INVESTIGATING MEASUREMENT MODALITIES FOR INTEGRATED USE IN OVERUSE INJURY RESEARCH

by

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Abstract

Running is a common form of exercise enjoyed throughout the world. Unfortunately, it poses a significant risk of injury, as up to 70% of runners suffer an overuse injury within a one-year period. Previous research indicates that three overarching factors contribute to overuse injuries in running – dosage, morphology, and movement patterns; however, the threshold of these three factors combining to result in injury is specific to each athlete, and not well understood.

To aid future research of overuse injuries, we conducted two studies using three different measurement modalities. In the first study, we used Inertial Measurement Units (IMUs) to monitor gait features of participants running on a treadmill. Using statistical parametric mapping of IMU signals, the system detected changes in strike pattern, stride length, bouncing gait, and genu valgus gait. With further validation and refinement, this system would allow for monitoring of these gait features in the wild, giving researchers greater insight into the development of overuse injuries.

The second study was a pilot study, in which we investigated motion of the tibia and relative motion of the talus and calcaneus with respect to the tibia during a hopping motion. We found that internal rotation of the talus after impact leads the internal rotation of the tibia. This grants plausibility to the theory of tibiotalar coupling – that internal rotation of the tibia is driven by that of the talus. After approximately 20% of stance, the talus moves into external rotation with respect to the tibia, and we hypothesize that it acts as a buffer to allow rotational independence between the rearfoot and tibia. Additionally, this study highlighted the inherent inter-subject variability in biomechanics studies. We have shown that there is benefit and necessity in subject-specific analysis over population-wide conclusions in biomechanics.

Finally, we discuss the implications of both studies to the future of overuse injury research. The research presented is a first step toward more comprehensive studies into overuse injury etiology. Using the approaches described, researchers may be able to use a combination of measurement modalities to encompass dosage, morphology, and movement patterns in a large-scale, prospective overuse injury study.
Co-Authorship

This thesis contains the original work of Liam Rodgers, completed under the supervision of Dr. Michael J. Rainbow. Chapter 2 was prepared as a manuscript which was submitted to the *Journal of Applied Biomechanics*. Portions of this chapter were presented at a joint meeting of the International Society of Biomechanics and American Society of Biomechanics in August of 2019, and were also submitted for presentation at the Canadian Society of Biomechanics in 2020. Co-authors for this chapter include Dr. Jessica C. Selinger, Dr. Ali Etemad, and Dr. Michael J. Rainbow. Experimental design, data collection, data analysis, and writing of the manuscript was completed by the author, in consultation with the co-authors. Chapter 3 was based on a data collection conducted by Sarah Kessler. Data analysis and writing of the chapter was completed by the author.
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Chapter 1

Introduction

Musculoskeletal human biomechanics is studied with a wide range of modalities. Laboratory-based studies that utilize modalities such as optical motion capture, force plate measurements, biplanar videoradiography, EMG, or ultrasound provide highly detailed human motion data. Studies in the laboratory also provide control over the independent and dependent variables that are being studied; however, they are often limited in their ecological validity for motions in the wild. Ecologically valid motions are particularly crucial in the study of overuse injuries since an accurate representation of motion is required to determine injury causation. Conversely, studies conducted using mobile monitoring methods collect ecologically valid data; however, it is often less detailed than data that can be collected in the lab.

The purpose of this thesis was to conduct preliminary research to allow the merging of ecologically accurate in the wild measurements with detailed high-fidelity biomechanical data including skeletal interactions. Merging several measurement modalities will allow researchers to gain further understanding of the etiology of overuse injuries in running by incorporating dosage, morphology, and specific motor patterns together in their research.

1.1 Overuse Injury Studies

Overuse injuries in running are a common problem among professional and recreational athletes alike, with up to 70% of runners sustaining an overuse injury in a one-year period [1]. This problem is compounded by the conflicting conclusions of previous research into injury causation. It is widely accepted that a subject-specific combination of factors including activity level (dosage), underlying structure, and motor pattern can push an individual over their injury threshold, or protect
them from injury [2]–[6]; however, predicting this threshold for an individual in hopes of preventing overuse injury remains a challenge.

Common overuse injuries in running include shin splints, stress fractures, Achilles tendonitis, and patellofemoral pain [5]–[7]. Unfortunately, despite the high instance of overuse injuries, longitudinal studies often do not see enough participants suffer from each of these injuries to warrant a separate analysis for each [8], [9]. Despite the likelihood of different root causes of each injury, an analysis is instead often completed of all overuse injuries together. Additionally, previous overuse injury studies have not reached consensus on injury causation, in part due to their methodology. Many motion-based studies are cross-sectional in nature [3], [6], [10], [11], and recruit participants who have already been injured [3], [6], [12], [13]. As such, the motions of participants in these studies may be symptomatic in nature, rather than causative. In the few prospective studies that have been conducted, the factors analyzed are often not all-encompassing. For example, studies may collect a ‘snapshot’ at the start and/or end of a study of the participants’ running mechanics [8], [10], [14]. Such an approach does not allow for analysis of changes in running mechanics over time, or changes in training regime. Further, depending on the data collection methodology, measurement and analysis of bone morphology may also not be incorporated in these studies.

