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The Comparison of Treadmill and Overground Running with the Use of Inertial Measurement Units (IMUs)

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ABSTRACT

The Comparison of Treadmill and Overground Running with the Use of Inertial Measurement Units (IMUs)

Griffin Moon

Treadmill running has historically been utilized in the laboratory to mimic outdoor running. Recent developments in portable technology, such as Inertial Measurement Units (IMUs) allow researchers to assess runners in their natural environment. **PURPOSE:** The primary purpose of the study was to compare peak tibial acceleration, stance time, stride frequency, and rearfoot eversion velocity between treadmill and overground running (asphalt, track, and grass). The secondary purpose was to test the reliability of IMU-based estimations of maximum rearfoot eversion velocity during treadmill running. **METHODS:** Twenty subjects (Age: 22.1 ± 2.0 yrs, Mass: 70.8 ± 11.9 kg, Height: 174.5 ± 10.0 cm, 8F/12M) participated. After consent and a warm-up period, IMUs were placed on the anteromedial aspect of the right distal tibia (1600Hz) and the posterior heel cap of the right shoe (1125 Hz) to record 3D linear accelerations and angular velocities. Subjects then ran three 30-meter trials on each overground surface (grass, track, and asphalt) at their self-selected speed. A timing system was used to ensure the same running speed was used between conditions. Subsequently, all subjects ran on a treadmill while their rearfoot motion was recorded through high-speed videography (240 Hz). Only one continuous trial was performed for each subject on the treadmill. In each of these trials, the variables of stride frequency, tibial acceleration, maximum rearfoot eversion velocity, and stance time were collected from the IMUs. Maximum rearfoot eversion velocity was the only variable processed from the rearfoot motion video data. A total of 30 steps from each condition were extracted and analyzed (10 steps from each trial for overground surfaces). In our statistical analysis, separate repeated measures ANOVAs or Friedman tests were performed on the mean and variability of each variable, depending on data normality, to examine differences between surface conditions. Post-hoc analysis was performed when appropriate through either Fisher's LSD ($\alpha = 0.05$) or Wilcoxon signed-rank tests ($\alpha = 0.008$). **RESULTS:** The means of stride frequency, peak tibial acceleration, and maximum rearfoot eversion velocity were significantly different ($p < 0.001$) between surfaces. Specifically, stride frequency was the fastest during treadmill running (1.39 (0.09) strides/sec) and slowest while running on grass (1.35 (0.07) strides/sec). Peak tibial acceleration was not different between the outdoor running conditions (asphalt: 11.0 (2.7) G, grass: 10.7 (3.4) G, and track: 10.6 (2.3) G), but significantly less during treadmill running (7.9 (1.6) G). Maximum rearfoot eversion velocity was lowest on grass (394.9 (256.5) deg/s) and greatest on track (623.5 (299.3) deg/s) and asphalt (620.7 (289.1) deg/sec). There was no difference in stance time between surfaces ($p = 0.231$). The variabilities of stance time, stride frequency, maximum rearfoot eversion velocity, and peak tibial acceleration were all found to be significantly different ($p < 0.001$) between running surface. Treadmill running presented with the lowest levels of variability in stride frequency (0.016 (0.005) strides/sec), maximum rearfoot eversion velocity (70.10 (35.58) deg/s), and peak tibial acceleration (0.79

(0.32) G) across all conditions. Due to this, all variables other than stance time were significantly less variable during treadmill running than any of the overground conditions. In contrast, running on grass displayed significantly larger variabilities across all conditions in stance time (0.014 (0.009) sec), stride frequency (0.029 (0.010) strides/sec), maximum rearfoot eversion velocity (123.71 (36.16) deg/s), and peak tibial acceleration (2.41 (1.21) G). Lastly, maximum rearfoot eversion velocities acquired from the heel mounted IMU obtained a moderate reliability with the high-speed videography (ICC = 0.739, CI: 0.322 – 0.899). **CONCLUSION**: A single IMU appears to only be moderately reliable when recording eversion velocities during running, so caution may be warranted when using a single IMU to estimate rearfoot motion during running. With the use of IMUs, treadmill running has been shown to be different than overground in terms of both mean differences and variability. As a result, laboratory-based studies on running biomechanics may not truly reflect the “natural” gait used on outdoor running surfaces.

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Griffin Moon

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Chapter 1: Introduction

In 2015, running was the third most popular form of physical activity following walking and weightlifting in the U.S. for individuals 15 years or older (1). Its popularity is strongly linked to the relatively inexpensive cost, low dependency on gym or athletic memberships, and the plethora of overall health benefits it provides (2). Running is protective against many different adverse physical health conditions (i.e. hypertension, dyslipidemia, insulin sensitivity, cancer) and has been shown to reduce all-cause mortality by 30-45% (2). Additionally, running appears to have a positive effect on mental health by helping manage depression and anxiety and by increasing self-esteem (3).

Not every aspect of running is beneficial. Runners are at an increased risk of sustaining a musculoskeletal injury. In fact, Chan et al. (4) reported that approximately 37-79% of runners will experience at least one injury annually. Approximately 80% of these running-related injuries (RRIs) are due to overuse and are located in various locations in the lower extremity (5). Some common injuries with a high incidence among runners are medial tibial stress syndrome (13.6-20%), Achilles tendinopathy (9.1-10.9%), plantar fasciitis (4.5-10%), and patellofemoral tendinopathy (5.5-22.7%) (6). Due to these high incidence rates corresponding to the percentage of runners that acquire these injuries, many clinicians and researchers have devoted a considerable amount of effort to uncover their exact etiologies (7).

Because the majority of these running injuries have been attributed to overuse, the identification of distinct causes or risk factors is of major importance for researchers in the field of running biomechanics (7). Bertelson et al. (7) asserted that through the study of the risk factors and potential causes for RRIs, injury prevention and treatment protocols can be developed. Some current risk factors that have been deemed significant in the etiology of injury are running form (i.e. foot strike pattern), footwear used, running surface, range of exposure, anthropometric characteristics (i.e. foot arch structure), and gait variability (7–13). The etiology of certain injuries still remains unclear, but it is believed that due to these known risk factors, some runners display kinematic or kinetic abnormalities that may increase their risk of specific injuries (5, 8, 14, 15). Examples of kinematic and kinetic alterations include increased impact loading, excessive pronation, and increased overall stiffness of the lower extremity during stance (16–18).

For several decades now, the gold standard measuring tools for these kinematic and kinetic variables have been video analysis and force plates (19–21). Despite the accuracy and reliability of these tools, they are often expensive to acquire and are restricted to use in a laboratory setting (19, 21). Furthermore, these tools are also limited in the number of continuous trials they can record unless the subject of interest is held stationary via treadmill running or additional equipment is used (22). These limitations have led some to use more portable devices that are wearable and allow for both laboratory and in-field data collection, such as accelerometers, gyroscopes, and inertial measurement units (IMUs) (19–21, 23, 24).

Accelerometers mounted on the lower leg, specifically, have been shown to provide researchers with reliable information of the amount of impact loading a runner experiences through the measurement of linear accelerations along the three axes of the tibia (25, 26). Some studies have even reported temporal features from these accelerometers, such as time of foot strike. (26–29). Gyroscopes have also been used in the detection of foot strike, but their main purpose currently in biomechanical research is to record angular velocities around both the mediolateral and anteroposterior axes to help researchers extract characteristics of foot motion throughout stance like foot strike angle and rearfoot eversion (22, 28, 30–32). When these two devices are combined in one unit, they make up a single device called an IMU. (31, 32).

The advent of these portable devices has allowed clinicians and researchers to put a larger emphasis on “real-world” analysis of a runner’s gait than ever before and has triggered a reoccurring question in the field of biomechanics, which is whether laboratory (i.e. treadmill) running is comparable to running overground or on natural surfaces (i.e. asphalt, sidewalk, grass, trail) (33–35). This debate originally started with the seminal work of Nelson et al. (36) and Nigg et al. (37), which both purported that a runner’s kinematics on a treadmill are not the same as what they would be while running overground. Treadmill running often causes a runner to shorten their stride, increase stride frequency, decrease stance time, and land with a flatter (e.g. more plantarflexed) foot (35, 37, 38). Possibly due to these temporal and kinematic changes, it has also been shown that treadmill running produces smaller joint moments and ground reaction forces compared to running overground (26, 34).

Significance

Even though the comparison of treadmill and overground running has been done by several prior studies, very few of these studies have analyzed the differences and similarities of the two forms of running with wearable devices (10, 33, 39). Milgrom et al. (10) measured the differences in the amount of tibial strain between treadmill and overground running by surgically implanting strain gauges into the mid-diaphysis of each subjects’ tibia. As hypothesized, they found that both tibial strain and strain rates were significantly higher when the subjects ran on the overground surface (asphalt) when compared to the treadmill.

Recently, Milner et al. (26) conducted a study similar to Milgrom et al. (10), but used shank-mounted accelerometers instead of strain gauges to analyze an additional overground surface (i.e. sidewalk and grass). Even though Milner et al. (26) measured linear acceleration instead of tibial strain, the findings still showed that overground running imposes more stress on the tibia during running due to the larger magnitudes of tibial acceleration. Strohrmann et al. (33) assessed the differences between treadmill and track running with six IMUs attached on different locations of the lower extremity and wrist in order to analyze kinematic changes on each surface that develop during fatigue. Overall, their most relevant finding they provided was that overground running (i.e. track surface) resulted in significantly lower step frequencies than running on a treadmill, which corresponds with previous research (34, 35).

Moreover, Benson et al. (39) showed that changes in stride regularity in both treadmill and sidewalk running can be detected by a single lower back-mounted IMU. Unfortunately, they did not provide any further insight on other biomechanical variables. Lastly, Tao et al. (38) used plantar pressure insoles to measure spatiotemporal differences between indoor track and treadmill running in 21 collegiate cross country runners. Similar to prior studies that did not use wearable technology to compare overground and treadmill running, Tao et al. (38) reported that treadmill running significantly increased stride frequency, decreased stance time and decreased swing time.

Even though these studies established differences between overground and treadmill running using some form of wearable technology, they all have their own limitations. As previously mentioned, Milgrom et al. (10) utilized strain gauges that were inserted into the tibia of each subject, which would not be ideal for quick and friendly collection. Despite Milner et al. (26) using an accelerometer that could easily be worn by subjects, the accelerometer itself had to be hard-wired into a data logger that was also worn by each subject. Although Tao et al. (38) utilized wireless plantar pressure insoles that encouraged a more natural running gait compared to the previously discussed studies, the authors did mention that some of the subjects thought the insoles were slightly uncomfortable, therefore, could have caused changes in running technique. Even though Benson et al. (39) and Strohrmann at al. (33) used newer forms of wearable technology that allow for convenient and wireless data collection, such as IMUs, they only reported changes in stride regularity and stride frequency.

Overall, there is a paucity of research concerning the similarities and differences between lab-based (i.e. treadmill) and real-world running biomechanics. Specifically, very few studies have utilized newer forms of wearable technology, such as IMUs, to answer this recurring question in the field of running biomechanics. Despite the studies done by Strohrmann et al. (33) and Benson et al. (39), there are have been no other studies performed, to the author's knowledge, that have used IMUs to compare treadmill and overground running.

Furthermore, there have only been a couple articles published on the comparison of different running surfaces (i.e. concrete, grass, synthetic track) using wearable technology. Zrenner et al. (40) used a foot-mounted IMU to calculate the differences in foot strike angle, frontal plane range of motion, pronation velocity, and sole angle velocity. They found that all of these variables were increased when running on harder and more rigid surfaces (i.e. asphalt, synthetic track). Boey et al. (41) compared the magnitudes of tibial acceleration between concrete, synthetic track, and trail and reported that tibial acceleration was significantly higher when subjects ran on the concrete surface. Other than these two articles, there are no current studies analyzing the differences between different overground running surfaces. Therefore, there is a need to investigate if wearable technology can detect other differences between overground running surfaces, such as changes in stance time or stride frequency.

Additionally, the reliability of the IMU to assess subtalar rearfoot motion (i.e. inversion and eversion) varies among recent published research. Some authors, such as Meinert et al. (42) and Zrenner et al. (31), claim that rearfoot range of motion and velocity measurements from IMUs are in high agreement with the 'gold standard' motion analysis using high speed videography, while Koska et al. (32) and Lederer et al. (43) urge future researchers to reference frontal plane kinematics from IMUs with caution due to a perceived overestimation of rearfoot motion. Shih et al. (23) only reported a moderate correlation ($r = 0.651$) between standard videographic motion analysis data and gyroscope-obtained angular velocities in the frontal plane during running. It should be noted that even though Shih et al. (23) and Lederer et al. (43) did not use an IMU in their study, their findings are still relevant since rearfoot motion from IMUs are derived from the gyroscope component of the devices (31, 32, 42). These conflicting findings in the reliability of IMU-based measurements in the frontal plane during running indicate that there is a need to validate these measures before one can confidently infer kinematic changes in individuals from these wearable devices.

Purpose

The main purpose of this study was to use IMUs to compare spatiotemporal and kinetic factors between treadmill running and running overground on various surfaces. These factors included peak tibial acceleration, stance time, stride frequency, and rearfoot eversion velocity. The secondary purpose of this study was to test the validity of rearfoot motion data recorded by an IMU during treadmill running through the comparison of high-speed video analysis.

Specific Aims

Specific Aim 1: To utilize IMU technology to compare the means and variability of peak tibial acceleration, stance time, stride frequency, and rearfoot motion in laboratory (i.e. treadmill) and real-world running (i.e. track, asphalt, and grass) settings.

> **Hypothesis 1a**: We hypothesize that there will be no differences in peak tibial acceleration between treadmill and overground running on any of the surfaces tested.

> **Hypothesis 1b**: We hypothesize that there will be no differences in stance time between treadmill and overground running on any of the surfaces tested.

> **Hypothesis 1c**: We hypothesize that there will be no differences in stride frequency between treadmill and overground running on any of the surfaces tested.

> **Hypothesis 1d**: We hypothesize that there will be no differences in rearfoot eversion velocity between treadmill and overground running on any of the surfaces tested.

Specific Aim 2: To validate rearfoot motion data from IMU technology with high-speed video analysis while running on a treadmill.

> **Hypothesis 2**: We hypothesize there will be no differences between the rearfoot eversion velocity recorded from a heel cap mounted IMU and video analysis.

Overall Design

This study compared running parameters from running on various overground surfaces and a motorized treadmill. This study is a within-subject design. Following informed consent, each subject was asked to run on three different overground surfaces (i.e. track, asphalt, grass) and on a treadmill while equipped with two inertial measurement unit (IMU) sensors on the shank and foot. During the treadmill running trials, two-dimensional rearfoot motion was also recorded and analyzed to compare rearfoot eversion velocity measures from a foot mounted IMU and those obtained by videography. Running speed for each subject was controlled by having the subjects run on the treadmill with the average speed they utilized in the overground running conditions, which was recorded by two timing gates. The following variables were calculated from the IMUs: peak tibial acceleration, stance time, stride frequency, mean rearfoot eversion velocity, maximum rearfoot eversion velocity. Only two variables were determined from the video analysis: mean rearfoot eversion velocity, maximum rearfoot eversion velocity. To compare between running surfaces, a within-subject analysis of variance was performed on the above variables, with surface (track, asphalt, grass, and treadmill) being the depending variable. To compare rearfoot eversion velocity obtained with the new technology (IMU) and high speed video analysis, a within-subject analysis of variance was performed on variable between the IMU and video data obtained from the same trials while the subject was running on the treadmill in the lab.

