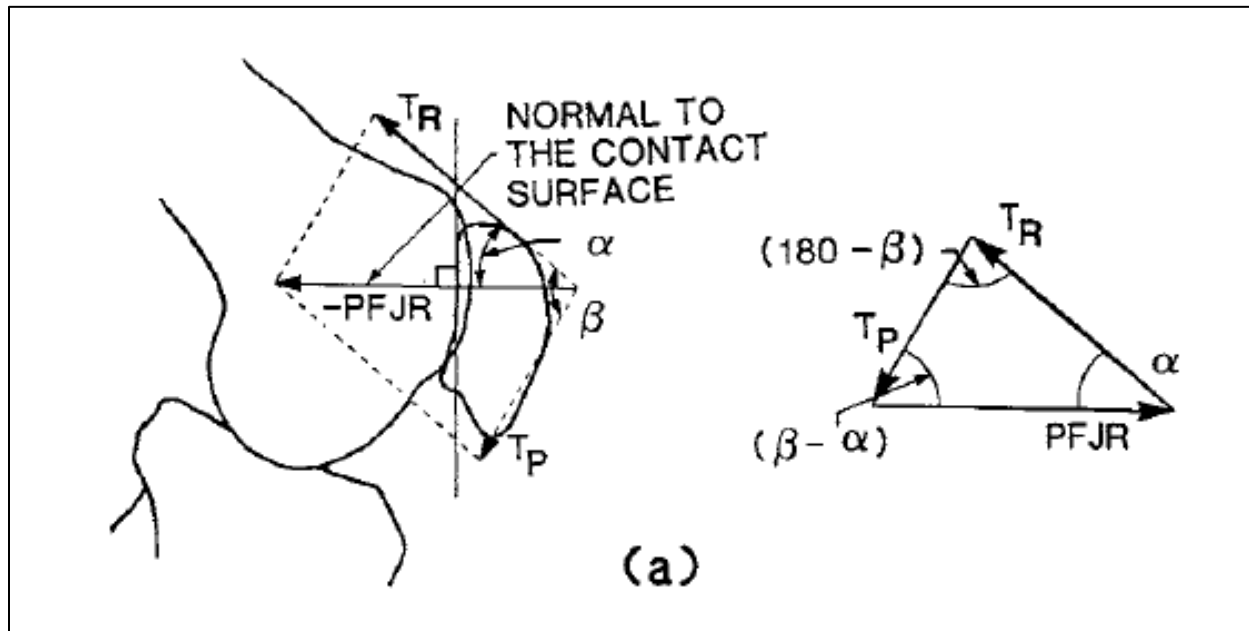
Figure 7. Ahmed et al. (1987)[39]

Tibio-Femoral Joint Force

Forces generated by muscle groups were determined during the stance phase of the

participants stride. Plantar flexor moment produced by the triceps surae muscles

(gastrocnemius and soleus) was used to calculate the gastrocnemius force. To find the triceps

surae force (TS) the planar flexor moment (At) was divided by the moment arm for the triceps

surae at the observed angular position of the ankle (ATd), assuming there is no co-contraction

by the dorsiflexors (Equation 1). Moment arm angle joint position curves from Rugg et al.

(1990) were used to determine the observed angular position of the ankle moment arm[40].

Gastrocnemius force (G) was then derived from the TS based on its proportion of the total

physiological cross-sectional area (PCA) of the triceps surae (equation 2)[37,38]. A was used to

represent the angle between G and the tibia as G applies to a large compressive load at the

knee by a small shear load.

Extensor moment at the hip that was observed during the stance phase was used to

calculate the hamstring force (H). Previous literature supports this method as it showed a string

association between hip extensor torque and hamstring EMG in early stance phase[18,41,42].

Hamstring PCA relative to the total Pca of the hamstring, gluteus maximus, and the hamstring

moment arm at the hip relative to the gluteus maximus moment arm was accounted for by H.

Equation 3 shows how the total hamstring proportion (Hp) to the hip extensor torque where

hamstring PCA and gluteus maximus (GM) PCA are the hamstrings and gluteus maximus PCAs

and GM moment arm and H moment arm are referred to as GMd and Hd. Previous literature

was used to obtain these constants where H PCA = 42.4 mm², GM PCA = 17.36 mm², Hd = 0.042

mm², and GMd = 0.047 m and moment arm values were determined from angular positions

between 30° and 90°[43]. These formula and constants can then be used to calculate the

proportion of hip extensor torque generated by the hamstrings which was found to be 0.63

(Hp). Equation 4 was used to calculate H where Het is used to refer to hip extensor torque. A

force of 0 is assumed for H while the hip torque was in the flexor direction which is supported

by EMG data in previous studies[44–46]. An angle of β was used to represent the angle of the force

direction generated by the hamstrings which was set parallel with the femur.

Observed net knee torque (Kt), H, G, and accountment of co-contraction of the knee

flexors was used to calculate the quadriceps force (Q). All muscles that crossed over the knee joint (Q, H, and G) were used in equation 5 to calculate Kt via inverse dynamics. The moment arms of the muscles were represented by Qd (0.035 m), Hd (0.032 m), and Gd (0.018 m) and all values were derived from previous literature through all angles of the knee position during the stance phase and then using the average. Q was then calculated using equation 8 and the direction of the quadriceps force was represented by φ, and was determined from literature as a function of knee angle[47,48].

The lateral collateral ligament is responsible for the loads in the frontal plane and was also used to determine the lateral support structure in the knee. The abductor moments from the quadriceps and lateral structures resisted the adductor loads that were placed on knee adductor moment. Observed net internal abductor moment arm was found from the product of Q and the frontal plane lever arm provided by the quadriceps abductor moment. The moment distributed to the lateral knee tissue was then found using inverse dynamics by dividing the torque by the moment arm found from position data through V3D. This force was considered parallel to the line of the tibia according to Messier et al (2011)[37].

The final calculations used to find tibiofemoral joint forces, both shear (Ks) and compressive (Kc), took the sum of all muscle forces (Q, H, and G), the force in the lateral support structure (Ls), and the joint reaction forces (Kz and Ky) found within the inverse dynamic's equations 7 and 8.

$$TS = At\ /\ ATd \qquad\qquad \text{Equation 1}$$

$$G = TS\ (0.319) \qquad\qquad \text{Equation 2}$$

$$Hp = [\text{Ham PCA}/\ (\text{Ham PCA} + \text{GM PCA})]\ (Hd/GMd) \qquad\qquad \text{Equation 3}$$

$$H = Hp\ (Het)/Hd \qquad\qquad \text{Equation 4}$$

$$Kt = Q(Qd) - H(Hd) - G(Gd) \qquad\qquad \text{Equation 5}$$

$$Q = (Kt + H(Hd) + G(Gd))\ /\ Qd \qquad\qquad \text{Equation 6}$$

$$Ks = G \sin \alpha - H \sin \beta + Q \sin \phi - Kz \sin \lambda + Ky \cos \lambda \qquad\qquad \text{Equation 7}$$

$$Kc = G \cos \alpha - H \cos \beta + Q \cos \phi - Kz \cos \lambda + Ky \sin \lambda + Ls \qquad\qquad \text{Equation 8}$$

Statistical Analysis

The following variables will be averaged over the five trials for each condition and runner.

- Running velocity

- Stride length

- Magnitude of the vertical GRF impact peak and rate of force application to this peak force

- Magnitude of the maximum vertical GRF and rate of force application to this peak force

- Magnitude of the maximum braking anteroposterior GRF

- Magnitude of the tibio-femoral compressive force at the time of the vertical GRF impact peak and rate of force application to this force

- Magnitude of the maximum tibio-femoral compressive force and rate of force application to this peak force

- Magnitude of the maximum tibio-femoral shear force and rate of force application to this peak force

- Magnitude of the patello-femoral compressive force at the time of the vertical GRF impact peak and rate of force application to this force

- Magnitude of the maximum patello-femoral compressive force and rate of force application to this peak force

Statistically significant differences over the two running modes (level and decline) and surface conditions (no mat and mat) will be determined using a two-way analysis of variance (ANOVA) with repeated measures and an alpha value being set at 0.05. In the event of a significant interaction effect, the effects of surface condition were tested independently in each running mode with standard Student's t-test.