An additional hurdle regarding the study of overuse injuries is to bridge the gap between biomechanics studies and epidemiological studies. While biomechanics-focused studies tend to contain a high amount of detail, often focusing on participant motion, kinetics, muscle activity, and/or bone interactions, they are often limited in the number of participants they include [3], [6], [15]. This can be problematic as it reduces the applicability of the findings of the study to the greater population. Conversely, epidemiological studies often include many participants; however, the data presented offers few biomechanically relevant details, for example relying on questionnaires regarding training regime, and not considering bone structure or underlying biomechanics of
motion [16], [17]. To reach a middle ground capturing the benefits of both styles of study, a large-scale study could be designed that monitors typical epidemiological factors such as training regime, as well as runner kinematics. To give further detail to the findings, participants representative of typical motions in the injured and uninjured groups could also be brought into a biplanar videoradiography (BVR) lab to provide additional details of underlying bone morphology and interactions.

1.2 Current Predictors of Overuse Injuries

Previous research into the causes of overuse injuries in running has determined that there are three groups of factors which can increase an athlete’s risk of injury: training regime, specific movement pattern, and biological structure [2]–[6]. While each of these factors has been shown independently to affect the risk of overuse injury, the injury threshold exceeded by their combination in an individual is not well established, and a multifactorial approach should be emphasized in overuse injury studies.

1.2.1 Training Regime

Training regime has been studied extensively as a risk factor of overuse injuries [18]–[22]. Specific factors have included amount and frequency of training, changes in running mileage, and running experience, among others. Overall, it has been shown that more experienced runners are at an increased risk of overuse injury [18], as well as runners with a history of overuse injury [9], [19]–[21]. It has been proposed that all overuse injuries in running could be considered a result of training regime errors – if a runner reduced their training load to below their injury threshold, perhaps to zero, they would not sustain an overuse injury [1]. While this logic cannot be disputed, it does not aid in the prevention of overuse injuries, as the injury threshold is different for every individual, and not participating in running may not be an acceptable prevention method for many. There is however, a consensus that longer runs, and more frequent runs both increase the risk of overuse injury because they do not allow time for the body to heal from the stress placed upon it.
Finally, environmental factors of training regime have been studied. For example, running surfaces can play a role in the onset of overuse injuries, with sand drastically increasing the relative risk of Achilles tendonitis [18].

1.2.2 Biological Structures

Biological structures remain a debated contributing factor to overuse injuries. There have been studies that suggest that a higher arch increases risk of injury [6], [23], [24], as well as studies that indicated that this was not a factor [25]–[27]. Similarly, research has been conducted that indicated risk of injury increases with greater ankle range of motion [6], [24], lower ankle range of motion [25], and that ankle range of motion does not change the risk of overuse injury [3], [28]. Even extremity asymmetry and alignment abnormalities have been both found [29]–[31] and not-found [25]–[27] to increase the risk of overuse injury.

In studying kinematic differences in gait patterns between males and females, it has been noted that female runners tend to have greater knee abduction as compared to males [32]. It is hypothesized that this may be a contributing factor in the increased likelihood of females to develop patellofemoral pain [33], [34]. It is important to note that biological and anthropometric differences alone may not be found consistently to change the risk of overuse injury. Due to the complexity of the musculoskeletal system, any such differences may be compensated by a change in joint kinematics of other joints, or a change in movement patterns. As such, a multi-factorial approach to the study of overuse injury causation is required to draw meaningful and actionable conclusions.

1.2.3 Movement Pattern

There is increasing agreement among biomechanics researchers regarding certain kinetic and kinematic variables and their links to overuse injury. For example, impact force and rate of loading at impact have been implicated to increase the risk of overuse injury [1], [3], [35]. In addition, both magnitude and rate of foot pronation have also been extensively studied as a risk
factor [6], [24], [27]. In the literature, it is found that an increased range of foot pronation during gait can be linked specifically to anterior knee pain, but not other overuse injuries [36]. This result highlights the need to research risk factors of each overuse injury independently, due to their varied causes.

Additionally, researchers have had success in retraining the gait of injured populations. Gait retraining has been found to reduce pain and improve function in individuals with patellofemoral pain [37], and osteoarthritis [38]. Although these examples of gait retraining occur after injury, gait retraining has also been used to improve knee alignment, resulting in a lower knee adduction moment [39]. Due to the relation between increased knee adduction moment and osteoarthritis development [40], it is expected that this retrained decrease in knee adduction moment will result in a decreased risk of osteoarthritis progression as well. The ability to alter joint kinetics, such as knee adduction moment, through gait retraining highlights the contribution of specific motion patterns to the development of overuse injuries, as injury-causing joint kinetics were improved by a change in body kinematics, and without a change in bone morphology.

1.2.4 Tibio-Talar Coupling

While the etiology of many injuries are not yet fully understood, previous research has linked pronation of the foot with internal rotation of the tibia and the development of patellofemoral pain [41], [42]. Previous work in this area has relied on skin-based landmarks for optical motion capture [36], [43], which are unable to measure motion of the talus. As such, the motion of the talus and contributions of the subtalar and talocural joints are not known, as talus motion cannot be observed with skin-mounted markers. It is believed that internal rotation of the talus drives the tibia into internal rotation as well, and that the rotation of these bones are coupled during early stance [44], [45].

