KNEE JOINT FORCES IN RELATION TO GROUND SURFACE STIFFNESS DURING RUNNING

By

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Abstract: Running shoes and surfaces have been developed to help enhance running efficiency and to reduce ground impacts by altering the total surface stiffness. However, to maintain running mechanics, an individual will increase leg joint stiffness while running across a more compliant ground surface and show an inverse effect when running across harder surfaces. Increasing leg stiffness causes landing impact forces to increase and may counteract the softer surface in terms of knee joint contact forces. Since the knee is an essential determinant for reducing impact forces and a primary site for changing leg stiffness, knowing more about knee joint forces while running on surfaces with different stiffnesses can be beneficial in developing injury prevention programs. It is our objective to determine the effect of surface stiffness on knee joint contact forces during running.

Seventeen healthy recreational heel strike runners were recruited and ran across a 15m track at a consistent pace (3.46m/s + 5%) on 3 ground conditions (hard floor with embedded force plate and 1 and 2 layers of shock absorbing mat). The study protocol took place over 2 days. On day 1, participants were able to practice running over the various ground conditions at the test speed, on day 2, data were collected. Five successful trials per surface condition were gathered and analyzed with focus being on

the knee joint and knee joint forces through musculoskeletal modeling. Data were statistically compared among surface conditions with a one way ANOVA, using three levels and alpha < 0.05.

Patello-femoral (PF) and tibio-femoral (TF) compressive forces were not significantly different between surface conditions. However, knee joint angular stiffness (P< 0.01), the rate to the vertical ground reaction force (vGRF) impact peak (P = 0.02), anteroposterior breaking force magnitude (P < 0.01), and TF shear force for both force magnitude (P < 0.01) and rate (P = 0.03) to the maximum forces were found statistically different (P < 0.05). All variables that were found to be significantly different decreased as the surface stiffness decreased.

Our hypothesis was partially supported for knee joint compressive loads but not for the shear loads. As the participants ran across the increasingly dampened surfaces their knee joint angular stiffness decreased. This is contrary to existing literature that suggests an inverse effect between surface stiffness and leg stiffness, which is closely related to knee joint stiffness. In addition, the rate to the vGRF impact peak and the breaking force magnitude decreased with the surface stiffness. Our data supported the idea that running across differing surface stiffnesses does not statistically alter knee joint compressive forces but can reduce knee joint shear forces. Future research should determine if this strategy is beneficial to a broader range of individuals including those with fore- and mid-foot strike patterns.

Knee Joint Forces in Relation to Ground Surface Stiffness during Running

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KNEE JOINT FORCES WHILE RUNNING ON DIFFERING GROUND SURFACE STIFFNESSES

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Chapter I. INTRODUCTION

People move around in the environment placing various loads on different joints. We can protect our joints by external mechanical tools that help dampen these forces in order to prevent injury. For instance, gymnasts use landing pads to soften force impacts and runners will use specific shoes with varying midsole thicknesses to absorb the shock from the repetitive foot strikes against the ground. As seen in Figure 1, Ferris et al. (1998) shows the representation of the human leg in motion during the support phase of running. The springed leg represents the elastic properties that our muscles, tendons, ligaments, and bones mimic while running and provides a visual of leg stiffness. The spring seen in the ground similarly represents the different ground surface stiffness levels that runners encounter on a daily basis and may provide dampening forces depending on the surface characteristics[31].

Ground stiffness levels that people transverse on a daily basis are variable. Examples of surfaces with high degrees of stiffness include concrete, tar, and running tracks and those with lower degrees of stiffness include grass and sand. The spring stiffness afforded by the lower extremity illustrated in Figure 1 can be modulated to adapt to various surfaces[27,29,30]. Tracks with greater compliance cause the foot to descend into the track and increase foot contact time, cause longer step length, and can affect the runner's athletic performance[49]. Consequently, Ferris et al. (1998) and Farley et al. (1996) suggest compliant tracks for mid- to long-distance running due to the experimental observation of less joint flexion in the lower limb during running on a softer surface. Reduced joint flexion subsequently results in lower joint moments and muscle force estimates needed to maintain ground contact force[8,27,28,31]. This fluctuating degree of lower extremity compliance is known as leg stiffness.

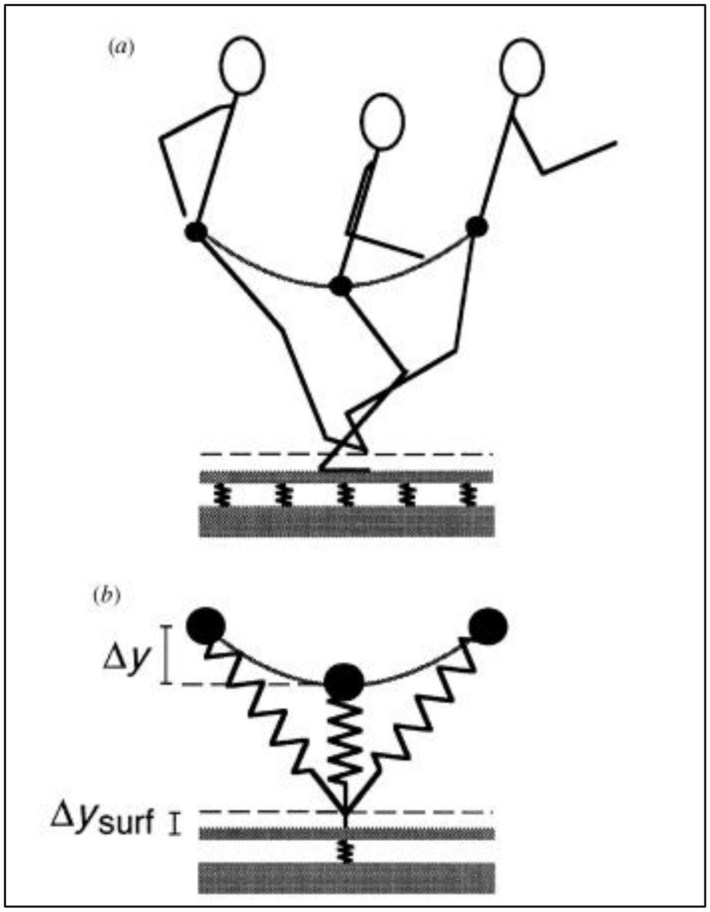

Figure 1: Ferris et al. 1998

Leg stiffness can be modulated by altering the stiffness of contributing joints within the extremity. Joint stiffness is commonly quantified as the change in joint torque relative to the change in angular position throughout the stance phase. Having a lower joint angular excursion relative to the applied torque indicates a stiffer motion than when the joint angular excursion is greater. In this manner, joint stiffness represents how much adjustment is needed at the joint level to determine how much muscle force is required to exert onto the ground contact force[31]. Potentially, this muscle or muscle-tendon activation associated with joint stiffness could indicate why injury rates are higher depending on the stiffness of the knee and ankle joints[27,33].

Harder, or stiffer surfaces, are generally easier to train on because they require less energy output than a more compliant surface that passively absorbs the energy without returning it to the runner[69]. However, the vertical ground reaction force is greater on these surfaces, up to 5 times body weight compared to tracks of more intermediate compliance, due to the lack of dampening properties in the ground and less energy dissipation which may play a role in injury risk especially in the heel strike running population[18,46,49]. Conversely, more compliant surfaces have been shown to enhance running performance; generally these surfaces have some elastic properties which enable the surface to passively store and then return potential energy to the runner[49]. Running on softer surfaces, particularly those that do not provide elastic properties, are associated with higher energy costs and increased heart rates which could result in higher risks of injuries due to fatigue[9,54,69,82].

In response to decreasing impact force and offering protection against environmental loads, shoes have been developed to act as an interface between our feet and the ground. Depending on the runner, the midsole is chosen based on the amount of cushioning needed during training and can potentially differ for races depending on race location. However, when these surfaces are all put together, the ground and the shoe midsole, the question arises whether there is any impact force reductions being elicited on joints while running across less stiff surfaces due to the effects of leg stiffness in relation to vertical GRF.

For example, with the addition of a softer midsole to a softer running surface, leg stiffness will increase to maintain running mechanics. This change in the running surface, to become more compliant, creates a different surface stiffness condition, and the running mechanics change for the runner to maintain running efficiency [29]. It is presumed that increasing the shock absorbing pad would decrease the load on the joints similar to how a gymnast uses mats to absorb energy and reduce forces as the athlete lands. However, as Ferris et al. (1998) has revealed, runners tend to increase leg stiffness on softer surfaces in order to maintain mechanics which may negate the shock-absorbing capacity of the ground interface[31].

The knee joint has been shown to be a primary force dampener while running and is most commonly injured amongst running populations[65,84,88]. Little research investigates the internal forces placed on the knee joint during running bouts however even less investigates the effects of surface stiffness changes to the knee joint. With forces up to 6 times body weight to the patello-femoral joint and 10 times body weight to the tibio-femoral joint, it would be of interest to determine if softer surfaces are biomechanically beneficial while running[71,94,95,98,100].

PURPOSE

The purpose of this study was to compare tibio-femoral and patello-femoral joint contact forces while running on three different surfaces of varying stiffness. The significance of this work is due to the knee being one of the most common sites of running overuse injury[89]. Many studies have investigated manipulating an individual's running mechanics or providing external equipment in order to reduce GRF magnitudes and rates to these magnitudes. Though, knee joint forces and how they are affected by these manipulations have been investigated, little have focused on the effects of ground stiffness changes on internal knee joint forces. Therefore, our long term goal is to determine if ground stiffness modulation has the potential to reduce the likelihood of injury brought about by impact forces or whether its role in knee joint health is negligible[52].

HYPOTHESIS

We hypothesized that varying the ground stiffness would not make a significant difference to the force placed on the knee across different ground stiffness levels due to the joint manipulating itself to maintain the body's natural running mechanics.

DELIMITATIONS

We studied trained recreational runners (10-30 miles/week) with a rear heel strike pattern who ran down a runway over various rubber mat surfaces to represent a variety of running environment stiffnesses. The experienced runner's learned stride allowed for a higher probability to run across varying surfaces while training. Heel strike runners were selected based on the assumption that they would show greater differences in knee joint forces than other foot strike patterns[37,84]. It was also assumed that participants were able to reach the requested 3.46m/s $\pm$ 5% speed within their first three strides before they came in contact with the force plate as well as were able to maintain a consistent velocity through 12m of the 15m track. Participants were instructed to continue past the 10m mark and not look at the floor while they were running to avoid unconsciously slowing down or aiming for the embedded force plate. With the use of timing gates, the runner's speed was able to be tracked and any trials that did not meet the set requirements were discarded. In addition, any speed effects on knee joint forces was able to be controlled. Participants practiced multiple trial runs on the various surfaces before data collection began to be accustomed to the different conditions. Any trials where the participant attempted to reach or target the force plate were not recorded as a successful trial. The purpose of the study was to determine a difference between surfaces as a range of three distinct stiffnesses not to quantify the surface stiffness value. For this reason, we allowed participants to wear their own running shoes for familiarization and to encourage consistency of their natural running mechanics.

LIMITATIONS

This study was limited by the accuracy of the musculoskeletal knee model that was used to calculate the joint patello-femoral and tibio-femoral forces (Figure 12). The model assumes the absence of several knee joint ligaments, no co-contractions by the hip flexors and abductors during stance, as well as a frictionless healthy joint. We were also limited to a 15m track runway for our running trials due to laboratory restrictions.

OPERATIONAL DEFINITIONS

Joint Stiffness: Change in angular position along a specific joint in relation to the joint torque placed on it.

Leg Stiffness: Change in leg spring compression in relation to the peak ground reaction force.

Surface Stiffness: Deformation of the spring depth into the ground as a force is placed on it.

Ground Reaction Force (GRF): Force exerted by the ground on a body in contact with it.

ΔL: Leg Spring Compression: change in length of the leg as the joints flex to absorb the GRF

Δy_{tot}: Vertical Displacement of the participant's center of mass

Δy_{surf} : Displacement of the ground surface as a mass makes contact with it

T_{peak}: Peak torque at the knee joint

$\Delta \theta$: Change in the knee joint angle

MVC: Maximum voluntary contraction measured using EMG signal

Chapter II. REVIEW OF LITERATURE

When investigating running over varying surfaces, it is important to not only understand the basic mechanics of running but also how an individual adjusts him or herself to maintain these running parameters in different running situations or environments. If mechanics are not changed or manipulated, excessive forces may be placed on joints as well as an offset to one's center of balance may be experienced which can lead to an increased risk of injury. A review of the running biomechanics and how they relate to lower extremity stiffness and knee joint loads is below. Following this, joint mechanics, injury components, and joint tissue load through musculoskeletal modeling will be discussed. Emphasis will be placed on the knee joint to develop a framework for changes in the knee loads when running across different ground stiffness levels.

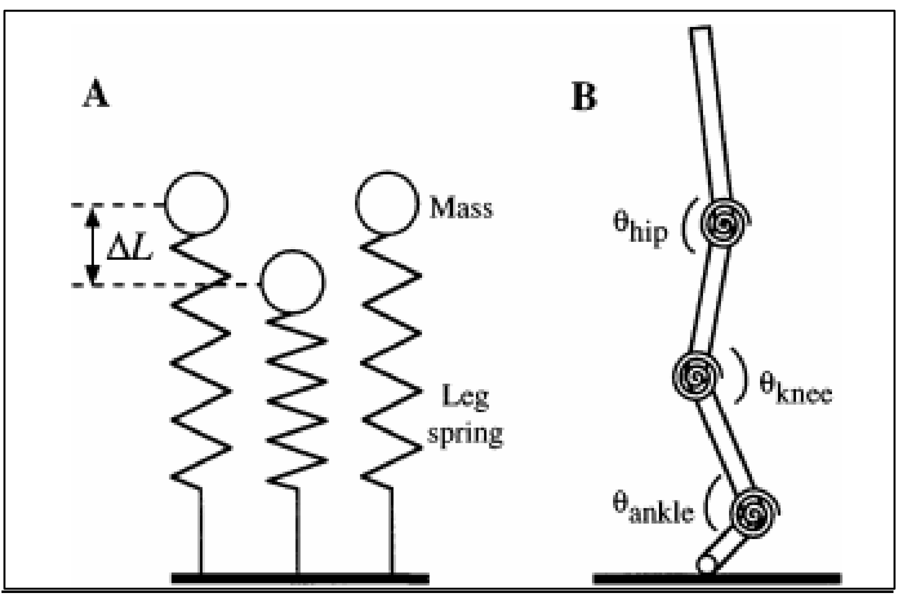

Figure 2: Farley et al. 1998

RUNNING BIOMECHANICS

McMahon et al. (1979) describes running as essentially a series of collisions with the ground[49]. As the foot strikes a surface, the downward vertical momentum within the individual is opposed by the vertical ground reaction force (GRF) which eventually stops the downward momentum and then creates upward momentum. Conceptually, McMahon and Greene (1979) describe running as a mass and spring to denote the dynamic properties of the leg muscles as well as the reflexes used to rebound back up and through the remainder of the stride and into the next (Figure 2 (A))[49]. As an individual runs, the lower extremity functions as the 'spring' and compresses after heel strike with the ground surface. This compressed 'spring' stores strain energy produced by the body tissues and the body's momentum, returning it to the individual when they rebound off the ground, increasing the energy of the runner.

Many studies have investigated different adjustments to a runner's biomechanics to be more effective across various surface conditions particularly focusing on reducing forces placed on the joints and limb. One area of focus has delved into foot strike pattern categorizing the different patterns into heel strike, mid-foot strike, and forefoot strike depending on which part of the foot makes contact with the ground first. Foot strike pattern alters the location of forces and loads placed on the body and can be associated with the type of footwear an individual wears[93]. Around 75% of runners have a heel strike pattern which indicates that their heel makes contact with the ground surface first before following through to toe off[33]. Sites for injury in this foot strike pattern are typically seen around the knee joint area believed to be related to force distribution directly up the bone structure of the lower limb caused by heel contact directly with the ground[17,60].

This impact is seen in a typical GRF curve which shows a sudden passive impact force on the body as the heel contacts the ground. GRF curves for a characteristically mid-foot strike and forefoot strike runner resemble a sinusoidal curve whereas heel strike patterns have an initial impact peak when the heel strikes the surface before continuing through with a similar sinusoidal pattern. This curve, which all foot strike patterns exhibit, is an active force resulting from muscle activation of the quadriceps, hamstrings, and gastrocnemius to propel off the ground.

The impact peak seen in a heel strike runner has been shown to increase with speed[39,42,47]. Step frequency, which is closely related to step length and speed, can also alter force magnitude. A greater stride frequency has also been shown to alter forces, particularly in the knee joint, and has been suggested as a possible effective strategy to reduce patello-femoral pain in runners by around 14% despite the extra steps required to go the same distance[44,97]. Stride frequency is also affected by the ground compliance and is closely connected to the amount of time the foot is in contact with the ground[49]. McMahon and Greene et al. (1979) found that the dampening properties of a more compliant surface (14.4 kNm^{-1} pillow track), compared to a stiffer board track (875 kNm^{-1}), caused the foot to have greater contact times with the ground and hindered the athlete's speed, whereas the stiffer surface showed a higher ability to enhance speed[49]. Ground reaction forces associated with a stiffer surface can reach up to 5 times an individual's body weight so an intermediate ground compliance would provide a more optimal running surface[49]. With the assumption based on the McMahon and Greene et al. (1979) pillow and board tracks, an intermediate track would produce a slight speed enhancement and shorter foot contact time while in addition producing less impact force on the body[49].

Contact time of the foot can be separated into two phases based on the GRFs; a breaking phase and a push off phase. During the breaking phase, the hamstrings produce the knee flexion prior to making initial contact with the ground and the quadriceps contract eccentrically to extend the knee extensor moments that prevent the athlete from collapsing on ground contact[60]. Positive extensor moment force and negative work is performed during this time by the knee and ankle joints and most of the shock absorption is performed by the quadriceps muscles[6,60,101]. This breaking force is overcome by a propelling force during the second half of the stance period, the push off phase, and overcomes the small amounts of velocity reductions caused by the breaking phase[16]. Muscles crossing the knee joint will then work concentrically to lift the body and extend the joints to create the push off phase, creating positive work by the ankle and knee joints[6]. The hip flexors behave in opposition to the ankle and knee joint work during these phases and produce positive work during the breaking phase and then negative work during the push off phase[6]. Faster speeds also show that the hip flexors will produce additional positive work for the last 15% of the push off phase, indicating additional pull to increase an individual's speed while running[6]. Winter et al. (1983) and Novacheck et al. (1998) indicate that the relative importance of each lower limb joint is unique to the running stride in that the hip supports the upper body and drives the lower limb into the swing phase, the knee absorbs mechanical energy, and the ankle will generate mechanical energy[60,101].

LEG STIFFNESS

Joint stiffness can be defined as the change in angular displacement along a specific joint in relation to the joint torque placed on it[5,27]. A stiffer motion or joint has low angular displacement relative to load compared to a more compliant or less stiff joint. In a similar way, surface stiffness can be defined as the deformation or compression in spring depth in the ground as a force is placed on it[27]. Surfaces that show stiffer properties and result in less deformation when a load is applied are typically seen on high traffic areas, for instance concrete, tar, or tracks that are made to withstand frequent use. Stiffer surfaces generally provide less potential energy back to the runner, resulting in a higher impact force when the foot makes contact with the ground[43,49]. Softer or less stiff ground surfaces, such as grass or sand, generally require more energy or power to traverse because the softer surface dissipates some of the person's energy[43,69]. Unlike joint stiffness, ground stiffness does not provide the same elastic properties as biological tissue and instead results in an effective shock absorber for impact forces with little energy transfer back to the athlete[18,24]. This excludes specialized "tuned" tracks for elite athletes or elastic surfaces which are created in order to return energy back to the individual. However, in these instances the surface density is altered and it can be assumed that an individual's joint stiffness would behave similarly to running on a softer surface[49]. As the foot makes contact with the ground, the ground stiffness level influences the amount of joint compression and affects running characteristics such as ground contact time, step length, and stride frequency[27,49]. Farley et al. (1996) showed that when the stride increases in frequency, the leg spring adjusts and becomes stiffer while running[28]. In addition, the knee and ankle increases extension, resulting in a more rigid stance, as an individual runs across lowering ground stiffness levels[31,34]. Ferris and Farley performed various studies examining the effects of traversing various surfaces and showed multiple times that surface stiffness affects the leg stiffness level[27,30,31]. In Figure 2 (B), leg stiffness is indicated by the various joint angular displacements that enable runners to dampen the variable GRF applied to the body upon foot strike[31].

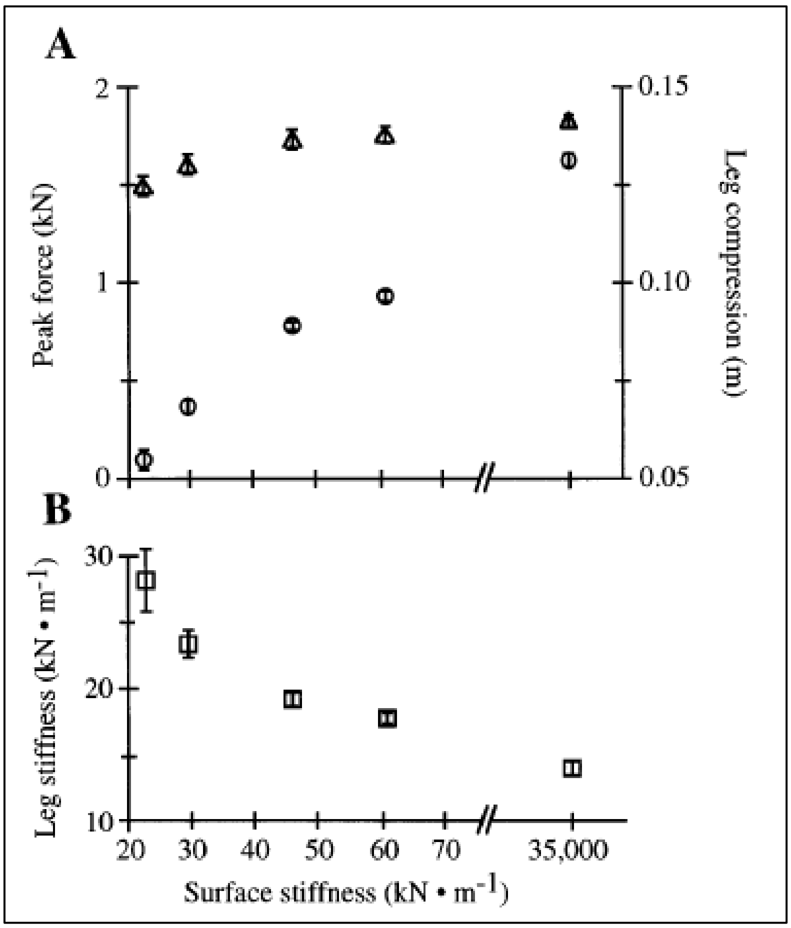

Figure 1: Farley et al. 1998 (Leg Stiffness (□), Leg Compression(O), Peak Force (Δ))

Vertical forces while running, such as GRF are known to differ based on the individual's speed and affects joint stiffness levels[36,57]. When an individual increases his or her speed, stride length increases and the center of gravity changes to a higher position. This results in greater impact forces and greater joint angular displacement[49]. Visualization of an individual increasing joint flexion to decrease the effects of vertical forces placed on the body can commonly be found when jumping from an increased height. DeVita et al. (1992) showed that as the knee flexion increased during a set height landing, resulting in decreased leg stiffness, the ground impact force lessened[24]. In addition, Farley et al. (1998) found that as surface stiffness increased, leg stiffness decreased (Figure 3 (B)) and leg compression, the vertical displacement of the participant's center of mass relative to the surface, increased. These data also showed peak GRFs had an inverse effect on the leg and increased as the surface increased[27] (Figure 3 (A); peak GRF (∆) and leg compression (O)). Overall, with a person's normal stride, the faster the running velocity, the more vertical forces were shown to be placed on the body. This is linked to leg stiffness which has also been shown to alter with surface stiffness changes[4].