Chapter IV. Results

This study's purpose was to compare the effect different surface compliances have on tibio-femoral and patello-femoral joint contact forces while running both level and downhill. We hypothesized that a more compliant ground surface will result in a reduced magnitude of forces experienced by the knee tibio-femoral and patello-femoral joints during downhill running due to the decreased impact forces that would occur in an altered declined run when compared to downhill running on harder surfaces. Where no interaction was found between the mat and no mat condition in decline running, we anticipate some interaction occurring between hard and soft surfaces in level running. Kinematic data, knee and limb parameters and ground reaction forces were used to evaluate this hypothesis by examining and comparing these variables across two different surface conditions in level and decline running to determine differences in running mechanics. Patello-femoral and tibio-femoral compression force as well as tibio-femoral shear force was also examined as the surface stiffness and decline level changed. The ground reaction forces, linear kinematics, knee and limb parameters, and knee contact forces are all broken down into subcategories later in this chapter.

Stride length

The interaction effect was not significant, $p < 0.107$, see figure 8. Participants ran with 11% longer strides in level compared to decline running, $p < 0.001$ but ran with stride lengths that were no statistically different in no-mat and mat conditions, $p < 0.152$.

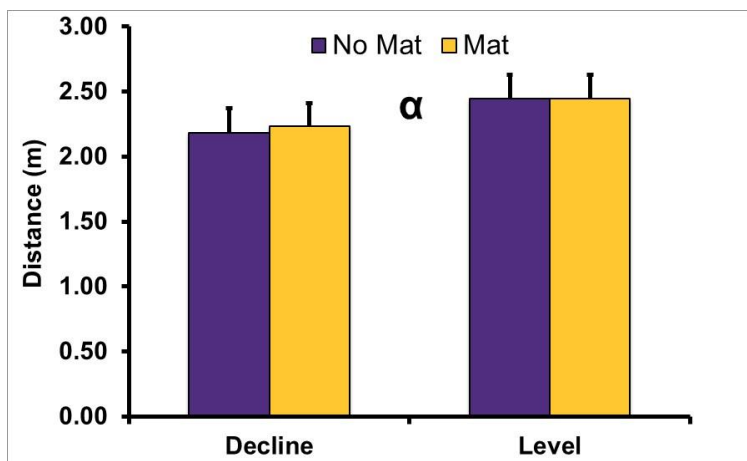Figure 8. Means, SDs for stride length. α: Decline < Level, F=22.1, p<0.001.

Running velocity

The interaction effect was not significant, p<0.256, see figure 9. Participants ran 10% faster in level compared to decline running, p<0.001 but ran at statistically identical speeds in mat and no-mat conditions, p<0.152.
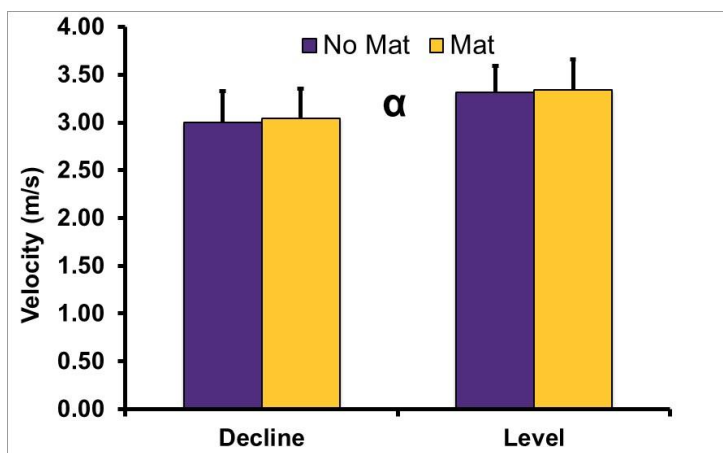
Figure 9. Means, SDs for running velocity. α: Decline < Level, F=23.8, p<0.001.

Magnitude of the vertical GRF impact peak and rate of force application to this peak force

The interaction effect for the magnitude of the vertical GRF impact peak was statistically significant, $p < 0.035$, see figure 10. The vertical impact peak increased 5% from no-mat to mat conditions, $p < 0.049$, in decline running but it decreased 4% from no-mat to mat conditions in level running, $p < 0.010$.
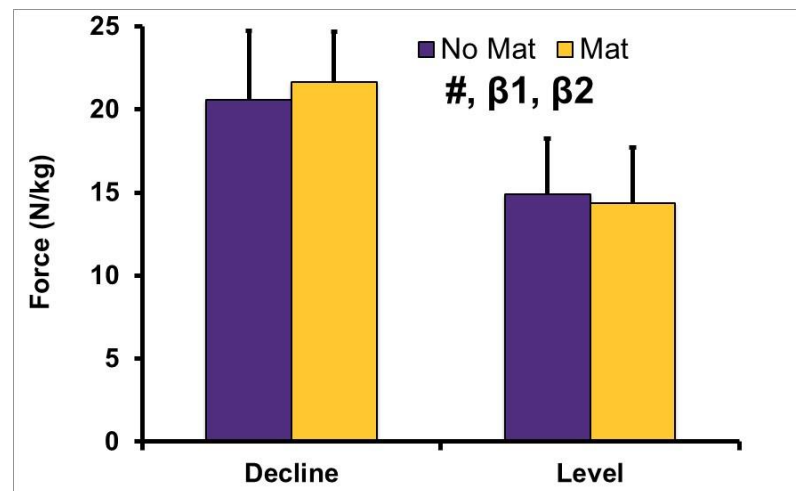
Figure 10. Means, SDs for the magnitude of vertical GRF impact peak #: significant interaction effect, $F = 5.45$, $p < 0.035$; β1: No-mat < Mat in Decline, $p < 0.049$; β2: No-mat > Mat in Level, $p < 0.010$.

The interaction effect for the rate of force application to the vertical GRF impact peak was not significant, $p < 0.230$, see figure 11. The rate of application for this force was 41% higher in decline compared to level running, $p < 0.001$ but was not significantly different between the mat conditions, $p < 0.070$.
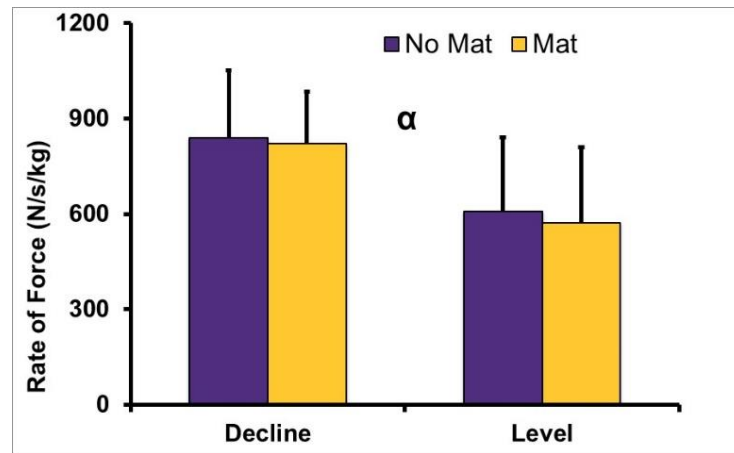
Figure 11. Means, SDs for the rate of force application to the vertical GRF impact peak.

α: Decline > Level, F=31.9, p<0.001.

Magnitude of the maximum active vertical GRF and rate of force application to this peak force

There were no statistically significant differences in the magnitude of the maximum

vertical GRF, see figure 12. The mean, SD values for this variable were 23.4 +/- 2.6 N/kg.
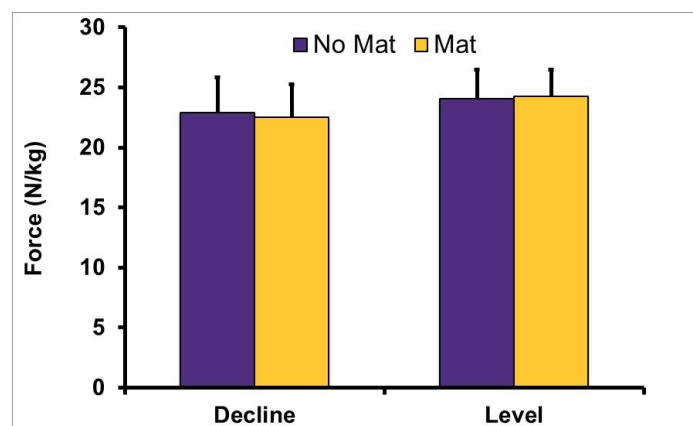
Figure 12. Means, SDs for the maximum vertical GRF