The tibia is the main weight-bearing bone in the lower leg and sits atop the talus at the ankle, forming the tibiotalar, or talocrural, joint (Figure 1.1). At the talocrural joint, the tibia glides,
predominantly through flexion and extension, on top of the smooth superior surface of the talus. Additionally, the talus articulates with the calcaneus below it, forming the subtalar joint. Foot pronation, which involves both the subtalar and talocrural joints, has been linked to an increase in tibial internal rotation during footfall in running [46]. Further, it has been suggested that increased tibial rotation has the potential to increase the risk of overuse injury due to the torque which this creates across the lower leg [6], [47]–[49]. Although the link between foot pronation and tibial internal rotation has been widely accepted in literature, recent studies have indicated differences between male and female populations [43]. Due to the disparity in the prevalence of patellofemoral pain between males and females [33], [34], the underlying mechanisms linking foot pronation to tibial rotation should be investigated further.
The talus is perhaps the least studied and understood of the 3 bones that form the ankle joint complex. This is in part due to the difficulties observing the talus in a non-invasive manner. While overall lower-leg motion can be treated as a proxy-measure for tibia motion, and calcaneal motion can be visually observed with external landmarks, the same techniques are not possible for the talus, due to its position within the ankle. Previously, detailed analyses of talus motion were limited to cadaveric studies; however, with the emergence of non-invasive techniques, such as BVR, in vivo talus motion can be studied in greater detail.

1.3 Measurement Systems for Analysis of Human Movement

My work used three different modalities for the analysis of human motion, each with its own specific advantages and limitations. Although optical motion capture is the most widely used system in biomechanics, inertial measurement units are an emerging technology whose use is increasing, and biplanar video radiography provides greater detail of bone interactions at a scale unachievable with other measurement systems. My research focuses on these measurement systems due to their potential to encompass all three overarching factors in the development of overuse injuries – dosage, morphology, and specific movement patterns.

1.3.1 Optical Motion Capture

Optical motion capture uses a series of infrared cameras, as well as reflective spherical markers affixed to the body. With known camera locations, the images of the reflective markers can be triangulated to determine their location in 3D space. With at least 3 markers on a body segment, its position and orientation can be determined. Although most studies use this technology for tracking of entire limb segments, there are some studies that have also used smaller markers...
and more refined marker placement in an attempt to gain further insight of specific motions, such as calcaneal eversion and inversion [36], [43]. Optical motion capture systems can also be easily coupled with additional measurement techniques, such as force plates and electromyography. Using this combination of measurement systems allows researchers to calculate joint forces and moments up the kinetic chain, as well as to monitor muscle activation. Additionally, using instrumented treadmills, researchers can reduce the number of variables in their experiments by controlling elements such as speed and slope.

Due to the nature of optical motion capture, it is typically a laboratory-based method of data collection. While modifications can be made to a lab environment, such as a treadmill for running-focused studies, this method of data collection is inherently less ecologically valid than data captured in the wild [50], [51]. Further, techniques used to affix marker clusters to participants can restrict motion and require participants to wear clothing which they may not typically wear during the activities being studied. While optical motion capture has led to great advances in biomechanics, the relation of these results to real-world activities has been questioned due to the abnormal environment and clothing requirements of the system.

### 1.3.2 Inertial Measurement Units

In recent years, advances in inertial measurement units (IMUs) have led to an increase in the use of wearable devices for analysis of human motion. IMUs designed for use in research are now capable of recording accelerometer, gyroscope, and magnetometer data in all 3 axes at greater than 500Hz. These devices can easily be used in any setting, inside or outside of a laboratory. Additionally, this measurement technique does not require specific clothing to be worn by the participants and can capture a larger quantity of data in a single trial. Overall, IMUs are praised for their ability to collect data in the natural environment of an activity.

These advantages are not without accompanying limitations. Unlike optical motion capture, where several points on a segment are tracked to present the overall motion, an IMU can
only record data of the exact location where it is placed. Additionally, while optical motion capture collects position information, IMUs collect acceleration, angular velocity, and magnetic field data of individual sensors. As such, certain biomechanical measures, such as joint angles, require sensor fusion, and a calibration of sensor relative positions. These sensor fusion techniques are sensitive to complications such as sensor drift throughout the trial, which is a common issue with gyroscopic readings. Additionally, if magnetometers are relied upon, they are sensitive to variations in the surrounding magnetic field, such as from electronics, machinery, motors, or large quantities of metal.

One of the main limitations of IMUs is their inability to accurately reproduce the same biomechanical measures that can be measured with optical motion capture. Due to the prevalence of optical motion capture in biomechanics, the focus of a new technology such as IMUs is on replication of previously used signals, such as joint angles, as opposed to analysis of the available signals from the new technology. Success has been seen with direct analysis of IMU signals in studies which have researched shank acceleration and its relation to tibial stress fractures [52]–[54]. In these studies, tibial acceleration has been measured directly using the accelerometer in an IMU. Similarly, a research approach using directly measured angular velocity data from an IMU gyroscope may lead researchers to conclusions which were unattainable through other measurement modalities. Because the human body is constantly managing and effecting angular motions around joint centres, angular velocities of limb segments can hold biomechanical meaning, as they are the link between muscle-driven angular accelerations, and end-effect limb positions. Additionally, the use of IMUs to measure angular velocities may provide biomechanics data from environments which cannot be studied using optical motion capture.

**1.3.3 Biplanar Video Radiography**

Biplanar video radiography differentiates itself from the previously mentioned measurement systems with the detail of information which it provides [55]. Unlike optical motion
capture or IMUs, which measure only segment motion, biplanar video radiography can measure motion of individual bones within the body in a non-intrusive way, and has previously been used to evaluate motions of the wrist [56], knee [57], and feet [58], [59]. Biplanar video radiography uses two x-ray sources to take videos of bone motion. With known locations and orientations of the cameras, computed tomography (CT) scans of the participant’s bones can be fit to the video frames to track the motion of each individual bone.

While this system provides greater detail of bone motions, the environment is restricted to a narrow runway or treadmill, and the capture volume is less than the size of a basketball. In addition, since x-rays and the required CT scans expose participants to a limited amount of radiation, a large number of trials cannot be collected for safety reasons.