Runners have demonstrated the ability to quickly alter their leg stiffness in order to adjust to new surfaces and to maintain their running mechanics[27,31]. Ferris et al. (1998) investigated joint stiffness as individuals ran across different surfaces and saw that not only did joints compress more as the ground surfaces became stiffer, they made a complete adjustment from stiff to soft or soft to stiff surfaces, by the first step[31]. Muller et al. (2010) also investigated reactions to different ground surfaces by measuring muscle pre-activation across uneven ground heights[56]. Ankle joint stiffness decreased significantly (from 8.14 Nm/degree during unperturbed (0 cm), or level without obstacles running to 6.33 Nm/degree during perturbed, uneven running with the highest step up

(15cm)) and adjusted itself based on the vertical height of the surface step. However, knee joint stiffness only showed a slight increase or decrease in stiffness depending on the uneven ground setup (Table 1)[56]. Muller's showed that the geometry of the leg segments at foot touchdown affects leg stiffness adjustment rather than a muscle pre-activation control process[56]. The spring mass model that McMahon and Greene developed is supported by showing that the body will automatically adjust its joint and leg stiffness levels based on the geometry of the leg at touchdown and enable us to adapt quickly to different surfaces. When looking at the role of each joint for the lower limb, the ankle was found to be the primary contributor to propulsion, landing adjustments, and body lift; the knee took on the primary role of shock absorption, force dampener, and leg stiffness; and the hip was the primary power for forward acceleration[4,19,20,28,65]. Though the foot plays a pivotal role in the quantity of force accumulation placed on the body, the knee appears to be an essential joint to protect the body from excessive forces.

Track Type	0	0/0	0/5	0/10	0/15
Ankle Joint Stiffness (Nm/°)	8.14 ± 1.31	8.01 ± 1.59	7.38 ± 1.73	7.21 ± 1.88	6.33 ± 1.67
Knee Joint Stiffness (Nm/°)	8.86 ± 1.35	9.38 ± 1.28	10.14 ± 3.10	9.44 ± 3.51	8.47 ± 3.00

Table 1: Adapted from Muller 2010
Ankle and knee joint stiffness while running across various track types. The different track types are regular level running (0), perturbed running without a step (0/0), perturbed running with a step of 5 cm (0/5), 10 cm (0/10), and 15 cm (0/15).

Another area that has been shown to affect leg stiffness is the design of shoe midsoles. As seen in various other sports, such as gymnastics and dancing, using an external tool to alter the ground cushioning or spring can be beneficial to the athlete and, in some cases, enhance their performance[18]. Shoe cushioning has been investigated extensively with a primary goal to lower the shock forces on the leg and provide stability while running or walking. Depending on the shoe midsole, leg mechanics will alter, and dynamic adaptations of the leg segments, rather than the realignment of the skeleton in response to the various stiffness shoes, contributes to the change in joint loading[13]. In addition, when investigating initial impact forces with different shoe midsoles, low ground surface stiffness levels were shown to lower or cancel out the impact force while running. This makes shoe cushioning not meaningful for more compliant surfaces though there was an intrinsic gain ((F_g – F_s) maximum) under 2% for softer surfaces while stiffer surfaces represented almost 10%[48]. Midsole cushioning has also shown to affect leg stiffness where, as the midsole cushioning increased, the joint stiffness of the leg became stiffer. Baltich et al. (2015) showed significant results amongst different footwear conditions which included vertical impact peak (Shoe Effect = 54.877, $p<0.001$), ankle joint stiffness (Shoe Effect = 55.409, $p<0.001$), and knee joint stiffness which were found to be significantly different between sex groups[5]. Females have shown to be at a greater risk for developing musculoskeletal injuries, particularly at the knee joint, however conclusions differ for various mechanical or conditional variables in terms of sex bias during running[5,94]. Despite the sex differences in joint stiffnesses, multiple footwear studies point to an increased vertical impact force in softer midsole shoes due to increased leg stiffness when running or walking (Figure 4)[5,75]. However, not all studies believe that this initial impact peak force is a reliable indicator of cushioning effects of midsole stiffness, and loading rate should be investigated as well when considering the harm in vertical GRFs[75].

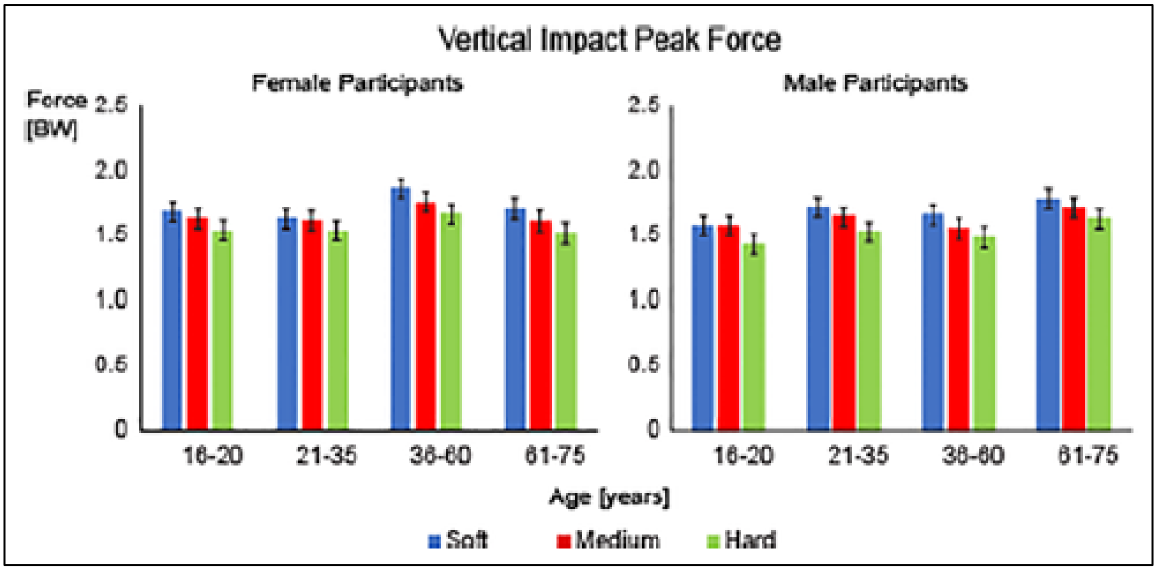

Figure 4: Baltich et al. 2015

When considering a running location, most individuals take into account primarily the ground properties before they begin (ex; street, sidewalks, trail, etc.). Secondary considerations then move to the shoe which will absorb force from the initial foot strike and provide cushioning and support for the duration of the run. It is commonly assumed that the shoe cushioning will create a softer landing which will produce decreased forces on joints. However, though shoe midsoles can decrease the initial impact force, the ground surface is changing depending on the shoe cushioning chosen. When combining the ground stiffness with the stiffness of the shoe midsole a new running surface is created upon which our running mechanics are modified in order to maintain consistent force dispersion. Since leg stiffness is manipulated based on the ground stiffness, one might assume that there is a consistent knee joint force across all surfaces whether it is combined with an external tool or not, and whether it is a stiffer versus softer ground surface.

JOINT AND TISSUE LOAD IN RUNNING THROUGH MUSCULAR SKELETAL MODELING

Running movements in the sagittal plane are predominately extensor while frontal plane movement is primarily abductor at the hip and knee and inversion at the ankle joints[66]. One of the largest variables of force and load distribution throughout the body while running is the individual's foot strike pattern on the ground surface. Cavanagh et al. (1980) has defined foot strike as the point of initial contact by the foot on the supporting surface[14]. Three types of patterns have been defined depending on this initial contact; heel strike, mid-foot strike, and fore foot strike patterns.

When the heel strikes the ground, also referred to as having a heel strike pattern, a sudden shock is sent through the body. This impact force is seen visually on a GRF curve and is affected by many factors including running velocity, stride length, and ground stiffness. In the distribution of the force throughout the body, muscles in the lower limb comprise a large factor involved in energy absorption[20]. Using the spring mass model which we have currently used to represent the leg while running, we can take that concept and put it on a smaller level within the leg itself. The muscles, tendons, and ligaments all act as individual springs and recoil upon heel strike then stretch as the foot leaves the ground[28]. At heel strike, the knee is the only joint that is flexed during the entire impact and is the most responsive when altering stride length[20].

Derrick et al. (1998) found that the tibia experiences greater impact forces during this phase and is matched with a greater energy absorption by the muscles crossing the knee joint[20].

In mid-foot strike patterned runners, the center of pressure is near the middle of the foot between the heel and metatarsal-phalangeal joint while the fore foot strike pattern places the center of pressure in the anterior third of the foot[14]. Though the heel strike pattern is the predominant strike pattern for runners, mid-foot and fore foot strike patterns are more commonly associated with faster running and sprinting on tracks due to a shorter ground contact time[39]. Unlike the heel strike pattern, which uses more musculature in the lower limb to resist the initial impact peak and shows a greater difference in leg stiffness, mid-foot and fore foot strike patterns put a greater force on the ankle joint through increased use of the soleus and peroneal muscles.

Due to the various lower limb musculoskeletal injuries, loading rates have been investigated and foot strike pattern has been noted as a possible explanation for higher load levels on specific joints as well as knee joint stiffness[41]. In variable step frequency testing, an individual can experience loads more than 2 times their body weight during the vertical impact peak when they decrease their frequency by 30%[41]. Increasing step frequency by as little as 10-15% helps lower the impact peak to around 1.7 times body weight and has been linked to a 16% decrease in patello-femoral forces[26,41,97]. Though shorter strides or an increased frequency will increase loading cycles, it is not as pertinent as strain magnitude while running[26]. Additionally, Vannatta et al. (2015) found that the change in foot strike pattern from heel to fore foot strike helped decrease patello-femoral joint stress by 27% and lowered quadriceps and hamstring forces while raising leg stiffness, gastrocnemius, and soleus muscle forces[92]. As many studies have shown, foot strike plays a large role in the determination of how the body changes its mechanics to alter the force distribution.

Many studies investigating knee joint forces have focused on changing the biomechanics of the individual's running mechanics[12,15,26]. With specific focus on common running injury loads on the lower limb, knee joint forces have been found to be up to 6 times a person's body weight for patello-femoral forces and up to 10 times body weight for tibio-femoral joint forces[71,94,96,98,100]. Decreasing these force magnitudes have shown to be successful with a shorter step length and step frequency[96,100]. Though these changes in biomechanical gait have shown to be beneficial, it is difficult for participants to adhere to these gait changes without going back to their habitual running stride. Changing the running surface stiffness requires less conscious changes for the runner and could help decrease loads to the joints. However, there are no studies that have investigated the internal forces of the knee as changes are made to the surface. This calls to question the benefits of running on softer surfaces, especially at the knee joint, as it could be a quick strategy to lower joint forces.

RUNNING RELATED INJURIES

Cook et al. (1995) put it bluntly when he noted that "even a slight biomechanical abnormality can induce injury" most of which will occur as a result from the high frequency of low impact forces at the heel strike phase of the stride[17,60]. Throughout all of the foot strike patterns, Rooney et al. (2013) found that the greatest peak forces were seen at the knee joint during the mid-stance phase (ranging from -11.9 – -13.2 times bodyweight) while Arampatzis et al. (1999) and Farley et al. (1996) found that the knee and ankle moment and mechanical power were affected at different velocities across a level consistent ground surface, with leg stiffness primarily affected due to the knee[4,28]. The knee was also found to be the primary energy absorber, whereas the hip and ankle joints were predominately the energy generators due to the amount of positive work done over one stride[70]. When stride length was affected, the knee absorbed slightly more energy than the ankle and the hip when stride length was affected (Figure 5)[20]. From this, we can see how the knee is sensitive to biomechanical changes and can be an area more prone to injury during running activities.

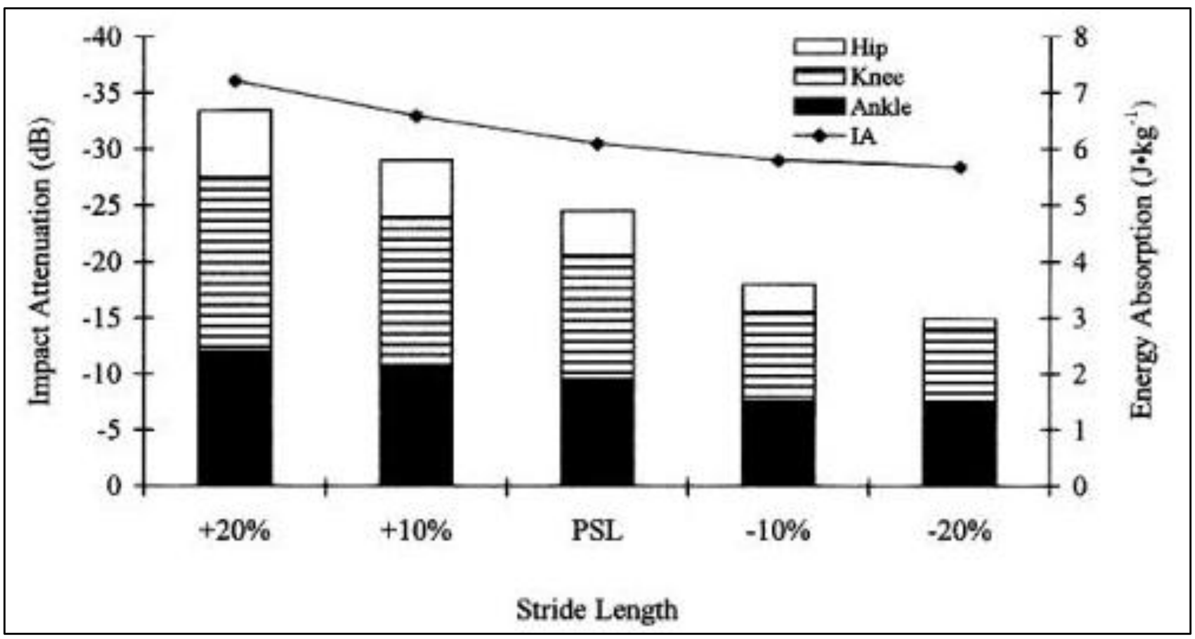

Figure 5: Derrick et al. 1998

With the growing popularity of leisure and recreational running, and with the repetitive nature of the sport, many lower limb overuse injuries have surfaced (varying between 2.5 to 12.1 injuries per 1000 hours of running and afflicting 79% of runners)[52,91]. Risk factors for overuse injury are brought about by two main factors; an individual's behavior and physiology. Possible behavioral reasons that cause injuries include improper training, warm up, stretching, shoe stability, and/or factors like overtraining, age, weight, hill running, and muscular imbalance[52,91]. Common running injuries include patello-femoral pain syndrome, iliotibial band friction syndrome, plantar faciitis, meniscal injuries, tibial stress syndrome, and Achilles tendinitis[52,84]. Most commonly, the knee joint has the highest percentage for injury (7.2-50%) followed by the lower leg (9.0-32.2%) and foot (5.7-39.3%)[90].

Many studies have been conducted to determine how to manipulate an individual's running biomechanics in order to potentially lower forces placed on the body. Willson et al. (2014) showed that with a shorter stride, patello-femoral joint stress decreased by 7.5% and Meardon et al. (2014) found that widening step width decreased shear stress, anterior tension, posterior compression, and medial compression on the tibia[50,97]. In addition, external tools have been created to decrease impact forces or enhance an athlete's performance such as shoe midsoles, track compliance, sprung floors, and floor mats[5,18,35,49,62]. With the consistent repetitive force that is placed on the lower limb while running, overuse injuries accounted for as many as 65% of runners per year causing them to cease running and seek treatment[52]. Staying active is important for maintaining a healthy lifestyle and function as we age. To be forced by an injury to cease normal activity can put individuals at risk depending on the degree of inactivity. By investigating factors that could affect knee forces, we can attempt to bring further awareness toward possible solutions for knee joint pains.

SUMMARY

As we run across various ground surfaces, at different speeds or with different running equipment, our running biomechanics change to account for the changing environment. Continuous pounding against the ground has led to the creation of external tools to help lower joint forces. Gymnasts will use mats to absorb some of the landing force when they dismount from a routine and dancers will use sprung floors for practice before a show. However, when shoe midsole cushioning was created to lower the accumulation of sudden GRFs for runners, many studies showed a greater initial impact peak force with greater cushioning but the loading rate decreased[5,75]. To offset the increased impact forces associated with increased GRFs, our leg stiffness will decrease to allow our muscles to dissipate forces and will alter depending on the amount of force placed on the joints. Leg stiffness and specific biomechanical variables of the running gait or environment have been manipulated which have led to lower joint loads. Greater compliant ground stiffness levels have been assumed to be a simple way to alter an individual's running stride to make the experience more advantageous for the joints however previous studies indicate that this may not be as beneficial as expected. As an individual runs across a more compliant surface, their leg stiffness becomes more rigid and increases GRF. More noticeably with heel strike runners, the impact peak force will increase, while the rate to this peak decreases. In addition, this impact peak force increases as the shoe midsole cushioning is increased. The potential of this greater repetitive force while running has led to many musculoskeletal injuries and is most commonly seen in the knee as an overuse injury. The knee joint is the primary joint to absorb energy and incur the highest levels of force. With 79% of runners per year

afflicted with overuse injuries, the knee needs to be examined further to determine quick and simple changes a runner can make to avoid wear on this already prone to injury joint[52,84,89].

It is unclear how knee joint forces are affected on varying surface stiffnesses though many individuals would assume that softer surfaces are more beneficial to the leg joints. Previous literature indicates that there is adequate information to question how knee joint forces are affected by ground stiffness, regardless of the shoe style chosen. It is our intention to investigate this question further. We hypothesize that the knee forces will stay relatively consistent throughout various running trials over different surfaces. Due to the importance of the knee during running, and other locomotor tasks, knowing more about the knee joint forces while running across different ground stiffnesses will allow for future improvements to injury prevention programs.

Chapter III. METHODS

This study aimed to determine whether a difference exists in knee joint forces while running on varying ground stiffness levels. We hypothesized that varying the ground stiffness would not make a significant difference to the force placed on the knee across different ground stiffness levels due to the joint manipulating itself to maintain the body's natural running mechanics. This chapter describes the participants' characteristics, instruments used, study procedures, data reduction, and statistical analysis.

PARTICIPANTS

Seventeen participants were recruited from the city of Greenville, North Carolina. This included individuals attending East Carolina University (ECU) for school and the city of Greenville residents. Recruitment was done using flyers, word of mouth, and website and ECU announcements. Participants were included if they meet the following inclusion requirements: 18 – 30 years old to reduce aging effects, recreational runner that trains between 10 and 30 miles per week, habitually heel strike runner, and were healthy and mobile during the time of the study. Healthy and mobile participants were defined as those who did not have any current or lingering musculoskeletal injuries or conditions and were free of pain in the lower limb. Participants were excluded if they had any current musculoskeletal injuries to the lower limb or previous surgeries that affected the knee joint, cardiovascular diseases, nervous system diseases, or any other major diseases that would affect their ability to perform the requested tasks, and did not display a heel strike running pattern which was determined based on the presence of an impact peak prior to the max ground reaction curve typical of a heel strike runner. Each participant was required to sign an informed consent form prior to participation and all procedures were approved by the ECU Institutional Review Board (Appendix B and C). Characteristics of each participant are displayed in Table 2 referencing participant sex, age, mass, height, and BMI, and average miles ran per week. Sex characteristic averages are displayed in Table 3.

Table 2: Participant characteristics

Sub ID#	Sex	Age (yr)	Mass (kg)	Height (M)	BMI (kg/m^2)	Miles/Week
1	M	24	78.6	1.81	24.1	25
2	F	22	55.9	1.62	21.3	11
3	F	30	76.7	1.71	26.2	14
4	F	24	54.3	1.67	19.5	11
5	F	19	72.0	1.75	23.5	12
6	M	24	65.2	1.78	20.6	17
7	F	18	51.9	1.65	19.1	15
8	M	18	58.3	1.86	16.9	13
9	M	19	88.9	1.79	27.7	12
10	F	20	60.6	1.69	21.2	10
11	M	19	66.9	1.78	21.1	12
12	F	20	54.5	1.57	22.3	12
13	M	24	75.2	1.82	22.7	12
14	M	20	89.1	1.77	28.6	11
15	M	21	98.0	1.83	29.3	10
16	F	21	54.6	1.65	20.1	10
17	F	19	59.6	1.70	20.6	10
Mean ± SD	9F, 8M	21 ± 3.1	68.3 ± 14.2	1.73 ± 0.08	22.6 ± 3.5	12.7 ± 3.7

Table 3: Sex characteristic averages (Mean ± SD)

Sex	Age (Yrs)	Mass (kg)	Height (m)	BMI (kg/m^2)	Miles/wk
Females	21.4 ± 3.7	60.0 ± 8.6	1.7 ± 0.1	21.5 ± 2.2	11.6 ± 1.8
Males	21.1 ± 2.5	77.5 ± 13.8	1.8 ± 0.0	23.9 ± 4.4	14.0 ± 4.9

INSTRUMENTS & SOFTWARE

Data collection was performed on a 15m track with an in-ground embedded force platform (AMTI Model LG-6, Newton, MA) with a sampling rate of 960Hz and a gain of 4000, covered with various rubber thicknesses to alter ground conditions (Consolidated Plastics Ultra Sponge Mat, Stow, OH). Each participant was timed as they ran with an infrared timing gate (Brower Timing Systems, Model IRD-T175, Salt Lake City, UT) to verify consistent speed and was tracked with eight Qualisys ProReflex MCU 240 motion capture cameras (Qualisys Medical AB, Gothenburg, Sweden) at 200Hz. In addition to the marker tracking, electromyography (EMG) data (Delsys Trigno Wireless EMG, Boston, MA) was gathered from the vastus lateralis, bicep femoris, anterior tibialis, and lateral gastrocnemius muscles along each participant's right leg to measure muscle activation. The EMG signal, as well as the motion camera and force platform data, were processed through Qualisys Track Manager Software (Innovision Systems Inc., Columbiaville, MI) (QTM) then exported to a compressed file that was analyzed in Visual 3D (C-Motion Inc., Rockville, MD) (V3D) program. Kinematic and kinetic data were then processed through Quick Basic (IBM Basmark QuickBasic, Cleveland, Ohio) to calculate specific muscle and joint measurements including tibio-femoral and patello-femoral forces for the lower limb in order to compare results. Prior to each data collection, participants were weighed and height was taken by a Seca 703 digital scale (Seca GMBN & C. Kg, Hamburg, Germany). Equipment was calibrated and maintained according to the factory protocols.