There were no statistically significant differences in the rate of force application to the maximum vertical GRF, see figure 13. The mean, SD values for this variable were 211 +/- 41 N/s/kg. The statistical comparison for running mode had a p-value of 0.053 and may be a type II error dues to our reduced sample size.
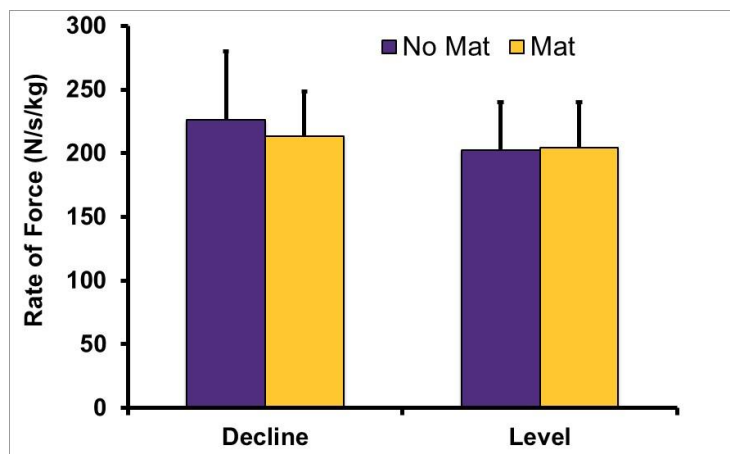
Figure 13. Means, SDs for the rate of force application to the maximum vertical GRF

Magnitude of the maximum braking anteroposterior GRF

The interaction effect was statistically significant, $p < 0.001$, see figure 14. The maximum braking force decreased 57% from no-mat to mat conditions, $p < 0.001$, in decline running but remained unchanged mat to no-mat conditions in level running, $p < 0.228$. With participants running at the 10° decline, the horizontal force vector has much more padding to work with in the x direction when compared to level. Because of this declination angle and the increase

volume of mat to dampen the braking force to attribute to this drastic decrease in braking force in the padded decline setting.
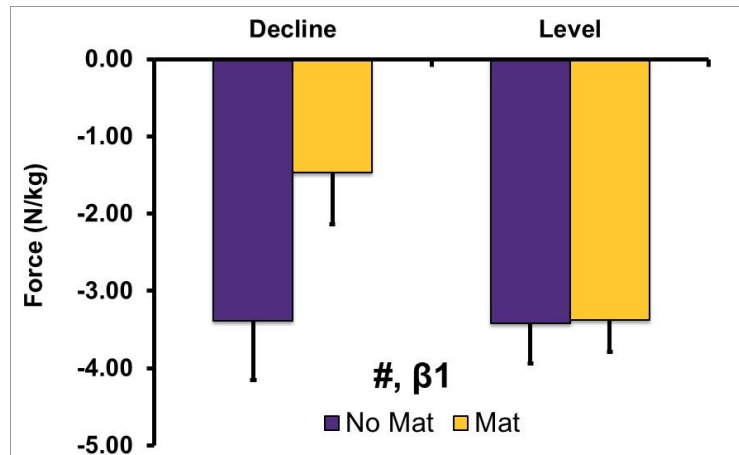
Figure 14. Means, SDs for the maximum braking anteroposterior GRF. #: significant interaction effect, F=238, p<0.001; β1: No-mat > Mat in Decline, p<0.001.

Magnitude of the tibio-femoral compressive force at the time of the vertical GRF impact peak

The interaction effect was not significant, p = 0.268, see figure 15. Participants experienced 10% higher forces in the mat conditions than no mat conditions during running, p<0.031. The comparison between decline and level running was not statistically significant, p<0.202.
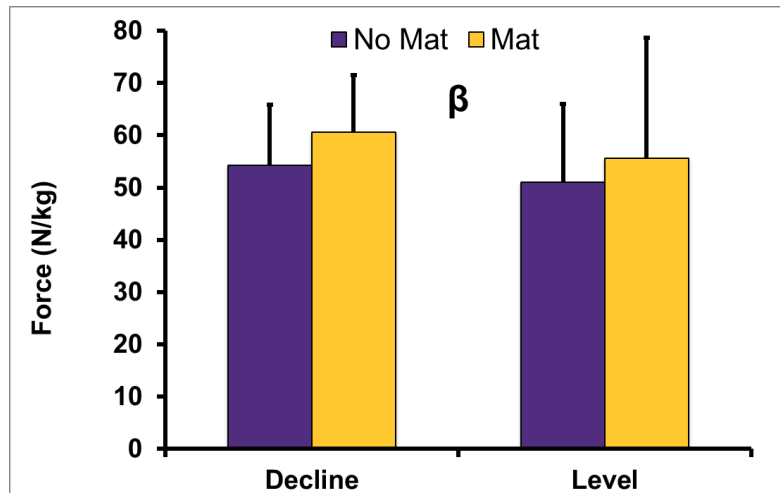
Figure 15. Means, SDs for the magnitude of the tibio-femoral compressive force at vertical GRF

impact β: No-mat < Mat, F=5.66, p<0.031

Rate of the tibio-femoral compressive force at the time of the vertical GRF impact peak

The interaction effect was statistically significant, p<.013, see figure 16. The rate of the

force application increased by 6%, from the no mat to mat condition in the decline setting,

p<0.011 and the difference between surface conditions was not significant for level running,

p<0.216.
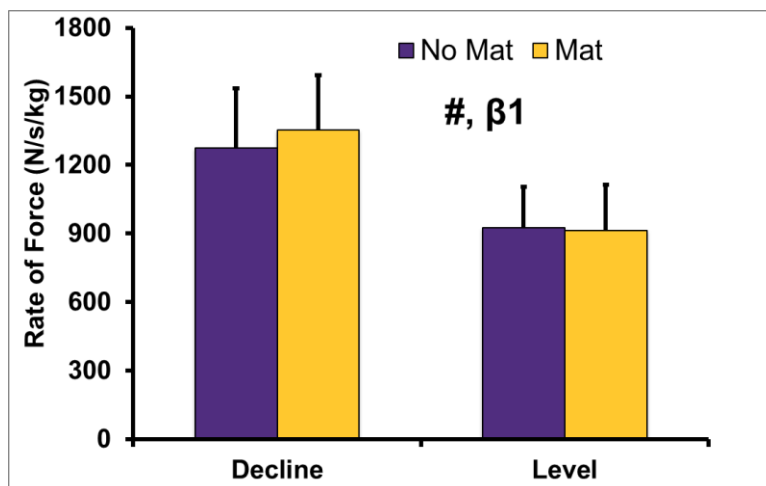
Figure 16. Means, SDs for the rate of tibio-femoral compressive force applied at GRF impact #: significant interaction effect, $F=7.49$, $p<0.013$, β1: No-mat < Mat in decline running, $p<0.011$

Magnitude of the maximum tibio-femoral compressive force

The interaction effect was statistically significant, $p<0.001$, see figure 17. The magnitude of the compressive force decreased by 5% from the no mat to mat condition in the decline setting, $p<0.023$. The magnitude of compressive force increased by 2% from the no-mat to mat condition in level running, $p<0.023$.
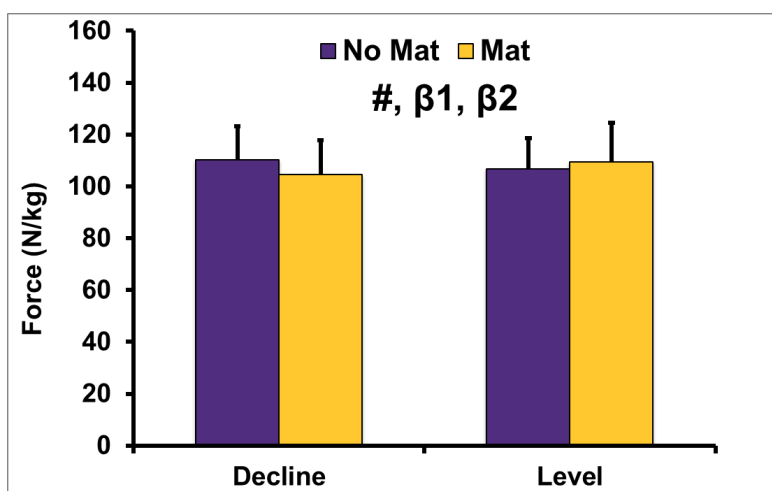
Figure 17. Means, SDs, for the maximum tibio-femoral compressive force #: significant interaction effect, F=12.4, p<0.001, β1: No-mat > Mat in decline running, p<0.023, β2: No-mat < Mat in level running, p<0.023.