Limitations of the Study

Limitations of this study included the control of running footwear, number of IMU sensors available, and the use of different treadmills. Footwear has been shown to affect a runner's kinematics and kinetics so there is a possibility that variables measured could be altered by one's habitual running footwear. This study was limited to two IMUs and after consideration of interested variables, it was decided to only include analysis of the right limb during this study. Kinematics and kinetics have been shown to change with different types of treadmills, especially when comparing motorized and non-motorized treadmills, so these changes were not accounted for in this study. However, only motorized treadmills were used for this study.

We did not control for a specific level of experience (i.e. weekly mileage) or pattern of injury. Our goal was to utilize new wearable technology (i.e. IMUs) to examine similarities or differences between in-field and laboratory running surfaces. Exclusion of participants based on running experience or other characteristics would not be relevant as this is a within-subject design to compare mechanics on different surfaces. We did ensure that all subjects were free of any cardiovascular conditions or musculoskeletal injury within the last six months.

Glossary of Terms

Achilles tendinopathy – An injury that can cause pain and swelling of the Achilles tendon, which connect calf muscles to heel bone

Iliotibial-band syndrome – An injury that causes pain or tenderness on the lateral aspect of the knee or thigh

IMU – Inertial Measurement Unit

Medial tibial stress syndrome – Pain over the anterior segment of the tibia (a.k.a. shin splints)

Patellar tendinopathy – An injury that can cause pain and tenderness inferior to the kneecap

Peak tibial acceleration (PTA) – the amount of acceleration the tibia experiences along its longitudinal axis

Plantar fasciitis – Inflammation of the soft-tissue that connects the heel bone to the toes

Rearfoot eversion – The movement of the heel or rear segment of the foot away from the midline

Rearfoot eversion velocity (EV) – The rate of change in the movement of the heel away from the midline

Stance time (ST) – The amount of time when the weight-bearing foot is in contact with the ground

Stride frequency (SF) – The number of strides within a given time interval

Stride length – Longitudinal distance between consecutive foot strikes of the same foot

Tibial acceleration – Surrogate measure of the impact forces experienced along the tibia

Chapter 2: Review of Literature

Running is a sport and a form of physical activity that has become enormously popular over the last few decades ever since its blossoming introduction as a leisure activity in the 1970s (44). It has grown at a pace of three to six million new participants every year until its peak in 2013 when the U.S. alone was estimated to have 19 million runners (2, 44). This growth in popularity has been attributed to a few different aspects of the sport. In addition to it being a convenient and inexpensive activity, it provides numerous health benefits. Running has been found to reduce the incidence of a variety of disease risk factors such as hypertension, dyslipidemia, insulin insensitivity, and obesity (2). As of result, people who run have a 45-70% decreased chance of developing cardiovascular disease (CVD) and a 30-50% reduced risk of cancer (2). Additionally, running has also been found to provide protection against conditions relating to mental and cognitive dysfunction like depression, anxiety, Alzheimer's disease, and Parkinson's disease (2, 45). With these findings, it comes with no surprise that runners have a 30-45% reduced risk of all-cause mortality and have a 3.2 years longer life expectancy than their non-running counterparts (2).

Although runners have a decreased risk of developing the previously discussed conditions, they have an increased risk of suffering from a variety of musculoskeletal injuries (2). Approximately 46.3 to 70 percent of runners will acquire some injury within a one year period (2, 46, 47). Furthermore, Van Gent et al. (48) reported that 19.4 – 75.3% of all runners will suffer at least one injury relating to the lower extremity during their participation in the sport. Due to this high prevalence, identification of the factors associated with running-related injuries has become a common area of study for many healthcare workers and researchers.

Common Running Injuries

A wide variety of musculoskeletal injuries have been reported among runners. Of particularly high incidence are: medial tibial stress syndrome (MTTS) (13.6-20%), Achilles tendinopathy (9.1-10.9%), patellar tendinopathy (5.5-22.7%), plantar fasciitis (4.5-10%), ankle sprains (10.9-15%), iliotibial-band syndrome (1.8-9.1%), tibial stress fractures (9.1%), and hamstring injuries (10.9%) (6). Interestingly, the occurrences of these injuries differ between runners of different levels and experience. For example, Arnold and Moody (5) reported that the annual injury rate for novice runners (27%) is considerably lower than more experienced long-distance and marathon runners, who report yearly injury rates of 32 and 52 percent, respectively. Furthermore, patellofemoral syndrome (anterior knee pain) has been reported in 15.6% of ultra-marathon runners, while only 5.5% of non-marathon runners suffer from this same injury (5, 6).

The etiologies of these injuries are often related to multiple different factors that do not occur in isolation (7, 49). Common risk factors include the footwear used, training regimens, anthropometric variances, running form, and even running surfaces (7). The study of these risk factors has provided great insight on why certain runners are susceptible to specific injuries and has allowed researchers and clinicians to mitigate the potential causes by formulating ways to limit the cumulative stress and strain that has been shown to increase injury rates (7).

A particular emphasis on the study of running form or running biomechanics has been made by several prior and current studies to further the understanding of these running-related injuries (7, 50). Unfortunately, the majority of running biomechanics research has been performed in a laboratory setting, which does not particularly simulate a runner's natural gait (9, 36, 37). With the advent of new wearable technologies such as inertial measurement units (IMUs), biomechanists are now able to study athletes in their real-world athletic environments

(50–52). A comparison of biomechanical movement parameters between laboratory and real-world settings is imperative because it enables biomechanists to determine the validity of previous results obtained in the laboratory.

Effect of Different Running Surfaces

Running on different surfaces and in different environments has been a topic of biomechanical research ever since the 1970s when the initial running craze began. Nelson et al. (36) performed one of the first studies to address the question of whether running on different surfaces truly made a difference in a runner's kinematics. They specifically examined the contrasting effects of running on a treadmill compared to overground. They reported that when individuals ran on a treadmill at a velocity of ~6.4 m/s, the stride length increased on average by 4.7%, which led to corresponding decreases in stride rate and increases in stance time when compared to overground running (36).

Subsequently, Nigg et al. (37) studied 22 rearfoot strikers on three different treadmills and a single overground runway. The treadmills, which varied in size and power, were implemented to test if different treadmills themselves could influence running kinematics. Their main finding was that treadmill running resulted in an average decrease in sole angle (i.e. angle between the sole of the shoe and ground at impact) of 7.9 degrees, meaning treadmill running consequently impelled runners to contact the treadmill with a flatter foot when compared to overground trials. In result, Nigg et al. (37) reported that 8 out of their 22 subjects converted from a rearfoot strike pattern to a midfoot strike pattern following the transition from overground to treadmill running.

Shortly after, Hardin et al. (9) examined the kinematic and kinetic changes of the hip, knee, and ankle of subjects running on a treadmill with three varying compliances while wearing two different types of shoes that were dissimilar in midsole stiffness. When midsole stiffness was controlled, subjects had greater hip and knee extension at impact while running on the lowest compliance (most stiff) setting. On the same setting, they also observed higher hip, knee, and ankle peak angular velocities throughout stance.

By the early 2000s, several more research groups were starting to examine the differences in running mechanics on different surfaces. Mok et al. (35) and Riley et al. (34) both reported decreases in stride length, stride time, and stance time during treadmill running, which contradicted what Nelson et al. (36) previously reported. Mok et al. (35) believed this was due to the backward motion or drag of the treadmill belt, which would assist in the extension of the leg during the stance phase and allow for shorter contact times. Consequently, the push off phase of running was shown to be altered during treadmill running as well. Mok et al. (35) observed that the ankle was less plantarflexed at toe off than what it is seen in overground running, which may explain why Riley et al. (34) reported reduced peak propulsion in anterior ground reaction forces in subjects running on a treadmill.

Interestingly, Riley et al. (34) also found reductions in peak medial ground reaction forces with no significant differences in vertical ground reaction forces in treadmill and overground running. This conflicts with Milgrom et al. (10), who used tibial implanted strain gauges to determine differences in tibial shock among treadmill and overground running conditions. Milgrom et al. (10) concluded that runners who habitually run overground experience 48-285% higher tibial axial strain rates than their “primarily treadmill” running counterparts. This was later confirmed by Milner et al. (26) who reported that sidewalk running produced 4.6 g more peak positive tibial acceleration than treadmill running. Even though these findings do not necessarily measure changes in vertical impact peak, they show that overground running exposes runners to higher tibial shock, which has been considered a surrogate measure for vertical impact loading (53–55).

These findings encouraged further research in the relationship of surface and running kinetics. Dixon et al. (15) conducted a study in which all their participants ran on three different overground surfaces that varied in stiffness. They found that running on the most compliant (least stiff) surface produced significantly lower loading rates without any significant changes in kinematics. Even though it was not considered to be statistically significant, Milner et al. (26) discovered similar trends in that average peak tibial accelerations during sidewalk running were higher in magnitude than when running on grass.

Although neither study reported alterations in participants' kinematics, some studies did report kinematic changes between various running surfaces (40, 56). Gruber et al. (56) found that when habitual rearfoot strikers are asked to run barefoot on a soft surface (i.e. foam matting), 80% of these runners will maintain a rearfoot strike pattern. When these same subjects converted to running on a hard surface (i.e. concrete floor), only 35% of the test subjects kept their habitual strike pattern, while 27.5% and 37.5% adapted a midfoot or forefoot strike pattern, respectively. Oddly enough, Zrenner et al. (40) found differing results in the changes foot strike pattern due to surface stiffness. They discovered that sole angle was directly related to surface stiffness, meaning their subjects, on average, were landing more on their heel on harder surfaces like tartan and asphalt, while landing less on their heel on softer surfaces such as grass or mulch.

These conflicting findings may be hard to justify since one study allowed runners to run shod while the other study only involved barefoot running, but they show that surface stiffness has a significant role on landing kinematics. Moreover, Zrenner et al. (40) also stated that along with their subjects having larger sole angles on harder surfaces, these rigid surfaces triggered higher values of frontal plane range of motion (i.e. eversion), eversion velocity, and sagittal range of motion at the ankle. Zrenner et al. (40) contributed these increased values to the runner's need to attenuate the elevated shock and ground reaction forces that are often associated with less compliant surfaces..

Overall, the lower extremity appears to adapt to the running environment or surface to what it is exposed, and also seems to have a collection of different mechanisms to adapt to them such as adjusting stride length, step frequency, or one's foot strike pattern (35–37, 56). Unfortunately, the specific changes in these adaptations still seem to be quite variant. The use of convenient wearable technologies such as IMUs can possibly bridge the gap between these studies with the selection of various biomechanical variables of athletes running on various surfaces.

Effects of Foot Strike Pattern

Because foot strike pattern will greatly influence tibial acceleration (54, 57), one of the key variables of this study, and because surface stiffness may alter foot strike pattern (8), a discussion of it is warranted in this thesis. There are three different types of foot-strike patterns in running: rearfoot, midfoot, and forefoot. The rearfoot strike (RFS) pattern, which is used by approximately 75 percent of all runners, is a strike pattern that involves the heel or the posterior 1/3 of the foot initializing contact with the ground with every step (58). The second most common strike pattern, used by 23 percent of runners, is the midfoot strike (MFS) pattern, which is defined when an individual lands on the ground with the middle 1/3 of their foot (58). Some may explain this by stating the posterior and anterior 1/3 of the foot lands simultaneously with every stride resulting in a more neutral ankle alignment at ground impact (59). The remaining 2 percent of runners use a strike pattern that involves them landing on the anterior 1/3 of their foot with every step, which is often called a forefoot strike (FFS) patten (58, 59).

Foot strike pattern has been studied extensively due to its potential role in the amount shock attenuation that occurs during the stance phase. Researchers speculate that the orientation of the foot at impact will determine how other parts of the human body absorb transient forces, which is an important aspect in the prevention of injury (8, 59, 60). Several studies have confirmed that rearfoot strikers experience higher impact forces and loading rates than forefoot strikers (8, 59, 60). These higher impact forces and loading rates are described by the impact transient peak, a landing characteristic predominately seen in the RFS pattern (60). Impact transient, seen in data from force plate measurements, is a kinetic variable indicating an abrupt deceleration of movement that is transformed into internal forces that propagate up the body (60). This propagation of force through the body is one reason why it is believed rearfoot strikers suffer from more running-related injuries than runners who implement other foot strike patterns (60). The initial discovery of this impact transient and its link to injury later contributed to the debate that all runners should shift to a non-RFS pattern to help avoid further injuries.

The purposeful conversion from a RFS pattern to a FFS pattern is deeply rooted with the idea that a reason people employ a RFS pattern is due to the emergence of the modern running shoe and that the RFS pattern is not a natural movement pattern for the foot during running (8). It is believed that before running shoes were manufactured, especially back in the primal times of human existence, a FFS pattern was the natural way to strike the ground (8). Some running enthusiasts and biomechanists believe that approximately 80 percent of endurance runners today use a RFS pattern because modern running shoes provide more cushion under the foot and specifically the heel, so runners are more apt to land posteriorly instead of landing on the anterior 1/3 of the foot (8, 60). Lieberman et al. (60) reported that since rearfoot strikers land more dorsiflexed than forefoot strikers, much of the translational energy of the foot is transmitted up into the lower extremity. Therefore, rearfoot strikers often experience higher frequencies of tibial stress fractures, plantar fasciitis, and knee-related injuries (59). Conversely, forefoot strikers are more efficient in terms of converting the translational energy of the foot into rotational energy so less shock is diffused into the lower limb (60). The FFS pattern usually coincides with larger degrees of knee flexion at foot contact, a reduction in step length, and greater values of rearfoot eversion excursion, which would all yield a softer impact with every step (8, 59). In result, much of the impact can be absorbed and mitigated through muscle actions and soft tissues instead of the bony mechanical structure of the lower leg. Unfortunately, this heightened demand on the muscles of the lower leg along with the repetitive striking of the forefoot segment leaves forefoot strikers more susceptible to Achilles tendinopathies and metatarsal stress fractures (8, 59).

Some believe that foot strike pattern may not be a fixed characteristic of each runner, though (8, 56). Hamill and Gruber (58) proposed that for endurance running, a RFS pattern may be ideal since it requires less metabolic energy to execute than a FFS pattern. They argued that forefoot striking is better suited for shorter bouts of running that require larger amounts of speed but would not be the most efficient when longer periods of running are required due to the higher demands of muscle action. Additionally, Daoud et al. (8) stated that a runner’s foot strike can change and be quite malleable depending on a multitude of factors such as running speed, running surface, footwear used, and fatigue. Overall, foot strike pattern has a large role in the foot’s ability to absorb the repetitive impacts of running and may even have an effect on the development of RRIs (59).