PROCEDURES

Participants were asked to come in twice for this study which was broken down into a learning phase and a data collection phase.

Learning Phase

On the first day the consent form was covered and directions were given of the tasks that each participant would perform over the course of the study. Once the consent form was signed, all questions had been answered, and participants understood the directions they were to perform, the remaining time was spent practicing running across the 15m track over the three surface conditions: force plate (highest stiffness), single mat layer (medium stiffness), and two mat layers (softest stiffness). Each condition had the participants run over an in-ground force platform, timed so they would hit the force plate with their right foot at the correct velocity. Infrared timing gates were set up 1.5m from the center of the force platform on each side so the speed of the participants was measured and they could practice running consistently at 3.46m/s $\pm$ 5% (7:44 min/mile). A similar fixed speed was used in other leg stiffness and surface running studies[4,11].

Running Conditions

There were three different running conditions ranging from a stiffer ground surface to a softer ground surface. Each was a distinct stiffness level to give the greatest opportunity for significant differences to be measured. The first condition (C1) was across the controlled stiff floor with no ground alterations. The second (C2) and third (C3) conditions were altered by laying down one and two layers of PVC sponge (shock absorbing, 100% closed cell, Shore00 65 mat) thickness' to create a mid-stiffness level ground (length 914.4cm x width 91.4cm x height 1.6cm) and a soft stiffness level ground (length 914.4cm x width 91.4cm x height 3.2cm). The rubber pieces were large enough to cover the majority of the track and the force plate so that at minimum, three strides of a participants' run were on the rubber mat prior to contacting the force plate. The specifications for the PVC sponge for one layer (C2) were rated at a manufacturer derived durometer value of Shore 00 65 $\pm$ 10. To create the softest stiffness ground condition (C3), a duplicate mat of the same PVC sponge specifications was layered on top of the first mat. An impact test with a known mass of 1kg was performed as well to determine if a difference was seen between mat layers. The mass was dropped from a consistent 12cm onto each condition at an increased capture frequency in the force plate to ensure the peak force was recorded (24K samples/s). Two trials were recorded and averaged for value verification as well as consistency. From C1 to C2, an 87.1% decrease (C1- 3671N, C2- 472N) was recorded and from C2 to C3, a 22.9% decrease (C3- 364N) was recorded for the initial impact GRF peak. It was determined that these percent decreases between conditions was optimal for this study, and showed that there was a difference between surface stiffness conditions. Figures 6-8 show the results from the weighted drop test. All figures are shown with the same frame count and the first 3 bounces of the weight.

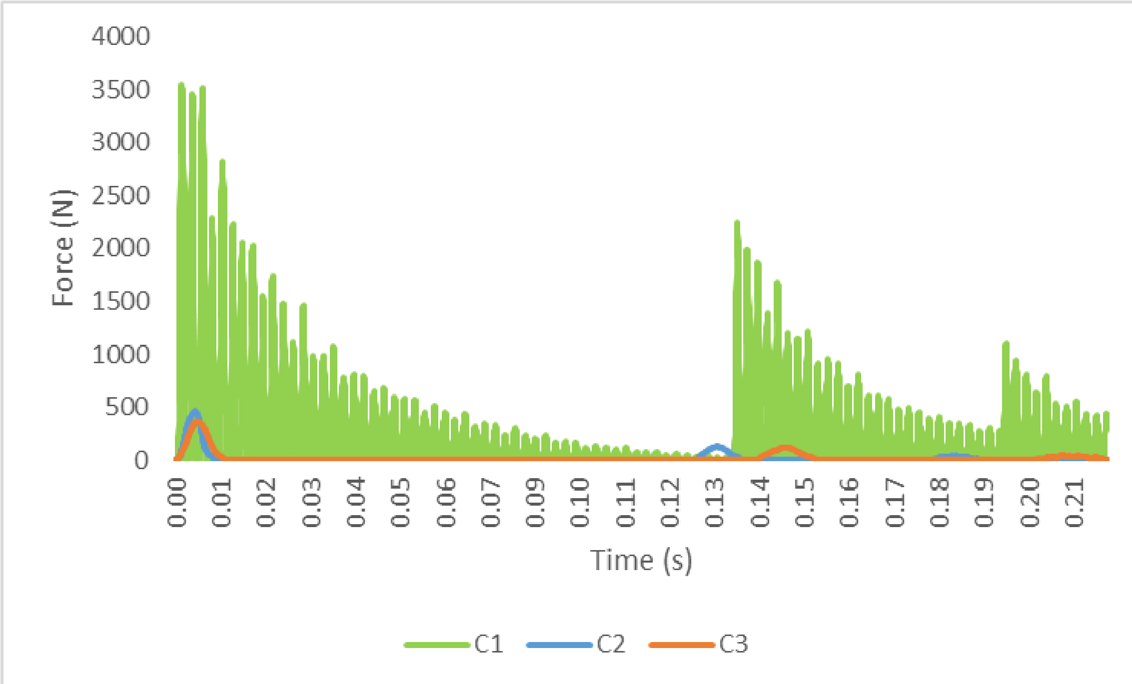

Figure 6: Weight drop GRFs for all three surface conditions (C1- Green, C2- Blue, C3- Orange)

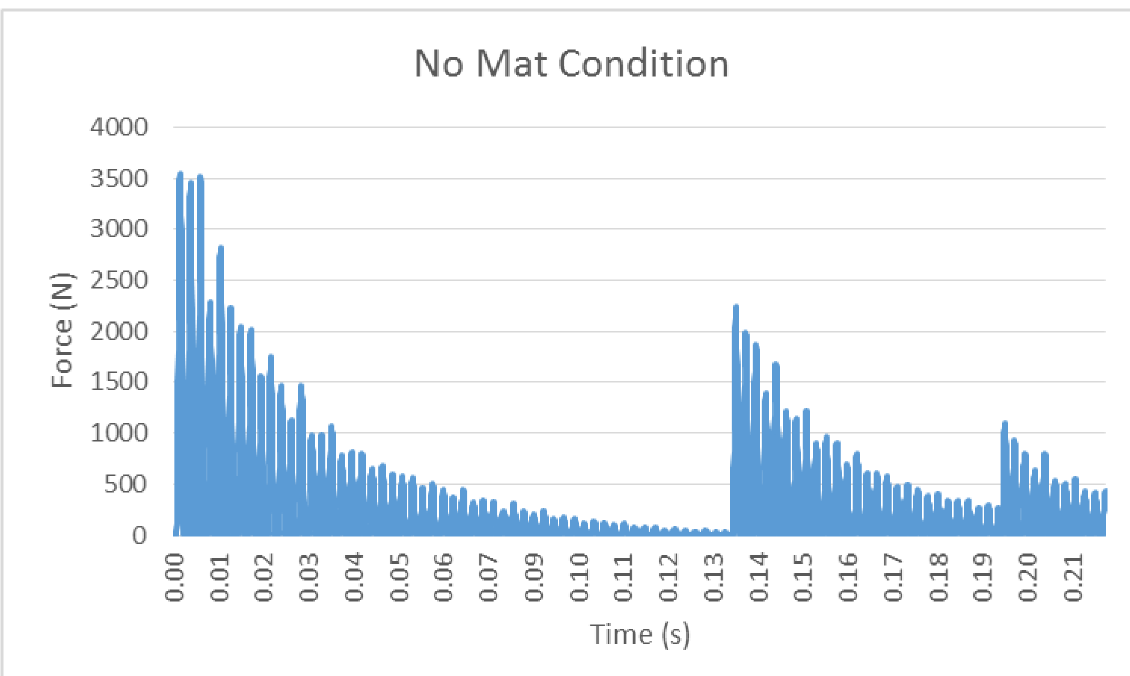

Figure 7: GRF for the first 3 bounces of the weight on condition 1

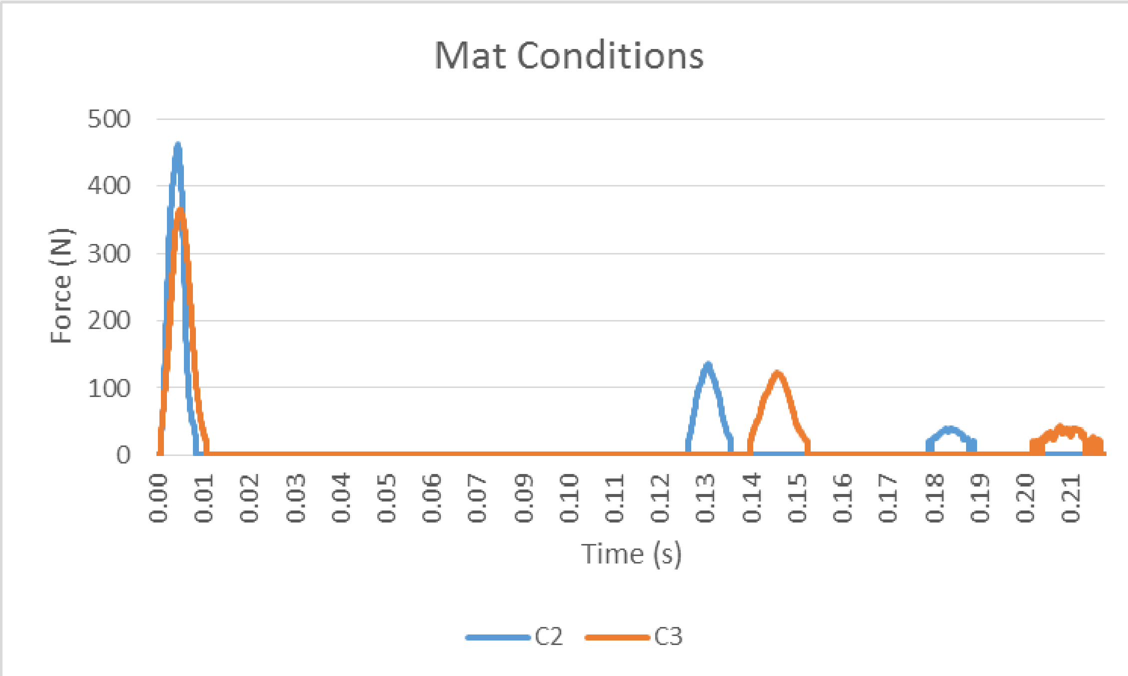

Figure 8: GRF for the first 3 bounces of the weight on conditions 2 and 3

Data Collection Phase

On day two, prior to the participants' arrival, the force plate was located in the QTM program with an L frame to mark the coordinates in the system and 4 placement markers to define the area of the platform. A 750.1mm T-wand was used to calibrate the testing area and camera capture location. Calibration trials were accepted with less than or equal to a 2mm average residual error per camera capture. The Delsys Trigno program was activated to verify that the electrodes were operational and were ready to process the signal data into QTM when applicable. In addition, 15 reflective markers, 3 reflective marker plates (for the thigh, leg, and foot), as well as 4 electromyography (EMG) electrodes and skin preparation supplies (adhesive scrub, alcohol wipes, and safety razor) were set out for the participants' arrival. Infrared timing gates were then set up on either side of the force plate and verified to work correctly by passing through the beam to start and stop the timer.

Upon arrival, each participant was asked to change into spandex shorts and a tight fitting shirt and then had their height and weight measured with their self-chosen shoes on. Participants then proceeded to have their skin prepared for attaching the EMG electrodes. After locating the muscle belly, the area was shaved and then an abrasive scrub was used to clean away dirt, lotion, and dead skin cells. The scrub was washed away using alcohol pads and then the electrodes were placed on the leg in the direction of the muscle fibers. The four EMG electrodes were placed on the bicep femoris, vastus lateralis, anterior tibialis, and lateral gastrocnemius. Once the electrodes were placed, the participant performed a maximum voluntary contraction (MVC) of the various muscles one at a time to verify that the Delsys Trigno program and QTM were picking up the EMG signal as well as to have a maximum contraction comparison to the gathered running data. Once the signals were verified and recorded, the electrodes were secured to the leg with a wrap so they would not move or fall off during the study. The participant then had the tracking and calibration markers put along their pelvis and right leg on the traditional anatomical palpable bony landmarks near the segment endpoints; right and left anterior superior iliac spine, posterior superior iliac spine, iliac crest, and greater trochanters, and lateral and medial femoral condyles and malleoli (Figure 9 A and B). Markers for the first and fifth metatarsal heads, and the right calcaneus were placed over the participants running shoes. To minimize clothing movement error for the pelvic markers while running, a bandage wrap was wrapped securely around the participant's pelvis and upper torso and the reflective markers were place on top of the bandage. The marker plates were then placed on the lateral thigh and leg, and dorsal foot. Prior to the running trials, the participants performed two, four second standing trials. The first standing trial included all tracking and calibration markers as well as the EMG electrodes. The second standing trial was performed without the calibration markers (right and left iliac crest and greater trochanters, lateral and medial femoral condyles and malleoli, and the first and fifth metatarsal heads) which was used to establish the participant's neutral joint angles and then to offset from the motion trials.

After the standing trials, each participant was allowed time to practice running at the required speed across the surface conditions prior to the data collection. Five successful running trials for each runner and each ground surface condition were collected. Successful trials were defined as the participant ran at the correct speed and was consistent throughout the trial run, their right foot struck the force plate in a normal stride, they ran the full length of the 15m track, and their tracking markers were visible for the entirety of the stride starting at toe off before the force plate and toe off from the force plate into the flight phase of the next stride. Running conditions were presented randomly between participants to avoid order effects (Figure 9 C). Once all trials were collected, the markers and electrodes were removed and participants were thanked for their participation and reminded that they could access their data at any date if they wished to do so.

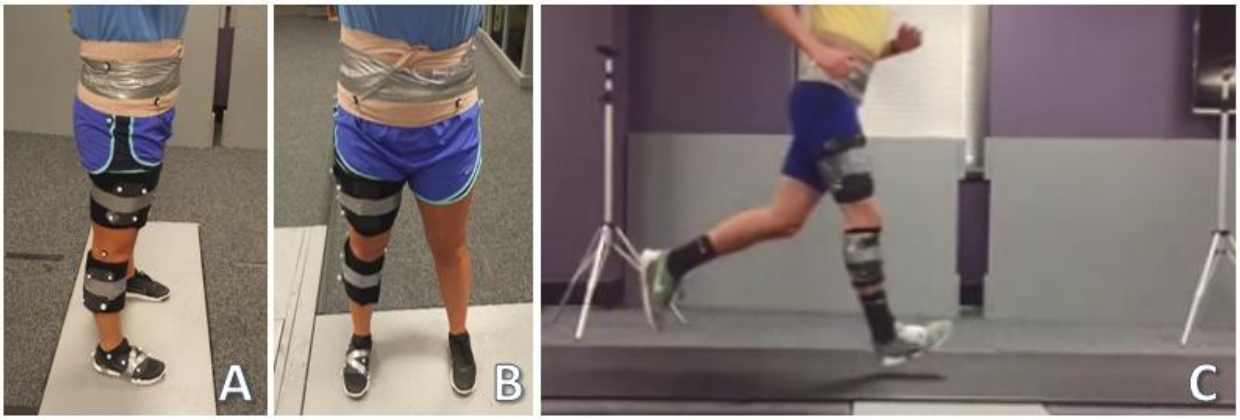

Figure 9: Marker and EMG placement from the sagittal (A) and frontal (B) plane. Visual representation of the protocol setup with participant running across condition 2 (C).

DATA REDUCTION

Each data trial gathered in QTM was cut to focus on the toe off of the stride before the force plate and a few frames past the toe off after the force plate. Markers points were identified based on their anatomical position during QTM processing to ensure all data points were visible and gap filled and were then exported as a C3D file. These files were used in the V3D program to create a leg and pelvis model to represent the participant as they ran across all trials based on their height and mass. Raw signals were filtered at 45Hz for force plate data and 6Hz for motion capture data to reduce noise caused by elements beyond the area of focus. Joint centers for the ankle and knee were found using the half way point between the lateral and medial marker points of each joint and were then used to calculate center of masses per leg segment. For the hip, the left and right greater trochanters were located and a quarter of the distance was taken medially from each marker to establish the joint socket. To create the leg segments (the pelvis, thigh, shank, and foot), at least three calibration marker points were used to establish the frontal plane and to calculate the distal endpoint to the proximal endpoint. Segment masses were assigned according the default settings established by the V3D program based on each participant's entered mass.

Joint angular position and velocity were calculated from the kinematic data using V3D and were used to determine kinetic data such as joint torque, work, and relative force via inverse dynamics across the different ground stiffness levels. All kinetic data was normalized to the participant's body mass which was gathered prior to data collection.

Lower Limb and Knee Stiffnesses

Lower limb joint angular displacements and joint torques were initially calculated from the V3D program. Leg stiffness (k_{leg}) (Figure 10) was calculated using the ratio of peak ground reaction force (F_{peak}) over leg spring compression (ΔL) (Equation 1). Leg spring compression was based on the vertical displacement of each participant's center of mass as estimated by the pelvis markers (ΔL)[31].

$$k_{leg} = \frac{F_{peak}}{\Delta L} \qquad \text{Equation 1}$$

Knee joint stiffness (k_{knee}) (Figure 11) was calculated while in this phase as the ratio of the peak torque (T_{peak}) at the knee joint, simultaneous as the knee is at its maximum flexion and the knee angular displacement from ground contact to the peak torque ($\Delta\theta$) (Equation 2)[64,72].

$$k_{knee} = \frac{T_{peak}}{\Delta\theta} \qquad \text{Equation 2}$$

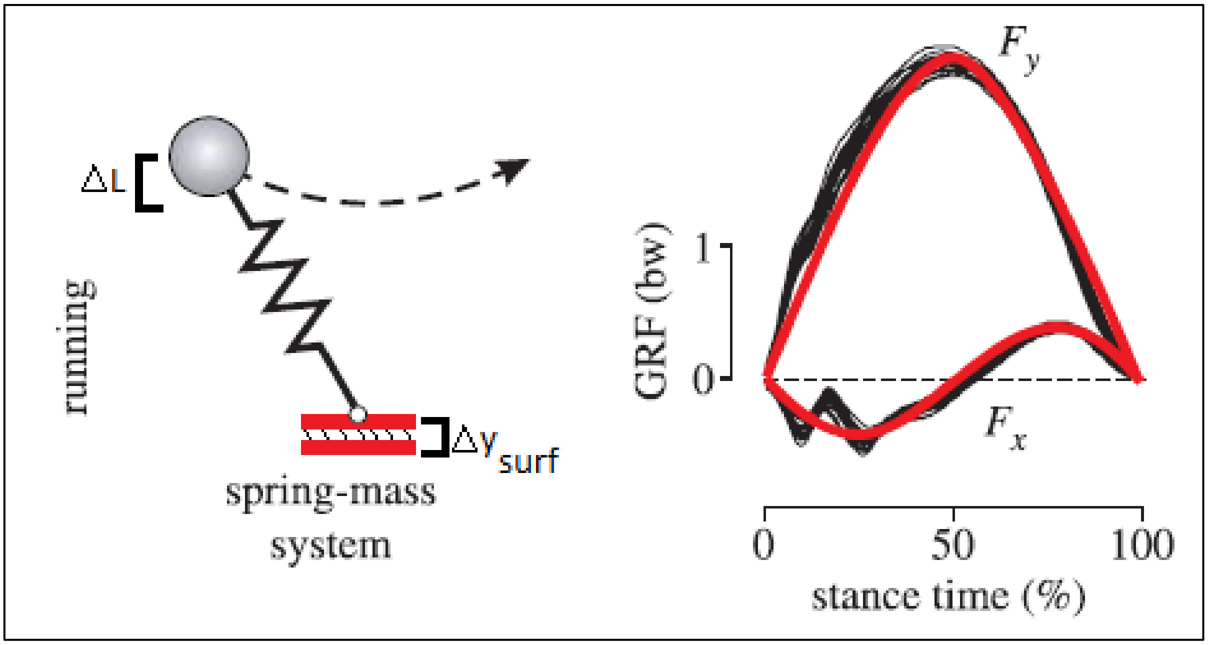

Figure 10: Adapted from Geyer et al. (2006); Standard running spring-mass model of a leg while in the stance phase on variable ground stiffness. Ground Reaction Force curve (red line) shows a typical predicted form of a runner during the stance phase. Forces are represented in the horizontal (Fx) and vertical (Fy) directions.

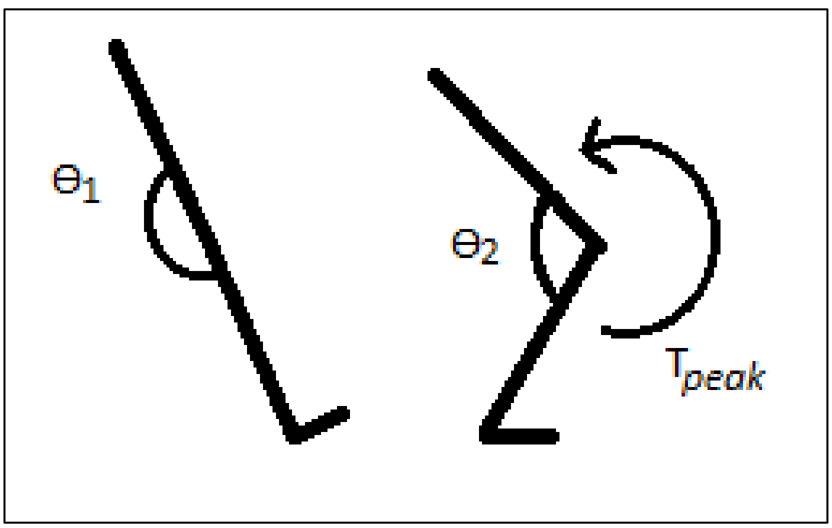

Figure 11: Knee joint stiffness at heel strike and at maximum flexion during stance phase.