Rate of the maximum tibio-femoral compressive force

The interaction effect was found to be not significant, p<0.192, see figure 18. There was no statistical significance between the mat to no-mat conditions, p<0.187. The rate of the forces applied increased by 82% from level to decline running, p<0.001.
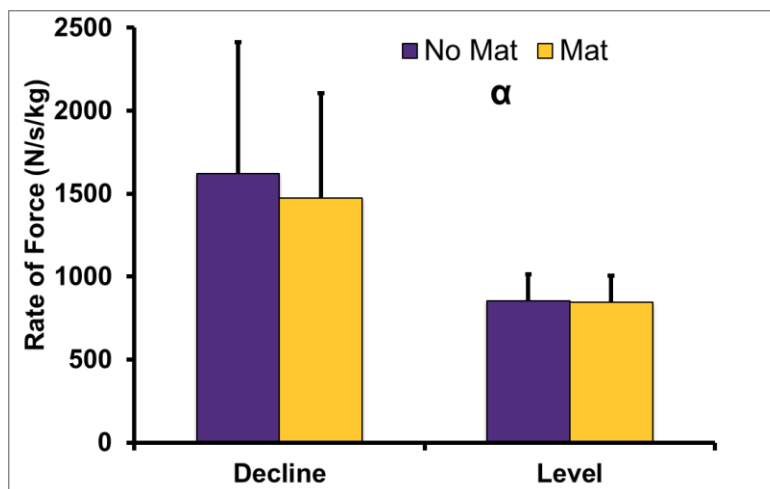
Figure 18. Means, SDs for the rate of maximum compressive force α: decline > level, F=15.3, p<0.001

Magnitude of the maximum tibio-femoral shear force

The interaction effect was statistically significant, p<0.001, see figure 19. The magnitude of the tibio-femoral shear forces decreased by 15% from the no-mat to mat conditions in

decline running, p<0.001. These forces remained statistically unchanged from the mat to no-mat conditions during level running, p<0.155.
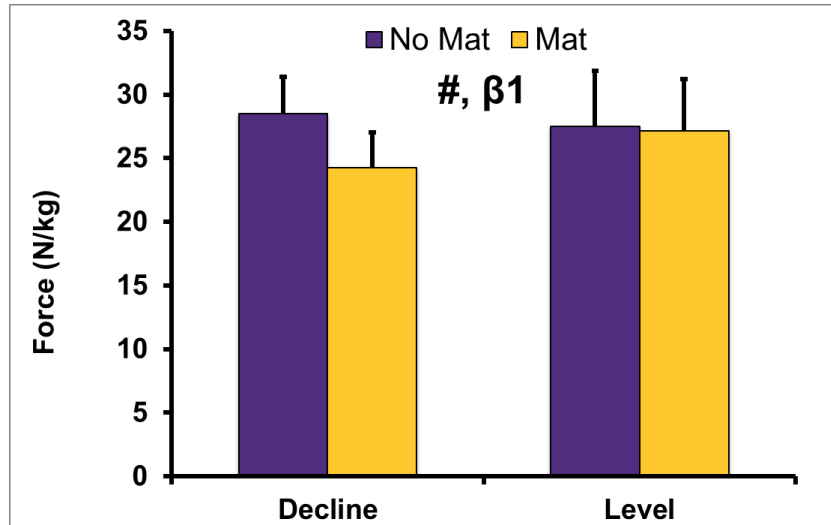
Figure 19. Means, SDs of the maximum tibio-femoral shear force #: significant interaction effect, F=54.0, p<0.001, β1: No-mat > Mat in decline running, p<0.000

Rate of the maximum tibio-femoral shear force

The interaction effect was found to not be significant, p<0.070, see figure 20 below. The rate of the forces applied decreased by 16% from the no-mat to mat conditions during running, p<0.029. There was no statistically significant change in the rate of the force applied between the no-mat and mat condition in level running, p<0.130.
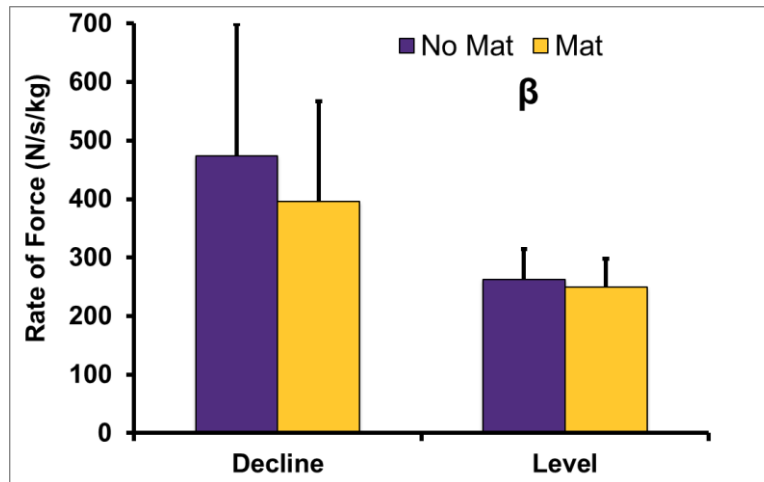
Figure 20. Means, SDs for the rate of tibio-femoral shear force applied β: No-mat > Mat, F=5.85,

p<0.029.

Magnitude of the patello-femoral compressive force at the time of the vertical GRF impact peak

The interaction effect was not statistically significant, p<0.269, see figure 21. There were no statistically significant differences between the no-mat to mat in the combined running modes, p<0.230, as well as between level and declined running, p<0.254. The mean, SD values for this variable were 12.4 +/- 8.2 N/kg.
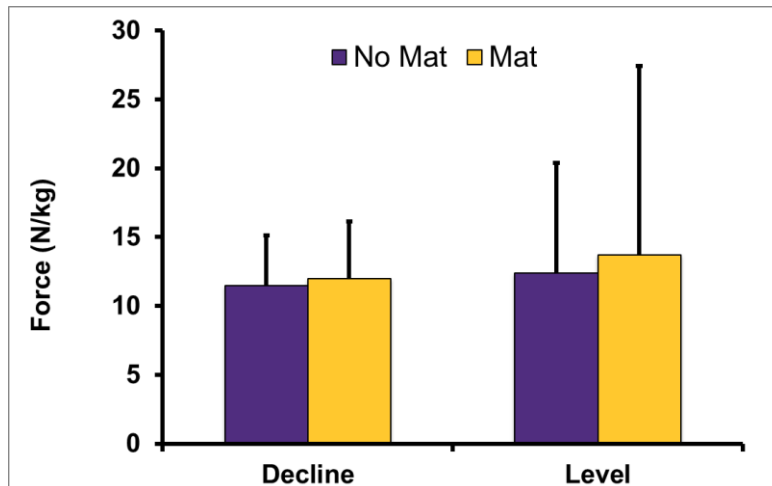
Figure 21. Means, SDs for the magnitude of the patello-femoral compressive force at GRF

impact

Rate of the patello-femoral compressive force at the time of the vertical GRF impact peak

The interaction effect was not statistically significant, $p < 0.246$, see figure 22. There were

no statistically significant differences between the mat and no-mat conditions, $p < 0.270$. The

rate of the forces applied at the time of the GRF impact peak increased by 30% from level to

decline running, $p < 0.011$.
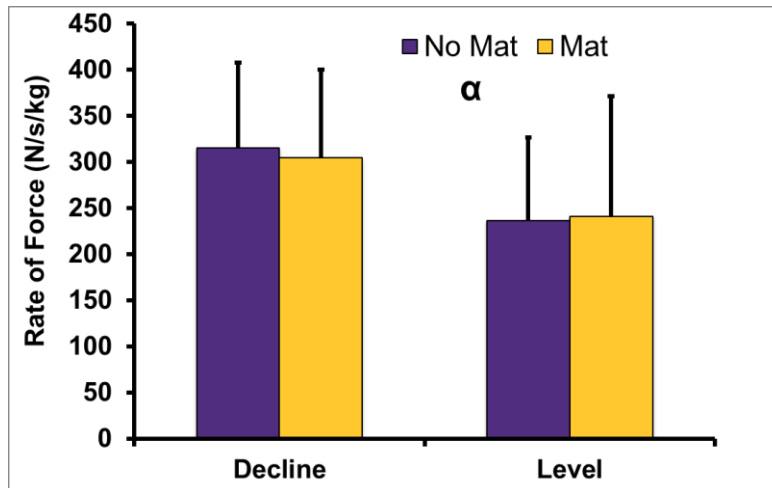
Figure 22. Means, SDs for the rate of patello-femoral force applied at GRF impact peak α:

Decline > Level, F=12.3, p<0.011.