While each measurement system presents distinct advantages over each other, few approaches have combined all three modalities to take advantage of each system’s strengths. For example, a greater understanding of ankle biomechanics during running could be gained if motions detected by IMUs outside of the laboratory environment could be mapped to typical optical motion capture measures, as well as detailed bone motions which can only be directly measured using laboratory-based technology.

1.4 Summary

Overuse injuries in running are a prevalent issue among recreational athletes. It is widely accepted that a combination of bone structure, running gait style, and exercise frequency all contribute toward overuse injuries; however, the interactions between these factors has not been studied extensively, and thus there are conflicting results regarding the causation and prevention of overuse injuries.

Current biomechanical measurement systems can all contribute toward further understanding of overuse injuries in different ways. Optical motion capture can be used to analyze running gait style in great detail at a segmental analysis level, but does not provide insight toward
bone structure, underlying bone interactions, or exercise frequency. IMUs can monitor running gait style with perhaps less detail than provided by optical motion capture, but they can monitor gait in the wild, and can provide details of running frequency and training regime when used over a long period of time. Finally, biplanar video radiography can provide details of underlying bone structure and joint interactions unlike either of the other systems.

The purpose of this thesis was to conduct preliminary work that begins integrating IMU gyroscopic data, optical motion capture, and biplanar videoradiography to ultimately map running biomechanics to overuse injury.

1.5 Objectives

To further our understanding of the factors involved in overuse injury, we performed one full study and one pilot study, which together used all three measurement systems. The first study serves as a proof-of-concept for the future use of IMUs to monitor running gait styles, and relate any changes in gait to changes in joint angles, which would typically be found using optical motion capture in the lab. The primary objective was to use statistical parametric mapping to detect changes in several gait parameters, such as strike pattern, stance width, and stride length, using a resultant angular velocity signal. The secondary objective of the study was to use the optical motion capture data to determine the biomechanical differences across these gait styles through analysis of joint angles. Thus, using a robust method of gait monitoring which could be conducted outside of the laboratory environment, a researcher can gain insight of a participants specific running style. The second study used an available data set of hopping motions to increase our understanding of talocrural joint motion during forefoot landing. The primary objective of this study was to observe if there was evidence of coupling between the internal rotation of the talus and tibia in early stance. Additionally, the variables used to evaluate coupling were evaluated on both a population level and a subject-specific level to draw initial conclusions on the generalisability of results in high-fidelity studies. Together, these studies lay the groundwork for future investigations relating various
running styles in the natural environment to their underlying bone interactions and potential for overuse injury using optical motion capture, IMUs, and biplanar video radiography.
Chapter 2

Can IMU-Based Resultant Angular Velocities Detect Common Changes in Running Gait?

2.1 Introduction

Running is one of the most common forms of exercise in the world. Unfortunately, it also involves a risk of overuse injuries [7], whose etiologies are not fully understood [1], [18], [60], [61]. Previous research has often focused on demographic variables [62], [63] or training regimen [62]–[64]; however, these studies have reached mixed conclusions regarding the risk factors associated with overuse injury. Large-scale, longitudinal studies focused on gait analysis prior to overuse injuries may shed light on how different running styles or changes in running styles predict running injuries.

Wearable sensors, particularly inertial measurement units (IMUs), are an attractive modality for conducting large-scale longitudinal studies on running injuries [52], [65]–[67]. IMUs are lightweight, non-intrusive, and capable of capturing continuous data at high-temporal resolution during a long bout of exercise in the wild. IMUs have previously been used to study shin splints and stress fractures using axial acceleration of the distal tibia [52]–[54]. These studies have given insights into stress fractures and highlight the potential for IMUs; however, running-related injuries vary [1], [6] and one of the best predictors of running injury remains previous injury [9]. Unfortunately, this predictor is unsatisfactory toward preventing new injuries, and thus further research into injury etiology is required.

Kinematic gait analyses of the lower extremity have used joint angles to differentiate numerous pathological gaits from asymptomatic gaits [6], [36], [68]–[70]. To compute joint angles with IMUs, sensor fusion techniques are required [71], [72]. Computing joint angles with sensor fusion techniques requires correcting for or eliminating sensor drift, inconsistent sensor placement,
changing magnetic fields, and asynchronous sensors [73]. Using prior information and biomechanical assumptions to account for these challenges, recent algorithms have calculated joint angles to within 5° when compared to optical motion capture [74].

While many studies have focused on joint angles in running [70], [75], [76], few have focused on segmental angular velocities [77], [78]. Angular velocity may be an important measure because it is proportional to power, and musculoskeletal joints are constantly absorbing or generating power during loading and propulsion. Moreover, IMUs are designed to directly measure angular velocities of a rigid body using a gyroscope, eliminating the issues surrounding sensor fusion. In a long-term, prospective study, the modality of data collection and analysis should be both convenient for the participants to use, as well as robust to inconsistent sensor placement. The joints of the lower extremity exhibit complex multi-planar motions and changes in sensor placement can introduce intra- and inter-subject variability; however, resultant angular velocity measurements of segments are largely robust to variations in sensor placement. Most importantly, the use of resultant angular velocities for analysis eliminates the need for calibration before collection, increasing the convenience and ease of use by participants in a long-term study.