Musculoskeletal Knee Model

Primary calculations specific for the knee joint focused on tibio-femoral and patello-femoral joint forces while in the stance phase. The movement kinematics, joint forces, and moments from inverse dynamics were entered into both biomechanical models to calculate the forces produced by the gastrocnemius, hamstrings, and quadriceps muscles and to determine the tibio-femoral and patello-femoral contact forces[23,52].

The DeVita et al. (2001) and Messier et al. (2011) musculoskeletal knee model (Figure 12) was used to calculate the compressive and anterior-posterior shear forces within the tibio-femoral area of the knee joint. This was done using inverse dynamics to determine joint reaction forces, moments, and kinematics to calculate forces generated by three major muscle groups (quadriceps, hamstrings, and gastrocnemius) as well as lateral support (lateral collateral ligament) surrounding the knee joint. The summation of these forces as well as the horizontal and vertical reaction forces allowed a determination for an estimate of the tibio-femoral force[21,51].

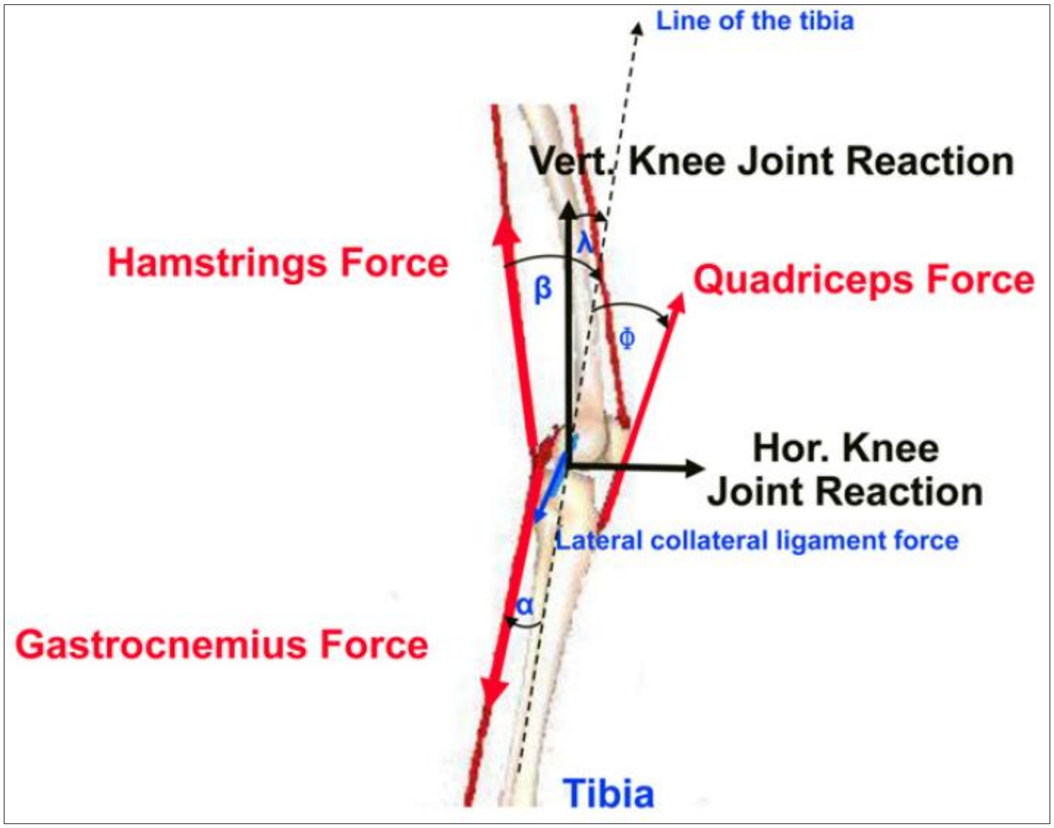

Figure 12: Messier et al. (2011)

Tibio-femoral Joint Force

All muscle group forces were determined during the stance phase of the individual's stride. The gastrocnemius was specifically determined from the plantar flexor moment produced by the triceps surae muscles (gastrocnemius and soleus muscles). The triceps surae force (TS) was a result of dividing the plantar flexor moment (At), assuming there is no co-contraction by the dorsiflexors, by the moment arm for the triceps surae at the observed angular position of the ankle (ATd) (Equation 3).

This observed angular position of the ankle moment arm, for all observed angles, was determined from the moment arm angle joint position curves from Rugg et al. (1990)

(0.051m)[68]. The 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 (0.319 a ratio of all PCAs of the gastrocnemius and soleus in order to get the most representative endpoint) (Equation 4)[21,51,103]. These methods were supported by EMG data of the gastrocnemius, and ankle plantar flexor torque[77,78,102]. The direction of G was identified as 3 degrees from the tibia and is expressed as α in the musculoskeletal knee model. This variable represents the angle between G and the tibia as G applies a relatively large compressive load but small shear load at the knee.

Hamstring force (H) was calculated from the extensor moment at the hip observed during the stance phase. This method was supported from previous literature that showed a strong association between hip extensor torque and hamstring EMG in the early stance[7,52,66,100]. The hip extensor torque was assumed to be produced by the hamstrings and gluteus maximus without co-contraction from the hip flexors. This assumption is supported by EMG measures and muscle force predictions in literature except that the rectus femoris does contract and produce some force during this time. However, predicted estimates of this force are low during the first half of the stance phase and have been assumed to produce a low amount of error for the H[3,32]. H accounted for both the hamstring PCA relative to the total PCA of the hamstring and gluteus maximus and the hamstring moment arm at the hip relative to the gluteus maximus moment arm. This total hamstring proportion (Hp) to the hip extensor torque was calculated (Equation 5) where Ham PCA and GM PCA are the hamstrings and gluteus maximus PCAs, and Hd and GMd refer to the hamstring and gluteus maximus moment arm. Values for these constants were obtained from literature and were Ham

PCA = 42.4 mm^2, GM PCA = 17.36 mm^2, Hd = 0.042 m, and GMd = 0.047 m (moment arm values were determined from him angular positions between 30° and 90°)[25,103]. This calculates the proportion of the hip extensor torque generated by the hamstrings, Hp, to be 0.63. H was then calculated (Equation 6) where Het refers to the hip extensor torque. H was assumed to be 0 while the hip torque was in the flexor direction which is supported by EMG data of the hamstrings[22,32,102]. The force direction generated by the hamstrings was set to parallel with the femur at an angle of β to the tibia.

Quadriceps force (Q) was calculated by taking the observed net knee torque, H, G, and accounted for co-contraction of the knee flexors. The observed net knee torque (Kt) was a function of all of the muscles that crossed over the joint (Equation 7) where Kt acquired from inverse dynamics, Q, H, and G are the forces by the quadriceps, hamstrings, and gastrocnemius, and Qd, Hd, and Gd are the respective moment arms for the muscles. Rewritten, the force of the quadriceps, Q, is then calculated (Equation 8). The moment arm values were derived from previous literature through all angles of the knee position during the stance phase with average values of: Qd = 0.035 m, Hd = 0.032 m, and Gd = 0.018 m[40,59,83,104]. Direction of the quadriceps force (φ) was determined from literature and was also a function of the knee angle[59,104].

Frontal plane loads, primarily provided by the lateral collateral ligament, were used to determine the lateral support structure in the knee. The external loads that were placed on the knee adductor moment are resisted by the abductor moments from the quadriceps and lateral structures. The product of Q and the frontal plane lever arm provided the quadriceps abductor moment which was taken away from the observed net internal abductor moment arm. Inverse dynamics were used to determine the moment distributed to the lateral knee tissues. The force in these tissues was found by dividing the torque by the moment arm, provided from the position data through V3D, and this force was considered to act parallel to the line of the tibia[51].

As a final calculation to find the tibio-femoral joint forces, both anterior-posterior shear (Ks) and compressive (Kc) forces were calculated by taking the sum of all the muscle forces (G, H, and Q), the force in the lateral support structure (Ls), and the joint reaction forces identified through inverse dynamics (vertical (Kz) and horizontal (Ky)) (Equation 9 and 10).

Ks was positive when the shear force was applied in an anterior load to the tibia and Kc was positive when the compressive force pushed into the tibia.

$$TS = At / ATd \qquad \text{Equation 3}$$

$$G = TS\ (0.319) \qquad \text{Equation 4}$$

$$Hp = [Ham\ PCA / (Ham\ PCA + GM\ PCA)]\ (Hd/GMd) \qquad \text{Equation 5}$$

$$H = Hp\ (Het)/Hd \qquad \text{Equation 6}$$

$$Kt = Q(Qd) – H(Hd) – G(Gd) \qquad \text{Equation 7}$$

$$Q = (Kt + H(Hd) + G(Gd)) / Qd \qquad \text{Equation 8}$$

$$Ks = G \sin \alpha - H \sin \beta + Q \sin \phi - Kz \sin \lambda + Ky \cos \lambda \qquad \text{Equation 9}$$

$$Kc = G \cos \alpha - H \cos \beta + Q \cos \phi - Kz \cos \lambda + Ky \sin \lambda + Ls \qquad \text{Equation 10}$$

Patello-femoral Joint Force

Calculations for the patello-femoral joint force were based on the methods used by Ahmed et al. (1987) (Figure 13)[1]. This force is primarily to determine the net force being placed on the knee joint by the patella, applied by the quadriceps muscles along the femur, focusing primarily on the rectus femoris (Tr), the patello-femoral joint reaction force (PFJR), as well as the tension of the ligamentum patellae (Tp). The knee joint angle was calculated through the position data and was represented by the angle β, assuming that the PFJR is normal to the contact surface. Both tension values are assumed to be equal, without account of friction on articular cartilage so the angle β was divided equally in half, split by the PFJR vector, which is represented as the angle value α. Forces were then calculated using basic trigonometry and the resultant force represented the net patello-femoral joint force. This method was supported by previous literature assuming the knee joint was healthy[1,96].

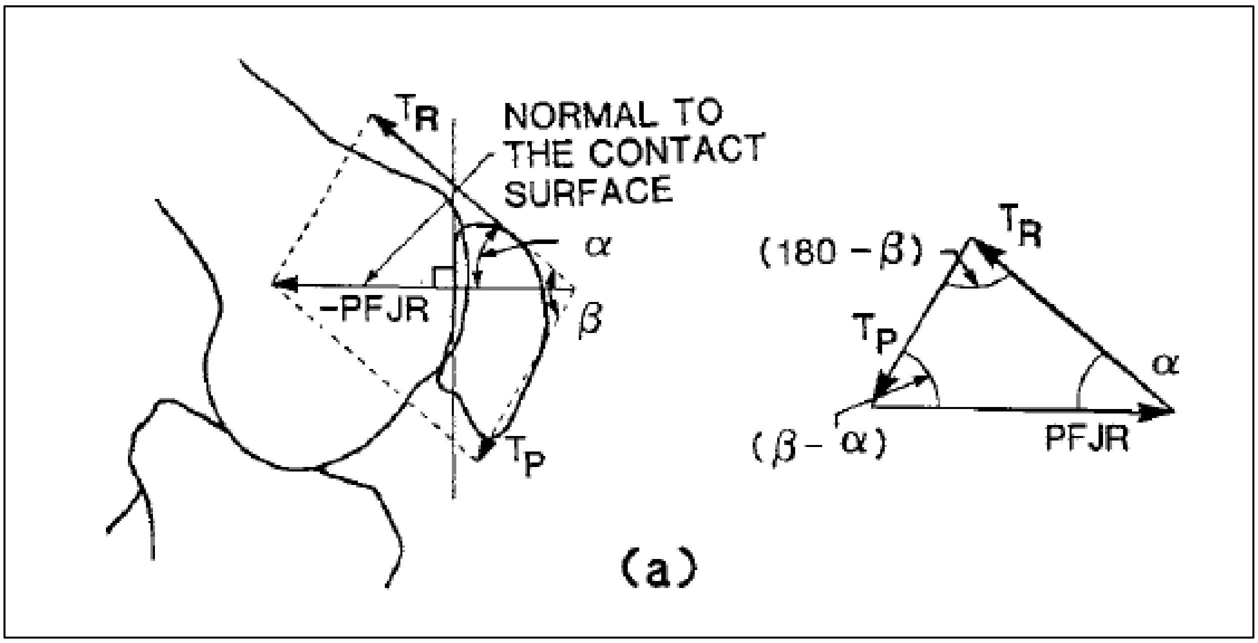

Figure 13: Adapted from Ahmed et al. 2015

Surface Stiffness

The manufacturer provided a Shore 00 65 ± 10% hardness value for the stiffness of the rubber mats used for this study. Table 4 provides a reference for comparable applications that similar hardness ratings are used for[10]. Comparatively, the present mat stiffness rating is similar to typical midsole cushioning found in common running shoes classified as medium or hard[38,58,85].

Hardness		**Density**		**Application**
Shore A	**Shore 00**	**g/cm^3**	**kg/m^3**	
8–14	42-56	0.06-0.10	60-100	Sport Padding
11-16	50-60	0.06-0.10	60-100	Molded Insole
12-18	-	0.08-0.11	80-110	Orthopedic Shoe Insole Cushion
13-19	50-63	0.11-0.15	110-150	Molded Insole
15-22	**60-72**	0.08-0.11	80-110	**Orthopedic Insole Cushion**
16-22	**60-70**	0.09-0.12	90-120	**Molded Insole**
	65			**Present Shock Absorbent Mat**
19-27	68-78	0.11-0.14	110-140	Orthopedic Insole Cushion

Table 4: Derived from Biron et al. (2012)

PILOT DATA

When performing original pilot data, data showed surprisingly consistent values for all surface conditions and trials for the mid-foot strike runner. The heel striker however, showed changes in values across trials as well as conditions which allowed for the assumption that other heel strikers would produce similar responses to surface stiffness changes.

Data from the heel strike runner showed similar maximum GRFs for all three ground conditions (C1: 24.0 N kg^{-1}, C2: 23.1 N kg^{-1}, and C3: 24.0 N kg^{-1}). However, initial impact peak as well as the force rate to the initial impact peak values showed more variation between conditions (Force/Force Rate: C1: 17.0 N kg^{-1}/ 783 Ns kg^{-1}, C2: 18.6 N kg^{-1}/ 776 Ns kg^{-1}, and C3: 18.2 N kg^{-1}/ 723 Ns kg^{-1}). This data shows a similar trend to the impact data found by Baltich et al. (2015) with softer midsoles producing greater impact forces[5]. The external force data gathered for the pilot work are conflicting, thus there is a need for internal force assessment.

Leg stiffness and knee joint angular stiffness showed little variation between conditions. However, slight differences were seen between the greatest stiffness trial (C1) and the rubber mat trials (C2 and C3). These data showed a potential difference between more rigid surface stiffness and those that are more compliant (Figure 14).

In addition, the force at the impact peak showed a greater difference between ground conditions for patello-femoral force, as well as tibio-femoral compressive force and shear force in the knee joint (Figure 15 & 16). Force values in the knee joint during maximum GRF showed a similar trend as the impact peak however values were not as differentiated as during the impact peak.

Though the hypothesis is based on one dataset and literature reviews, the results provide enough information to question if there is significance difference in knee forces while running over various surface stiffnesses and if the type of forces differ based on the ground conditions.

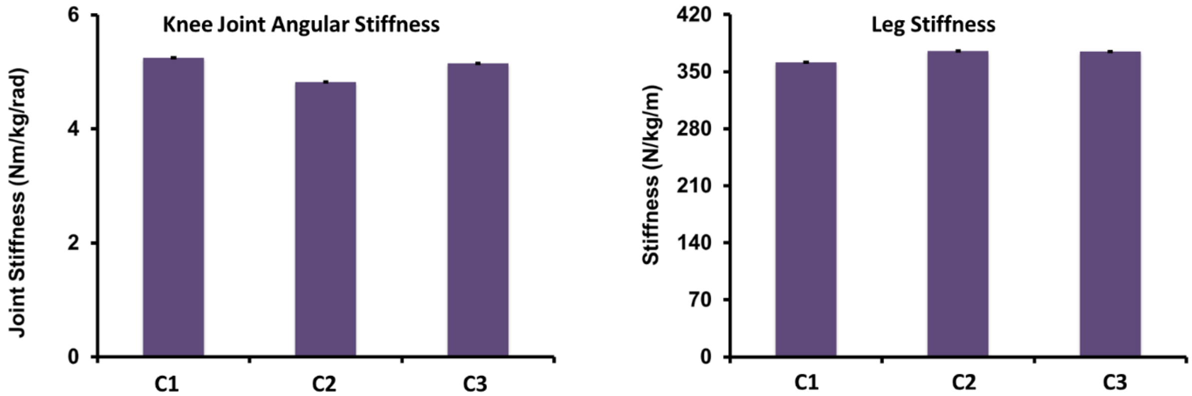

Figure 14: Pilot data knee joint angular and lower extremity stiffnesses

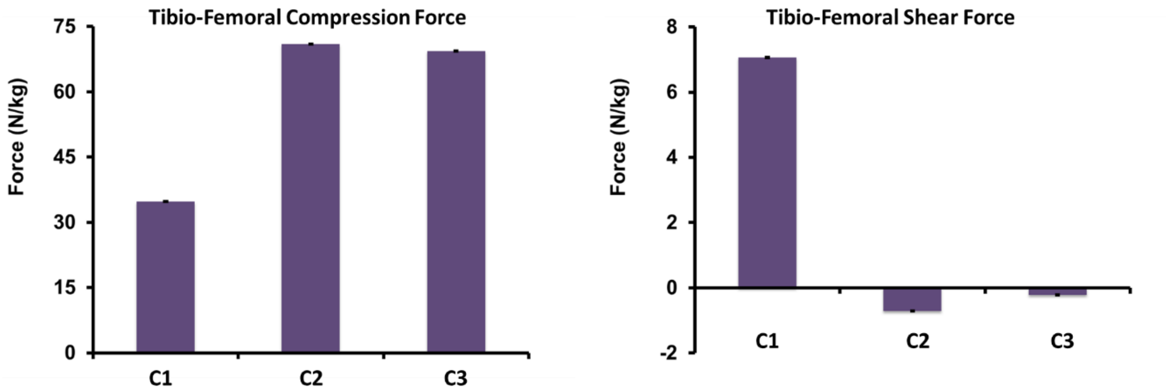

Figure 15: Pilot data knee joint tibio-femoral forces during initial impact peak

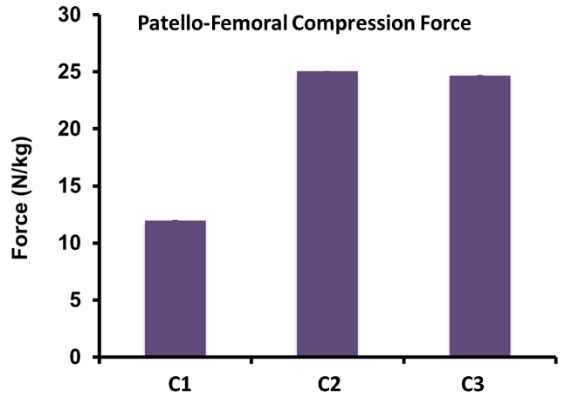

Figure 16: Pilot data knee joint patello-femoral compression force during initial impact peak

STATISTICAL ANALYSIS

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

- magnitude of and rate to the vertical impact GRF
- magnitude of and rate to maximum vertical GRF
- magnitude of the maximum braking anteroposterior GRF
- patello-femoral compressive force at the time of the vertical impact GRF and rate to this force
- maximum patello-femoral compressive force and rate to this force
- tibio-femoral compressive force at the time of the vertical impact GRF and rate to this force
- tibio-femoral shear force at the time of the vertical impact GRF and rate to this force
- maximum tibio-femoral compressive force and rate to this force
- maximum tibio-femoral shear force and rate to this force

- knee joint angular stiffness
- leg stiffness
- other selected explanatory variables

A one way ANOVA was be used to identify statistically significant differences over the three surface conditions (stiffest, medium, soft) for each variable. The alpha value will be set to less than 0.05 to find any significance between the various ground conditions, and t-tests for post hoc testing will be used to identify precise statistically significant differences if the omnibus test is significant.

Due to the lack of research investigating surface stiffness changes on knee joint forces an a priori power calculation could not be conducted. Present sample size was however substantially larger than those used by Ferris and Farley in their surface stiffness studies which had approximately six people in each study[27-31]. It was acknowledge that a low power score could increase the statistical probability of a type II error.

Chapter IV: RESULTS

This study proposed to compare tibio-femoral and patello-femoral joint contact forces while running on three different surfaces of variable stiffness. Our hypothesis was varying the ground stiffness would not make a significant difference to the force placed on the knee across different ground stiffness levels due to the joint manipulating itself to maintain the body's natural running mechanics. To evaluate this hypothesis, kinematic data, GRFs, and knee and limb parameters were examined and compared across three surface stiffness levels to determine differences in running mechanics. In addition, specific knee joint forces were evaluated as the surface stiffness changed. Knee joint forces that were focused on included patello-femoral compression force, tibio-femoral compression force, and tibio-femoral shear force.

Each category is detailed further in this chapter and is broken down into subcategories which include linear kinematics, GRFs, knee and limb parameters, and knee contact forces. Knee contact forces are further detailed into patello-femoral compression force, tibio-femoral compression force, and tibio-femoral shear force subsections. All data and statistical significance values ($p < 0.05$) for mentioned variables, determined from a one-way ANOVA test and t-tests for post hoc comparisons, can be seen in Appendix A.

LINEAR KINEMATICS

While reviewing kinematic data, speed was held within 0.5% error to the target speed for each condition. An omnibus F- test identified no significant difference between all conditions for the identified kinematic variables. As such, stride length, stride rate, and negative displacement of the pelvis while the individual was in stance phase remained similar between each condition. Knee joint flexion may have increased slightly as the surface stiffness decreased though these values were small enough in difference to be considered negligible and not statistically significant (a 0.22% increase from C1 to C2 and a 0.67% decrease from C1 to C3). Kinematic data are represented in Table 5 laying out the values seen in each condition.