Magnitude of the maximum patello-femoral compressive force

The interaction effect for the magnitude of the maximum patello-femoral compressive force was found to be statistically significant, p<0.001, see figure 23. The magnitude of these forces decreased by 15% from the no-mat to mat condition in decline running, p<0.001. These forces increased by 3% from no-mat to mat conditions in level running. p<0.053 although this may have been a Type II statistical error.
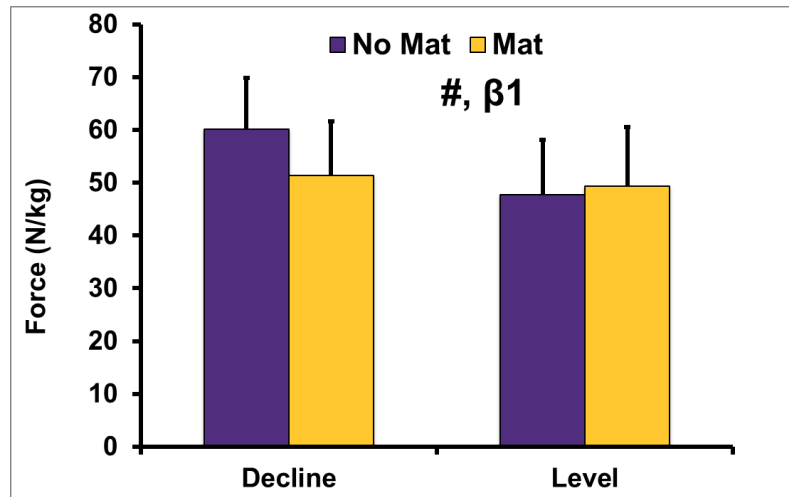
Figure 23. Means, SDs, for the maximum patello-femoral compression #: significant interaction effect, F=25.3, p<0.001, β1: No-mat > Mat in decline running, p<0.001.

Rate of the maximum patello-femoral compressive force

The interaction effect for the rate of the maximal patello-femoral compressive force was not statistically significant, p<0.071, see figure 24. The rate of the forces applied during the stance phase increased by 107% from level to decline running, p<0.001. The statistical test for surface stiffness was not significant, p<0.097.
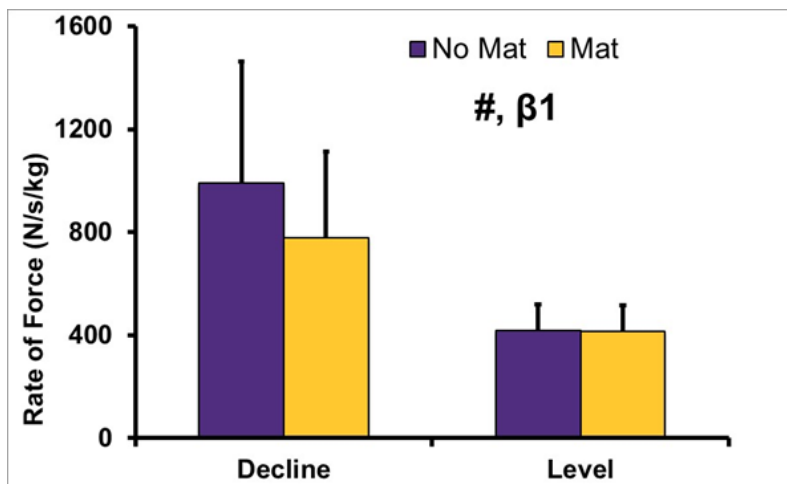
Figure 24. Means, SDs for the rate of maximum patello-femoral force applied #: significant interaction effect, F=4.89, p<0.043, β1: No-mat > Mat in decline running, p<0.028

Knee joint angular stiffness

The interaction effect for the knee joint angular stiffness was significant, p<0.001, see figure 25 below. The angular stiffness decreased by 25% from the no-mat to mat condition in decline running, p<0.000 but was statistically unchanged in level running, p<0.139.
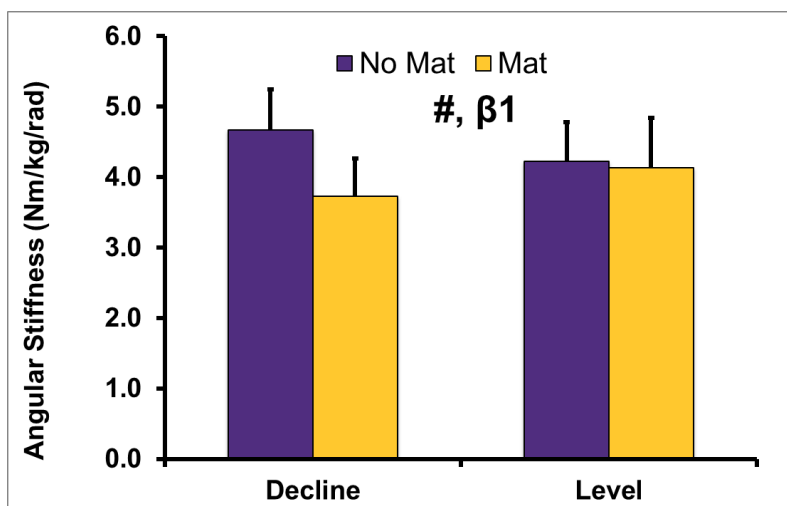
Figure 25. Means, SDs for knee joint stiffness #: significant interaction effect, F=38.6, p<0.001,

β1: No-mat > Mat in decline running, p<0.000.

Summary

These results show how mat and no mat running vary in many ways, ranging from different magnitudes of GRF impact peak to braking anteroposterior GRF. Even within these conditions, there are significant differences that occur between the no-mat and mat conditions. These results show how a softer running surface could have an impact on running mechanics and forces that occur in the body in decline running, but not in level. The following table highlights the overall results of the study. These include a number of interaction effects with surface stiffness being significant in decline running but not as much in level. The overall secondary outcome is in the absence of interaction effect, several force variables were larger in decline vs level.

Summary of Results	No Interaction		Interaction	
	Decline vs Level	No Mat vs Mat	Decline	Level
Kinematics				
Stride Length	Level 10% > Decline			
Running Velocity	Level 10% > Decline			
Kinetics				
Vertical GRF Impact Peak			Mat 5% > No Mat	No Mat 4% > Mat
Rate of Impact Force	Decline 41% > Level			
Maximum Vertical GRF				
Rate of Maximum Vertical GRF				
Maximum Ant/Post GRF			No Mat 57% > Mat	
Tib-Fem Compression at Impact Peak		Mat 10% > No Mat		
Rate of Tib-Fem Compression at Impact Peak			Mat 6% > No Mat	
Maximum Tib-Fem Compression			No Mat 5% > Mat	Mat 2% > No Mat
Rate of Maximum Tib-Fem Compression	Decline 82% > Level			
Maximum Tib-Fem Shear			No Mat 15% > Mat	
Rate of Maximum Tib-Fem Shear		No Mat 16% > Mat		
Pat-Fem Compression at Impact Peak				
Rate of Pat-Fem Compression at Impact Peak	Decline 30% > Level			
Maximum Pat-Fem Compression				
Rate of Pat-Fem Compression at Impact Peak	Decline 30% > Level			
Maximum Pat-Fem Compression			No Mat 15% > Mat	
Rate of Maximum Pat-Fem Compression			No Mat 22% > Mat	
Knee Joint Angular Stiffness			No Mat 25% > Mat	

Key: Entry in No-Interaction area indicates a non-significant interaction and the main effects of locomotion type and surface condition.

Entry in interaction area indicates a significant interaction effect and the No-Mat vs Mat comparison results in Decline and Level running.

Color Key		Level > Decline	Decline > Level	Mat > No Mat	No Mat > Mat

Table 1. Summary of the results significant differences.

Chapter V. Discussion

The purpose of this study was to compare patello-femoral and tibio-femoral joint contact forces while running on two different surfaces of varying stiffnesses in decline and level running. It was hypothesized that a more compliant ground surface would result in a reduced magnitude of tibio-femoral and patello-femoral forces during downhill running due to the decreased impact forces that would occur in the altered decline run when compared to downhill running on harder surfaces. The following chapter is divided into the following sections: 1) Development of the Hypothesis, 2) Validation of the Knee Joint Musculoskeletal Model, 3) Discussion of the Results, 4) Limitations, 5) Future Directions, and 6) Conclusions.

Development of the Hypothesis

Human locomotion is a continuous process that involves the foot going through a swing and stance phase for multiple cycles. During the swing phase, the leg muscles prepare the foot and limb for contact with the ground as the leg experiences higher net joint torques. As the runner enters the stance phase, the foot makes contact with the ground as the ground produces and equally opposing reaction force on the foot. These opposing forces sends a shock up the limb as the muscles, tendons, and ligaments of the leg absorb and redistribute this shock from initial impact. In order to dampen these forces, the joints of the leg flex and then the muscles reposition the joints so that they can generate enough power to push off the ground and begin the swing phase of the next step. This process is repeated for every step, resulting in

a large volume of forces being applied and absorbed by the body during running, resulting in a common cause of injury in runners[49].