Large-scale collections of time series data are frequently analyzed with machine learning [68], [69], [79], [80] or data reduction [3], [72], [75], [81], [82] techniques. While powerful, machine learning techniques can limit the ease with which one can relate results to the participant’s running style, and interpret physics-based signals such as joint angles or power. Similarly, data reduction can concisely indicate focused changes in biomechanical data, but does not allow for complete interpretation of the time-series curve. Statistical Parametric Mapping (SPM) [83] is a method developed specifically for the comparison of multi-dimensional data, and is gaining popularity in biomechanics [84]–[86]. SPM allows for identification of the areas of time-series data that differ between data sets without the use of data reduction or machine learning techniques. Additionally, SPM indicates how the data curves differ from each other, and to what significance
level. Using SPM, resultant angular velocities of lower-extremity segments may also be able to detect changes in running patterns and differentiate gait styles similar to machine learning approaches [68], [69], [79], while preserving a physically interpretable signal.

The purpose of this study was to determine if SPM can be used on IMU-based resultant angular velocity signals of the shank and foot to identify regions of the gait cycle that indicate common running gait features. We hypothesized that there would be time points in the gait cycle when each gait feature tested, such as stride length or strike pattern, would be detected by a significantly different ($\alpha<0.05$) resultant angular velocity in at least 80% of subjects by either foot- or shank-mounted IMUs. We validated the resultant angular velocity curves measured by the IMUs with synchronized optical motion capture, and then mapped the identified gait style changes to corresponding changes in joint angles.

2.2 Methods

2.2.1 Data Collection:

We recruited ten runners (5F/5M, age 20-26 years) to participate in the study. All participants were recreational runners, who ran a minimum of 10 km/week over the previous three months and had no history of lower limb injury. Participants ran 8 separate bouts, each approximately five minutes (> 150 gait cycles) emphasizing certain gait features in each bout. The running gait features included modifications to stride length, strike pattern, and stance width, as well as a genu-valgus gait, and a ‘bouncing’ gait. Participants were instructed how to modify their gait (e.g. “In this trial, focus on always landing on your forefoot”) ahead of each 5-minute bout. Other features of gait were open to what the participant felt was most comfortable. For example, during the short strides trial, the participant was free to either rearfoot-strike or forefoot-strike, as the focus was solely on the short stride length. The order of the trials was randomized for each participant. All running was done on an instrumented treadmill (AMTI, Watertown, MA) at a self-selected comfortable running speed. An instrumented treadmill was used so that optical motion
capture data could be collected to validate the IMU data, and to control for variables such as speed and terrain. Participants were instructed to hop on the stationary treadmill 3 times at the start and end of each running bout for system synchronisation purposes.

To determine each participant’s comfortable running speed, they began walking on the treadmill at 1.5 m/s. The treadmill speed was increased by 0.1 m/s every 10 seconds until the participant stated that they were at a comfortable speed. After running at this speed for 30 seconds, the treadmill speed was increased by 25%. The speed was then lowered by 0.1 m/s every 10 seconds until the participant again indicated that they were at a comfortable speed. The mean of the two participant-indicated speeds was used as the test speed for the participant.

We used optical motion capture (Qualysis, Gothenburg, Sweden) and IMUs (iMeasureU, Auckland, New Zealand) to record participant motion at 250Hz and 500Hz, respectively. Each participant was instrumented with four IMUs: one on each foot and one on each shank (Figure 2.1). The foot IMUs were affixed to the superior midfoot using the shoelaces, and the shank IMUs were affixed to the distal medial tibia using Velcro straps provided by the manufacturer. Optical motion capture markers on the foot were placed on the metatarsal heads, medial and lateral malleoli, and posterior of the shoe (Figure 2.1a). Additionally, markers were placed on the femoral epicondyles, anterior and posterior superior iliac spine, and lateral pelvis. Marker clusters on the lateral leg were used for tracking of the shanks (Figure 1b) and thighs, and a cluster on the lower back was used for tracking of the torso.
2.2.2 Detection of Areas of Common Significance:

All data was imported into Matlab (MathWorks, Natick, MA), where we calculated the magnitudes of the resultant angular velocities based on the IMU gyroscopic data for both the feet and shank segments. Using the force platform data with a threshold of 100N to detect foot-strike, the resultant angular velocities from both measurement systems were normalized to a percent gait-cycle time scale.

To determine the areas of common significance in the gait cycle, the resultant angular velocity curves were analyzed in pairs of gaits using an SPM 2-tailed t-test ($\alpha=0.05$) [83]. The gait pairings for analysis consisted of opposing gait features (short vs long strides, rearfoot vs forefoot strike, and narrow vs wide stance), as well as a comparison of the genu-valgus and bouncing gaits to a regular running gait. These pairings allowed for identification of regions of interest for each gait feature.

Figure 2.1: Locations of optical motion capture markers (blue circles) and IMUs (orange rectangles) used to track motion of the a) feet and b) shanks.
We conducted the 2-tailed t-tests for each gait pair on an individual-participant-leg basis (i.e. 10 participants gave 20 angular velocity curves for each segment - feet and shanks), and then identified the areas of common significance across the population. Areas of common significance were defined as the areas in the gait cycle in which at least 80% of the limbs analyzed indicated a significant difference between the gaits based on their SPM analysis. This would be shown as at least 80% of the individual limb lines, shown in dotted blue, being either above or below the SPM threshold lines, shown in dashed red (Figure 2.2b). The level of agreement across the population, expressed as a percent of analyzed limbs which showed significant differences in the SPM analysis, was recorded for the areas of common interest. This analysis was completed for foot-mounted and shank-mounted IMU data separately.