Table 5: Linear Kinematic Variables

	Condition 1	Condition 2	Condition 3
Observed Speed (m/s ± SD):	3.47 ± 0.06	3.44 ± 0.07	3.45 ± 0.07
Stride Length (m ± SD):	2.49 ± 0.13	2.47 ± 0.11	2.48 ± 0.12
Stride Rate (m/s ± SD):	1.40 ± 0.07	1.40 ± 0.07	1.40 ± 0.07
Negative Pelvis Displacement in Stance Phase (m ± SD):	-0.07 ± 0.01	-0.07 ± 0.01	-0.07 ± 0.01
Knee Joint Maximum Flexion (degrees ± SD):	-44.8 ± 5.5	-44.9 ± 5.6	-45.1 ± 6.4

GROUND REACTION FORCES

When comparing values between vertical GRF magnitude at the impact peak force, the maximum vertical GRF, and the rate to the maximum vertical GRF there was little difference between the conditions (Figure 17, 18, and 19). However, an omnibus F-test identified a significant difference ($p = 0.02$) between the softest surface condition and the stiffer surface conditions for the rate to the vertical GRF impact peak (C3 was 11.54% lower than C1 ($p= 0.02$) and 7.54% lower than C2 ($p< 0.01$)) (Figure 18).

Maximum anteroposterior breaking force was also investigated and an omnibus F-test revealed a significant difference between conditions ($p < 0.01$). Post hoc tests indicated that maximum anteroposterior breaking force decreased as the surface stiffness decreased (C3 was 9.01% lower than C1 ($p < 0.01$) and 8.18% lower than C2 ($p < 0.01$)) (Figure 19 and 20).

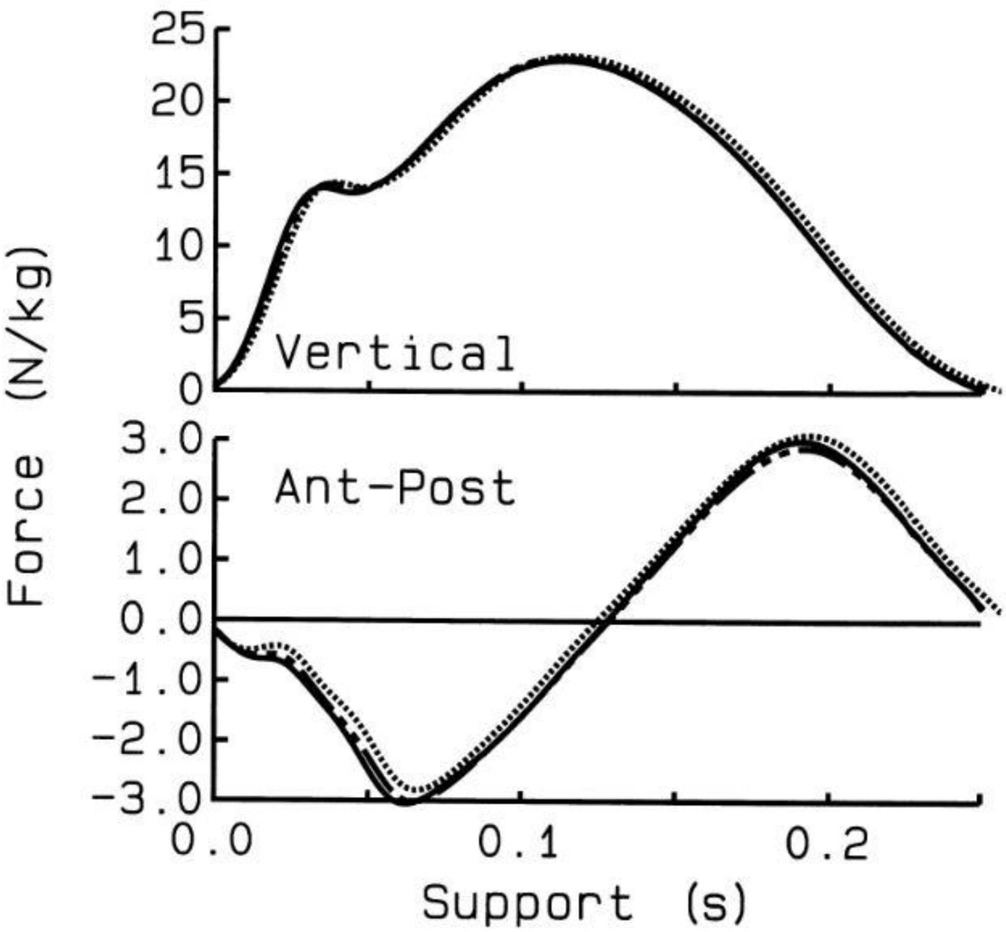

Figure 17: Ground reaction forces during stance for C1-no mat (solid line), C2- one mat (dashed line), and C3- two mats (dotted line).

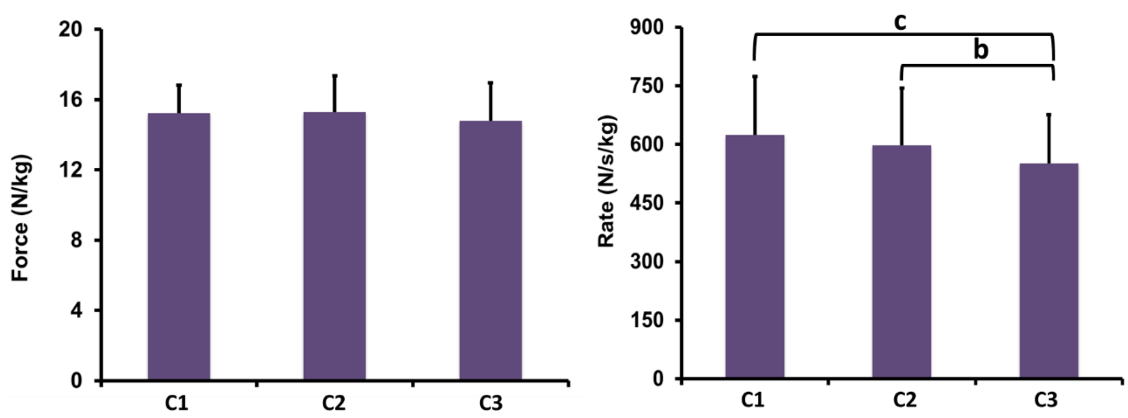

Figure 18: Maximum vertical ground reaction force at the vertical impact peak force and rate to the vertical impact peak force for over each condition. Significant differences were found between C2 and C3 (b) ($p < 0.01$) and between C1 and C3 (c) ($p = 0.02$) for rate to the impact peak.

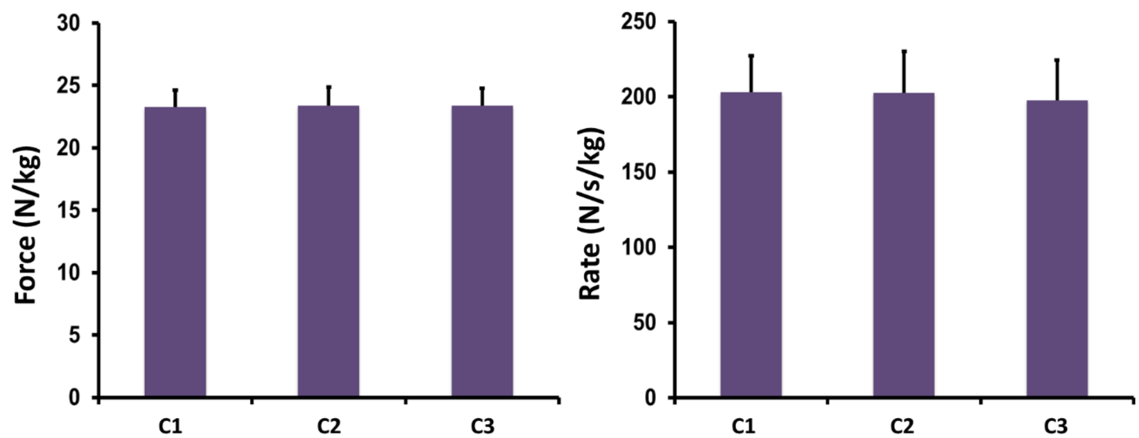

Figure 19: Maximum vertical ground reaction force during stance and rate to the maximum force for over each condition.

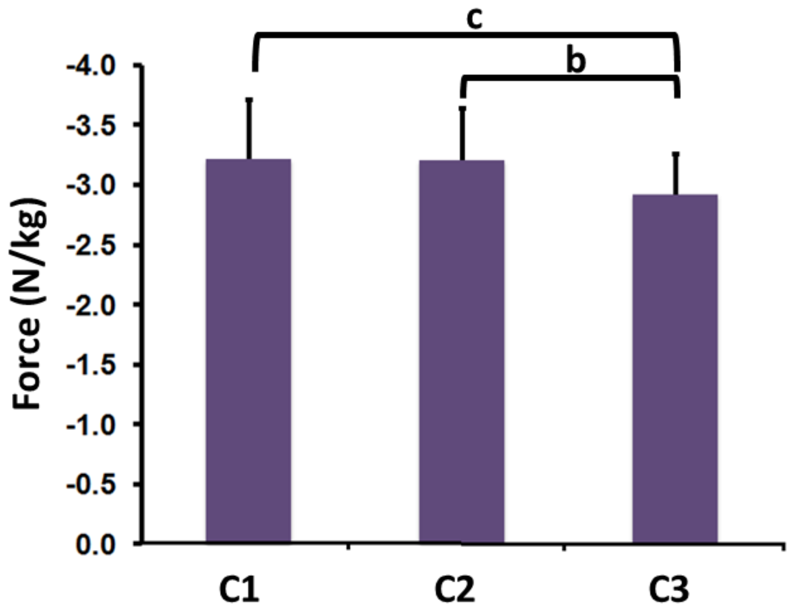

Figure 20: Maximum anteroposterior breaking force for each condition. Significant differences were found between C2 and C3 (b) (p< 0.01) and between C1 and C3 (c) (p< 0.01).

KNEE AND LIMB PARAMETERS

An omnibus F-test identified a significant difference in knee joint angular stiffness between conditions (p< 0.01). The post hoc tests showed that knee joint angular stiffness decreased as the surface stiffness decreased (C3 was 6.93% lower than C1 where C1 (p< 0.01) and 4.49% lower than C2 (p= 0.02)). Though not found to be statistically significant, knee joint maximum torque decreased as the surface stiffness decreased; a reduction of 0.69% from C1 to C2 and 2.41% from C1 to C3. Vertical leg stiffness decreased as well and was 2.19% lower from C1 to C3 as well as C2 to C3, however these differences were also not statistically significant (both p= 0.07). The data parameters for the knee and limb are represented in Table 6 outlining the values seen in each condition.

Table 6: Knee and Limb Variables.

	Condition 1	Condition 2	Condition 3
Knee Joint Maximum Torque (Nm/kg ± SD):	2.90 ± 0.29	2.88 ± 0.29	2.83 ± 0.29
Knee Joint Angular Stiffness (Nm/rad/kg ± SD):	**5.34 ± 0.51 a**	5.10 ± 0.61	**4.97 ± 0.64 b**
Leg Stiffness (N/m/kg ± SD):	365 ± 53	365 ± 58	357 ± 57

Significant differences ($p < 0.05$) between a: C1 vs C2 ($p = 0.02$) and b: C1 vs C3 ($p < 0.01$).

KNEE CONTACT FORCES

Below outlines the results derived for the following knee contact forces; patello-femoral compressive forces, tibio-femoral compressive forces, and tibio-femoral shear forces.

Patello-Femoral Compression Force

The patello-femoral compression force magnitudes and rates to the vertical impact peak and maximum force were determined through an omnibus F-test to not be statistically significant. At the vertical impact peak, patello-femoral compressive force exhibited slight reductions in value from C1 to C2 (a 2.76% decrease ($p = 0.27$)) and from C1 to C3 (a 3.45% decrease ($p = 0.26$)), decreasing as the surface stiffness decreased (Figure 21 and 22). The rate of the patello-femoral compression force to the vertical impact peak force showed no relationship to the maximum vertical impact peak force and was greater for the stiffer surface condition (C3 was 4.60% lower than C1 ($p = 0.17$) but 5.35% greater than C2 ($p = 0.42$)) (Figure 21 and 22). An increase was seen in maximum patello-femoral force as the surface stiffness decreased (C3 was 2.39% lower than C1 (p= 0.24) and 1.26% lower than C2 (p= 0.33)), though this increase was determined not to be statistically significant (Figure 23). A difference was seen between C2 and C3 for the rate to the maximum patello-femoral force (a 7.44% decrease (p< 0.01)) as the surface stiffness decreased though an omnibus F-test did not find a significant difference between conditions (p= 0.12) (Figure 23).

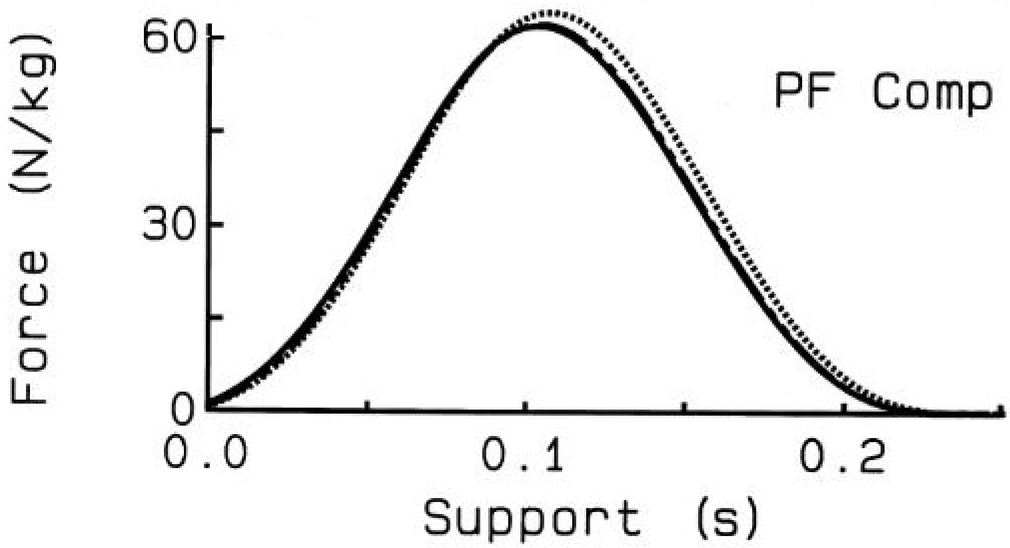

Figure 21: Patello-femoral force during stance for C1-no mat (solid line), C2- one mat (dashed line), and C3- two mats (dotted line).

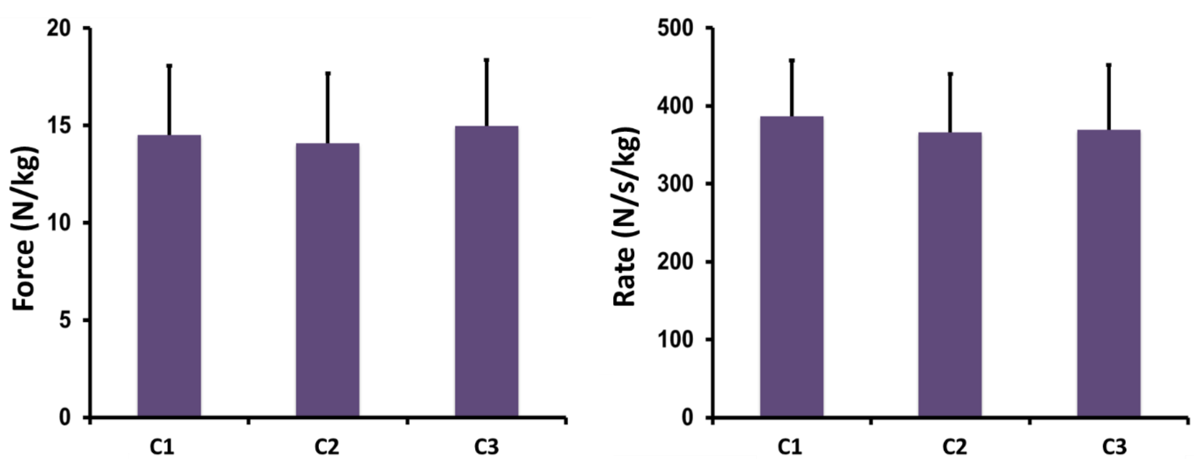

Figure 22: Maximum patello-femoral force at the vertical impact peak and rate to the vertical impact peak for over each condition.

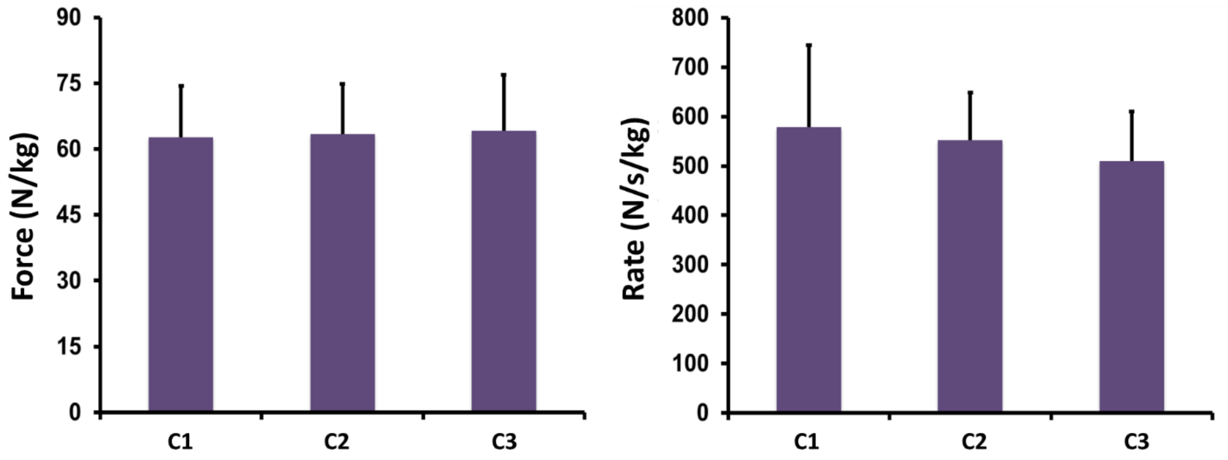

Figure 23: Maximum patello-femoral force and rate to the maximum force for over each condition.

Tibio-Femoral Compression Force

Tibio-femoral compressive force at the impact peak and the rate to this force were not statistically significant among conditions (Figure 24 and 25). Though not significant, the linear trend of the tibio-femoral compression force at the vertical impact peak force showed a slight increase as the surface stiffness decreased (a 1.48% increase from C1 to C2 ($p= 0.27$) and a 3.34% increase from C1 to C3 ($p= 0.16$)) and the rate to these values decreased as the surface stiffness decreased (C3 was 3.91% lower than C1 ($p= 0.14$) and 2.25% lower than C2 ($p= 0.21$)) (Figure 25). Similar to tibio-femoral compression force magnitude at the vertical impact peak force, maximum tibio-femoral compression force showed slight increase (C3 was 1.64% greater than C1 ($p= 0.18$) and 0.81% greater than C2 ($p= 0.32$)). The rate to the maximum tibio-femoral compressive force decreased as the surface stiffness decreased (C3 was 11.76% lower than C3 ($p= 0.07$) and 4.82% lower than C2 ($p= 0.26$)) (Figure 26). Though not statistically significant (both p= 0.16), small differences were seen between C1 and C2 in maximum tibio-femoral compression force (p= 0.04) as well as a difference between C2 and C3 or the tibio-femoral compression force rate to the maximum force (p< 0.01) (Figure 26).

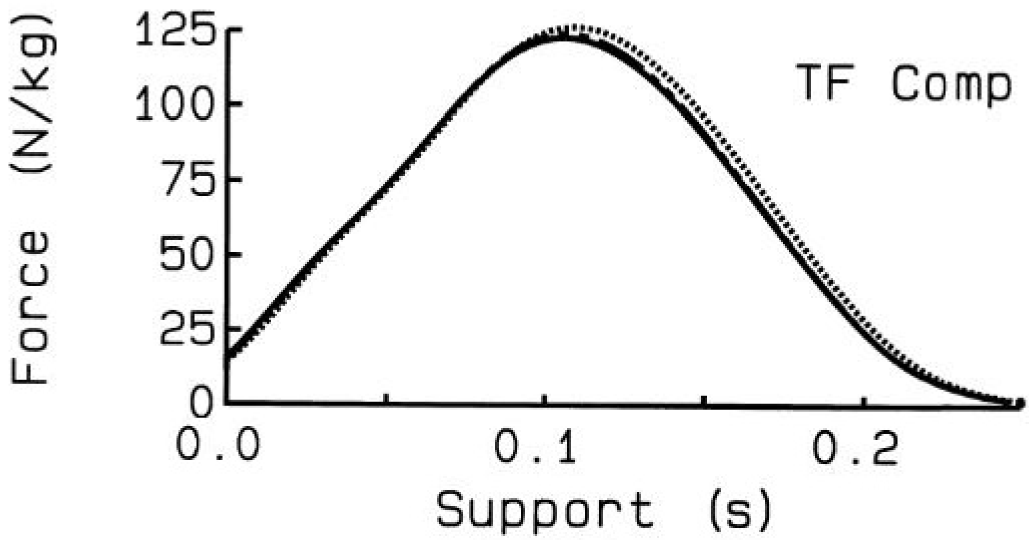

Figure 24: Tibio-femoral compression force during stance for C1-no mat (solid line), C2-one mat (dashed line), and C3- two mats (dotted line).

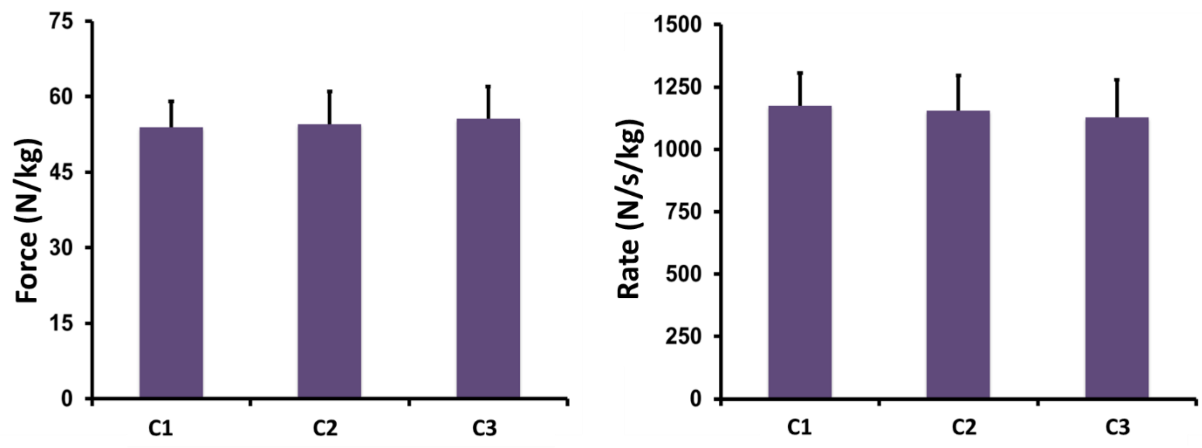

Figure 25: Maximum tibio-femoral compressive force at the vertical impact peak force and rate to the vertical impact peak force for over each condition.