With these opposing ground reaction forces being a commonality in injuries experienced by runners, reducing these forces is commonly investigated in running studies[9,14,24,25,36,50,51]. One particular aspect of these ground reaction forces is the magnitude of the vertical ground reaction force, and the rate at which this force is applied. Some strategies for runners manipulating this magnitude and rate is by altering the biomechanics of their gait cycle. This can be done in various ways such as altering foot strike pattern, running velocity, stride length, stride velocity, shoe midsole, running surface stiffness, and incline or decline level[1,9,11,52]. Each of these methods have been shown to alter the vertical ground reaction forces experienced by the body in different ways. It has been shown that a forefoot strike pattern results in a sinusoidal vertical GRF curve while a heel-strike pattern results in an initial impact peak followed by a similar sinusoidal pattern. Different midsoles in the shoe can have an effect on the magnitude and rate at which the vertical GRF is applied to the body as well. Softer midsoles have been shown to produce greater vertical impact peaks than those that are harder[1,25,53]. Whether it's the midsole of the shoe or the surface run on, the leg modulates its stiffness to accommodate the running conditions, demonstrating the complexity of human locomotion and the injuries associated with it.

During running, leg stiffness acts in a similar way to a spring-mass system where the leg is a linear spring acting on the ground[26,28]. With the foot making contact with the ground, the leg compresses like spring as the opposing vertical GRF is applied. This compression translates into a recoil force as the leg decompresses in the last half of the stance phase, allowing for a

stronger propulsion into the next step. Joints in the leg can affect the overall level of stiffness as a more flexed joint would reduce the stiffness of the leg. Increasing flexion in both the hip and ankles have been shown to reduce the magnitude of the vertical GRFs put on the body while also decreasing overall stiffness of the leg[28]. Although there is a reduction in the vertical GRF, the reduction in leg stiffness results in more work being required by the muscles to propel the leg to the next stride. Although the hip and ankle can affect leg stiffness, the knee joint has the greatest impact on overall stiffness and energy usage of the leg throughout the stance phase[12]. While manipulating joints to decrease leg stiffness can be done, the stiffness of the surface run on can also change the overall stiffness of the leg. Many studies have found that decreasing the surface stiffness can lead to an increase in leg stiffness to maintain a constant level of stiffness between the ground and the runner[11,24,26,28,36]. The knees role of being the primary force dampener leads to the knee being the most common site for overuse injuries and the question leading from this is how can surface stiffness relate to injury prevention.

While the alteration of surface stiffness can change the patello-femoral and tibio-femoral forces, the grade of decline can have a significant effect on the joint forces as well. The greater the decline, the higher the GRF impact peaks put on the body, leading to higher knee joint forces which could potentially cause higher rates of injury[11,14,54]. Reducing the surface stiffness in decline running might appear to be a solution to reducing injury, but the increase in leg stiffness has been shown increase joint stiffness[24,35]. With the knee having the role of primary force dampener in the leg, manipulations to leg stiffness can have a larger effect on the knee. There is still a lot of uncertainty on how these manipulations in decline running impact the knee and help prevent injury.

Many studies investigate how manipulation to a runner's mechanics can decrease vertical GRFs and how these decreased forces can ultimately reduce the number of injuries. While these manipulations can affect the GRFs put on the body, they can also affect knee joint forces, however these effects are still unclear. Because the knee joint is crucial during running, knowing how the knee copes with different ground stiffnesses could allow for improvements in injury prevention in the future. This study therefore investigated GRFs and knee joint loads during running on two surface stiffnesses in level and decline running.

Validation of the Knee Joint Musculoskeletal Model

Our hypothesis focused on forces that occur in the knee as individuals ran over a soft and hard surface in level and decline settings. The musculoskeletal knee joint model used to calculate tibio-femoral and patello-femoral forces was originally created and published by DeVita and Hortobagyi in 2001 and Messier et al. 2011, and has been validated in multiple knee focused studies since[16,38,41,48,55]. The same musculoskeletal knee joint model has been used to calculate tbio-femoral shear and compression forces and has been validated in multiple studies since[26,32,42,56,57]. Joint reaction forces and torques were used to model muscles of the legs and were derived using inverse dynamics, allowing for further calculations to define tibio-femoral and patello-femoral contact forces, as described above in Chapter III Methods.

While we cannot truly validate the current knee joint force predictions, we can compare our findings to previous studies investigating loads and forces in the knee using procedures similar to the ones described in Chapter III. Bergmann, 2014 used instrumented knee implants

to accurately measure the forces that occur during running and found that tibio-femoral compressive forces measured around 5,000 N for jogging. These results were similar to the results obtained in this study where tibio-femoral compression averaged about 7,500 N for runners in level running. While the results in this study were over 2,500 N higher on average than Bergmann 2014, we predict this difference in magnitudes is due to higher running velocity since running velocity is directly related to joint laods[8,14,58]. Furthermore, Rooney and Derrick measured tibio-femoral compressive forces in forefoot and heal strike runners and found an average of 12.89 times their bodyweight[53]. These results are similar to this study where the average tibio-femoral compression in level running was found to be 11.02 x BW. The higher results found by Rooney and Derrick is likely due to the higher running velocity used in this study. Across multiple studies that investigated knee contact forces, the tibio-femoral compression ranged from 6 to 13 BW, which contains the result of 11.02 BW observed in this study[8,26,41,42,53].

Tibio-femoral shear forces have been calculated using similar calculations to the ones described in Chapter III in previous studies but is less common than tibio-femoral compression and patello-femoral compression. Findings from four previous studies found a range of 1.6 – 3.98 BW for tibio-femoral shear force[26,59–61]. This study calculated an average tibio-femoral shear force of 2.79 BW, which falls within the range of values found in previous studies.

Patello-femoral compression has been investigated in multiple previous running studies using similar calculations to the ones described in Chapter III. Patello-femoral compression averaged 4.94 BW in the level conditions in this study and was averaged at 4.5 BW in four other studies[16,21,26,32]. Patello-femoral compression values used to compare here are the level, control

values provided so that the experimental variables used in these studies are not taken into account. Although some values found in the literature are slightly lower than the 4.94 BW found in this study, running speed or running surface (treadmill or over ground) could play a role in these slight differences. Despite not being able to directly test eh validity of our model, the similarity between our data and published results give us confidence that our data are reasonable.

Discussion of the Results

Running Kinematics

Participants had an 11% longer stride during the level running conditions when compared to decline. This difference between conditions is expected since the running velocity is also greater in the level conditions. It is common for runners speed to increase as their step length increase[58]. This can also relate back to runners instinctually reducing their stride and speed during decline running to reduce the magnitude of forces put on their body. Because reducing stride length has been shown to decrease the magnitude of ground reaction forces put on the body, runners might have habitually developed a shortened stride for decline running[8,11,54,62].

Running velocity was intended to remain the same between the no-mat and mat conditions. Although infrared timing gates were intended to be used to control for all participants, the timing gates available did not work properly for everyone. When the timing gates were not used to control for velocity, the average velocity of a posterior hip marker was

calculated in Qualisys after the trial was completed. The hip markers were used to calculate velocity because they go through the same motions as the persons torso and remained relatively constant. Despite controlling for participants running speed, a 10% increase in running speed was recorded for level running when compared to decline running. One possibility for this difference could be because participants were instructed to focus on performing a controlled run for the decline setting. With the ramp being 15m long, participants might not have focused on performing a clean and controlled run, instead they may allow gravity to pull them down the length of the ramp, resulting in an accelerated, uncontrolled run. This possible outcome was avoided by instructing participants to focus on controlling their run as if they were running down a long decent. By emphasizing the control of the run, participants might have run at a slower pace.

Another possibility for a slower velocity in the decline settings could be because participants have developed a habit of running at a slower pace while running downhill to reduce the forces put on their body. By reducing the forces put on their body during downhill running, they are ultimately reducing their risk for injury[11]. As they build up this habit over their runs, it would be expected to translate into this study where participants are mimicking running down a slope. Although there was a 10% difference in running velocity between level and decline running, there were no significant differences between the no-mat and mat conditions in either setting. This lack of significance is ideal for this study because although we have different speeds occurring between the level and decline conditions, our goal was to keep running velocity as consistent as possible in every condition. By having no differences from no-

mat to mat, variables discussed later in results could be different because of the running surface, and not the running velocity.

Ground Reaction Forces

The vertical GRF impact peak increased by 5% in the mat condition for decline running when compared to the no mat condition. This was an interesting finding for this study as surface stiffness has been shown to remain constant despite the a softer running surface[28,63]. Previous literature has not had any significant findings with different surfaces in decline running and although we have a higher GRF impact peak, this is not caused by a stiffer leg because our findings found the leg to be less stiff in the decline mat condition than the no mat condition. Previous research by Ferris et al. (1998) conducted their research in the level setting, but not in a decline setting, making it difficult to give a direct comparison between eh two studies, although our level running data is in agreement. With decline running being biomechanically different from level running, it can be difficult to determine what caused this 5% increase in GRF impact peak. One plausible cause for this increase in the decline setting could be the 1% increase in step length that is seen in the mat condition in decline running. Previous studies have instigated how step length can affect GRFs and they found that an increase step length results in an increased GRF[8,23,52]. These findings would suggest that the 5% increase in vertical GRF in the matted, decline condition could be attributed to the 1% increase in step length that occurred during this condition.