Tibial Acceleration/Deceleration

Bone fatigue fractures, commonly known as stress fractures, have been reported in approximately 50 percent of all runners and military recruits (53, 54). Interestingly, 33-55 percent of stress fractures occur in the tibia (53, 54). The exact cause of stress fractures is still quite nebulous, but several reports claim that stress fractures, especially tibial stress fractures, are due to the repetitive impact forces of running (55). When the foot strikes the ground, it is forced to decelerate until its velocity reaches zero (54). When this abrupt cease in movement of the foot and the lower limb occurs, large forces (i.e. ground reaction forces) ascend through the body. Due to the proximity of the foot and tibia to the impact of foot strike, a large magnitude of these ground reaction forces are transferred to these structures before attenuation occurs. This results in a variable that has been recently introduced as tibial acceleration or tibial shock in the field of running biomechanics (54).

Prior to the inception of tibial acceleration as a kinetic variable, impact loading of the lower extremity was typically assessed by force plate measures such as vertical average loading rate, vertical instantaneous loading rate, and impact peak (41). Even though tibial acceleration has not been shown to be strongly related with impact peak during running (54), several studies have established a positive correlation between tibial accelerations and vertical loading rates making tibial acceleration a surrogate measure for impact loading (53–55). Since tibial accelerations or increased impact loading rates have been proposed as the cause of these running-related stress fractures (41, 53, 55, 57), recent research has investigated factors that exacerbate and mitigate these tibial accelerations.

Firstly, running velocity is an important determining factor of the magnitude of tibial acceleration. Sheerin et al. (54) stated that for every 1.0 m/s increase in running velocity, there is a corresponding increase of 34 percent in tibial acceleration. Secondly, Milner et al. (53) reported that individuals with a history of tibial stress fractures and who consequently present with larger than normal tibial accelerations often have higher joint stiffnesses throughout the gait cycle. Essentially, runners who land with a stiffer lower extremity are at a greater risk of developing these stress fractures because they are not actively absorbing the impact of foot contact. Much of the impact is being attenuated through the passive components of their lower limb (i.e. bone, tendon, ligaments, heel pad) rather than eccentric loading of the musculature (54, 55). Thirdly, some believe that converting from a RFS pattern to a non-RFS pattern (i.e. FFS, MFS) may be beneficial for reducing vertical loading rate, which has been shown to be directly linked to tibial acceleration (54, 57). When a runner uses a RFS pattern, they produce an impact transient from the abrupt impact of the foot and the ground, which is typically not seen in a non-rearfoot striker (57). Without the presence of an impact transient, the vertical loading rates are minimized, potentially leading to less tibial shock (53–55).

Another proposed factor is surface condition (41, 54, 61, 62). Running on harder surfaces such as concrete often causes more shock to travel up the lower extremity than softer surfaces like trail or grass (41, 54). Additionally, Montgomery et al. (62) found that their subjects experienced smaller tibial accelerations while running on a non-motorized treadmill compared to overground. Some runners have tried to simulate these same effects by wearing maximalist running shoes, which provide more cushion in the sole of the shoe to provide softer impacts, but these do not always yield the same results as running on softer or more compliant surfaces (54).

Lastly, fatigue may have a role in the elevation of impact loading rates (55). Reenalda et al. (55) found that following of an exhausting and prolonged run, tibial accelerations significantly increased on average by 7 percent. Reenalda et al. (55) speculated this was most likely due to kinematic changes that occurred throughout the run. They reported their participants had significantly lower degrees of hip and knee flexion during mid-stance by the end of the run, which would inhibit each runner's ability to properly attenuate impact forces (55). In result, tibial acceleration has been shown to be a reliable metric for the amount impact loading a runner experiences and can be altered by a multitude of factors, such as running speed, foot strike pattern, surface compliance, and fatigue (41, 53–55, 57).

Rearfoot Motion in Running

As discussed in previous sections, the manner in which the foot and ankle interact with the ground at foot strike and through the stance phase has an enormous effect on how the body will absorb the forces of impact with every stride (8, 54, 57, 59, 60). A healthy runner will most efficiently attenuate these impact forces by striking the ground with the plantar surface of the lateral aspect of their heel, while the ankle is slightly dorsiflexed and the forefoot is supinated (63). At approximately 20 percent of stance phase, the subtalar joint will allow the foot to begin to pronate and the calcaneus to evert (16, 63). After the foot reaches maximum pronation at about 35-40 percent of stance and the ankle reaches its peak dorsiflexion at approximately 50 percent of stance, the foot starts to shift more into a supinated position (63). Between 70-90 percent of the stance phase, the subtalar joint returns to a neutral alignment where it then continues to supinate while the ankle begins to plantarflex to initiate the push off phase (63). One thing to keep in mind is that this mostly describes the motions of the foot and ankle for a rearfoot striker (63). A forefoot striker displays different kinematics, although these occur primarily at foot strike. Due to the foot landing more anteriorly, healthy forefoot strikers land in a more plantarflexed ankle position along with a more adducted foot (59, 64). Bruening et al. (64) observed that due to this more plantarflexed ankle position at impact, a forefoot striker may land with a more inverted calcaneus. Despite these landing differences, the remaining periods of stance seem to be quite similar in these different foot strike patterns.

Regardless of what foot strike pattern a runner uses, pronation of the foot is a healthy, natural mechanism that allows the body to absorb the forces of impact and limit impact loading (14, 65). Unfortunately, this same mechanism can become excessive in some individuals, which may lead to injury. In fact, excessive pronation or overpronation has been linked to running related injuries such as medial tibial stress syndrome, Achilles tendinitis, plantar fasciitis, patellofemoral disorders, iliotibial friction syndrome, and lower extremity stress fractures (16, 66, 67). Hintermann and Nigg (16) stated that ankle joint eversion, an indirect measure of pronation, was typically 2-4 degrees larger in injured runners compared to non-injured runners. Despite this, they eluded that overpronation in isolation may not always be detrimental because 40-50 percent of runners that overpronate do not acquire overuse injuries (16). This has led some to think that foot motion alone does not predispose one to injury. Perhaps there is some other motion or interaction that is going on in the lower extremity that leaves some overpronators in pain and injured, while others are unaffected. Some believe that overpronation or excessive rearfoot eversion may be detrimental to susceptible runners because these motions often produce abnormal amounts of internal tibial rotation that then cause increased external rotations of the femur (16, 65, 66, 68). These coupling actions put undue strain on the knee and other structures in the lower extremity due to the torsional forces that are created from these counter rotations (16, 66). It is possible that some overpronators may avoid injury due to their ability to minimize these internal tibial rotations during stance unlike their injured counterparts.

In order to mitigate overpronation and related injuries, several shoe manufacturers have designed footwear to minimize rearfoot motion or pronation, which are often called motion control running shoes (65, 67, 69). This specific type of shoe is designed such that the lateral and/or medial midsoles are reinforced with materials that do not have the same deformation rates as normal running shoes, so when the foot starts to overpronate, it is instantly supported and pronation should cease (65). Jafarnezhadgero et al. (68) reported that on average, the motion control shoe they used decreased peak rearfoot eversion by 3 degrees, whereas Meinert et al. (42) observed a mean decrease of 30 degrees/second in maximum pronation velocity in subjects wearing an anti-pronation shoe. This effect appears to only be seen in overpronators or runners that have an excessive amount of rearfoot eversion (67, 70, 71). Silva et al. (67) found that runners with pronation or supination within the normal ranges of motion experienced no significant changes to their rearfoot motion while wearing motion control shoes. Additionally, Malisoux et al. (71) reported significantly reduced injury rates only in runners that were classified as overpronators and not runners with normal amounts of pronation. Hintermann and Nigg (16) proposed that although medial support or motion control shoes indeed prevent excessive pronation, they may consequently lead to external rotation of the tibia when the motion control is not needed. Overall, medial arch support or motion control running shoes can resolve the issue for some runners, but it may not be the panacea for minimizing all injury.

Stride Variability in Running

It was once thought that variability in one's gait was a sign of dysfunction or a byproduct of acting out a novel motor pattern (72). This belief was constructed by the concept of "end-point" variability, which states that a biological system is to be considered robust and less prone to deterioration when the variability of the system is minimal (72). What this would mean in terms of running is that healthy runners would present with low variabilities of gait variables (i.e. stride time, stance time, joint coupling, stride length), while injury-prone runners would display high variabilities in their gait.

Ironically, some research has shown that this relationship in stride variability and injury is quite the opposite (12, 13, 72). Mann et al. (12) conducted a study where they separated their subjects either into a control group (no injury in the last 12 months) or RRI group (had an injury within last 12 months) and had them run on a treadmill at their preferred running speed. All subjects in the RRI group were considered healthy by the time they participated in the study, so they were essentially runners that had recovered from an injury. While the subjects of this study ran on the treadmill, their stride index was measured, which is a metric that indirectly determines foot strike pattern by calculating the percentage of a runner's sole that initiates contact with the ground. The findings of this study showed the runners in the control group that have remained completely healthy displayed with significantly larger degrees of variability in their foot strike pattern than the previously injured runners (12). An additional study done by Meardon et al. (13) noted similar findings when they had previously injured and healthy participants run around a 300 meter track until volitional fatigue. Even though both the previously injured and healthy runners presented with increases in stride time variability throughout their run, it was reported that the previously injured runners had a lower magnitude of variability when compared to the healthy runners.

These findings have led some researchers to believe that stride variability is beneficial to runners, since variability in movement may be a mechanism to allow repetitive loads on the body to be dispersed equally across different tissues, rather than all forces being attenuated in a similar pattern and location (12, 72). The only study that did not find higher variability in healthy runners' gaits was a study done by Heiderschiet et al. (73) where they found runners with a history of patellofemoral pain displayed with higher variabilities in stride length compared to healthy runners. This result was discussed by Hamill et al. (72) through the scope of the "loss of complexity" hypothesis, which predicts that variability is advantageous for human movement because without it, our tissues become burdened with the repetitive actions that occur during walking and running. According to this hypothesis, insufficient amounts of variability would lead to dysfunction in either athletic performance or in everyday movement. Hamill et al. (72) does admit though, that too much variability can also lead to dysfunction or injury, but unfortunately there appears to be no single optimal value of stride variability for all individuals. Either way, stride variability in running seems to be beneficial, rather than detrimental, through mechanisms that are not entirely understood quite yet.

Uses of Wearable Technology

Recently, there has been an increasing amount of research involving the use of IMUs in human gait analysis (51). This surge in the utilization of these IMU sensors is mainly due to their ability to measure kinematic and kinetic parameters in a more cost-effective and convenient manner compared to the standard tools of biomechanical analysis such as motion capture and force plates where data collection is constrained to a laboratory setting (50–52). The components of these sensors that allow for these measurements to occur typically include an accelerometer, gyroscope, and a magnetometer, all of which record data along the x, y, and z-axes of the device (50). How this translates to the use of IMUs in gait and other biomechanical analyses is by allowing researchers and clinicians the ability to measure linear accelerations, angular velocities, and orientation of certain body segments and joints at a given time during some motion or activity (52). Among all the activities that have been analyzed with IMUs, running has been one of the commonly studied considering the popularity of the sport and the need to understand the potential causes of RRIs (50–52).

Some variables that have been determined in running so far with the use of IMUs have been tibial acceleration, stride frequency, stance time, and rearfoot motion (25, 26, 43, 74–76). Prior to the use of IMUs and accelerometers, researchers and clinicians were limited in the ways they could quantify impact loading in running and other activities during in-field collection. Impact loading was often calculated using data recorded from force plates or instrumented treadmills, which are commonly restricted to laboratory settings (19, 21, 41). However, authors of recent studies have asserted that impact loading can be more conveniently determined by calculating tibial acceleration from shank-mounted IMUs or accelerometers (25, 41, 53, 61, 77).

Tibial Acceleration

To assess tibial acceleration with an IMU or accelerometer, the device is commonly affixed to the anteromedial portion of the distal tibia, with the vertical axis of the device aligned with the longitudinal axis of the tibia (18, 25, 53, 78, 79). Once data are collected, researchers often define tibial acceleration by averaging the magnitudes of the positive acceleration peaks throughout a given trial (18, 25, 41, 53, 61). These peak positive accelerations are the maximum linear accelerations either recorded along the longitudinal axis of the tibia or from the resultant of all 3-axes of the sensor (25, 53). These values represent impact loading of a given subject without the subject needing to practice striking a force plate or having access to a specialized instrumented treadmill (18, 53–55).

Stride Frequency and Duration

As previously mentioned, stride frequency and duration have been shown to be determined with an IMU by calculating the time between consecutive foot strikes (76, 80, 81).

Bergamini et al. (81), who measured stride duration with an IMU mounted on the lower back reported good agreement between their IMU estimations and the reference system (i.e. force plates) by stating the mean difference between the two systems was 0.005 seconds. Lee et al. (76) found similar results when they compared the stride durations calculated from a sacral-mounted IMU and motion analysis data. Although the correlation between the IMU and motion analysis systems weakens with increasing running speeds, the overall bias between these two systems was 0.0002 seconds with a correlation coefficient of 0.99.

Even though Bergamini et al. (81) and Lee et al. (76) specifically reported stride duration collected by IMUs, their results also show that stride frequency can be accurately measured with IMUs since stride frequency is simply the reciprocal of stride duration (80). Regardless of which variant is of interest, the process of determining stride duration or frequency with IMUs is the same (76, 80, 81). Once all foot strike events are detected in each gait cycle, stride duration can simply be determined by averaging the time between consecutive foot strike events and reciprocating these durations to yield stride frequency (80).

Stance Time

To determine stance time from IMU-based data, accurate detection of both foot strike and toe off is a necessity. Foot strike is defined as the instant when the foot initiates contact with the ground, while toe off is the moment when the foot leaves the ground to begin the swing phase. The method used to detect these events is often unique to a certain study since there is not a singular correct approach used in the field to locate foot strike and toe off in running with IMU sensors. For example, Schmidt et al. (82) considered the local minimum that occurred before each impact peak (> 5g) in the vertical accelerometer data as foot strike and the local minimum that was approximately 90 ms after each foot contact as toe off. In contrast, Chew et al. (82)

classified the first local minimum in anteroposterior accelerometer data as foot strike, while toe off was considered to be the following local minimum in the same signal. Once each foot strike and toe off event are detected, stance time can be simply be calculated by finding the elapsed time between these two events (22, 82). Overall, IMU's and other wearable technology have been determined reliable for calculating stance time in running (22, 24, 82). Both Chew et al. (82) and Schmidt et al. (24) showed that the mean difference between IMU-based calculations of stance time and the stance time determined with optical analysis was 6.09 ± 6.19 milliseconds (ms) and 4.3 ± 3.2 ms, respectively.