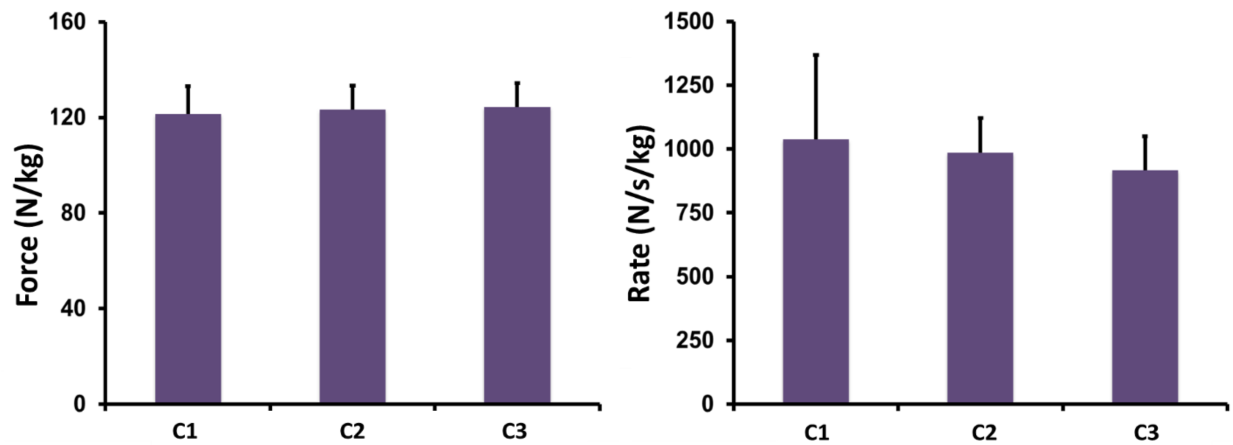

Figure 26: Maximum tibio-femoral compressive force and rate to the maximum force for over each condition.

Tibio-Femoral Shear Force

Values for the tibio-femoral shear force at the vertical impact peak between trials and among participants varied between positive and negative values. This caused the tibio-femoral shear force rate to the vertical impact peak to cancel out and made the data not reliable for comparison. These values will not be featured in this thesis. However, an omnibus F-test revealed a significant difference between conditions for both tibio-femoral shear force magnitude as well as rate to the maximum force percentile of the stance phase (both p= 0.03). The maximum tibio-femoral shear force decreased as the surface stiffness decreased and an omnibus F-test (p< 0.01) identified a significant difference with post hoc testing results showing differences between C1 and C2 (p< 0.01, a decrease of 2.75%), C2 and C3 (p= 0.04, a decrease of 2.83%), and C1 and C3 (p< 0.01, decrease of 5.50%) (Figure 27 and 28). The tibio-femoral shear force rate to maximum force also exhibited a decrease as the surface stiffness decreased and post hoc testing revealed differences between C1 and C3 (p< 0.01, a decrease of 18.3%) and C2 and C3 (p< 0.01, a decrease of 10.39%) (Figure 28) were significant.

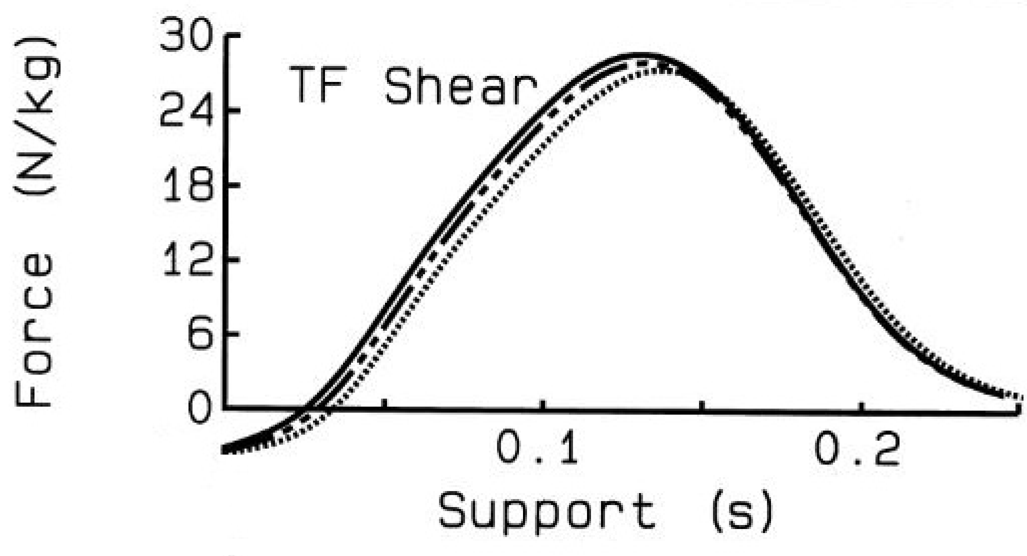

Figure 27: Tibio-femoral shear force during stance for C1-no mat (solid line), C2- one mat (dashed line), and C3- two mats (dotted line).

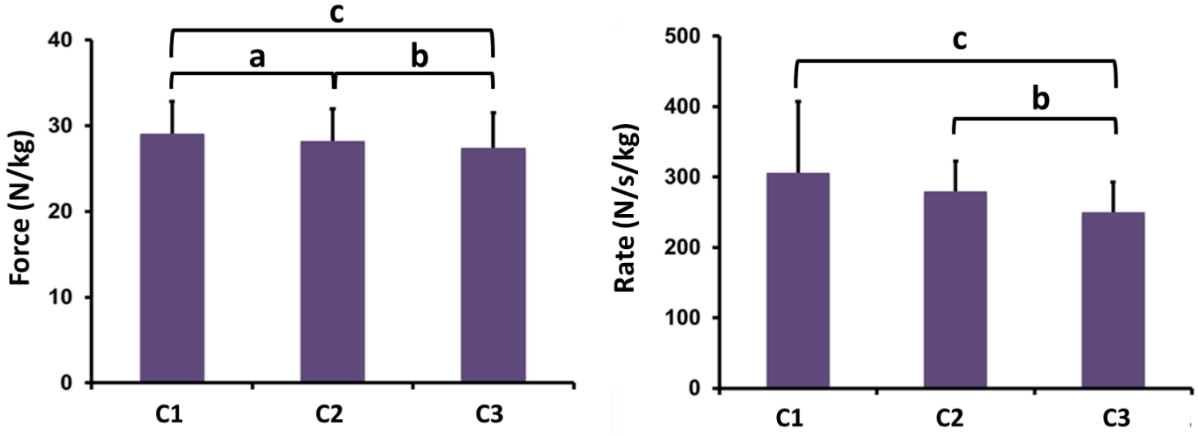

Figure 28: Maximum tibio-femoral shear force and rate to the maximum force for over each condition. Significant differences were found between C1 and C2 (a) ($p < 0.001$), C2 and C3 (b) ($p < 0.04$), and C1 and C3 (c) ($p < 0.003$) for maximum force and between C2 and C3 (b) ($p < 0.00$) and C1 and C3 (c) ($p < 0.01$) for rate to the maximum force.

SUMMARY

Knee joint angular stiffness, vertical GRF rate to vertical impact peak force, anteroposterior breaking force magnitude, and tibio-femoral shear force magnitude and rate to magnitude were the only variables to be found statistically different ($p < 0.05$). All statically significant variables decreased as the surface stiffness decreased.

Knee joint angular stiffness exhibited significant differences ($p < 0.01$) between C1 and C2 ($p < 0.02$) and C1 and C3 ($p < 0.01$), a 4.49% and 6.93% reduction. Vertical GRF rate to the vertical impact peak force showed significant differences between the most compliant surface condition and the stiffer conditions (C3 was 11.54% lower than C1 ($p < 0.015$) and 7.54% lower than C2 ($p < 0.009$)). Maximum anteroposterior breaking force showed C3 was 0.09% lower than C1 ($p < 0.01$) and 0.09% lower than C2 ($p < 0.01$). Tibio-femoral shear force was the only knee joint force which showed significant differences between conditions. Tibio-femoral shear force magnitude was statistically significant ($p < 0.01$) between all conditions (C1 and C2 ($p < 0.01$, decrease of 2.75%), C2 and C3 ($p < 0.04$, decrease of 2.83%), and C1 and C3 ($p < 0.01$, decrease of 5.50%)). The rate to maximum tibio-femoral shear force also exhibited a significant difference ($p < 0.03$) between C2 and C3 ($p < 0.01$, decrease of 10.4%) and C1 and C3 ($p < 0.01$, decrease of 18.3%).

All remaining variables were determined to be insignificant across the varying surface stiffness levels through an omnibus F-test. Kinematic data were consistent over all three surface stiffness levels and though vertical GRF rate to the vertical impact peak was significant, the magnitude to the impact peak, the maximal GRF as well as the rate to the maximum vertical GRF showed little difference between conditions. Knee joint torque decreased as the surface decreased as well as leg stiffness, however these value differences were also minor and not significant. Patello-femoral compressive force rate to maximum force between C2 and C3 ($p < 0.01$) and tibio-femoral maximum compression force between C1 and C2 ($p < 0.04$) as well as tibio-femoral compression force rate to the maximum force between C2 and C3 ($p < 0.01$) exhibited differences between conditions. Though these force data indicated differences between conditions, they were not found to be statistically significant through the omnibus F-test.

These results could be caused by a type II error due to low statistical power based on our low sample size.

Chapter V: DISCUSSION

The purpose of this study was to compare tibio-femoral and patello-femoral joint contact forces while running on three different surfaces of varying stiffness. It was hypothesized that varying the ground stiffness would not make a significant difference to the force placed on the knee across the different ground stiffness levels due to the joint manipulating itself to maintain the body's natural running mechanics. This 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

The process of human locomotion involves the foot going through a swing phase and stance phase over many cycles. While running, the swing phase of the leg experiences higher net joint torques and the muscles of the limb prepare for foot contact with the ground. Once the runner has contacted the ground, the stance phase, the ground will produce an equally opposing reaction force which sends a shock from the impact up the limb and dissipates through the muscles, tendons, and ligaments of the leg. The leg flexes at the joints to assist with dampening the forces from the ground and the body then modifies its position so that it can generate energy and power to extend into the next step. This repetitive application of loads to the joints over multiple cycles is a common cause for injury[60].

As individuals run, running mechanics alter to enable them to run as effectively as possible through various environments by the first step[30]. A common variable investigated amongst running studies involves the magnitude of the vertical GRF and how this affects the runner. Current strategies to decrease this force involve changing the runner's biomechanics in ways such as foot strike pattern, running velocity, stride length and frequency, shoe midsole cushioning, and varying running surface stiffness levels. It is known that changing foot strike patterns alters the vertical GRF and forefoot and mid-foot strike patterns produce a sinusoidal vertical GRF curve. Nevertheless, heel strike runners have an initial impact peak which occurs as the heel strikes the surface before continuing with a similar sinusoidal pattern. Aid to decrease vertical GRF has been assumed by decreasing surface stiffness, as seen in gymnastic landing mats[53]. However, when this application has been carried over to shoe midsole cushioning, the softer midsoles resulted in runners having greater vertical impact peak forces[5]. When applied to ground surface stiffness, the vertical GRF appears to decrease and running mechanics alter by changing the leg stiffness in relation to the surface stiffness[27].

Leg stiffness while running has been related to a spring-mass system which compares the leg as a linear spring with a pointed body mass[27]. As the foot strikes the surface, the downward vertical momentum is opposed by the vertical GRF which causes the leg joints to compress, storing strain energy. The leg will then rebound, or spring up through the remainder of the stride and continue into the next stride providing more energy to the runner. The more extended the joints are during the stance phase of the stride, the more rigid the leg stiffness is. Landing studies investigating the vertical GRF saw that increasing knee flexion decreased landing impact forces[24]. Though leg stiffness studies show greater flexion in the ankle and some indicate is the main joint contributor to leg stiffness when surface stiffness levels change, the knee joint assumes to be the primary bearer of impact forces which can reach up to 6-10 times the subject's body weight[4,27,34,71,96,100]. In addition, the knee joint is also a primary force dampener during running trials [65]. Multiple studies have shown that increasing the compliancy of the surface decreases vertical GRF, though will inversely increase leg stiffness which results in increased vertical GRF[27,38]. This opposing effect between surface stiffness and leg stiffness leads to the question of whether running on a softer surface provides advantageous force reductions to the leg joints. More importantly, since the knee joint's primary role during running has been found as a force dampener, and is often prone to overuse injuries, is this inverse effect eliminating beneficial effects from ground cushioning to the knee?

Many studies focus on manipulating running mechanics in order to decrease vertical GRFs which could lead to a decrease in injury risk. In addition, knee joint forces and how they are affected by these manipulations have been investigated though the effects of ground stiffness changes on knee joint forces remains unclear. Due to the importance of the knee joint during running, knowing more about the knee joint forces while running across different ground stiffnesses will allow for future improvements to injury prevention strategies.

VALIDATION OF THE KNEE JOINT MUSCULOSKELETAL MODEL

Our hypothesis focused on knee joint forces as individuals ran over varying surfaces. The musculoskeletal knee joint model used to calculate these forces, originally created and published in DeVita and Hortobagyi et al. (2001), has been previously used in multiple studies and has proven similar results to other knee joint focused studies[44,45,50,80,99]. Inverse dynamics was used for determining joint reaction forces and torques which was then used to model muscles of the leg. This allowed for further calculations to define the tibio-femoral and patello-femoral joint contact forces. The results from this study showed similar values for compression forces across studies using healthy recreational runners using different knee joint models across similar running speeds (Figures 29 and 30).

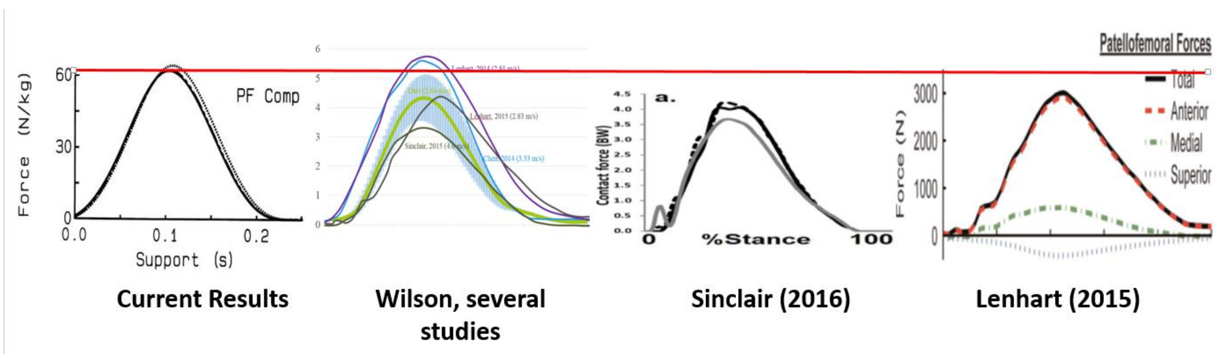

Figure 29: Patello-femoral compression force curve results compared to other running studies[15,44,45,80,97]. All graphs are fitted to the same scale on the vertical axis. The red line indicates the peak value from this study and how it compares across all graphs.

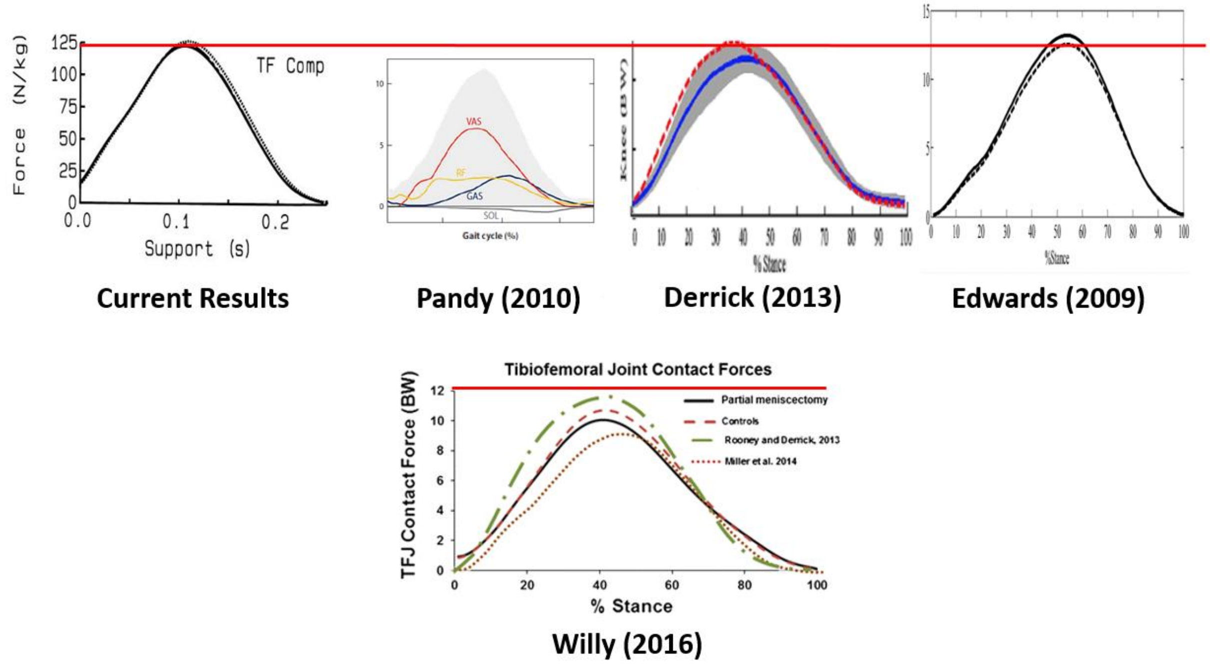

Figure 30: Tibio-femoral compression force curve results compared to other running studies[26,61,67,99]. All graphs are fitted to the same scale on the vertical axis. The red line indicates the peak value from this study and how it compares across all graphs.

Tibio-femoral shear force has commonly been measured in landing studies and has more recently been investigated in running though in the medial-lateral movement[74,86,87]. Few studies have focused on the anterior knee joint shear force and though we believe our results to be correct, validation of this force is not available at this time against other studies.

DISCUSSION OF THE RESULTS

The participants of this study were instructed to run 15 meters across varying surface stiffness levels at a consistent, moderate pace for recreational runners (3.46m/s) which was similar to other running investigations[4,11,34,76]. Though only speed was controlled, running kinematics such as stride length and stride rate were consistent over each condition. In addition, the negative displacement of the hip and vertical GRF magnitudes at the initial impact peak and maximum vertical GRF remained consistent which reflected in similar leg stiffness values across the conditions. Based on the findings of multiple studies conducted by Ferris and Farley showing an inverse effect on leg stiffness relating to surface stiffness, our results unexpectedly disagreed and our leg stiffness values did not show any significant difference across the varying conditions[29,31]. However, because vertical GRF magnitudes did not show significant differences, it would appear that leg stiffness is primarily affected by, or affects, this force accumulation. Alternatively, the sensitivity of GRF impact peak magnitude may have been negated due to the speed control in this study which could also explain the lack of differentiation between conditions[36]. The rate to the vertical impact peak force was significant which shows a potential effect caused specifically from the change in surface stiffness as it decreased. The results seen in the vertical GRF and leg stiffness could support the idea stated by Farley et al. (1998) that changes to the stride must be occurring more distally and follow the results of Gunther et al. (2002) and Seyfarth et al. (2001) that the angle and velocity of the ankle joint coming towards the ground reduces demands on the limb and prevents overextension of the knee joint[27,34,73]. This points to the importance of not just investigating the vertical position and force at foot contact, but variables in the horizontal plane as well. It is possible that surface stiffness alterations affected ankle joint function and through this effect altered the knee joint responses. However, ankle joint torques were nearly identical in all our conditions (data not reported here) and so it is unlikely that ankle joint musculature changed the knee joint loads across the conditions. In addition, the ankle joint typically is stiffer in heel strike runners and more compliant at the knee joint which was supported by our knee joint stiffness values, though not in the inverse relationship that was anticipated with the ground stiffness[37]. However, Shorten et al. (2011) similarly rejected the idea of using initial vertical GRF magnitude at the impact peak as a means of comparing cushioning effects in shoes and suggests the rate to the force magnitude as a main differentiator[75]. Our results indicate this holds true to ground stiffnesses as the cushioning of the ground surface increases, the rate to the vertical impact peak force decreases.

Though little changes were seen in the maximum vertical GRF magnitude and rate to this maximum force, anteroposterior GRF was significantly different across the conditions and this translated to the tibio-femoral shear force magnitude and rate to this force across the conditions as well. Anteroposterior breaking force magnitude and knee joint angular stiffness was significantly different between conditions ($p< 0.01$). These results indicate there are changes occurring at the knee depending on the surface condition. Surprisingly, the knee joint angular stiffness results contradict multiple studies indicating an inverse effect between surface stiffness and leg stiffness, which was assumed to be closely related to knee joint stiffness[27,29]. Sinclair et al. (2015) however, also found opposing results while investigating midsole stiffnesses and saw that leg stiffness and knee joint stiffness increased while running barefoot rather than with conventional shoes (leg joint stiffness: 610.2 Nkg/m^2 vs. 460.2 Nkg/m^2, knee joint stiffness: 7.07 Nm/kg/rad vs. 5.88 Nm/kg/rad)[79]. Sinclair associated these higher stiffness ratings to the short stance times for the barefoot condition, though this did not appear to affect leg stiffness in our study but could help account for the decrease in knee joint stiffness[55,79]. Rather than seeing an overall leg mechanical alteration to the differing surfaces in our results, the rate to the initial impact vertical GRF decreases as well as the breaking force as the surface stiffness decreased. These two force events both occurred within the first 25% of the support phase indicating these effects of altering surface stiffness were associated with the impact portion of the support phase. In addition, knee joint angular stiffness was our main mechanical alteration which decreased as the surface stiffness decreased. Due to all kinematic data showing similar results, these values indicate a slower increase in forces which also allows the limb to not need to break as much to maintain speed. The knee not only benefits from this decreased rate to force magnitude at, and shortly after, heel contact, but also through the remainder of the stride by slowing down the accumulation of shear forces, as well as shear force magnitude. Since the shoe midsole stiffness's varied for this study based on participant preference, these benefits are independent of shoe type and could be seen as a response based on surface stiffness.