While the decline mat condition had a 5% increase, the level mat condition has a 4% decrease in GRF impact peak. Instead of attributing the increased GRF to leg stiffness, the increased GRF could be a result of increased muscle activation in the mat conditions. Because the softer surface is less stable than the hard surface, the leg joints could create a greater muscle force to stabilize the joints. While the magnitude of the forces varied no mat and mat conditions slightly, the rate of application of these forces was 41% higher in decline compared to level. This drastic difference is likely due the increased GRF impact peak in decline running while the stride velocity was relatively similar across all conditions. This would cause the slope to be relatively similar but because the decline setting had a much greater magnitude at GRF impact peak, the rate is much higher because it must reach that higher magnitude within the same window of time as the level.

There were no statistically significant differences in the maximum GRF across all the conditions. The maximum GRF is commonly seen in the middle of the stance phase where the runner is pushing off of the ground to propel them into their next step, also called the propulsion phase. There were not any significant differences in the propulsion phase across all conditions. The same is also seen for the rate to the maximum GRF. This is expected because the maximum GRF is relatively similar for all conditions and the stride rate is similar for all conditions. Because neither of these are significantly different, we would not expect the rate to the maximum GRF to be significantly different for any of the conditions. These findings are in agreement with previous literature where the propulsion phase showed similar GRFs despite the level of decline[6,11,64].

The maximum braking anteroposterior GRF had a significant interaction (p< 0.001) where the magnitude decreased 57% from no-mat to mat condition in decline running. The level condition did not differ between the no-mat and mat condition and was found to have similar magnitudes as the no-mat condition in decline running. This lack of significance between level and decline was unexpected as previous literature has found braking force to be exponentially greater in decline running as the runner works more to stop themselves from losing control while running[11,63]. While only 6.2 cm thick in the level condition, the addition of the decline means when acting straight down on the mat, there is actually more mat available to provide cushion than the level condition with the mats. Because this force is the braking force, it is acting in the opposite direction that the participant is running. Due to this force acting in the opposite direction, there is much more padding for the runner to utilize in lowering their braking GRF. This drastic decrease in braking GRF magnitude could play a vital role in protecting runners against the harsh and rapid impacts that occur in decline running. As this force acts in the opposite direction of the runner, a shear force could occur on the runner. The reduced braking force seen could help prevent injury associated with tibio-femoral shear forces as these put a large amount of strain on the knee. With this decrease in braking forces, we also see a decrease in the tibio-femoral shear forces in the decline mat condition.

Knee Contact Forces

We hypothesized that a more complaint running surface would result in more interactions in the decline setting than level. If there was no interaction between surface conditions in decline running, interactions between the running surface and declination angle

were observed. Out of the twenty-one variables reviewed, eight of them showed significant interactions between running surface in the decline setting and five of them were knee contact forces.

Tibio-femoral compression at GRF impact peak had a 10% higher magnitude in mat conditions for both level and decline running. This main effect could be attributed to an increase in muscle forces surrounding joints to provide more stability on the less stable surface. This increase in compressive forces could allow for a stronger propulsion into the next step as a softer surface will dissipate more of the forces that act against the leg, helping push it forward. Because of this dispersion of reacting forces, the leg could allow for more compression so that it can have a greater recoil, helping the runner continue to run at the same pace and accommodate for the softer surface. While there was a higher incidence of tibio-femoral compression at GRF impact peak in the mat condition, a reduced magnitude of forces was seen for maximum tibio-femoral compression and tibio-femoral shear force in the matted, decline condition. Tibio-femoral shear force decreased the most with an 18% reduction in forces from no-mat to mat in the decline setting. This substantial reduction could be a result of the decreased anteroposterior braking force seen in this condition. With the mats offering more cushion in the anteroposterior direction in the decline setting, it is logical that the shear force would decrease as a result. This reduction could help runners reduce their risk of injury while running downhill as this reduction is significant. A 17% reduction in maximum patello-femoral compression during the stance phase was also seen in the mat condition of decline running. While we see a decrease in the mat condition in decline running, it is important to note that the patello-femoral compression increased by 52% in the decline setting when compared to level.

This result is similar with the increase that was observed by Ho et al. where they found a 35% increase in patello-femoral compression during decline running[65]. This could possibly be due to the runner falling a greater distance in the decline condition than the level. This increased fall distance can play a role in the increased impact that occurs in decline running, which is why it is important to try and reduce these forces. Previous literature has shown decline running to have higher vertical GRFs as well as significant increase in patello-femoral compression which could lead to injury. This reduction in these knee contact forces in decline running could play a role in preventing injury in runners since decline running has been shown to have an increase in forces when compared to level[11,65].

Overall, we observed an increase in the rate of the force application in the decline setting when compared to level. This observation was expected as gravity, and a greater falling distance would cause these forces to be applied much faster. Some force variables are higher in the decline setting as well, so if the runners were running at a relatively similar velocity, a much higher force is applied in about the same time as it would in the level condition. This would then be seen by showing a much higher rate of application because although the time frame of the application is similar, the decline setting has a much higher magnitude of forces to reach within that same time frame. While the increased rate of force application was expected in decline running, there was some observed reduction in force application within the decline setting. The mat condition showed to have decrease in rate of force application for maximum patello-femoral compression, tibio-femoral shear force, and tibio-femoral compression. This reduction in rate of application corresponds with the magnitude of the forces experienced for these variables because all of these listed has a decreased magnitude of these forces as well.

This finding suggests that the rate of the application is less because the forces are applied to a lesser magnitude within the same timeframe as the other condition. This suggests that rate of force application is not a direct change of ground surface but corresponds with the magnitude of the forces experienced. This would mean that to help runners reduce knee contact forces and decrease the rate of the forces applied, they would ultimately use a softer surface in decline running to reduce the magnitude of the forces which would also result in decrease in the rate of application.

Knee joint angular stiffness

Angular stiffness decreased by 25% from no-mat to mat conditions in decline running. Previous literature has shown that decrease surface compliance results in a stiffer leg to maintain a constant level of stiffness between the ground and the runner[28,60]. While this data demonstrates an increase in leg stiffness as the surface stiffness decreases in level, our data shows the opposite effect occurring in decline running. Decline running in the two conditions had the same amount of joint flexion but decline running has a 20% lower joint torque and therefore, less knee joint angular stiffness. Data not reported above, but both conditions flexed the same amount but with a smaller load there was more compliance.

Limitations

Participants for this study were recruited and it was assumed that these individuals truthfully filled out the questionnaire regarding their health and running history. This study also tested recreational runners who ran between 5-30 miles per week. This lack of running

experience could cause participants to have a less precise running form than experienced runners. This recruitment of recreational runners was primarily because students of East Carolina University were more willing to participate than student athletes who were more experienced runners. The area of recruitment also restricted the availability of high-level runners who might be more experienced than those tested.

The lab was limited in the length of the running track used for this study as it was only 15 meters in length. The ramp used also limited our testing procedures as the ramp only tested at one angle of 10 degrees. This constant decline angle did not allow us to observe how the variables measured changed with the decline angle. The length of the ramp also restricted the runner's ability to adjust to the decline before hitting the force plate. Participants were only able to perform one step before their designated foot made contact with the force plate. Perhaps a longer decent could have allowed participants to find their natural rhythm before hitting the force plate in this study.

Testing procedures also only tested at one speed in both level and decline settings. Incorporating more than one speed could show differences that occur at different speeds between the conditions. We were also restricted on surface stiffness based on the thickness of the mats used. We were also only able to use the ramp and floors stiffness as the stiff condition and were not able to have more than one stiff condition. Having more than one soft and hard surface could help further understand how surface stiffness plays a role in the variables tested.