Rearfoot Motion

Rearfoot motion is another interesting feature that has been studied with the use of wearable technology such as IMUs and gyroscopes (23, 32, 40, 42, 43, 83). The most common outcome variables that are evaluated through the recording of these devices are eversion and eversion velocity during the stance phase of running. Data for both of these variables are typically acquired from placing either an IMU or a gyroscope on the heel cap of a runner's shoe or by placing a sensor on the forefoot segment where it rests over the shoelaces (23, 32, 83). These placements have been shown to be popular locations since the device (i.e. IMU, gyroscope) can be aligned with anteroposterior axis of the foot, which would allow the sensor to record frontal plane movements like eversion and eversion velocity (23, 32, 83). The reliability of these frontal plane measurements has been quite mixed throughout the current literature. Some authors, Koska et al. (32) and Lederer et al. (43), found that rearfoot motion measured with these wearable devices often overestimate the true movements of the foot, which led to them to conclude that all researchers should interpret their findings of both eversion and eversion velocity with caution anytime they use devices like IMUs or gyroscopes. Other researchers like

Mitschke et al. (83) and Meinert et al. (42) have found strong correlations ($0.7 < r \leq 0.9$) between rearfoot motion data from wearable devices and their reference systems. Shih et al. (23) actually found that eversion velocities recorded by a forefoot mounted gyroscope only presented with a moderate correlation ($r = 0.651$) with the true forefoot motion recorded with a 3D Vicon motion analysis system. Despite this, Shih et al. (23) also reported that the gyroscope and Vicon systems were highly correlated ($r = 0.975$) when measuring the total ankle ROM throughout each stance phase. Due to these conflicting results in the current literature, it's difficult to determine if wearable technology can accurately measure rearfoot motion in running at this time.

Chapter 3: Methods

Subjects

Twenty runners (8 female, 12 male) with varying levels of experience (i.e. weekly mileage (4.3 mi (SD 5.6 mi), sport participation) and running form (foot strike pattern) were included in this study. This gave the study a broad spectrum of subjects to better represent that wide range of different types of runners (i.e. amateur, competitive). All subjects were 18 years or older (age: 22.1 yr. (SD 2.0 yr.); height: 174.5 cm (SD 10.0 cm); mass: 70.8 kg (SD 11.9 kg)) and were free from any musculoskeletal injuries and cardiopulmonary conditions that may have limited their ability to participate in the study. All subjects were either current students at West Virginia University or residents of the Morgantown, West Virginia area and were recruited by word of mouth.

Procedures

Data collection for this study occurred in two different locations at West Virginia University. All the overground running conditions were recorded at a public track facility that is located on the edge of the Evansdale campus, while the treadmill running conditions were held on the 8th floor of the Health Sciences building. The public track facility was deemed as a good location for the overground running conditions since it contained all three running surfaces (track, asphalt, and grass) that were selected for this study. Since two distinct locations were used, data collection was split into two sessions that occurred on the same day. The first session always consisted of the subjects running on the three selected overground surfaces, which meant the second session was designated for the treadmill running condition. Prior to the first session, each subject was asked to wear athletic/running attire for all the running conditions (i.e. running shoes, shorts, athletic top).

Once each subject arrived at the track facility for the first session of data collection, they were briefed on the risks and procedures of the study, which was followed by the completion of an informed consent form. During this briefing, all subjects were encouraged to ask questions or bring forth any concerns they may have about the nature of the study. After consent was obtained, each subject was given an additional form that asked for their anthropometric data (e.g., age, height, weight, gender) along with a few survey questions that would give the researchers more insight on each subject's running experience (i.e., weekly mileage, common running surface used, prior injuries, running lessons).

Following the completion of the survey, subjects were equipped with two Vicon (Culver City, CA) Blue Trident IMU sensors on the right leg that comprise of three separate components: accelerometer (± 16g, ± 200g), gyroscope (± 2000 deg/s), and magnetometer (± 4900 µT). One of the tri-axial IMU sensors was placed on the antero-medial aspect of the distal tibia and was secured with a propriety strap that was provided by Vicon. This shank mounted IMU was oriented so the y-axis of the sensor was aligned with the longitudinal axis of the tibia. The second IMU was placed on the heel cap of the right shoe so the y-axis of the sensor extended along the mediolateral axis of the foot, while the z-axis closely represented the anteroposterior axis of the foot. This sensor was held in place with masking tape which spanned across the heel cap of the shoe.

After ensuring both sensors were securely attached to the shank and shoe, each subject performed 3 running trials at a self-selected speed for all three overground surfaces (i.e., track, grass, asphalt). The order of the overground surfaces was randomized for each subject by using a

Latin Squares design. The running speed of the first trial for the first overground surface was used as a reference for all other trials throughout the study for each subject. This was controlled by asking the subject to select a speed that they would use as a typical running pace. Since the running experience among subjects was quite varied, different verbal cues were used to capture this typical running pace. Some of the experienced runners selected the pace they would use for a 5k, while other subjects had to find their pace by selecting a speed that they either considered to be a slow to moderate run (average speed among subjects, 3.63 m/s (SD 0.44 m/s)). The subject then performed a practice trial by running a total of 40 meters on the first overground surface, while the elapsed time between the 10 and 40-meter mark was recorded by a Dashr timing system (Dashr 2.0, Lincoln, NE) to determine their running velocity. This 30-meter interval was selected because it gave each runner an ample amount of time to perform approximately 10 gait cycles within each trial for all overground surfaces. The subject was instructed to run 10 meters prior the timing gaits to ensure the subject was utilizing a stable gait during the 30-meter window.

Once the running speed of the subject was determined, the subject was asked to maintain that same running speed for each of the trials on all three overground surfaces. Following each trial, the running speed was calculated to verify whether the subject was running with the correct speed. If their running speed was more than ± 5% of their initially preferred speed, the subject was asked to redo the trial. During this time, the foot strike pattern of each subject was visually inspected and was documented to assist in the analysis of each subject's data. All subjects were encouraged to take breaks in between each trial to prevent the onset of fatigue. During all recorded trials, the accelerometer component of each IMU sensor recorded at a sampling frequency of 1600 Hz, meanwhile the gyroscope component recorded at 1125 Hz. The data acquired from the magnetometer were not used in this study. This format was used for each subject that participated in this study.

For the second session of data collection, all subjects ran on a motorized treadmill that was located in the Human Performance Lab on the 8th floor of the Health Sciences building. All subjects were asked to wear the same running shoes they wore in the overground trials to help prevent any possible changes in their gait or shock attenuation that can be attributed to their footwear. Once each subject arrived at the Human Performance Lab, two Vicon IMU sensors were placed on the right lower extremity in the same locations and orientations that were used in the overground running conditions.

Following this, four reflective markers were placed on the posterior leg and running shoe heel cap according to the method outlined by Clarke et al. (84). This was done by placing an anthropometer (Lafayette Instrument Company, Model 01291) on the head of the fibula and the medial point on the leg directly across from the fibular head, and a line bisecting the knee joint line was dropped to locate and mark the midpoint of the subject's right leg at the musculotendinous intersection of the gastrocnemius and the Achilles tendon. This location was used for the first reflective marker on the posterior leg segment. An additional marker was then placed 10 cm below this first marker to represent the midline of the lower leg. Each subject was then asked to kneel on a chair facing the back to allow the subtalar joint to be placed in a neutral position. Two markers were then placed on the back of the shoe so there would be a marker both on the superior and inferior aspect of the heel cap along the midline of the calcaneus.

At this point, each subject was asked to mount a motorized treadmill and begin running at a self-selected pace for one minute in order to warm-up and familiarize themselves with the treadmill. Even though some research has shown that setting a treadmill at 1% grade can better represent overground running in a physiological perspective, previous articles done to compare overground and treadmill running have employed a level treadmill (0% incline) setup (37, 39, 85). Additionally, Tao et al. (38) found that level and 1% grade treadmill running displayed no significant differences in spatiotemporal variables. Due to this, each subject in this study ran on a treadmill with a 0% incline. A camera recording at 240 Hz (Apple, iPhone XR, Cupertino, CA) was positioned 2 m behind the treadmill so it could capture the rearfoot motion of each subject's right foot during every stance phase. Specifically, the camera was placed on a floor-mounted tripod so that the focal point of the camera was 7 cm above the height of the treadmill belt, or approximately the same height of the rearfoot markers. Given that the typical stance width is 2-7 cm (86), the camera was positioned approximately 2 cm to the right of the center of the treadmill belt in order to be located directly behind the foot at foot strike. We also ensured that the camera was directed perpendicular to the subject's frontal plane by using a carpenter's square aligned with treadmill belt.

After this warm-up period, the speed of the treadmill was increased to the average speed the subject used in the overground running trials. The subject then ran at this speed for 1 minute, while the data recorded from the accelerometer component of each IMU sensor were still sampled at 1600 Hz and the gyroscope component was recorded at 1125 Hz. Unlike in the overground running trials, this single 1-minute bout was the only trial necessary to capture the proper amount of gait cycles to compare with the overground conditions. In effect, 30 steps occurring in the middle of this trial were extracted for analysis.

Data Analysis

As mentioned earlier, the running distance used in the overground trials was selected to ensure at least 10 gait cycles could be analyzed for each subject during every trial. With that said, 10 consecutive steps were extracted from each trial for all three overground running surfaces. As a result, 90 steps were processed for each subject for the overground conditions. Furthermore, 30 steps were extracted from the treadmill condition for each subject, meaning a total of 120 steps were analyzed for each subject in this study. All data from the IMU sensors were processed in MATLAB R2020b (Mathworks, Inc., Natick, MA) through a series of custom-made functions. These functions were used to determine the temporal and kinetic variables that were previously outlined in Specific Aim 1 (i.e. stance time, stride frequency, rearfoot eversion velocity, peak tibial acceleration). All data from the video capture were exported and in Kinovea video analysis software (Version 0.9.3).

Stride Segmentation

All data were segmented into strides in each of the trials in order to calculate the outcome variables (stance time, stride frequency, and rearfoot eversion velocity) for this study. This process began by extracting the data from the y-axis of the gyroscope component of the heel cap mounted IMU sensor. The gyroscope component was used in this study because the researchers found that it presented a smoother and more consistent signal than the accelerometer. This gave the researchers the ability to identify certain features across different types of runners (i.e. rearfoot, forefoot strikers) with more ease and accuracy. Due to this, all variables other than peak tibial acceleration were calculated from the analysis of the gyroscope component of the heel cap mounted IMU sensor.

Segmentation of the data into strides began with identification of swing phase for each stride. The method to do this was guided by Zrenner at al. (31) and Falbriard et al. (19). It started by finding the beginning of swing phase through the recognition of a large negative peak in the angular velocity data (obtained via the gyroscope) around the Y axis of the IMU, which was coincident with the mediolateral axis of the runner. We began by first determining the global minimum for each corresponding trial. A trial is comprised of the data from each condition (grass, track, asphalt, and treadmill), thus multiple steps were included in each trial. From this global minimum, a percentage of its magnitude was set as a threshold and used to find all data points that fell below it. There was not an exact threshold used across subjects due to each subject having varying magnitudes in their own gyroscope data. This step was implemented so all swing phases within each trial could be identified accurately.

However, because this thresholding technique identified all data points below the threshold (i.e. more negative), a collection of data points were identified in each swing phase which were not necessary for the identification of foot strike and toe off. Therefore, the algorithm used in this study only returned the data points that were one sample below the pre-defined threshold. This yielded one data point that proceeded maximum swing angular velocity and another data point that instantly followed this same swing phase (Fig. 1). The data point that proceeded maximum swing angular velocity was used to find toe off events. This was done by establishing a window of interest that ended at this data point that proceeded maximum swing angular velocity and started ~400 samples prior. Once this window of interest was set, the global maximum within this range was located and labelled as toe off. In a similar fashion, the foot strike events were located by establishing a window of interest that started at the data point that instantly followed this same swing phase and ended ~190 samples afterward. This window of interest was then used to find the global maximum within this set range, which corresponded to foot strike. Effectively, foot strike events were identified as the first global maximum following swing phase, while toe off events were the global maxima prior to the following swing phase in the Y axis gyroscope data, which can be seen in Figure 1.

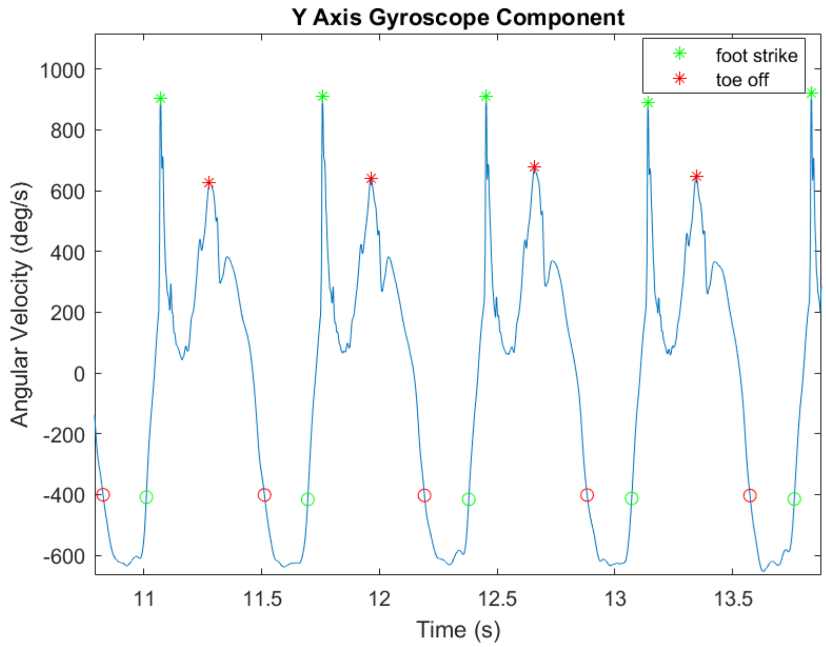

***Figure 1**. This figure shows the result of the stride segmentation method along with the location of both foot strike and toe off events. As shown, the circle icons are the data points that were used to identify the location of swing phase and were also used to establish the windows of interest to find the maxima that would correspond to foot strike (green asterisk) and toe off (red asterisk) in each gait cycle.*

Stance time

Once the locations or indices of all foot strike and toe off events were determined, a custom-made function was used to calculate the time of stance for each gait cycle. Since the location of each event was known, stance time was simply calculated by subtracting the time values that corresponded to the i^{th} foot strike and toe off throughout the trial. Following the determination of each individual stance time, all these stance times were averaged to find the mean stance time for the treadmill and overground conditions.

Stride Frequency

Stride frequency was determined in a similar process. Since stride time is defined as the elapsed time between two consecutive foot strikes of the same foot, stride frequency was simply calculated as the reciprocal of this stride time (80). In order to compute stride frequency for this study, a function was used to first calculate the duration of each individual stride. Once all the stride durations were determined, these values were reciprocated and then averaged to yield the stride frequency of the overall trial.