With the ground stiffness decreasing, longer stance times occurred in C3 compared to C1 and C2 (C1- 0.252s, C2- 0.254s, C3- 0.259s, $p < 0.01$). Longer stance times are associated with an affected running performance and may increase cumulative load due to having a larger linear impulse per step [49,63]. Our results contradict the findings of Ferris et al. (1998) which showed no difference in stance time but did see changes in leg stiffness while individuals adjusted for different surface stiffness levels[31]. McMahon et al. (1979) did show longer foot contact times on more compliant surfaces when investigating varying running track surfaces, though found that step length increased by a factor of 1.6 and running speed was not significantly affected by the softer track surface[49]. Since significant changes in leg stiffness did not occur for our participants, though longer rates to the vertical impact peak were observed, these stance longer times on the more compliant surfaces appear to primarily benefit our runners approaching the impact peak. This could also help increase the storage of strain energy due to the knee stiffness decreasing while the tendons of the knee stretch with the decreasing surface stiffness levels. This has the potential to assist with counteracting the effects of greater amounts of muscle activity being used when landing on harder surfaces[27].

Our results also differed from other studies in regards to shoe cushioning relating to vertical GRFs. Baltich et al. (2015) found that softer midsoles showed greater impact forces in heel strike runners whereas our floor conditions showed no difference amongst the varying surface stiffnesses[5]. Knee joint stiffness was found to be significant in both studies, however Baltich also showed a greater relationship between ankle joint stiffness and midsole cushioning where ankle stiffness increased as the midsole stiffness decreased[5]. Though the knee joint was determined to be the main force dampener, the ankle joint appears to not only be the primary joint for landing adjustments and propulsion, but also the first set of defense against unnecessary GRF accumulation to the knee[65].

It can be inferred that the close to vertical orientation of the tibia as the heel strikes the ground closely relates the vertical GRF with the compressive knee joint forces and the anteroposterior GRF to tibio-femoral shear force. No significant differences were found between surface conditions for both patello-femoral and tibio-femoral compressive forces. In addition to a non-significant relationship between conditions for leg stiffness, this could lead to the conclusion that compressive forces in the knee are primarily affected by the compliance of the knee joint as the heel contacts the ground. Since step length was similar in all conditions and speed was controlled, indicating a comparable heel strike stride over each condition, this reinforces that compressive forces were not as sensitive to the surface stiffness changes as anticipated. However, our results showed beneficial responses to running on softer surfaces for tibio-femoral shear force and was seen to be the primary force dominated by surface stiffness changes. This could also be partially caused by a variable amount of extra cushion in the horizontal direction as the heel made contact with the mat surfaces. Unlike with vertical compression of the mat which has a measured 1.6cm height for C2 or 3.2cm height for C3, the horizontal compression changes each time the foot strikes the ground and may have offered additional cushion to our more compliant surfaces. These data partially support our hypothesis that varying the ground stiffness would not make a significant difference to the force placed on the knee across different ground stiffness levels. However, this was not for the reason that was assumed as it was anticipated that knee joint stiffness would be manipulated to maintain the body's natural running mechanics. Our results suggest that the horizontal plane is just as sensitive as the already assumed vertical plane when investigating knee joint forces.

Sex comparisons were not run on our data because gender differences were not our primary focus. Arguably, our knee joint force results could be affected based on sex differences in running relative to structure and size. Females have a higher tendency to develop patello-femoral pain syndrome as well as iliotibial band friction syndrome in comparison to males[84]. Studies that have investigated sex differences while running have yielded contradicting results in regards to compressive knee joint forces. Sinclair et al. (2015) found that females have significantly greater knee extension moment, knee abduction moment, patello-femoral contact force, and patello-femoral contact pressure which could account for the increase risk of injury[81]. However, Almonroeder et al (2016) found knee extension moment, patello-femoral joint stress and reaction force to be significantly greater in males, while females showed significantly greater differences in hip adduction and internal rotation[2]. In addition, Willson et al. (2015) found no significant differences at the knee joint between sexes during an exhaustive run[94].

Participant characteristics were similar with the exception of a 22% increase in mass in males than females which could have also caused discrepancies in our data due to a bottoming out effect for the heavier males. In this incidence, some of the heavier runners wouldn't have the same effects as a lighter runner and results would not be consistent. When comparing the heaviest runner (98.0kg) to the lightest (51.9kg), the lightest runner had lower values for leg and knee joint stiffness, vertical GRF at impact peak and the rate to the maximum force, knee joint flexion, and the rates to the knee joint forces at both peak forces across the ground surfaces, whereas the anteroposterior GRF and knee joint torque were higher, and knee joint force magnitudes were on average lower at impact peak but higher for the maximum joint forces. These results could indicate that the lighter runner did experience more effects from the cushioning and compensated with greater leg and joint stiffness. Contradictory, this compensation could be caused by weight effects of the individual rather than our surface changes. Though, in relation to knee joint forces, the lightest runner experienced an increase in maximum forces as the surface stiffnesses decreased whereas the heavier runner showed a decrease as the surface stiffness decreased. This relationship could also be explained by differences in running experience. On average, our lighter runner spent more time running and ran further distances than our heavier runner per week. Due to this difference in experience, between our participants as well as within our study, that experience could play a factor on results. It could be inferred that those with more experience (data not reported) can feel and respond to surfaces differently which, comparing between our lightest and heaviest runner, could be more detrimental to the knee joint.

LIMITATIONS

Participants were recruited in Greenville, North Carolina area primarily around the East Carolina University campus and it was assumed that these individuals truthfully filled out the questionnaire about their health and running activity. This study was limited by the length of the track that the participants ran across. Due to laboratory limitations, the track was 15m in length and participants were required to reach the required speed 3m prior to the embedded force plate. In addition to the limited track length, the mat layers were limited on length to fit within the track restrictions. Debatably, a longer track could have allowed the participants to have a longer chance to run on a new surface, however Ferris et al. (1999) found that the center of mass was unaffected by the change in surface stiffness with a smooth transition. Thus, upon first step onto the differing conditions, it was assumed that any mechanical alterations the participant needed to make were done so on their first step.

Ground conditions were limited to surfaces that were not unnaturally springy and mimicked outdoor surface stiffnesses as closely as possible. Due to motion camera set up and the embedded force plate, the study was limited to collecting data within the East Carolina University Biomechanics Laboratory. Participant involvement was limited as well to heel strike runners. Though natural stride was encouraged and screened for the customary impact peak in the vertical GRF curve on the first day the participants came to the lab, some changed their stride on the different surfaces while gathering data. Trials missing an impact peak were deemed as not successful. Participants could have learned to change their stride in order to complete the study protocol though the required foot strike should have been similar to their own natural stride.

In order to encourage natural running mechanics, participants were allowed to wear their own running shoes for the protocol. However by doing so, the study focus was limited to only those that wear shoes while running. Though it has been shown that minimalistic footwear does not necessarily change running mechanics, we did not feel confident that our heel strike runners would feel comfortable running across the embedded force plate at our specified speed[93]. For this reason, participants were not asked to run over the surface conditions barefoot and were requested they wear their normal running shoes. Conversely, running barefoot across the different surface conditions could have shown larger force values or differences in force attenuation.

The accuracy of the musculoskeletal knee model that was used to calculate the joint patello-femoral and tibio-femoral forces was limited on certain biological criteria. The absence of several knee ligaments as well as no co-contractions by the hip flexors and abductors during stance was assumed. In addition, our model presumed the knee joint to be frictionless and healthy.

Our participant's data was analyzed as a single group rather than separately by sex. Running discrepancies have been noted between sex and lower compressive knee joint loads are seen in those with a lower body mass[51,84]. Our male participants were an average of 22% greater in mass than our female participants. Despite this difference, the purpose for this study was to investigate changes amongst surface stiffnesses rather than sex differences so was not analyzed. In addition, our sample size may have yielded low statistical power, however, this allows for future studies to investigate gender differences on varying surface stiffness levels.

FUTURE DIRECTIONS

Future studies would benefit from investigating all foot strike patterns not just heel strikers. Our assumption was that the initial impact of the heel would cause more differentiation with knee joint forces however this was not the case. Recruiting all foot strike patterns will determine if the shear force is beneficial to all runners. Another area of interest would be to investigate knee joint forces as participants ran at their own preferred speed as well as a specified speed. Some of our participants felt the selected speed was too slow for the distance they were traveling and allowing for a preferred speed would encourage a more natural running gait not bound by speed constraints. In addition, to create a more natural environment, our study would be beneficial to be performed outside on natural running surfaces. This would require a more mobile laboratory set up with motion cameras, however using pressure insoles, this study could be taken into a real world setting to investigate joint forces.

Another avenue that would benefit from further investigation would be determining the driving force of injury; external, internal, or sensory factors. A possible discussion point mentioned previously was experienced runners were able to better sense the environmental changes and were able to respond differently than those who were novice runners. Possibly the one and two mat layers were not substantial enough to elicit a sensory or perceptional recognition of the altered environment. This lack of sensation may be the source of the steady consistent source of the leg stiffness values and may have not been able to engage the cortical commands to producing running. If an individual has a current knee joint injury, determining the stimulus for which the injury was derived from, whether it stems from subconscious sensory cues causing muscular or biomechanical changes or external environmental or individual alterations which cause changes in running mechanics, would be beneficial for future injury prevention programs.

CONCLUSIONS

In conclusion, our hypothesis was partially supported for knee joint compressive loads but not for the shear loads. As the participants ran across the increasingly softer surfaces their knee joint angular stiffness decreased. This is contrary to existing literature that suggests an inverse effect between surface stiffness and leg stiffness, which is closely related to knee joint stiffness. In addition, the rate to the vertical GRF impact peak and the anteroposterior breaking force magnitude decreased with the surface stiffness. Our data supported the idea that running across differing surface stiffnesses does not statistically alter knee joint compressive forces but can reduce knee joint shear forces. Future research should determine if this strategy is beneficial to a broader range of individuals including those with fore- and mid-foot strike patterns.

Chapter VI. REFERENCES

1. Ahmed AM, Burke DL, Hyder A. Force analysis of the patellar mechanism. *J Orthop Res*. 1987;5(1):69-85. doi: 10.1002/jor.1100050110 [doi].

2. Almonroeder TG, Benson LC. Sex differences in lower extremity kinematics and patellofemoral kinetics during running. *J Sports Sci*. 2016:1-7. doi: 10.1080/02640414.2016.1225972 [doi].

3. Andersson EA, Nilsson J, Thorstensson A. Intramuscular EMG from the hip flexor muscles during human locomotion. *Acta Physiol Scand*. 1997;161(3):361-370. doi: 10.1046/j.1365-201X.1997.00225.x [doi].

4. Arampatzis A, Bruggemann GP, Metzler V. The effect of speed on leg stiffness and joint kinetics in human running. *J Biomech*. 1999;32(12):1349-1353. doi: S0021-9290(99)00133-5 [pii].

5. Baltich J. Increased vertical impact forces and altered running mechanics with softer midsole shoes. *PloS one*. 04;10(4):e0125196; e0125196.

6. Belli A. Moment and power of lower limb joints in running. *Int J Sports Med*. 02;23(2):136; 136-141; 141.

7. Bergmann G, Bender A, Graichen F, et al. Standardized loads acting in knee implants. *PLoS One*. 2014;9(1):e86035. doi: 10.1371/journal.pone.0086035 [doi].

8. Biewener AA. Scaling body support in mammals: Limb posture and muscle mechanics. *Science*. 1989;245(4913):45-48.

9. Binnie MJ, Dawson B, Arnot MA, Pinnington H, Landers G, Peeling P. Effect of sand versus grass training surfaces during an 8-week pre-season conditioning programme in team sport athletes. *J Sports Sci*. 2014;32(11):1001-1012. doi: 10.1080/02640414.2013.879333 [doi].

10. Biron M. *Thermoplastics and thermoplastic composites.* William Andrew; 2012.

11. Boey H, Aeles J, Schutte K, Vanwanseele B. The effect of three surface conditions, speed and running experience on vertical acceleration of the tibia during running. *Sports Biomech*. 2016:1-11. doi: 10.1080/14763141.2016.1212918 [doi].

12. Bowersock CD, Willy RW, DeVita P, Willson JD. Reduced step length reduces knee joint contact forces during running following anterior cruciate ligament reconstruction but does not alter inter-limb asymmetry. *Clin Biomech (Bristol, Avon)*. 2017;43:79-85. doi: S0268-0033(17)30049-9 [pii].

13. Boyer KA, Federolf P, Lin C, Nigg BM, Andriacchi TP. Kinematic adaptations to a variable stiffness shoe: Mechanisms for reducing joint loading. *J Biomech*. 2012;45(9):1619-1624. doi: 10.1016/j.jbiomech.2012.04.010 [doi].

14. Cavanagh PR, Lafortune MA. Ground reaction forces in distance running. *J Biomech*. 1980;13(5):397-406. doi: 0021-9290(80)90033-0 [pii].

15. Chen YJ, Powers CM. Comparison of three-dimensional patellofemoral joint reaction forces in persons with and without patellofemoral pain. *J Appl Biomech*. 2014;30(4):493-500. doi: 10.1123/jab.2011-0250 [doi].

16. Clark KP, Weyand PG. Sprint running research speeds up: A first look at the mechanics of elite acceleration. *Scand J Med Sci Sports*. 2015;25(5):581-582. doi: 10.1111/sms.12520 [doi].

17. Cook SD, Kester MA, Brunet ME, Haddad RJ,Jr. Biomechanics of running shoe performance. *Clin Sports Med*. 1985;4(4):619-626.

18. Davidson PL, Wilson SJ, Wilson BD, Chalmers DJ. An approach to modeling impact energy absorption by surfaces. *J Appl Biomech*. 2009;25(4):351-359.

19. Debaere S, Delecluse C, Aerenhouts D, Hagman F, Jonkers I. Control of propulsion and body lift during the first two stances of sprint running: A simulation study. *J Sports Sci*. 2015;33(19):2016-2024. doi: 10.1080/02640414.2015.1026375 [doi].

20. Derrick TR, Hamill J, Caldwell GE. Energy absorption of impacts during running at various stride lengths. *Med Sci Sports Exerc*. 1998;30(1):128-135.

21. DeVita P, Hortobagyi T. Original research functional knee brace alters predicted knee muscle and joint forces in people with ACL reconstruction during walking. *JOURNAL OF APPLIED BIOMECHANICS*. 2001;17(4):297-311.

22. DeVita P. The selection of a standard convention for analyzing gait data based on the analysis of relevant biomechanical factors. *J Biomech*. 1994;27(4):501-508. doi: 0021-9290(94)90025-6 [pii].

23. DeVita P, Fellin RE, Seay JF, Ip E, Stavro N, Messier SP. The relationships between age and running biomechanics. *Med Sci Sports Exerc*. 2015. doi: 10.1249/MSS.0000000000000744 [doi].

24. DeVita P, Skelly WA. Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Med Sci Sports Exerc*. 1992;24(1):108-115.

25. Duda GN, Brand D, Freitag S, Lierse W, Schneider E. Variability of femoral muscle attachments. *J Biomech*. 1996;29(9):1185-1190. doi: 0021929096000255 [pii].

26. Edwards WB, Taylor D, Rudolphi TJ, Gillette JC, Derrick TR. Effects of stride length and running mileage on a probabilistic stress fracture model. *Med Sci Sports Exerc*. 2009;41(12):2177-2184. doi: 10.1249/MSS.0b013e3181a984c4 [doi].

27. Farley CT, Houdijk HH, Van Strien C, Louie M. Mechanism of leg stiffness adjustment for hopping on surfaces of different stiffnesses. *J Appl Physiol (1985)*. 1998;85(3):1044-1055.

28. Farley CT, González O. Leg stiffness and stride frequency in human running. *J Biomech*. 1996;29(2):181-186. doi: http://dx.doi.org/10.1016/0021-9290(95)00029-1.

29. Ferris DP, Farley CT. Interaction of leg stiffness and surfaces stiffness during human hopping. *J Appl Physiol (1985)*. 1997;82(1):15-22; discussion 13-4.

30. Ferris DP, Liang K, Farley CT. Runners adjust leg stiffness for their first step on a new running surface. *J Biomech*. 1999;32(8):787-794. doi: S0021-9290(99)00078-0 [pii].

31. Ferris DP, Louie M, Farley CT. Running in the real world: Adjusting leg stiffness for different surfaces. *Proc Biol Sci*. 1998;265(1400):989-994. doi: 10.1098/rspb.1998.0388 [doi].

32. Glitsch U, Baumann W. The three-dimensional determination of internal loads in the lower extremity. *J Biomech*. 1997;30(11-12):1123-1131.

33. Goss DL, Gross MT. A review of mechanics and injury trends among various running styles. *US Army Med Dep J*. 2012:62-71.

34. Gunther M, Blickhan R. Joint stiffness of the ankle and the knee in running. *J Biomech*. 2002;35(11):1459-1474. doi: S0021929002001835 [pii].

35. Hackney J, Brummel S, Newman M, Scott S, Reinagel M, Smith J. Effect of reduced stiffness dance flooring on lower extremity joint angular trajectories during a ballet jump. *J Dance Med Sci*. 2015;19(3):110-117. doi: 10.12678/1089-313X.19.3.110 [doi].

36. Hamill J, Bates BT, Knutzen KM, Sawhill JA. Variations in ground reaction force parameters at different running speeds. *Human Movement Science*. 1983;2(1–2):47-56. doi: http://dx.doi.org/10.1016/0167-9457(83)90005-2.

37. Hamill J, Gruber AH, Derrick TR. Lower extremity joint stiffness characteristics during running with different footfall patterns. *Eur J Sport Sci*. 2014;14(2):130-136. doi: 10.1080/17461391.2012.728249 [doi].

38. Hardin EC, van den Bogert AJ, Hamill J. Kinematic adaptations during running: Effects of footwear, surface, and duration. *Med Sci Sports Exerc*. 2004;36(5):838-844. doi: 00005768-200405000-00016 [pii].

39. Hasegawa H, Yamauchi T, Kraemer WJ. Foot strike patterns of runners at the 15-km point during an elite-level half marathon. *J Strength Cond Res*. 2007;21(3):888-893. doi: R-22096 [pii].

40. Herzog W, Read LJ. Lines of action and moment arms of the major force-carrying structures crossing the human knee joint. *J Anat*. 1993;182 (Pt 2)(Pt 2):213-230.

41. Hobara H, Sato T, Sakaguchi M, Sato T, Nakazawa K. Step frequency and lower extremity loading during running. *Int J Sports Med*. 2012;33(4):310-313. doi: 10.1055/s-0031-1291232 [doi].

42. Keller TS, Weisberger AM, Ray JL, Hasan SS, Shiavi RG, Spengler DM. Relationship between vertical ground reaction force and speed during walking, slow jogging, and running. *Clin Biomech (Bristol, Avon)*. 1996;11(5):253-259. doi: 0268003395000682 [pii].

43. Kerdok A, Biewener A, McMahon T, Weyand P, Herr H. Energetics and mechanics of human running on surfaces of different stiffnesses. *Journal of Applied Physiology*. February 2002;92(2):469-478.

44. Lenhart RL, Thelen DG, Wille CM, Chumanov ES, Heiderscheit BC. Increasing running step rate reduces patellofemoral joint forces. *Med Sci Sports Exerc*. 2014;46(3):557-564. doi: 10.1249/MSS.0b013e3182a78c3a [doi].

45. Lenhart RL, Smith CR, Vignos MF, Kaiser J, Heiderscheit BC, Thelen DG. Influence of step rate and quadriceps load distribution on patellofemoral cartilage contact pressures during running. *J Biomech*. 2015;48(11):2871-2878. doi: http://dx.doi.org/10.1016/j.jbiomech.2015.04.036.

46. Lieberman DE, Venkadesan M, Werbel WA, et al. Foot strike patterns and collision forces in habitually barefoot versus shod runners. *Nature*. 2010;463(7280):531-535. doi: 10.1038/nature08723 [doi].

47. Lieberman DE, Warrener AG, Wang J, Castillo ER. Effects of stride frequency and foot position at landing on braking force, hip torque, impact peak force and the metabolic cost of running in humans. *J Exp Biol*. 2015;218(Pt 21):3406-3414. doi: 10.1242/jeb.125500 [doi].

48. Ly QH, Alaoui A, Erlicher S, Baly L. Towards a footwear design tool: Influence of shoe midsole properties and ground stiffness on the impact force during running. *J Biomech*. 2010;43(2):310-317. doi: 10.1016/j.jbiomech.2009.08.029 [doi].

49. McMahon TA, Greene PR. The influence of track compliance on running. *J Biomech*. 1979;12(12):893-904. doi: http://dx.doi.org.jproxy.lib.ecu.edu/10.1016/0021-9290(79)90057-5.

50. Meardon SA, Derrick TR. Effect of step width manipulation on tibial stress during running. *J Biomech*. 2014;47(11):2738-2744. doi: 10.1016/j.jbiomech.2014.04.047 [doi].

51. Messier SP, Legault C, Loeser RF, et al. Does high weight loss in older adults with knee osteoarthritis affect bone-on-bone joint loads and muscle forces during walking? *Osteoarthritis Cartilage*. 2011;19(3):272-280. doi: 10.1016/j.joca.2010.11.010 [doi].

52. Messier SP, Legault C, Schoenlank CR, Newman JJ, Martin DF, DeVita P. Risk factors and mechanisms of knee injury in runners. *Med Sci Sports Exerc*. 2008;40(11):1873-1879. doi: 10.1249/MSS.0b013e31817ed272 [doi].

53. Mills C, Yeadon MR, Pain MT. Modifying landing mat material properties may decrease peak contact forces but increase forefoot forces in gymnastics landings. *Sports Biomech*. 2010;9(3):153-164. doi: 10.1080/14763141.2010.524244 [doi].

54. Moran KA, Clarke M, Reilly F, Wallace ES, Brabazon D, Marshall B. Does endurance fatigue increase the risk of injury when performing drop jumps? *J Strength Cond Res*. 2009;23(5):1448-1455. doi: 10.1519/JSC.0b013e3181a4e9fa [doi].