2.2.3 Validation with Optical Motion Capture:

We also used Matlab to calculate resultant angular velocity curves for the feet and shanks based on optical motion capture data. We synchronized the angular velocity curves by minimizing the RMS difference of the curves during a distinct motion (3 hops) at the start and end of each 5-minute trial. We then calculated the difference between the IMU- and optical motion capture-based mean angular velocity curves, as well as the difference between the IMU- and optical motion capture-based standard deviations for each gait feature dataset.

The same methodology for determining areas of common significance from the IMU data was repeated using optical motion capture data. We then calculated the difference between measurement systems in the start and end locations of the areas of common significance with a level of agreement ≥90%. The mean and standard deviation in the difference of start and end locations was calculated across all gait features.

2.2.4 Relation to Joint Angles:

To determine how the gaits biomechanically differ, we calculated joint angles in Visual3D (C-Motion Inc, Germantown, MD) for the ankles, knees, and hips based on optical motion capture
data of 150 strides of each gait style from each participant. Torso angle and position were also calculated.

These biomechanical measures were compared using an SPM 2-tailed t-test. Areas of common significance were found for the joint angles to identify methods of gait style changes that were common across participants.

2.3 Results

We first validated that the participants responded to the instructions as expected by comparing optical motion capture derived data signals from Visual3D determined to be indicative of the instructed change (Table 2.1). After validating that participants’ gait changes were made as expected, we analyzed the resultant angular velocity data. Based on the IMU data, SPM 2-tailed t-tests detected changes in all gait features that we tested. The foot mounted IMUs detected areas of common significance for every gait feature (Figure 2.2b). Notably, the stance-width feature has a much smaller area of common significance than other gait features tested and is only common to 90% of participants. We also found areas of common significance based on the shank mounted IMUs for every gait feature tested except stance width (Figure 2.2c).

<table>
<thead>
<tr>
<th>Gait Pairing</th>
<th>Indicative Signal</th>
<th>Change</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rearfoot vs Forefoot Strike</td>
<td>Foot angle relative to floor at strike</td>
<td>20.9 ± 7.4° more plantarflexed foot in Rearfoot Strike</td>
</tr>
<tr>
<td>Short vs Long Strides</td>
<td>Total stride length</td>
<td>13.7 ± 5.3 cm longer in Long Stride</td>
</tr>
<tr>
<td>Narrow vs Wide Stance</td>
<td>Stance width calculated by consecutive strike positions</td>
<td>8.0 ± 3.7 cm wider in Wide Stance</td>
</tr>
<tr>
<td>‘Bouncing’ vs Regular Gait</td>
<td>Maximum torso centre of mass height</td>
<td>3.7 ± 1.7 cm higher in ‘Bouncing’ Gait</td>
</tr>
<tr>
<td>Genu Valgus vs Regular Gait</td>
<td>Maximum lateral foot position during swing</td>
<td>11.7 ± 7.5 cm wider swing in Genu Valgus Gait</td>
</tr>
</tbody>
</table>

Note: Signals averaged across all 1500 gait cycles used for analysis (150 cycles x 10 participants)
Figure 2.2: Results of the Statistical Parametric Mapping analysis of resultant angular velocities, as measured by IMU. a) Means ± 1 SD of the resultant angular velocity of the foot during regular gait (black) and bouncing gait (red), and b) the SPM result of this comparison. The solid black line indicates the population-wise SPM results, and the dotted blue lines illustrate the SPM results for each participant. Values farther from zero indicate increased significance, and an SPM value equivalent to p=0.05 is indicated with the red dashed line. Areas of the curve in which at least 80% of individual subjects were found to have a significant difference between gait styles are indicated with the greyscale bar at the bottom of the lower plot. This corresponds with the area of common significance of the bouncing gait indicated in c) foot mounted IMU areas of common significance. The same workflow was used to create d) shank mounted IMU areas of common significance.
The mean resultant angular velocities of the both the feet and shanks showed similarity between the IMU system and the optical motion capture system (Figure 2.3). Across all gaits and participants, the mean resultant angular velocities of the foot were 9.8°/s (1.2% of the velocity range) faster as measured by IMU vs optical motion capture. Conversely, the resultant angular velocities of the shank were 16.35°/s (2.8% of the velocity range) slower on average as measured by IMU. Further, the standard deviations of the resultant angular velocities were an average of 3.4°/s (3.8% of the standard deviation) larger as measured by IMU for the foot segment, and 15.3°/s (19.2% of the standard deviation) smaller as measured by IMU for the shank segment.
To further quantify the similarity between the optical motion capture data and the IMU data, we compared the locations of the areas of common significance determined by each system. Using the optical motion capture data of the foot to determine areas of common significance found areas of common significance starting 0.85% ± 0.8% gait cycle later and an ending 0.5% ± 2.8% gait cycle earlier than the same analysis with IMU data, averaged across all gait features. Based on the shank data, the optical motion capture analysis found areas of common significance that start...
1.14% ± 1.9% gait cycle later and end 0.3% ± 1.1% gait cycle earlier than the IMU analysis, averaged across all gait features.

Analysis of the hip, knee, and joint angles, based on optical motion capture, shows that for each gait feature, many participants modified their gait in the same manner (Figures 2.4-2.8). In many cases, the locations where the knee angles differ between gaits is at the edges of where the shank angular velocity differs, and likewise for the relation between ankle angles and feet angular velocities. For example, a comparison of the stride length styles shows increases in the hip, knee, and ankle sagittal plane joint angles across all subjects when taking longer strides (Figure 2.5).