Rearfoot Eversion Velocity (IMU)

The data obtained from the z-axis of the gyroscope were used to calculate rearfoot eversion velocity because it closely represents the anteroposterior axis of the foot. Because the gyroscope recorded synchronously along all three axes, the time of each foot strike and toe off were appropriate to identify all the stance phases throughout a given trial. These event locations were used in the z-axis gyroscope data to determine the mean eversion velocity for each stance phase along with the average maximum eversion velocity of the IMU sensor throughout the trial. This was done by using a function that would establish a moving window that was constrained by the i^{th} foot strike and toe off to calculate the mean eversion velocity within that data window. From there, the same function was used to quantify the global maximum in each window, which corresponded to maximum eversion velocity (Fig. 2).

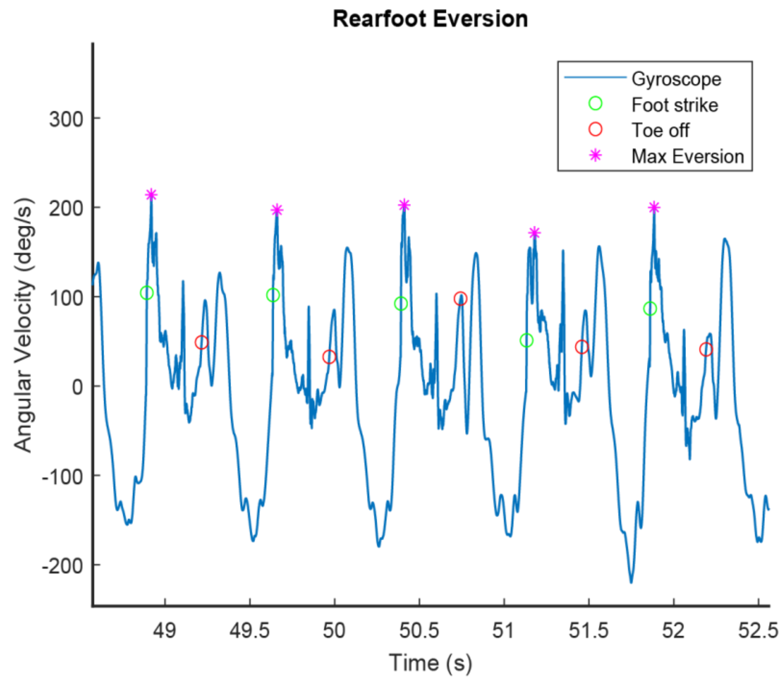

***Figure 2**. Identified points of maximum rearfoot eversion velocity (pink). Foot strike (green) and toe off (red) events are shown to help display where maximum rearfoot eversion velocity occurred in each stance phase.*

Rearfoot Eversion Velocity (Video)

All video data that captured the rearfoot motion of each subject running during the treadmill condition were imported into Kinovea video analysis software (Version 0.9.3). Kinovea is a free, open-source software that provided the researchers the ability to analyze the changes in the rearfoot angle throughout each stance phase. This rearfoot angle was quantified by measuring the angle of the intersection of a line connecting the two heel cap markers and a line connecting the two markers that were placed on the Achilles tendon. During each stance phase in the treadmill running condition, the change in this rearfoot angle was measured by using the ‘angle’ tool and tracking function that is native to Kinovea. The ‘angle’ tool acts as a digital goniometer that was specifically used to measure the rearfoot angle on a frame-by-frame basis during each stance phase. The tracking function was used in conjunction with the ‘angle’ tool so all rearfoot angles could be saved and stored into a csv file, which allowed for exportation of each subject’s data. Since this study is interested in mean and maximum rearfoot eversion velocity and the data recorded via Kinovea was expressed as angles (degrees), each subject’s data was derived once in order to obtain angular velocities (deg/s) of the rearfoot segment. Following this, the mean and maximum angular velocity were calculated for each stance phase, which were later used to determine the overall mean and maximum rearfoot eversion velocity for each given trial.

Peak Tibial Acceleration

As previously mentioned, peak tibial acceleration was the only variable that was determined from the use of the shank mounted IMU sensor. We decided to only calculate peak axial tibial acceleration (accelerations along the vertical axis of tibia) within this study due to peak resultant acceleration (resultant accelerations of all three axes of the tibia) being consistently larger, which can lead to overestimations of the amount of shock the tibia experiences axially. (26). Therefore, only the data recorded along the y-axis of the shank mounted IMU sensor were analyzed. For clarity of writing, peak axial tibial acceleration will be referred to simply as peak tibial acceleration. Prior to the calculation of peak tibial acceleration, all data were filtered with a 4th order Butterworth filter with a cutoff frequency of 70 Hz. Following this, a function was used to quantify the peak tibial acceleration within each stance phase along with the mean peak tibial acceleration throughout a given trial. This was done by first converting all measures from the IMU sensor from m/s^2 to units of g. A threshold that varied between subjects was then used to isolate the impact peaks within the accelerometer signal and to form clusters of data where theoretically an impact peak would reside. These clusters of data surrounding each impact peak were used to find the maximum point, which corresponded to the peak tibial acceleration (Fig. 3). After each stance phase contained a labeled maximum point, the magnitudes of these maxima were averaged to yield the mean peak tibial acceleration for each trial.

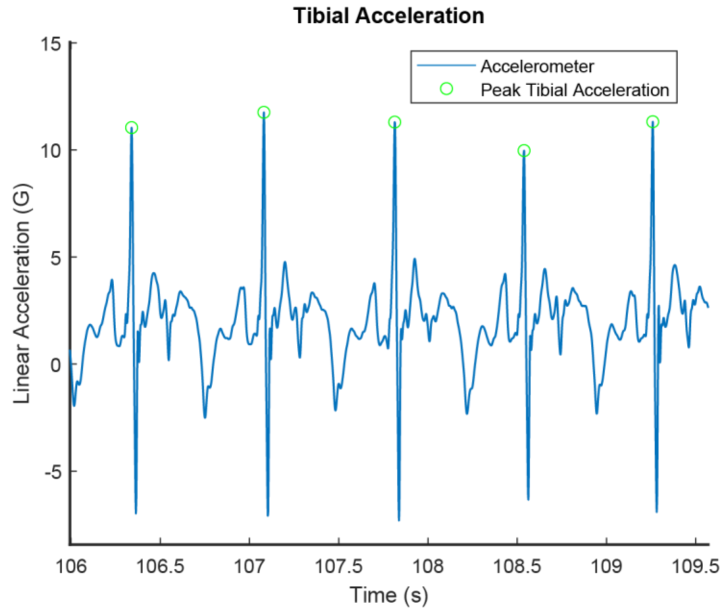

***Figure 3**. The locations of the peak tibial acceleration (green) during each stance phase.*

Statistical Analysis

All statistical analyses were performed in SPSS software (IBM, version 21, Armonk, NY). Descriptive statistics, including measures of central tendency (means, medians, other percentiles) and dispersion (standard deviations, ranges) were computed for continuous data such as age. For each of the measured variables, graphical displays including histograms, box plots, and Q-Q plots were produced to observe the distribution of the data. Prior to any of the statistical tests, normality and sphericity assumptions were tested on both the means and standard deviations of each variable. The normality assumption was checked by using a Shapiro-Wilk test on each condition's data with an alpha level set of 0.05. Sphericity was assessed by using

Mauchly's test. If the sphericity assumption was not met, a Greenhouse-Geisser correction was used to determine within-subjects effects.

Specific Aim 1:

Our first Specific Aim was to determine if running surface had any significant effect on the means or variability of the outcome measures of this study. The independent variable for the Specific Aim 1 was surface, which had four levels: track, asphalt, grass, and treadmill. The dependent outcome measures for this aim were stance time, stride frequency, maximum rearfoot eversion velocity, and peak tibial acceleration. Due to the within-subject design of the study, a repeated-measures ANOVA for each running condition was performed to detect any significant differences in the outcome variables whenever the normality and sphericity assumptions were met. If the data were found to not be normally distributed, the non-parametric version of a repeated-measures ANOVA (i.e. Freidman test) was used. Regardless of which test was used, the significance value was set at 0.05. If significance was found from the repeated-measures ANOVA, Fisher's LSD post-hoc analysis was performed to determine which conditions were significantly different from one another ($\alpha = 0.05$). If significance was found following a Friedman test, a series of Wilcoxon signed-rank tests were used to assess the interactions between conditions. Since there were four total conditions, meaning six comparisons were performed, a Bonferroni correction was used to protect from Type I error. Therefore, the alpha level for each of the Wilcoxon signed-rank tests was set to 0.008. This process was also performed on the standard deviations of each variable for every subject to assess intrasubject variability across conditions.

Specific Aim 2:

Our second Specific Aim was to determine if maximum rearfoot eversion velocity was significantly different when determined with high-speed videography or a heel mounted IMU. We assessed this in two ways. To test for the reliability of IMU derived maximum rearfoot eversion velocity values when compared to the standard high-speed videography, an intraclass correlation coefficient (ICC) and its 95% confidence intervals were calculated based on mean-rating (k = 2), absolute-agreement, and a 2-way mixed-effects model. Afterword, a Bland-Altman plot was generated with 95% limits of agreement (LoA) to visually assess the agreement between the two systems.

Chapter 4: Results

Specific Aim 1:

Specific Aim 1 was to discover if running surface (track, asphalt, grass, treadmill) had any significant effect on the IMU derived variables (i.e., stance time, stride frequency, maximum rearfoot eversion velocity, peak tibial acceleration) in this study. Specifically, we examined the means and within subject variability associated with each of the four surfaces.

Comparison of Subject Means

The data distributions of stance time, stride frequency, and peak tibial acceleration were found to be normal, therefore, a repeated measures ANOVA was conducted to examine the between-surface effects on these variables (Table 1). In contrast, the normality assumption was not met for the means of maximum rearfoot eversion velocity, so a Friedman test was used instead (Table 2). Overall, there were significant differences found between running surface and all these variables other than stance time ($p = 0.321$). Specifically, the means of stride frequency ($p < 0.001$), maximum rearfoot eversion velocity ($p < 0.001$), and peak tibial acceleration ($p < 0.001$) were found to be significantly different between at least two of the running surfaces.

As seen in Table 3, stride frequency was approximately 2% faster in the treadmill running condition compared to all the overground surfaces. Even though the differences in stance time between the overground surfaces were quite marginal, the average stride frequency used on the track was significantly faster than what was used on both asphalt and grass. Despite this difference, asphalt and grass running did not present with significantly different stride frequencies in this study ($p = 0.312$).

Also shown in Table 3, grass running seems to present with considerably less maximum rearfoot eversion velocities when compared to the other running surfaces. Grass was shown to have a 25% percent lower average for maximum rearfoot eversion velocity than treadmill running and a 36% lower average than both track and asphalt running. No differences were seen between track and asphalt surfaces (p = 0.681).

Lastly, there were also significant differences in peak tibial acceleration between the treadmill and all three of the overground surfaces. On average, overground running produced 26% higher peak tibial accelerations than treadmill running. As expected, there were no differences found in peak tibial acceleration between the individual overground surfaces (Table 3).

Table 1. Results from the repeated measures ANOVA performed on the variable means*. Final column includes the p-values associated with each post-hoc comparison. Significance value was set to 0.05.*

Variable Name	ANOVA p-value	Interactions	Fisher's LSD p-value
Stance Time (s)	0.321	Track – Asphalt	0.436
		Track – Grass	0.574
		Track – Treadmill	0.141
		Asphalt – Grass	0.818
		Asphalt – Treadmill	0.218
		Grass - Treadmill	0.253
Stride Frequency (stride/s)	< 0.001	Track – Asphalt	0.008 *
		Track – Grass	0.001 *
		Track – Treadmill	0.047 *
		Asphalt - Grass	0.312
		Asphalt – Treadmill	0.001 *
		Grass – Treadmill	0.001 *
Peak Tibial Acceleration (G)	< 0.001	Track – Asphalt	0.055
		Track – Grass	0.793
		Track – Treadmill	< 0.001 *
		Asphalt - Grass	0.436
		Asphalt – Treadmill	< 0.001 *
		Grass – Treadmill	< 0.001 *

***Table 2**. **Results from the Friedman test performed on the means of the maximum rearfoot eversion velocity data**. Since multiple comparisons were done through a series of Wilcoxon signed-rank tests, the significance value was adjusted to 0.008.*

Variable Name	Friedman p-value	Interactions	Wilcoxon p-value
Maximum Rearfoot Eversion Velocity (deg/s)	< 0.001	Track – Asphalt	0.681
		Track – Grass	< 0.001 *
		Track – Treadmill	0.002 *
		Asphalt - Grass	< 0.001 *
		Asphalt – Treadmill	0.002 *
		Grass – Treadmill	0.002 *

***Table 3**. **Means (standard deviations) of each of the IMU derived variables for each running surface**. The superscripts by each of the standard deviations represent the significant interactions between the conditions. (Superscript A – Asphalt, G – Grass, T – Track, TM – Treadmill). Example. Superscript A represents that condition is significantly different from the asphalt condition.*

Variable Name	Track	Asphalt	Grass	Treadmill
Stance Time (s)	0.212 (0.017)	0.213 (0.017)	0.213 (0.020)	0.216 (0.024)
Stride Frequency (stride/s)	1.37 (0.7) $^{A\ G\ TM}$	1.36 (0.07) $^{T\ TM}$	1.35 (0.07) $^{T\ TM}$	1.39 (0.09) $^{T\ A\ G}$
Maximum Rearfoot Eversion Velocity (deg/s)	623.52 (299.39) $^{G\ TM}$	620.72 (289.06) $^{G\ TM}$	394.92 (256.48) $^{T\ A\ TM}$	531.07 (243.29) $^{T\ A\ G}$
Peak Tibial Acceleration (G)	10.60 (2.25) ^{TM}	11.01 (2.65) ^{TM}	10.73 (3.41) ^{TM}	7.93 (1.64) $^{T\ A\ G}$

Comparison of Within Subject Variability

The data representing the within subject variability for stance time, stride frequency, maximum rearfoot eversion velocity, and peak tibial acceleration across all running surfaces in this study did not meet the normality assumption needed to appropriately run a repeated measures ANOVA. In effect, a series of Friedman tests were conducted on each of the variables (Table 4). The results of these tests showed there were significant differences between the intrasubject variability of stance time ($p < 0.001$), stride frequency ($p < 0.001$), maximum rearfoot eversion velocity ($p < 0.001$), and peak tibial acceleration ($p < 0.001$) across the different running surfaces (Table 4).