55. Morin JBJ. Effects of altered stride frequency and contact time on leg-spring behavior in human running. *J Biomech*. 2007;40(15):3341; 3341-3348; 3348.

56. Muller R, Grimmer S, Blickhan R. Running on uneven ground: Leg adjustments by muscle pre-activation control. *Hum Mov Sci*. 2010;29(2):299-310. doi: 10.1016/j.humov.2010.01.003 [doi].

57. Munro CF, Miller DI, Fuglevand AJ. Ground reaction forces in running: A reexamination. *J Biomech*. 1987;20(2):147-155.

58. Nigg BM, Bahlsen HA, Luethi SM, Stokes S. The influence of running velocity and midsole hardness on external impact forces in heel-toe running. *Journal of Biomechanics*. 1987;20(10):951-959. doi: http://dx.doi.org/10.1016/0021-9290(87)90324-1.

59. Nisell R. Mechanics of the knee. A study of joint and muscle load with clinical applications. *Acta Orthop Scand Suppl*. 1985;216:1-42.

60. Novacheck TF. The biomechanics of running. *Gait Posture*. 1998;7(1):77-95.

61. Pandy MG, Andriacchi TP. Muscle and joint function in human locomotion. *Annu Rev Biomed Eng*. 2010;12:401-433. doi: 10.1146/annurev-bioeng-070909-105259 [doi].

62. Perez-Soriano P, Llana-Belloch S, Morey-Klapsing G, Perez-Turpin JA, Cortell-Tormo JM, van den Tillaar R. Effects of mat characteristics on plantar pressure patterns and perceived mat properties during landing in gymnastics. *Sports Biomech*. 2010;9(4):245-257. doi: 10.1080/14763141.2010.537675 [doi].

63. Petersen J, Sorensen H, Nielsen RO. Cumulative loads increase at the knee joint with slow-speed running compared to faster running: A biomechanical study. *J Orthop Sports Phys Ther*. 2015;45(4):316-322. doi: 10.2519/jospt.2015.5469 [doi].

64. Powell DW, Williams DS,3rd, Windsor B, Butler RJ, Zhang S. Ankle work and dynamic joint stiffness in high- compared to low-arched athletes during a barefoot running task. *Hum Mov Sci*. 2014;34:147-156. doi: 10.1016/j.humov.2014.01.007 [doi].

65. Qiao M, Abbas JJ, Jindrich DL. A model for differential leg joint function during human running. *Bioinspir Biomim*. 2017;12(1):016015-3190/aa50b0. doi: 10.1088/1748-3190/aa50b0 [doi].

66. Rooney BD. Joint contact loading in forefoot and rearfoot strike patterns during running. *J Biomech*. 09;46(13):2201; 2201-2206; 2206.

67. Rooney BD, Derrick TR. Joint contact loading in forefoot and rearfoot strike patterns during running. *J Biomech*. 2013;46(13):2201-2206. doi: 10.1016/j.jbiomech.2013.06.022 [doi].

68. Rugg SG, Gregor RJ, Mandelbaum BR, Chiu L. In vivo moment arm calculations at the ankle using magnetic resonance imaging (MRI). *J Biomech*. 1990;23(5):495-501.

69. Sassi A, Stefanescu A, Menaspa' P, Bosio A, Riggio M, Rampinini E. The cost of running on natural grass and artificial turf surfaces. *J Strength Cond Res*. 2011;25(3):606-611. doi: 10.1519/JSC.0b013e3181c7baf9 [doi].

70. Schache AG, Blanch PD, Dorn TW, Brown NA, Rosemond D, Pandy MG. Effect of running speed on lower limb joint kinetics. *Med Sci Sports Exerc*. 2011;43(7):1260-1271. doi: 10.1249/MSS.0b013e3182084929 [doi].

71. Scott SH, Winter DA. Internal forces of chronic running injury sites. *Med Sci Sports Exerc*. 1990;22(3):357-369.

72. Serpell BG, Ball NB, Scarvell JM, Smith PN. A review of models of vertical, leg, and knee stiffness in adults for running, jumping or hopping tasks. *J Sports Sci*. 2012;30(13):1347-1363. doi: 10.1080/02640414.2012.710755 [doi].

73. Seyfarth A, Gunther M, Blickhan R. Stable operation of an elastic three-segment leg. *Biol Cybern*. 2001;84(5):365-382. doi: 10.1007/PL00007982 [doi].

74. Shimokochi Y, Ambegaonkar JP, Meyer EG. Changing sagittal-plane landing styles to modulate impact and tibiofemoral force magnitude and directions relative to the tibia. *J Athl Train*. 2016;51(9):669-681. doi: 10.4085/1062-6050-51.10.15 [doi].

75. Shorten M, Mientjes MIV. The ‘heel impact’ force peak during running is neither ‘heel’ nor ‘impact’ and does not quantify shoe cushioning effects. *Footwear Science*.

2011;3(1):41-58. http://dx.doi.org/10.1080/19424280.2010.542186. doi: 10.1080/19424280.2010.542186.

76. Silder A, Besier T, Delp SL. Running with a load increases leg stiffness. *J Biomech*. 2015;48(6):1003-1008. http://search.proquest.com/docview/1674478509?accountid=10639. doi: http://dx.doi.org/10.1016/j.jbiomech.2015.01.051.

77. Simonsen EB, Dyhre-Poulsen P, Voigt M, Aagaard P, Fallentin N. Mechanisms contributing to different joint moments observed during human walking. *Scand J Med Sci Sports*. 1997;7(1):1-13.

78. Simonsen EB, Dyhre-Poulsen P, Voigt M, Aagaard P, Sjogaard G, Bojsen-Moller F. Bone-on-bone forces during loaded and unloaded walking. *Acta Anat (Basel)*. 1995;152(2):133-142.

79. Sinclair J, Atkins S, Taylor PJ. The effects of barefoot and shod running on limb and joint stiffness characteristics in recreational runners. *J Mot Behav*. 2015:1-7. doi: 10.1080/00222895.2015.1044493 [doi].

80. Sinclair J, Richards J, Selfe J, Fau-Goodwin J, Shore H. The influence of minimalist and maximalist footwear on patellofemoral kinetics during running. *J Appl Biomech*. 2016;32(4):359-364. doi: 10.1123/jab.2015-0249 [doi].

81. Sinclair J, Selfe J. Sex differences in knee loading in recreational runners. *J Biomech*. 2015;48(10):2171-2175. doi: 10.1016/j.jbiomech.2015.05.016 [doi].

82. Small K, McNaughton LR, Greig M, Lohkamp M, Lovell R. Soccer fatigue, sprinting and hamstring injury risk. *Int J Sports Med*. 2009;30(8):573-578. doi: 10.1055/s-0029-1202822 [doi].

83. Spoor CW, van Leeuwen JL. Knee muscle moment arms from MRI and from tendon travel. *J Biomech*. 1992;25(2):201-206.

84. Taunton JE, Ryan MB, Clement DB, McKenzie DC, Lloyd-Smith DR, Zumbo BD. A retrospective case-control analysis of 2002 running injuries. *Br J Sports Med*. 2002;36(2):95-101.

85. Trudeau MB, VON Tscharner V, Vienneau J, Hoerzer S, Nigg BM. Assessing footwear effects from principal features of plantar loading during running. *Med Sci Sports Exerc*. 2015;47(9):1988-1996. doi: 10.1249/MSS.0000000000000615 [doi].

86. Tsai LC, Ko YA, Hammond KE, Xerogeanes JW, Warren GL, Powers CM. Increasing hip and knee flexion during a drop-jump task reduces tibiofemoral shear and compressive forces: Implications for ACL injury prevention training. *J Sports Sci*. 2016:1-7. doi: 10.1080/02640414.2016.1271138 [doi].

87. Valenzuela KA, Lynn SK, Noffal GJ, Brown LE. Acute effects of foot rotation in healthy adults during running on knee moments and lateral-medial shear force. *J Sports Sci Med*. 2016;15(1):50-56.

88. van der Worp MP, de Wijer A, van Cingel R, Verbeek AL, Nijhuis-van der Sanden MW, Staal JB. The 5- or 10-km marikenloop run: A prospective study of the etiology of running-related injuries in women. *J Orthop Sports Phys Ther*. 2016;46(6):462-470. doi: 10.2519/jospt.2016.6402 [doi].

89. van Gent RN, Siem D, van Middelkoop M, van Os AG, Bierma-Zeinstra SM, Koes BW. Incidence and determinants of lower extremity running injuries in long distance runners: A systematic review. *Br J Sports Med*. 2007;41(8):469-80; discussion 480. doi: bjsm.2006.033548 [pii].

90. van Gent RN, Siem D, van Middelkoop M, van Os AG, Bierma-Zeinstra SM, Koes BW. Incidence and determinants of lower extremity running injuries in long distance runners: A systematic review. *Br J Sports Med*. 2007;41(8):469-80; discussion 480. doi: bjsm.2006.033548 [pii].

91. van Mechelen W. Running injuries. A review of the epidemiological literature. *Sports Med*. 1992;14(5):320-335.

92. Vannatta CN, Kernozek TW. Patellofemoral joint stress during running with alterations in foot strike pattern. *Med Sci Sports Exerc*. 2015;47(5):1001-1008. doi: 10.1249/MSS.0000000000000503 [doi].

93. Willson JD, Bjorhus JS, Williams DS,3rd, Butler RJ, Porcari JP, Kernozek TW. Short-term changes in running mechanics and foot strike pattern after introduction to minimalistic footwear. *PM R*. 2014;6(1):34-43; quiz 43. doi: 10.1016/j.pmrj.2013.08.602 [doi].

94. Willson JD, Loss JR, Willy RW, Meardon SA. Sex differences in running mechanics and patellofemoral joint kinetics following an exhaustive run. *J Biomech*. 2015;48(15):4155-4159. doi: 10.1016/j.jbiomech.2015.10.021 [doi].

95. Willson JD, Ratcliff OM, Meardon SA, Willy RW. Influence of step length and landing pattern on patellofemoral joint kinetics during running. *Scand J Med Sci Sports*. 2015;25(6):736-743. doi: 10.1111/sms.12383 [doi].

96. Willson JD, Ratcliff OM, Meardon SA, Willy RW. Influence of step length and landing pattern on patellofemoral joint kinetics during running. *Scand J Med Sci Sports*. 2015;25(6):736-743. doi: 10.1111/sms.12383 [doi].

97. Willson JD, Sharpee R, Meardon SA, Kernozek TW. Effects of step length on patellofemoral joint stress in female runners with and without patellofemoral pain. *Clin*

Biomech (Bristol, Avon). 2014;29(3):243-247. doi: 10.1016/j.clinbiomech.2013.12.016 [doi].

98. Willy RW, Bigelow MA, Kolesar A, Willson JD, Thomas JS. Knee contact forces and lower extremity support moments during running in young individuals post-partial meniscectomy. *Knee Surg Sports Traumatol Arthrosc*. 2017;25(1):115-122. doi: 10.1007/s00167-016-4143-9 [doi].

99. Willy RW, Meardon SA, Schmidt A, Blaylock NR, Hadding SA, Willson JD. Changes in tibiofemoral contact forces during running in response to in-field gait retraining. *J Sports Sci*. 2016;34(17):1602-1611. doi: 10.1080/02640414.2015.1125517 [doi].

100. Willy RW, Meardon SA, Schmidt A, Blaylock NR, Hadding SA, Willson JD. Changes in tibiofemoral contact forces during running in response to in-field gait retraining. *J Sports Sci*. 2015:1-10. doi: 10.1080/02640414.2015.1125517 [doi].

101. Winter DA. Moments of force and mechanical power in jogging. *J Biomech*. 1983;16(1):91-97. doi: 0021-9290(83)90050-7 [pii].

102. Winter DA, Yack HJ. EMG profiles during normal human walking: Stride-to-stride and inter-subject variability. *Electroencephalogr Clin Neurophysiol*. 1987;67(5):402-411.

103. Yamaguchi GT, Sawa A, Moran D, Fessler M, Winters J. A survey of human musculotendon actuator parameters. *Multipbe Muscle Systems: Biomechanics and Movement Organization*. 1990:717-773.

104. Yamaguchi GT, Zajac FE. A planar model of the knee joint to characterize the knee extensor mechanism. *J Biomech*. 1989;22(1):1-10. doi: 0021-9290(89)90179-6 [pii].

APPENDIX A: DATA TABLES

DATA VALUES

	Condition 1	**Condition 2**	**Condition 3**
Linear Kinematics			
Observed Speed (m/s ± SD):	3.47 ± 0.06	3.44 ± 0.07	3.45 ± 0.07
Stride Length (m ± SD):	2.49 ± 0.13	2.47 ± 0.11	2.48 ± 0.12
Stride Rate (m/s + SD):	1.40 ± 0.07	1.40 ± 0.07	1.40 ± 0.07
Negative Displacement of the Pelvis in Stance Phase (m ± SD):	-0.07 ± 0.01	-0.07 ± 0.01	-0.07 ± 0.01
Knee Joint Maximum Flexion (degrees ± SD):	-44.8 ± 5.5	-44.9 ± 5.6	-45.1 ± 6.4
Knee and Limb Parameters			
Knee Joint Maximum Torque (Nm/kg ± SD):	2.90 ± 0.29	2.88 ± 0.29	2.83 ± 0.29
Knee Joint Angular Stiffness (Nm/rad/kg ± SD):	5.34 ± 0.51	5.10 ± 0.61	4.97 ± 0.64
Leg Stiffness (N/m/kg ± SD):	365 ± 53	365 ± 58	357 ± 57
GRF			
Vertical GRF Maximum at Impact Peak (N/kg ± SD):	15.2 ± 1.57	15.3 ± 2.07	14.8 ± 2.14
Vertical GRF Rate to Impact Peak (N/s/kg ± SD):	624 ± 150	597 ± 146	552 ± 125
Vertical GRF at Maximum Force (N/kg ± SD):	23.3 ± 1.37	23.4 ± 1.52	23.4 ± 1.37
Vertical GRF Rate to Maximum Force (N/s/kg ± SD):	202 ± 24.4	202 ± 27.9	197 ± 27.1
Maximum Anteroposterior Force (N/kg ± SD):	-222 ± 68	-220 ± 62	-202 ± 53
Patello-femoral Compression Force			
Force at GRF Impact Peak (N/kg ± SD):	14.5 ± 3.55	14.1 ± 3.57	15.0 ± 3.38
Rate to Force at GRF Impact Peak (N/s/kg ± SD):	386.8 ± 71.6	366.1 ± 74.8	369 ± 83.1
Maximum Force (N/kg ± SD):	62.7 ± 1.71	63.4 ± 11.4	64.2 ± 12.7
Rate to Maximum Force (N/s/kg ± SD):	579 ± 165	551 ± 96.5	510 ± 101
Tibio-Femoral Compression Force			
Force at GRF Impact Peak (N/kg ± SD):	53.9 + 5.15	54.7 + 6.48	55.7 + 6.25
Rate to Force at GRF Impact Peak (N/s/kg ± SD):	1175 + 132	1155 + 141	1129 + 150
Maximum Force (N/kg ± SD):	122 + 11.6	123 + 10.1	124 + 9.95
Rate to Maximum Force (N/s/kg ± SD):	1037 + 331	987 + 135	915 + 135

Tibio-Femoral Shear Force			
Maximum Force (N/kg + SD):	29.1 + 3.78	28.3 + 3.73	27.5 + 4.04
Rate to Maximum Force (N/s/kg + SD):	306 + 101	279 + 43.0	250 + 42.7

STATISTICAL SIGNIFICANCE

Statistical significant relationships are distinguished by bold, starred (*) values (p< 0.05).

Values indicated with (**) show significant difference amongst the conditions (p< 0.05).

T-test values may show significance between conditions however are not noted as significant due to the insignificant F-value ratio represented by the probability value (P Value).

	F-Test	T-tests		
		C1 vs C2	C2 vs C3	C1 vs C3
Linear Kinematic Variables				
Stride Length (m + SD):	0.182	0.082	0.271	0.082
Stride Rate (m/s + SD):	0.436	0.449	0.425	0.481
Negative Displacement of Pelvis in Stance Phase (m + SD):	0.120	0.244	0.082	0.057
Knee Joint Maximum Flexion (degrees + SD):	0.346	0.302	0.349	0.186
Knee & Limb Parameters				
Knee Joint Maximum Torque (Nm/kg + SD):	0.283	0.358	0.164	0.198
Knee Joint Angular Stiffness (Nm/rad/kg + SD):	**0.005 ****	**0.022 ***	0.088	**0.002 ***
Leg Stiffness (N/m/kg + SD):	0.242	0.485	0.072	0.133
GRF				
Vertical GRF Maximum at Impact Peak (N/kg + SD):	0.242	0.443	0.109	0.170
Vertical GRF Rate to Impact Peak (N/s/kg + SD):	**0.015 ****	0.087	**0.015 ***	**0.009 ***
Vertical GRF at Maximum Force (N/kg + SD):	0.417	0.286	0.483	0.367
Vertical GRF Rate to Maximum Force (N/s/kg + SD):	0.182	0.435	0.108	0.073
Maximum Anteroposterior Force (N/kg + SD):	**< 0.001 ****	0.450	**0.001 ***	**0.001 ***

Patello-femoral Compression Force				
Force at GRF Impact Peak (N/kg + SD):	0.271	0.270	0.098	0.264
Rate to Force at GRF Impact Peak (N/s/kg + SD):	0.240	0.086	0.415	0.169
Maximum Force (N/kg + SD):	0.343	0.154	0.332	0.237
Rate to Maximum Force (N/s/kg + SD):	0.122	0.252	**0.001 ***	0.055
Tibio-Femoral Compression Force				
Force at GRF Impact Peak (N/kg + SD):	0.285	0.269	0.236	0.162
Rate to Force at GRF Impact Peak (N/s/kg + SD):	0.247	0.216	0.206	0.137
Maximum Force (N/kg + SD):	0.288	**0.038 ***	0.319	0.184
Rate to Maximum Force (N/s/kg + SD):	0.156	0.260	**0.003 ***	0.072
Tibio-Femoral Shear Force				
Maximum Force (N/kg + SD):	**< 0.001 ****	**0.001 ***	**0.040 ***	**0.003 ***
Rate to Maximum Force (N/s/kg + SD):	**0.034 ****	0.133	**0.000 ***	**0.013 ***

APPENDIX B: IRB LETTER OF APPROVAL

EAST CAROLINA UNIVERSITY
University & Medical Center Institutional Review Board Office
4N-70 Brody Medical Sciences Building· Mail Stop 682
600 Moye Boulevard · Greenville, NC 27834
Office **252-744-2914** · Fax **252-744-2284** · **www.ecu.edu/irb**

Notification of Initial Approval: Expedited

From:	Social/Behavioral IRB
To:	Victoria Price
CC:	
	Paul DeVita
Date:	5/23/2016
Re:	UMCIRB 15-002387 Knee Forces while Running on Different Surfaces

I am pleased to inform you that your Expedited Application was approved. Approval of the study and any consent form(s) is for the period of 5/20/2016 to 5/19/2017. The research study is eligible for review under expedited category # 4, 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 continuing review/closure application to the UMCIRB prior to the date of study expiration. 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
FoamRunning- Spring2016-Phone Interview -Initial Health Survey.docx	Surveys and Questionnaires
Informed Consent Form - Foam Running - Spring 2016.docx	Consent Forms
Study Protocol- IRB.docx	Study Protocol or Grant Application
Subject Recruitment Announcement.docx	Recruitment Documents/Scripts
Surface Running Study Flyer 2.docx	Recruitment Documents/Scripts
Surface Running Study Flyer-Victoria.pub	Recruitment Documents/Scripts

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

APPENDIX C: INFORMED CONSENT FORM

East Carolina University

Informed Consent to Participate in Research

Title of Research Study: Biomechanics and Running Stiffness

Principal Investigator: Victoria Price

Institution/Department or Division: Department of Kinesiology

Address: 332 Ward Sports Medicine Building

Telephone #: 252-737-4563

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 you 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 and loads while running across three surfaces stiffnesses ranging from relatively hard (force plate) to relatively soft (thick rubber) at a consistent speed (7:30 min/mile). 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?

I meet the inclusion criteria and have no apparent contraindication to participating in the study. Inclusion criteria are 18-35 years old, experienced runner, non-smoker, healthy, free of skeletal, nervous, muscular, and psychological impairments, and a BMI below 35.

Are there reasons I should not take part in this research?

I understand I should not volunteer for this study if I am a smoker, under 18 years of age or over the age of 35, 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?

I 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?

On the learning day, I will be asked to complete a short health survey to determine eligibility as well as cover the consent form and any questions that may arise.

I will wear my own running shoes. I will have several minutes to run on the lab runway and become more comfortable with the testing surfaces.

On the data collection day, I will be measured for my height and weight. I will wear my own running shoes and wear tight fitting clothes provided by the researchers during the running trials. Once the tight fitting clothes are on the researchers will place small reflective balls for motion capture purposes on my hips, right thigh, right knee, right leg, and right foot. The researcher will also place small electromyography (EMG) electrodes on my right thigh and leg for muscle activity purposes. I will have several minutes to run on the lab runway and become more comfortable with the testing instrument and different surface conditions.

Once the reflective markers and EMG electrodes have been placed I will perform three sets of running trials on hard, moderate, and soft running surfaces. I will perform about 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 or low voltage EMG electrodes. I might experience some leg muscle soreness or joint pain the day after the running trials but this is typically quite rare. I also may experience temporary dry skin in areas where the EMG electrodes were placed due to the removal of dead skin and cleaning of the area. East Carolina University, the principal investigator, and all other personnel associated with this project accept no responsibility with side effects experienced during participation.

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?

I 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 me 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 I took part in this research and may see information about me: Victoria Price, the main investigator, Paul DeVita, the study coordinator, Covey Clunan and Cannon Vick the undergraduate research assistants assisting with the test.

How will you keep the information you collect about me secure? How long will you 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, I will be identified in the data records with an identity number. My name and code number will not be identified in any subsequent report or publication. The main investigator, study coordinator, and the research student will be the only persons 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?

I 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. I may contact the Principal Investigator, Victoria Price at 252-737-4563 (Monday-Friday 9am-5pm).

If I have questions about my rights as someone taking part in research, I may call the Office for Human Research Integrity (OHRI) at phone number 252-744-2914 (days, 8:00 am-5:00 pm). If I would like to report a complaint or concern about this research study, I may call the Director of the OHRI, 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 you to read the following and if you agree, you should sign this form:

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

Victoria Price, study director

Person Obtaining Consent (PRINT) **Signature** **Date**