![Figure 2.4: Comparison of hip (a-c), knee (d-f), and ankle (g-i) angles in rearfoot striking (black) and forefoot striking (red) gaits, derived from optical motion capture data. Standard deviation clouds (± 1 SD) are represented by the dashed lines. Areas of the curve in which at least 80% of individual subjects were found to have a significant difference in joint angles between gait styles are indicated with the greyscale bar at the bottom of the respective plots. Note particularly the increase in ankle plantar flexion during late swing and early stance in the forefoot striking pattern (g).]
Figure 2.5: Comparison of hip (a-c), knee (d-f), and ankle (g-i) angles in short strides (black) and long strides (red) gaits, derived from optical motion capture data. Standard deviation clouds (± 1 SD) are represented by the dashed lines. Areas of the curve in which at least 80% of individual subjects were found to have a significant difference in joint angles between gait styles are indicated with the greyscale bar at the bottom of the respective plots. Note the increase in range of motion of all 3 joints in the sagittal plane for the long strides as compared to the shorter strides (a,d,g). Additionally, there is an increase in hip abduction/adduction range in the longer strides (b).
Figure 2.6: Comparison of hip (a-c), knee (d-f), and ankle (g-i) angles in regular (black) and bouncing (red) gaits, derived from optical motion capture data. Standard deviation clouds (±1 SD) are represented by the dashed lines. Areas of the curve in which at least 80% of individual subjects were found to have a significant difference in joint angles between gait styles are indicated with the greyscale bar at the bottom of the respective plots. In the bouncing gait there is an increase in ankle dorsiflexion during early stance, as well as plantarflexion during late stance (g). Additionally, an increase in hip and knee extension during late stance and flexion during swing in the bouncing gait is seen (a,d).
Figure 2.7: Comparison of hip (a-c), knee (d-f), and ankle (g-i) angles in regular (black) and genu valgus (red) gaits, derived from optical motion capture data. Standard deviation clouds (± 1 SD) are represented by the dashed lines. Areas of the curve in which at least 80% of individual subjects were found to have a significant difference in joint angles between gait styles are indicated with the greyscale bar at the bottom of the respective plots. Particularly of note is the decrease in knee adduction during swing (e), increase in ankle eversion during stance (h), and increased range of ankle internal and external rotation in the genu-valgus gait (i).
Figure 2.8: Comparison of hip (a-c), knee (d-f), and ankle (g-i) angles in short strides (black) and long strides (red) gaits, derived from optical motion capture data. Standard deviation clouds (± 1 SD) are represented by the dashed lines. Areas of the curve in which at least 80% of individual subjects were found to have a significant difference in joint angles between gait styles are indicated with the greyscale bar at the bottom of the respective plots. The main differences to note are the increases in hip abduction (e) and ankle eversion (h) throughout the gait cycle in the wide stance gait, as well as the increase in ankle internal rotation during swing (i).

2.4 Discussion

In this study we used SPM to identify areas of common significance in the running gait cycle based on IMU resultant angular velocity curves. The approach in this study is novel in that it is a compromise between a robust method of feature detection and maintaining relatability to lower extremity biomechanics.

Both the foot-mounted and shank-mounted IMUs were able to detect changes in all gait parameters tested, with the exception of stance width which was only detected with the foot-mounted IMUs. Additionally, there was excellent agreement between resultant angular velocities measured from IMUs and computed with optical motion capture. We showed that each gait feature change possessed common changes in segment angular velocities and joint angles across participants. These results explain the specific motion pattern modifications that occur when a runner changes each feature of their gait.
For all gait comparisons except stance width there are sections of the gait cycle where the angular velocity changes were detected in 100% of subjects, surrounded by buffer zones with above 80% of subjects (Figure 2.2). This indicates consistency across subjects and applicability of the results to a greater population of recreational runners without need for subject-specific calibration.

Additionally, if the sensor results from the foot-mounted and shank-mounted IMUs were combined, it will increase the ability to detect a change in gait features. For example, if a person shows a change in angular velocity of the foot during midstance as well as a change in angular velocity of the shank at toe-off, it increases the certainty that they have begun a bouncing gait compared to if only one of these changes was seen. Accordingly, changes in the resultant angular velocity of the feet and shanks in focused areas of the gait cycle can be used to identify changes in strike pattern, stride length, genu valgus, and bouncing gaits.

Similarity was shown between the IMU and optical motion capture systems in the resultant angular velocity means and standard deviations of both the feet and shanks. This indicates that the variation captured is a result of physical variation in running gait, not variability within a measurement system. The only area of the feet data with notable differences between systems is in late stance. This is likely due to the IMU placement on the superior midfoot. While the IMU captures the motion of this exact location on the foot, the optical motion capture model – based on markers on the ankle, heel, and metatarsals – captures the motion of the entire foot as if it were a rigid body. In late stance, when the runner begins to lift their heel, the motion captured by the two systems will differ due to deformation of the foot. The mean angular velocities of IMU-based and optical motion capture-based shank data are also similar; however, on average the IMU data had a smaller standard deviation. This is likely due to the attachment method of the optical motion capture marker cluster leading to soft-tissue and Velcro-movement artifact during running.