***Table 3. Results from the Friedman tests performed on the variability of each subject for each of the running surfaces**. Since multiple comparisons were done through a series of Wilcoxon signed-rank tests, the significance value was adjusted to 0.008.*

Variable Name	Friedman p-value	Interactions	Wilcoxon p-value
Stance Time (s)	< 0.001	Track – Asphalt	0.117
		Track – Grass	0.003 *
		Track – Treadmill	0.433
		Asphalt – Grass	0.002 *
		Asphalt – Treadmill	0.204
		Grass - Treadmill	0.004 *
Stride Frequency (stride/s)	< 0.001	Track – Asphalt	0.017
		Track – Grass	0.001 *
		Track – Treadmill	0.001 *
		Asphalt - Grass	0.315
		Asphalt – Treadmill	< 0.001 *
		Grass – Treadmill	< 0.001 *
Peak Tibial Acceleration (G)	< 0.001	Track – Asphalt	0.502
		Track – Grass	0.002 *
		Track – Treadmill	< 0.001 *
		Asphalt - Grass	0.003 *
		Asphalt – Treadmill	< 0.001 *
		Grass – Treadmill	< 0.001 *
Maximum Rearfoot Eversion Velocity (deg/s)	< 0.001	Track – Asphalt	0.232
		Track – Grass	0.007 *
		Track – Treadmill	0.002 *
		Asphalt - Grass	0.023
		Asphalt – Treadmill	< 0.001 *
		Grass – Treadmill	< 0.001*

Analysis of the intrasubject variability between running surfaces revealed multiple significant findings (Table 5). One of these significant findings was seen in stance time variability. Stance time was approximately 36% more variable while the subjects ran on grass compared to any other condition in this study. Surprisingly, stance time variability did not differ between any of the other running surfaces.

As for stride frequency, the intrasubject variability was 28% higher on grass than what it was for track. Additionally, there was a 45% reduction in stride frequency variability when subjects ran on the treadmill compared to the grass surface. The difference between treadmill and the other running surfaces (i.e. track, asphalt) was also found to be significant, but not nearly as profound as the comparison to grass. Treadmill running presented with 24% and 33% less variability than track and asphalt running, respectively. There were no significant differences seen between track and asphalt ($p = 0.017$) or asphalt and grass ($p = 0.315$).

Similarly, the results for the intrasubject variability for maximum rearfoot eversion velocity was the lowest when subjects ran on the treadmill and highest while running on grass. Treadmill running had a 43%, 32%, and 29% lower variability in maximum rearfoot eversion velocity when compared to grass, asphalt, and track, respectively. The variability in maximum rearfoot eversion velocity was shown to be 21% higher during grass running compared to when the subjects ran on the track. No differences in intrasubject variability were seen between track and asphalt ($p = 0.232$) or asphalt and grass ($p = 0.023$).

Among the four variables calculated in this study, the intrasubject variability in peak tibial acceleration presented with the most significant differences between conditions. The interaction between track and asphalt was the only comparison that was found not to be significantly different ($p = 0.502$). Like what was seen in the variability of a couple other variables in this study, the variability in peak tibial acceleration during treadmill running was significantly lower than all other running surfaces. Treadmill running resulted in a 67% lower variability in peak tibial acceleration when compared to grass running, while approximately a 50% reduction between both track and asphalt. Similar to some of the other findings discussed, running on grass once again displayed the highest intrasubject variability across all surfaces.

Grass running was found to have a 33% and 35% higher variability in peak tibial acceleration than asphalt and track, respectively. Overall, there seems to be a constant trend throughout the results of the within subject variability for most of the calculated variables. Perhaps other than stance time variability, all other parameters were noticeably less variable when the subjects ran on the treadmill, while less consistent on grass.

Table 5. Means (standard deviations) of the variability seen between conditions from the IMU derived variables*. The superscripts by each of the standard deviations represent the significant interactions between the conditions. (Superscript A – Asphalt, G – Grass, T – Track, TM – Treadmill).*

Variable Name	Track	Asphalt	Grass	Treadmill
Stance Time (s)	0.008 (0.004) ^{G}	0.010 (0.006) ^{G}	0.014 (0.009) $^{T\ A\ TM}$	0.009 (0.007) ^{G}
Stride Frequency (stride/s)	0.021 (0.005) $^{G\ TM}$	0.024 (0.006) ^{TM}	0.029 (0.010) $^{T\ TM}$	0.016 (0.005) $^{T\ A\ G}$
Maximum Rearfoot Eversion Velocity (deg/s)	98.05 (52.43) $^{G\ TM}$	103.27 (44.78) ^{TM}	123.71 (36.16) $^{T\ TM}$	70.10 (35.58) $^{T\ A\ G}$
Peak Tibial Acceleration (G)	1.57 (0.63) $^{G\ TM}$	1.62 (0.69) $^{G\ TM}$	2.41 (1.21) $^{T\ A\ TM}$	0.79 (0.32) $^{T\ A\ G}$

Specific Aim 2:

Specific Aim 2 was to determine the agreement of the maximum rearfoot eversion velocity values acquired from an IMU and high-speed videography. As shown in Table 6, the ICC value was determined to be 0.739, which can be considered a moderate reliability for the IMU estimations (87). The calculated 95% confidence interval was shown to range from 0.322 – 0.899, meaning the level of reliability of the IMU derived maximum rearfoot eversion velocity values ranged from poor to good when compared to the same variable collected via videography (87).

Table 6. Results of intraclass correlation coefficient (ICC) generated to test for the reliability of IMU derived maximum rearfoot eversion velocity.

Type	Intraclass Correlation	95% CI – Lower Bound	95% CI – Upper Bound	Significance
Single Measures	0.586	0.192	0.817	0.003
Average Measures	0.739	0.322	0.899	0.003

To better represent the agreement between the IMU and high-speed videography, the Bland-Altman plot (Fig. 5) below shows the dispersion of the differences between the two systems. The horizontal black line, which is oriented at approximately 12.67 deg/s, represents the average difference (bias) between these two methods. As depicted by the red-dashed lines, it is shown that all discrepancies between the IMU and high-speed videography other than one subject were captured within the 95% limits of agreement (LoA).

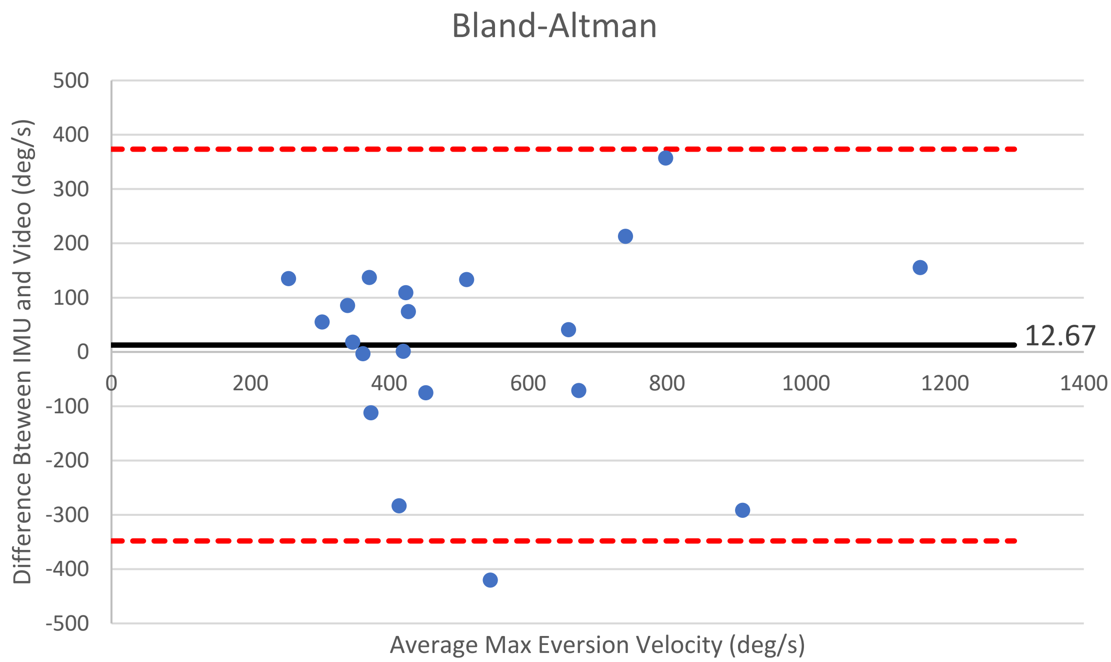

***Figure 5**. Plotting the agreement between the IMU and high-speed videography estimations of maximum rearfoot eversion velocity. The bias between the two systems (solid black line) was found to be 12.67 deg/s. The limits of agreement (red-dashed lines) were calculated by bias ± 1.96*SD.*

Chapter 5: Discussion

The primary purpose of this study was to use Inertial Measurement Units (IMUs) to compare spatiotemporal and kinetic variables during running on different surfaces. The surfaces used for this study were track, grass, asphalt, and treadmill, while the specific variables of interest were stance time, stride frequency, maximum rearfoot eversion velocity, and peak tibial acceleration. These surfaces were selected to represent common environments used by leisure and competitive runners during their training. The variables that were calculated for this study have been previously determined with IMUs and other forms of wearable technology and are considered basic metrics that have been used in the field of running biomechanics to provide insight on any changes that may be occurring in one's gait (22, 24, 26, 31, 40, 80).

The comparison of running surfaces has been a popular area of study in biomechanics and as a result, biomechanists and some clinicians understand that running surface does have a significant influence on the investigated variables in this study and several others (9, 26, 34, 36). A majority of the knowledge around this subject has been gained through the instrumentation of motion capture, force plates, and for some, instrumented treadmills. These are all pieces of equipment that require significant amounts of financial backing and are often restricted to indoor/laboratory settings (20–22). Due to this, much of the original work done to investigate how running surface effects one's running mechanics has focused on the comparison of treadmill and overground running, which in most cases involved participants running in a controlled area of a laboratory.

Even though a running path in a laboratory may resemble a sidewalk, track, or trail, it may still not perfectly reflect the mechanics of a runner when they are running on their habitual surface outdoors. This thought and some advancements in wearable technology led the field of running biomechanics outside the laboratory to analyze runners in their natural settings for the first time (i.e. track, grass, asphalt, trail). Unfortunately, there is currently a limited amount of research done using forms of wearable technology like IMUs, accelerometers, and gyroscopes. This may be a result of the rapid emergence of these devices, which has not given all researchers time to fully explore the devices' capabilities in their fields of interest.

The research that has been conducted with running and wearable technology shows great promise since biomechanical variables like running speed, stance time, flight time, rearfoot eversion velocity, stride length, stride frequency, and tibial acceleration have been derived from these devices (21, 26, 40, 50, 82, 88, 89). Despite some of these articles analyzing running biomechanics with the use of IMUs and wearable technology, very few have used these forms of technology to revisit the question of whether overground and treadmill running are different from each other (26, 33, 39–41). Therefore, this study was created to answer this question and to potentially fill a gap in the current research of using wearable technology in the study of running biomechanics.

Specific Aim 1*:*

Due to this study testing multiple running surfaces, Specific Aim 1 consisted of the effects of each running surface on stance time, stride frequency, maximum rearfoot eversion velocity, and peak tibial acceleration. We will address each hypothesis separately.

<u>Surface Effects on Stance Time</u>

We hypothesized there would be no significant difference in stance time when analyzing across the different running surfaces. This hypothesis was accepted following the repeated measures ANOVA testing, meaning running surface had no significant effect on stance time for the subjects in this study (p = 0.321). This finding agrees with some of the current research (9, 90), while disagrees with others (35, 36, 38). To simplify the comparisons, the findings between treadmill and all overground surfaces will be discussed first. Nelson et al. (36) and Mok et al. (35) both reported differences in stance time when comparing treadmill and overground running, but differed in the direction of the difference. Nelson et al. (36) reported his subjects experienced longer stance times when running on a horizontal treadmill, while Mok et al. (35) showed that stance time actually decreased during treadmill running and was longer for overground running. Mok et al. (35) believes this finding is explained by the motion of the treadmill belt, since it pulls the runner's foot backward passively, which may result in an accelerated stance phase and a quicker transition into swing. It should be said that the stance changes reported by Nelson et al. (36) were only seen when the subject ran at that fastest condition for the study (6.4 m/s). There were no significant differences in stance time for the slower running speeds (3.35, 4.88 m/s).

When discussing the effects of overground surfaces (i.e. track, grass, asphalt) on stance time, the only study that could be found that reported different stance times was done by Hollis et al. (88), who acquired stance time from a heel-mounted IMU and found that their subjects had significantly longer stance times while running in grass than what they did on track. Similar to the study done by Nelson et al. (36), this finding was only relevant when the subjects ran at the fastest speed that was required in the study. This trend may show that even though stance time may be invariant across surfaces at slower speeds, faster running speeds may lead to more variability in an individual's stride.

Apart from these few articles, all other studies performing the comparison between overground surfaces and treadmill running have found no significant differences in stance time (34, 90, 91). The agreement between these studies and the current one may be explained by the fact that stance time may be more related to running speed rather than surface compliance or stiffness. It has been believed for a considerable amount of time in the field of running biomechanics that stance time is negatively correlated with running speed (92) and since this study ensured each subject maintained the same constant speed throughout the running conditions, it would be expected that the stance time would remain the same.

<u>Surface Effects on Stride Frequency</u>

We hypothesized there would be no significant differences in stride frequency across running surfaces. This hypothesis was rejected due to the significant p-value generated from the repeated measures ANOVA ($p < 0.001$). Even though most of the comparisons performed between surfaces were found to be significant, the main finding was treadmill running presented with a faster stride frequency than any of the overground conditions (i.e. track, grass, asphalt). This seems to be an overlapping finding in a couple of studies that looked at the differences between stride frequency for overground and treadmill running, albeit the reported differences between the two types of running were slightly larger than what was seen in this study (34, 38). These larger differences in stride frequency may simply be explained by the average running speed being faster in these studies compared to the current one (3.6 m/s vs 3.8 m/s; 3.6 m/s vs 4.5 m/s). Nelson et al. (36) noticed stride frequency decreased when a runner exercised on a treadmill, but this was only shown when the runner ran at a relatively fast pace (6.4 m/s).

Even though our study's findings differ from Nelson et al. (36), the increase seen in stride frequency in the treadmill condition can be attributed to two features of treadmill running that were highlighted in a study done Riley et al. (34). Treadmill running has been shown to present with a minor, but significant, reduction in peak propulsion when compared to overground running, which would logically hinder a runner's ability to rapidly transition into swing phase.

This result could lead to a slower cadence (i.e. stride frequency) since there is less force propelling the runner into their next step. According to Riley et al. (34), this is not truly the case. Stride frequency increases because treadmill running allows for a significantly larger plantarflexion moment around the ankle, allowing for a larger spring-like effect of the lower extremity than what is seen on overground surfaces. This spring-like effect would, in turn, provide a runner an enhanced ability to propel into their stride. Another explanation for this increased stride frequency during treadmill running is that the treadmill belt pulls the runner's foot backward during each stance phase, therefore, promoting a quick and smooth transition into the following step, while this process in overground running is actively controlled via hip and knee extension (35).

To the author's knowledge, there are no current studies that analyzed the differences in stride frequency between multiple overground surfaces, so the stride frequency results between the overground surfaces will be discussed without the comparison of previous studies. The paucity of work done comparing stride frequency between surfaces like track, asphalt, and grass may be a result of running cadence being quite comparable when running overground. Despite stride frequency being significantly faster on track than what was observed on both asphalt and grass, the specific differences were rather minimal. For example, when the subjects of this study ran on the track, they only presented with a 0.02 and 0.01 stride/s difference between grass and asphalt, respectively.