Conclusions

This study hypothesized that a more compliant running surface would result in a reduced magnitude of knee contact forces and that this effect would be larger in decline running. It was observed that the mat was generally effective in reducing loads that occurred in the knee during decline running. While there was some reduction in forces in the level mat condition, the mat proved most effective at reducing forces in the decline setting. This suggest that a softened surface might be more important for reducing forces in decline running rather than level. Decline running also showed an increase in magnitude of vertical GRF impact peak, and tibio-femoral compressive force at GRF impact peak. This demonstrated just how impactful decline running can be as these loads create unwanted loading in the knee. These three increased variables were also seen with level having a slightly longer stride length and faster running velocity. As both of these variables have been shown to increase forces that occur during running, we could infer that the forces seen in decline running would have increased even more if running velocity and stride length were identical in this study. This study showed how decreasing the surface stiffness in decline running can cause a reduction in forces that occur in the knee during running[11,14]. As it has been shown in this study and previous studies, decline running can cause an increase in GRFs, leading to an increase in knee contact forces which can be harmful to the knee. Because decline running is so impactful, reducing the knee contact forces could prove beneficial at reducing runners' risk of injury. This study found that decreasing surface stiffness in decline running leads to a general reduction in knee contact forces, among other variables, potentially leading to a reduction in overuse injuries experienced by runners.

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EAST CAROLINA UNIVERSITY
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Notification of Initial Approval: Expedited

From:	Biomedical IRB
To:	Paul DeVita
CC:	
	Paul DeVita
Date:	7/26/2019
Re:	UMCIRB 18-002877
	Knee Forces In Level & Non-Level Running

I am pleased to inform you that your Expedited Application was approved. Approval of the study and any consent form(s) occurred on 7/25/2019. The research study is eligible for review under expedited category # 4,6,7. The Chairperson (or designee) deemed this study no more than minimal risk.

Changes to this approved research may not be initiated without UMCIRB review except when necessary to eliminate an apparent immediate hazard to the participant. All unanticipated problems involving risks to participants and others must be promptly reported to the UMCIRB. The investigator must submit a Final Report application to the UMCIRB prior to the Expected End Date provided in the IRB application. If the study is not completed by this date, an Amendment will need to be submitted to extend the Expected End Date. The Investigator must adhere to all reporting requirements for this study.

Approved consent documents with the IRB approval date stamped on the document should be used to consent participants (consent documents with the IRB approval date stamp are found under the Documents tab in the study workspace).

The approval includes the following items:

Name	Description
Knee Forces - Level and Non-level Running - Informed Consent R1 - Spring 2019 (1).docx	Consent Forms
Knee Forces - Level and Non-level Running - Initial Health Survey - Spring 2019.docx	Surveys and Questionnaires
Knee Forces - Level and Non-level Running - Study Protocol R1 - Spring 2019.docx	Study Protocol or Grant Application
Subject Recruitment Announcement R1.docx	Recruitment Documents/Scripts
Surface Running Study Flyer 2 R1.docx	Recruitment Documents/Scripts
Surface Running Study Flyer-Joey.pub	Recruitment Documents/Scripts

The Chairperson (or designee) does not have a potential for conflict of interest on this study.

Informed Consent to Participate in Research
Information to consider before taking part in research
that has no more than minimal risk.

Title of Research Study: Knee Forces In Level & Non-Level Running
Principal Investigator: Paul DeVita
Institution/Department or Division: Department of Kinesiology
Address: 332 Ward Sports Medicine Building
Telephone #: (252) 737-4616

Researchers at East Carolina University (ECU) study problems in society, health problems, environmental problems, behavior problems and the human condition. Our goal is to try to find ways to improve the lives of me and others. To do this, we need the help of volunteers who are willing to take part in research.

Why is this research being done?
The purpose of this research is to compare knee joint forces while running downhill, level, and uphill and on soft and hard surfaces. This research will increase our knowledge about running injuries and possibly identify whether softer running surfaces provide significant force saving benefits.

Why am I being invited to take part in this research?
You meet the inclusion criteria and have no apparent contraindication to participating in the study. Inclusion criteria are 18-25 years old, experienced runner, non-smoker, healthy, free of skeletal, nervous, muscular, and psychological impairments, and a BMI below 28.

Are there reasons I should not take part in this research?
You understand you should not volunteer for this study if I are a smoker, under 18 years of age or over the age of 25, have suffered a serious injury to my legs or back, had or have a medical condition (for example diabetes or asthma), surgery on my legs, take medications that cause dizziness, or have any kind of heart condition.

What other choices do I have if I do not take part in this research?
You can choose not to participate.

Where is the research going to take place and how long will it last?
The research procedures will be conducted in Biomechanics Laboratory, Room 332, Ward Sports Medicine Building at East Carolina University, Greenville, North Carolina. The research will take place in two sessions lasting approximately 1-2 hours.

What will I be asked to do?
You will visit the lab on two different days. On the first day, you will be asked to complete a short health survey to determine eligibility as well as read and sign this consent form. You can ask any questions you think of.

you will wear your own running shoes. You will then run on the different sloped surfaces without and with the padded mat. This may take about 20 minutes and is done so that you become more comfortable with the testing area.

On the second, data collection day, you will be measured for height and weight. You will wear my own running shoes and wear tight fitting clothes provided by the researchers during the running trials. Once you put on the tight fitting clothes, the researchers will place small reflective balls for motion capture purposes on my hips, right thigh, right knee, right leg, and right foot. You will also have small muscle recording electrodes placed on my thigh and calf muscles. You will have several minutes to run on the lab runway and become more comfortable with the testing instruments and different surface conditions. You will then perform three sets of running trials on hard, and soft running surfaces at each incline. You will perform about five to seven trials at each speed with a trial being one run across the room.

What possible harms or discomforts might I experience if I take part in the research?
There are no documented risks and side effects associated with the marker placement. You might experience some leg muscle soreness or joint pain the day after the running trials but this is typically quite rare. If any muscle soreness or joint pain is experienced, the principle investigator would advise the participant to rest.

What are the possible benefits I may experience from taking part in this research?
This research will be used to explore running injuries and determine force saving effects of running on softer surfaces for the knee joint. There may be no personal benefit from my participation, but the information gained by doing this research may help others in the future.

Will I be paid for taking part in this research?
You will not be compensated for my time and participation.

What will it cost me to take part in this research?
There will be no cost to you in order to participate in this research.

Who will know that I took part in this research and learn personal information about me?
To do this research, ECU and the people listed below may know that you took part in this research and may see information about me: Joey Casadonte, the sub-investigator, and Paul DeVita, the primary investigator. Also, the University & Medical Center Institutional Review Board (UMCIRB) and its staff have responsibility for overseeing my welfare during this research and may need to see research records that identify me.

How will we keep the information we collect about me secure? How long will we keep it?
Data files will be kept for 5 years after the study is completed. The investigators will keep my personal data in strict confidence by having my data coded. Instead of my name, you will be identified in the data records with an identity number. Your name and code number will not be identified in any subsequent report or publication. The main investigator and sub-investigator will be the only people who know the code associated with my name and this code as well as my data will be kept in strict confidence. The computer file that matches my name with the ID number will be encrypted and the main investigators will be the only staff that knows the password to this file. The data will be used for research purposes.

What if I decide I do not want to continue in this research?
You may stop my participation at any time during the study. There will be no penalty for withdrawing from the study.

Who should I contact if I have questions?
The people conducting this study will be available to answer any questions concerning this research, now or in the future. You may contact the Principal Investigator Paul DeVita, or the sub-investigator Joey Casadonte at (252) 737-4616 (Monday-Friday 9am-5pm).

Is there anything else I should know?
Your information or biospecimens collected as part of the research, even if identifiers are removed, will not be used or distributed for future studies.

Will I receive anything for the use of my private identifiable information or identifiable biospecimens?
Your information or biospecimens collected as part of the research, even if identifiers are removed, will not be used or distributed for future studies.

Will my identifiable biospecimen be used for whole genome sequencing?
Whole genome sequencing is the process of determining the complete DNA sequence of an individual at a single time. However, further analysis must usually be performed to provide any biological or medical meaning of this sequence. For this research, whole genome sequencing [choose one: will/will not] occur.

If you have questions about my rights as someone taking part in research, you may call the Office of Research Integrity & Compliance (ORIC) at phone number 252-744-2914 (days, 8:00 am-5:00 pm). If you would like to report a complaint or concern about this research study, you may call the Director of the ORIC, at 252-744-1971.

I have decided I want to take part in this research. What should I do now?
The person obtaining informed consent will ask me to read the following and if you agree, you should sign this form:

- You have read (or had read to me) all of the above information.
- You have had an opportunity to ask questions about things in this research you did not understand and have received satisfactory answers.
  - You know that you can stop taking part in this study at any time.
  - By signing this informed consent form, you are not giving up any of your rights.
  - You have been given a copy of this consent document, and it is yours to keep.

_____ _____
Participant's Name (PRINT)   **Signature**   **Date**

Person Obtaining Informed Consent: I have conducted the initial informed consent process. I have orally reviewed the contents of the consent document with the person who has signed above and answered all of the person's questions about the research.

Joey Casadonte
_____
Person Obtaining Consent (PRINT)   **Signature**   **Date**