The marginally faster stride frequency during track running is probably related to the material properties of the track surface itself. Unlike asphalt and grass, the specific purpose of track (tartan) is to provide a surface that is beneficial for runners. All experienced or competitive runners know that track provides a noticeable amount of "spring" in one's step that is not apparent in any other outside running surface. When considering this "spring" effect in terms of stride frequency, it may be plausible that due to this additional amount of propulsion, a runner could indirectly maintain a faster cadence than compared to other overground running surfaces.

Surface Effects on Maximum Rearfoot Eversion Velocity

We hypothesized there would be no significant differences in maximum rearfoot eversion velocity across the four running surfaces. We rejected this hypothesis due to there being multiple significant differences found between the running surfaces in this study ($p < 0.001$). Similar to previous studies that have specifically assessed maximum rearfoot eversion velocity across running surfaces, this study showed maximum rearfoot eversion velocity appeared to have a direct relationship with surface compliance (40, 88). Essentially, the more rigid or less complaint the running surface, the larger the magnitude of maximum rearfoot eversion velocity.

This concept was first mentioned by Zrenner et al. (40) and Hardin et al. (9) when addressing why runners typically pronate or evert more on less compliant surfaces, but it also agrees with a common belief about rearfoot motion (i.e. eversion velocity, pronation) in the field of running biomechanics (16, 66). Generally, it is believed that pronation is one of the mechanisms our lower extremity uses to help attenuate the transient forces we humans experience with every step during both walking and running (16, 66). When we are exposed to rigid surfaces while running, it is conceivable to believe most runners will have a larger pronation range of motion or experience faster eversion velocities in order to help protect themselves from injuries that may be associated with chronic forces on the body (40).

This concept or relationship was evident in this study in that grass running (most compliant) on average produced a maximum rearfoot eversion velocity of 394.92 deg/s, while asphalt running (least compliant) had an average of 620.72 deg/s. This is about a 36% difference in the amount of rearfoot eversion velocity between the two conditions. Contrary to our expectations though, when the subjects ran on track and asphalt, they did not present with significantly different maximum rearfoot eversion velocity (623.52 deg/s vs 620.72 deg/s), despite track being a more compliant surface. No other studies could be found that analyzed the differences in this variable between track and asphalt, so it is difficult to conclude if this an accurate finding.

Hollis et al. (88) examined the differences in maximum rearfoot eversion velocity between track and grass and reported similar results. The only difference between this study and the referenced one is the mean difference between the two running surfaces. This study calculated the mean difference in maximum rearfoot eversion velocity between track and grass to be 228.6 deg/s, while Hollis et al. (88) reported their mean difference as 143 deg/s. It is difficult to interpret why our study found a larger difference between the two running surfaces. Hollis et al. (88) did not mention their subjects' average running speed, which may have been a factor in the differences seen between their and our study. Either way, both this study and Hollis et al. (88) agree that with grass being a more compliant surface than track, it would be expected for grass to have significantly lower maximum rearfoot eversion velocities.

When our subjects ran on the treadmill, maximum rearfoot eversion velocity was on average 136.15 deg/s larger than during grass running, but approximately 90 deg/s slower compared to either track or asphalt. Even though the exact stiffness of the treadmill used in our study in not known, it is plausible that the compliance of the treadmill lies between what is perceived for grass and the other two overground surfaces (i.e. track, asphalt). Hence, the results from the treadmill running condition appear to agree with the overall trend that surface compliance determines maximum rearfoot eversion velocity magnitude.

Surface Effects on Peak Tibial Acceleration

We hypothesized there would be no significant differences in peak tibial acceleration across the four running surfaces. This hypothesis was rejected, meaning running surface had a significant effect on peak tibial acceleration in this study ($p < 0.001$). Similar to other studies that have assessed tibial acceleration across different running surfaces, the main finding of this study was that treadmill running presented with significantly lower levels of peak tibial acceleration than any of the overground surfaces ($p < 0.001$) (26, 62, 93). Fu et al. (61) was the only study that did not observe any differences in peak tibial acceleration between treadmill and overground running, which was possibly due to them placing the accelerometer in a different location on the tibia. Most studies elected to place the IMU/accelerometer on the anteromedial aspect of the distal tibia (26, 93), while Fu et al. (61) strapped their device around the tibial tuberosity.

The rationale behind the increased peak tibial accelerations observed in overground running still seems to be quite ambiguous to current researchers (26, 54, 94). Milner et al. (26) specifically attempted to use tibial accelerations from laboratory overground running and the cadence used by their subjects and found these variables could only explain 50% and 21% of the variance seen in peak tibial acceleration on sidewalk and grass, respectively. Furthermore, Milner et al. (26) explicitly states that when cadence was removed from the regression analysis and the same predictions were made, there were no significant differences in the variances explained. This essentially shows that cadence does not have a significant role in determining the discrepancy between overground and treadmill running for peak tibial acceleration.

Surprisingly, the actual determinants of the magnitude of peak tibial acceleration still remain unknown even when comparing between overground surfaces. Initially, it was thought that peak tibial acceleration was indirectly related to surface compliance, since this is a relationship that has been noted with ground reaction forces in running (15, 94). In other words, it was believed that soft or compliant surfaces would present with lower magnitudes of peak tibial acceleration since the surface itself would attenuate some of the forces for the runner (15). Unfortunately, surface compliance has been determined to explain less than 10% of the variance in peak tibial acceleration (95). This would explain the results seen in this study and others where the differences in peak tibial acceleration between grass and other overground surfaces (i.e. track, asphalt) were found not to be statistically significant (26, 41, 61).

Fortunately, the divergence from the original belief that surface compliance predicts peak tibial acceleration has given way to other explanations, one involving the differences seen in stride length between different running surfaces. Riley et al. (34) was able to show that treadmill running is associated with significantly shorter stride lengths compared to overground running. This would effectively allow the foot of the runner to land more under their center of mass, which would in turn give the joints of the lower extremity (i.e. ankle, knee, hip) the ability to eccentrically control the impact of each step.

In contrast, when stride length is elongated or increased (observed in overground running) (34), the foot lands more in front of the center of mass and predisposes the lower extremity to a position that is not optimal for landing. In other words, the longer stride lengths seen in overground running allow the foot to travel too far in front of the runner, so when they initiate impact with the ground, the joints of the lower extremity are not in a prime position to control the eccentric action of the running stride (54).

Alterations in stride length, of course, is just one determinant or reason for why peak tibial acceleration may be different between surfaces. Potthast et al. (95) tested the effect of sagittal plane knee angle and muscle activation on the magnitude of peak tibial acceleration and found that higher degrees of knee extension and muscle pre-activation led to the greatest mitigations in the amount of accelerations the tibia experiences. Furthermore, this study also discovered that pre-activation levels of the muscles of the knee can explain up to 35% of the variance seen in peak tibial acceleration across surfaces that vary in compliance (95). To allow for full transparency, one of the limitations of this referenced study was the peak tibial acceleration values were recorded during static, non-running movements so the results may not fully be transferrable to how the lower extremity reacts normally during the running motion (95).

In addition, Montgomery et al. (62) reported that higher percentages of muscle co-contractions of the rectus femoris and semitendinosus lead to significant reductions in peak tibial acceleration when subjects ran on three different running conditions (motorized treadmill, non-motorized treadmill, overground). Surprisingly, Montgomery et al. (62) did not note any significant differences between the motorized treadmill and overground running. This was attributed to these two running conditions presenting with similar levels of co-contraction of the rectus femoris and semitendinosus muscles (62).

Essentially, what these studies show is peak tibial acceleration magnitude during running is a byproduct of multiple interactions and cannot be strictly explained by surface, kinematics, or muscle activation (54, 62, 95, 96). Since this current study did not record any direct measurements of these factors, it is difficult to conclude if any of these determinants had any effect on the subjects in this study.

Surface Effects on Within Subject Variability

We will now discuss the results relating to our hypotheses in terms of within subject variability. Moderate levels of stride variability have been considered protective against injury and dysfunction (12, 13, 72). The amount of variability in one’s gait has been used to predict the risk of an individual to develop certain ailments even though what is classified as "normal" variability is currently not established. The optimal range of variability is likely to be movement and subject specific, so the threshold for where injury occurs, and healthy tissues persist is unknown (13, 72). Many of the studies that have investigated this have simply compared the variabilities of stride variables (i.e. stride time, stance time) and noted that healthy runners most often present with higher variabilities than injured runners (12, 13, 73). It is understood that too much variability in one's stride can also lead to injury or dysfunction, but what is considered excessive still needs to be investigated (72). This study decided to analyze the variability of a few biomechanical variables to see if running surface has any effect on the consistency of one's gait.

The findings of within subject variability of stance time, stride frequency, maximum rearfoot eversion velocity, and peak tibial acceleration will all be addressed in this section. As seen in Tables 4-5, significance differences in within subject variability were found for stance time, stride frequency, maximum rearfoot eversion velocity, and peak tibial acceleration across the different running surfaces ($p < 0.001$, all variables).

Although each significant finding is important and should be discussed, the main focal point of this section will be on the difference seen between treadmill and overground running. Treadmill running was shown to have significantly lower within-subject variability in stride frequency, maximum rearfoot eversion velocity, and peak tibial acceleration compared to all overground conditions, which is a finding that seems to be consistent among other studies that performed this same comparison (97, 98). Surprisingly, the intrasubject variability observed in stance time for treadmill was comparable to both track and asphalt ($p = 0.433$, $p = 0.204$), but was significantly lower than what was recorded for grass (avg. variability treadmill = 0.009 sec, grass = 0.014 sec).

Overall, intrasubject variability of three of the four variables in this study was significantly less when the subjects ran on the treadmill compared to any of the overground surfaces. Even though the number of studies in the current literature that address the reason for this occurrence is small, there have been some proposed explanations that could provide insight on the results in this study. Initially, it was thought that treadmill running or locomotion presented with a more stable gait because of the constant and predictable speed the treadmill belt (97, 98). This was eventually refuted by Wheat et al. (98) after they showed their subjects displayed no significant differences in coordination variability when they ran on a non-motorized treadmill (no constant belt speed) and conventional treadmill.

In result of this finding, Wheat et al. (98) and others have speculated that the differences in gait variability between treadmill and overground surfaces are due to other factors like the absence of wind during indoor running and alterations in the somatosensory information received by the runner (97, 98). Aside from these, some researchers have proposed that the minimal changes in visual stimuli related to treadmill running can be another reason for this phenomenon (97–100). It should be noted that during the data collection process of this study, it was not uncommon for subjects to express to the researchers that they "felt" like they ran faster on the treadmill, despite the speed being set to their self-selected pace that was used for all the overground conditions. This is an interesting point since some research has shown that "optical flow" or the movement of images in the central and peripheral aspects of one's vision can act as a modulator for speed used in locomotion (97–99). Essentially, when the flow of images or objects does not match up with the perceived walking or running speed, it is innate for an individual to either increase or decrease their speed to synchronize the speed sensed both by the visual system and the proprioceptors in the body.

Of course, when this synchronization of sensory information is occurring, the load on the nervous system can effectively increase leading to having less "attention" strictly focused on a singular task (97, 101). In terms of this study, it would nearly be expected for treadmill running to present with lower intrasubject variability compared to overground since treadmill running does not require the subjects to account for any changes in their optical flow during their run. This would theoretically allow them to be more conscious of their gait, therefore, resulting in more consistent measurements of stride frequency, maximum rearfoot eversion velocity, and peak tibial acceleration.

This idea of the level of sensory information dictating stability in one's gait can even be applied to explain the high variabilities seen in all four variables in this study when the subjects ran on grass. The track facility used in this study was open to the public, so when the subjects performed their overground running trials, it is plausible that they were receiving large amounts of sensory stimuli. They were focused not only on the location of their next step, but also tracking objects (i.e. timing gaits) and unfamiliar people that were occupying their visual field. In addition, their lower extremity was presumably still processing the same amount of somatosensory information it typically would while running on a treadmill. This could mean that running overground, especially on grass since it was probably the most inconsistent surface, requires larger allocation speeds from the nervous system than running on a treadmill where most external factors are controlled or static. Consequently, the lower extremity allows for more variability in the kinematics of the runner, which would result in more variability in the outcome variables of this study.

Implications of Results

Considering this study found multiple differences between running surface and a few biomechanical variables, the real-life implications of these findings deserve a bit of discussion. Even though this study did not assess injury specifically, this study did measure variables that have been associated with certain running-related injuries (RRIs). Therefore, the differences found in this study may add to the volume of research that has investigated the effect of surface on injury. Although the subsequent paragraphs of this section may be concentrated with recommendations for runners to possibly mitigate or avoid injury, it must be understood that this study was not a clinical study so the implications of the results should be considered with caution and not as absolute truth. Clinical recommendations and treatments should be reserved for clinicians (i.e. physical therapists, athletic trainers, medical doctors) that have specialized training in the matter of running and RRIs.

With that said, based off the findings of this study, specific surfaces may be used as a training tool to help certain runners avoid or limit injury. To start, this study found significantly higher stride frequencies when the participants ran on a treadmill compared to any of the overground conditions. Additionally, peak tibial acceleration was also shown to be significantly reduced while running on a treadmill. Both increases in stride frequency and reductions in peak tibial acceleration have been associated with decreases in both the rate and magnitude of the forces a runner experiences (53–55, 102). These reductions in overall impact forces are crucial for all runners considering heightened impact forces have been associated with RRIs, especially the injuries relating to the bony structures of the lower extremity (53, 54). In effect, running on a treadmill may be used by any runner that may have a predisposition to bony structure injuries (i.e. shin splints, stress fractures) or naturally high impact forces.

Fortunately, some potential benefits could also be derived from running on overground surfaces. In terms of what surface would be best for runners that tend to overpronate or are affected by abnormalities in their rearfoot motion, running on grass appears to be a good surface to implement in one's normal training regimen. The reasoning behind this implication is that the participants in this study presented with the lowest values of maximum rearfoot eversion velocity when they ran on a grass surface. Even though this study did not measure the absolute changes (i.e. angles) in the runners' pronation, the assumption that slower rearfoot motion/eversion would benefit overpronators by allowing them to control the motion of the foot with every step over a longer period of time rather during quick impacts. Now the effect of running on harder surfaces such as asphalt and track is quite the opposite. Since running on these harder, overground surfaces present with significantly higher levels of maximum rearfoot eversion velocity, one can believe that this would benefit runners that suffer from soft-tissue injuries (i.e. Achilles pain, plantar fasciitis) since the muscles and tendons of the feet would not have to eccentrically control the load of impact over a prolonged period of time.

The real-life implications of the results of the variability of the different biomechanical variables in this study are a bit more unclear. Overall, the general understanding of the effect of variability in one's gait has on the chance on developing a running-related injury is still ambiguous. Even though some have postulated that having lower levels of gait variability is a result or precursor to injury (12, 13), very large levels of variability are likely to have a negative effect on a runner's health as well (72). In addition, there does not seem to be a standard value of what is considered "good" levels of gait variability, which makes it difficult to conclude which running surface would benefit most runners. Given variability in all the variables in this study were the least consistent when running on grass, there is a general temptation to say running on grass would provide the most benefit when it comes to the prevention of injury. Unfortunately, due there not being a standard for what is considered “normal” or “healthy” levels of variability, there is an equal chance that the amount of gait variability on grass may be excessive, which could inadvertently lead to injury instead of preventing them. In result of this predicament, the best form of advice for runners based off the results of this study is to try to include balanced amounts of running that limit variability (i.e. treadmill running), but also encourage it (i.e. grass running).

Specific Aim 2:

The purpose of Specific Aim 2 was to test the reliability of rearfoot motion data acquired from a heel mounted IMU. The validity of the IMU device was tested by first calculating maximum rearfoot eversion velocity from the gyroscope component of the IMU and comparing those values with what was determined via high-speed videography. There are currently inconsistent findings on the accuracy of IMUs and gyroscopes to approximate foot kinematics, especially in the frontal plane (23, 32, 42). This Specific Aim was conducted to contribute to the limited amount of research done on using wearable technology to estimate rearfoot motion in running (19, 23, 31, 32, 42, 43, 83).

We hypothesized there would be no difference in maximum rearfoot eversion velocity calculated from an IMU and the values determined from high-speed videography. According to the ICC value (0.739) obtained in this study, it is fair to conclude this hypothesis was partially supported. This intraclass correlation is considered to represent a moderate reliability for the IMU maximum rearfoot eversion velocity values when compared to videography (87). This finding is comparable to a few of the other studies that have analyzed the agreement between IMU or gyroscope predicted values of rearfoot motion data and other reference systems (23, 32, 43). Specifically, Shih et al. (23) reported a moderate correlation ($r = 0.651$) between the frontal plane data they acquired from a gyroscope that was attached to the dorsal aspect of the foot and their motion analysis system. Additionally, Lederer et al. (43) noted a strong correlation ($0.85 < r < 0.88$) between the maximal pronation velocity values determined by a heel mounted gyroscope and the values recorded by their motion analysis system. Surprisingly, there is one previous study that actually reported very strong correlations ($r > 0.9$) between the peak eversion velocity values of a heel mounted gyroscope and an electrogoniometer they used while comparing the stiffness of running shoes (83).

Even though these studies are relatable to the current study because they all evaluated the reliability of rearfoot motion data from wearable technology, it is difficult to truly compare the results for specific reasons. First, all these studies tested the validity of their gyroscope-based measures by calculating the correlations between the gyroscope data and their motion analysis system, while this study elected to use intraclass correlation coefficients instead to understand the agreement between the IMU device and our high-speed videography data. Secondly, the methods used to calculate rearfoot motion (i.e. maximum rearfoot eversion velocity) with the wearable forms of technology differed across studies. Some of the previous studies acquired eversion data from their reference systems by placing markers on the shoe of their subjects and recording the displacement of these markers from a global coordinate system. Alternately, this study and Mitschke et al. (83) recorded eversion in respect to the subtalar joint by calculating the angle of the rearfoot segment in reference to the tibia.

Lastly, the mean differences (bias) between the IMU device and the high-speed videography will be discussed. According to other studies, it is common for IMU-based frontal plane motion to overestimate what is seen in motion analysis (32, 43). This was also found in this study and can be observed in the Bland-Altman plot (Fig. 5) that was generated in the results section of this paper. The mean difference is represented by the bolded, black line that extends across the graph and is set at approximately 12.67 deg/s on the y-axis. This mean difference between the IMU device and high-speed videography was calculated by obtaining the maximum rearfoot eversion velocity value from the IMU and subtracting it by the value determined from the video data. Essentially, this shows that the IMU on average overestimated the magnitude of maximum rearfoot eversion velocity by 12.67 deg/s. This would correspond to about 2 percent of the average rearfoot eversion velocity recorded across all subjects.

One of the studies that reported overestimations on eversion velocity did not explicitly report the mean difference between their two systems (32), but fortunately the other study found that addressed this same result did provide the root mean square error between the gyroscope they used to calculate maximum pronation velocity and their motion capture system (mean RMSE: 43.6 deg/s ± 17.0 deg/s) (43). Since the calculation of root mean square error is different than what was used to determine the bias between the IMU and high speed videography in this study, it difficult to conclude if the observed bias is considered acceptable or relatable (43). Overall, a heel mounted IMU appears to present with moderate reliability in terms calculating maximum rearfoot eversion velocity during running and displays with moderate, but also an inconclusive amount of agreement with high-speed videography. Therefore, maximum rearfoot eversion velocities collected from an IMU remain to be estimations that should be interpreted with caution (32).

Limitations

This study had several limitations. First, we did not control for each subject's footwear during their participation in this study. We did ensure the same running shoes were used during the overground and treadmill running conditions since changes in footwear can alter some of the variables calculated in this study (i.e. maximum rearfoot eversion velocity, peak tibial acceleration) (54, 65, 67, 69). We also did not recruit a specific type of runner, meaning we did not attempt to enroll participants based off characteristics such as habitual foot strike pattern, average weekly mileage, injury pattern, or years of experience. The only inclusion criteria for this study were all participants had to be 18 year or older, free of any musculoskeletal injury in the last 6 months and have no history of any cardiopulmonary conditions.

Another limitation of this study relates to the availability of certain equipment. At the start of data collection, we only had access to two IMUs, which were used to record data from the right leg of each subject. If more IMUs were available, this study could have expanded the number of potential variables that could have been collected. Also, the validation of the IMUs in this study was done by comparing the IMU-based eversion velocity data to 2D high-speed videography, while most other studies had access to a 3D motion capture system (23, 32, 43). We are not implying the results of the reliability testing of the IMU would have been different if another form of video analysis was done, but all eversion angles were determined by a free, open-source software that might not detect changes in position to the same precision level as a gold-standard motion capture system.

The timing system used in this study was only designed to record reliable speeds up to about 37 meters. This limited the distance each subject could run for the overground trials, which inherently limited the number of steps that could have been analyzed in each trial. This study also did not test for all possible running surfaces due to time restrictions. Track, grass, and asphalt were selected since they were all in proximity to each other at a public track facility and are some of the most common running surfaces used. Therefore, we did not account for other running surfaces such as trail, sand, turf, or concrete. Furthermore, only a standard motorized treadmill was used in this study, so the results from the treadmill running condition may not reflect the findings that would be seen on a non-motorized, self-paced treadmill (62, 98).

An additional limitation of this study is we did not assess whether running surface had any effect on the variability of each runner's speed. Even though we controlled for running speed by constraining all subjects to ±5% of their self-selected speed, it is plausible that the subjects could have consistently ran at the upper range of this allowed speed interval on some surfaces and then at the lower range on other surfaces. Theoretically, this could result in a subject potentially running 10% faster or slower between two different running surfaces, which could have affected our results. However, to confirm that this did not occur, we performed a one-factor ANOVA with speed as the dependent variable and running surface as the independent variable. The results from the one-factor ANOVA showed no significant differences in running between the conditions ($p = 0.999$) so we are confident that running speed had no significant effect on the within-subject differences we observed in this study.

Lastly, this study did not investigate the relationship between running-related injuries and running parameters. Even though some of the variables in this study have been used to differentiate between injured and non-injured runners, we did not analyze the effect of previous injuries on each subject's biomechanics. Additionally, due to this study not accounting for injury status, we were not able to truly test if certain running surfaces increase or decrease a runner's risk of developing a given injury. Therefore, the results from this study may not be used to help explain the causes of specific running-related injuries.

Future Work

Through the development and implementation of this study, there have been a few unanswered questions or problems that could be transformed into future studies. First, despite the volume of research done on the determinants of peak tibial acceleration, there still remains a level of uncertainty of what truly causes peak tibial acceleration to be elevated or reduced in certain conditions of running (54, 95, 96). Future studies could attempt to investigate this topic further since peak tibial acceleration has appeared to gain a high level of appearance in the field of running biomechanics. Second, this study was able to show that across different running surfaces, treadmill running appeared with the lowest levels of variability in three of the four variables collected. Even though there are some current studies discussing the effects of running surface on injury (103–105), no studies could be found on what surfaces would be the most suggested for the mitigation of running-related injuries. Future studies can investigate whether these low levels of variability in treadmill running fall within the optimal range of stride variance for most individuals. If so, treadmill running could be beneficial for at risk runners.

Third, a follow-up study of this current one could be performed on the analysis of gait asymmetry while subjects run on different surfaces. Similar to stride variability, gait asymmetry has been a metric used to detect or predict pathological gaits and risk of injury, respectively (106). Even though gait asymmetry has been shown not to be significantly different between overground surfaces (91), gait asymmetry has been reported to be higher in overground running than when subjects run on a treadmill (107). With the advancements of wearable technology like accelerometers and IMUs, the comparison between treadmill and overground running could be performed easily by comparing the symmetry index of a given variable from each side of the lower extremity (106, 108). Furthermore, IMUs could be used to measure differences in gait asymmetry between injured and non-injured runners. For example, runners that run on surfaces that promote asymmetry (i.e. crowned roads) have been documented to have an increased incidence of injuries like Achilles tendinitis (109). In this instance, IMUs could be used to measure the degree of asymmetry in these affected runners and then be utilized again after medical treatment to assess whether any excessive amounts of asymmetry have been removed prior to the athlete returning to their normal training regimen.

Lastly, through the review of current studies that have used IMUs and gyroscopes to reliably analyze rearfoot motion during walking and running, it is recommended that some future studies are devoted to developing methods or procedures that allow for more consistent and valid data acquired from these wearable devices. Some articles have presented with some levels of success in this topic (110), but most do not seem to be confident in the reliability of rearfoot motion data from wearable technology (32, 43).

Conclusion

The purpose of this study was to compare multiple running surfaces through the use of inertial measurement units and to assess whether these devices can collect reliable rearfoot eversion velocity data during running. Based on the findings, this study was able to show there are some fundamental differences in a runner’s gait when they run on different surfaces. Most of these differences appeared to occur between treadmill and overground running, but significant differences were found within overground surfaces, as well. Generally, treadmill running tends to lead to a faster stride frequency, lower stride variability, and significantly reduced magnitudes of impact loading.

Even though the differences seen between overground surfaces were generally smaller than what was observed between overground and treadmill running, there were differences, nonetheless. This shows that even when it comes to overground running, there is not one surface that is the same. Each running surface will have its own effect on a runner's gait and will provide both benefits and detriments in terms of a runner's health. Possibly the best form of action to prevent overuse injuries in runners is to encourage the use of a multitude of running surfaces to promote variability in one's training.

Lastly, IMUs have been shown to present with moderate levels of reliability when recording rearfoot eversion velocity, but this does not mean this same level reliability will be seen with other frontal plane variables (i.e. pronation). Calculating kinematics with IMUs is still an upcoming field in the realm of wearable technology in running biomechanics, so progress in the methodology of recording frontal plane movements with IMUs needs to be a focus for future research.

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Appendix

Dear Participant,

This letter is a request for you to take part in a research project that will be investigating the differences in treadmill and overground running. This project is being conducted by Griffin Moon in the Department of Human Performance and Applied Exercise Science at WVU under the supervision of Dr. McCrory, an Associate Professor in the Department of Human Performance and Applied Exercise Science, to fulfil requirements for a Master's Degree in thesis track of the Exercise Physiology program.

If you decide to participate, you will be asked to run on three different overground/outside surfaces (e.g., track, grass, asphalt) and a motorized treadmill. You will first be given a brief survey that will ask you about your running experience (i.e. average weekly mileage, typical running surface, sport participation). Following the survey, two inertial measurement units (IMUs) will be attached to your right lower leg and shoe. You will then be asked to run 3-5 trials of 40 meters on each overground surface at a self-selected pace. We will then place 4 markers/stickers on your right calf and shoe prior to your run on a motorized treadmill. These markers will be used to help us record your foot and lower leg movements during your run. The treadmill speed will be set at the same speed you used for the overground surfaces and you will be asked to maintain this pace for 1-minute. During this time, we will have a video camera placed behind the treadmill so we can record your foot motion with every step. Once you complete the treadmill run, we will measure specific dimensions of your feet such as foot length, arch height, and arch rigidity. Following the foot measurements, your participation in this study will be complete. Your participation in this project will take approximately 1.5 - 2 hours to complete. You must be 18 years of age or older to participate and free from any musculoskeletal injuries in the past 6 months. You also need to be free from any cardiovascular or pulmonary conditions that may inhibit your ability to perform multiple bouts of running.

Your involvement in this project will be kept as confidential as legally possible. Your research records and test results may be subpoenaed by a court order or may be inspected by the study sponsor or federal regulatory authorities without your additional consent. All your data will remain anonymous through the assignment of a subject number, so specific information such as name, date, or birth will not be used as identification in this study. You will not be asked any questions that could lead back to your identity as a participant. Video footage obtained during the study will only record your feet and lower legs so your identity will remain protected. If you are a student, your class standing will not be affected if you decide either not to participate or to withdraw. West Virginia University's Institutional Review Board approval of this project is on file. You are free to withdraw your consent to participate in this study at any time.

If you have any questions about this research project, please feel free to contact me at 304-657-8175 or by e-mail at grmoon@mix.wvu.edu or Dr. McCrory at jmccrory@hsc.wvu.edu. If you have any questions about your rights as a research participant, please contact the WVU Office of Human Research Protection by phone at 304-293-7073 or by email at IRB@mail.wvu.edu.

I hope that you will participate in this research project, as it could help us better understand whether new forms of technology like IMUs can detect differences between running on different surfaces. Thank you for your time and consideration.

Sincerely,

Griffin Moon

Participant Survey/Questionnaire

Subject Number: ______________________________

Age: _________

Height: _________

Weight: _________

Gender: Male Female

Average Weekly Mileage: ______________________________

Circle the surfaces you most commonly run on:

Concrete Sidewalk Asphalt Grass/Turf Treadmill Trail

Mark on the line corresponding to either YES or NO for the following questions:

Yes	NO	
___	___	Have ever competed in sports that involve large amounts of running (i.e. track & field, cross country, soccer, basketball, field-hockey, tennis, rugby)?
___	___	Do you currently run for competition or leisure?
___	___	Have you ever received running lessons to improve or change your running form?
___	___	Have you ever experienced any running-related injuries (i.e. shin splints, knee pain, Achilles pain)? If so, please list them below.
___	___	Do you have any current lower extremity injuries?
___	___	Have you experienced any lower extremity injuries in the last 6 months?

Data Collection Form

Subject Number: ____________________

Date: ____________________

Self-Selected Speed: ____________________

Speed Range (± 5%): ____________________

Foot strike pattern: ____________________

Surface Speeds

	Track	Grass	Asphalt	Combined Surfaces
Trial 1 (m/s)				
Trial 2 (m/s)				
Trial 3 (m/s)				
Avg. Speed (m/s)				

1 m/s = 2.237 mph

Treadmill Speed: ____________________