Across all gait features tested, including features not detected by the IMUs, changes in joint angles were found using the optical motion capture data. By comparing the changes in joint angles
across participants, common strategies of gait change were identified (Figures 4-8). Specifically, we identified that runners plantarflexed more in late swing during forefoot striking, increased their sagittal plane range of motion in all lower-limb joints during long strides, increased sagittal plane range of motion in all lower limbs as well as plantarflexed the ankle earlier during a bouncing gait, everted the ankle and abducted the knee more during genu-valgus gait, and ab ducted the hip more during a wide stance gait. Given that changes in joint angles were detected in 100% of participants for each gait style change, this could allow researchers to statistically relate a change in a gait feature, as detected by IMU, to changes in the participant’s joint angles without the need for an optical motion capture collection. For example, if a change from a regular gait to a bouncing gait was detected based on the IMU resultant angular velocity data, it is likely that the participant extended all lower limb joints more during late stance, as well as flexed their knees more during the swing phase. With this mapping, researchers may be able to obtain a detailed representation of a participant’s gait or gait changes without the need for in-lab optical motion capture techniques. Additionally, using a bilateral system would allow for detection of asymmetry in the aforementioned gait features.

A comparison to previous work is difficult given the methodologies used in this study. Many studies with IMUs do not focus on angular velocities, and either focus on joint angles and segment positions [71], [72], [81], or focus on detection of gait features, often from a machine learning approach [68], [69], [79], [80].

One noteworthy implication of this approach is the ability of a simple system of IMUs to detect changes in several different gait features simultaneously. For example, due to the differences in areas of common significance in the shank IMU data for detecting bouncing gait and detecting stride length, if a runner changed both of these parameters simultaneously, the changes could both be detected by their respective areas of common significance. This ability could be further
improved by combining the detection areas of both the shank-mounted and feet-mounted IMUs; however further statistical analysis of the data would be needed.

Although the IMU setup used in this study was only able to detect a change in stance width using the foot IMU in a very limited area of the gait cycle, an otherwise placed IMU may have been able to do so using the same methodology. For example, the thigh is more adducted throughout gait when running with a narrower stance (Figure 8). Additionally, angular position data of the torso indicates a greater range of torso twisting and leaning in the wide stance running style as compared to the narrow stance. To go through a greater range of motion over the gait cycle would require an increased angular velocity, which may be detected with SPM. Both these results indicate the potential for detection of stance width if the placement of IMUs were expanded to include thigh-mounted or torso-mounted IMUs.

While this approach is promising, our experimental paradigm prevents extrapolation of these results to in the wild studies. To capture optical motion data, collection was conducted on a treadmill in a laboratory environment. This is paradoxical given the motivation of the study; however, it was necessary for validation of the measurement and gait change detection techniques with optical motion capture. Additionally, due to the laboratory-based protocol, all testing was conducted on level ground at a constant, self-selected pace for each individual. Outside of a laboratory environment, these variables cannot be controlled, and their variation will affect the results of the gait change detection techniques used. This could be addressed in future studies by further validation outside of the laboratory environment using prescribed running routes with known slopes, as well as tracking participant speeds via GPS. Finally, while all changes except stance width were detected in 100% of participants, the changes were intentionally contrived, and therefore may be exaggerated. During natural gait changes, the differences in gait styles may not be as pronounced and may be more difficult to detect. In summary, this proof of concept study has validated the feasibility of the system to detect the desired changes; however, further validation in
an outdoor environment is required before the approach can be used for a large-scale prospective study.

In addition to the limitations of the protocol, it is important to note that the gait styles examined were not mutually exclusive. For example, when comparing the ankle dorsiflexion angles of the genu-valgus gait or bouncing gait to the regular gait, the first 5% and final 20% of the gait cycle indicate similar characteristics to that of the comparison between rearfoot and forefoot strike. This may indicate that participants were more likely to forefoot strike when running in the genu-valgus or bouncing gait styles.

In conclusion, the results of this study have shown that IMU-measured resultant angular velocities of the shanks and feet are a valid modality to monitor and detect intentional changes in several running gait features. These angular velocities have been validated by optical motion capture data, and the changes detected by the IMU data can be mapped to changes in various joint angles across the population. With further validation of this approach, which allows for feature detection while retaining biomechanical interpretability, researchers may be able to monitor the gait patterns of participants in the wild and gain a greater understanding of how changes in gait lead to overuse injury.
Chapter 3

Relative Movement of the Tibia, Talus, and Calcaneus During a Forefoot Landing Motion

3.1 Introduction

Running is a common form of physical activity throughout the world and is often not thought of as a dangerous activity. Despite this, as high as 70% of runners have been reported to sustain an overuse injury from running in a one-year period [1]. While there is a wide array of overuse injuries, many with etiologies not yet fully understood, a link has been drawn between patellofemoral pain and excessive pronation of the foot during early stance [41], [42]. Previous research has used skin-based landmarks to measure calcaneal eversion and tibial internal rotation [36], [43]; however, the contributions of the talocrural and subtalar joints to the coupling of calcaneal eversion and tibial internal rotation are not known. Similarly, it is unknown how athlete-specific factors, such as width variation of the talar dome, may contribute to an athlete’s susceptibility to patellofemoral pain. The talar dome is the superior surface of the talus, which the tibia articulates on. This surface is arch shaped when viewed from the medial or lateral directions, to allow the talus to articulate in plantar and dorsiflexion relative to the bottom of the tibia. Additionally, the talar dome tends to be narrower toward the anterior end of the dome than the posterior end. Statistical shape modelling has shown that one of the principal components of variation in talus shape is the amount of narrowing of the talar dome toward the anterior end of the articular surface [87], [88].

The most common site of running related injuries is the knee, predominantly due to diagnoses of patellofemoral pain [63], [89]. Additionally, female runners and older runners are at a greater risk of this injury [63], [89], [90]. As with most overuse injuries, the causation of
patellofemoral pain is multifactorial – involving training regime, bone structure, and specific movement patterns [3]–[5]. Previous research has found various correlations between the development of patellofemoral pain and pronation of the foot. Although findings are inconsistent, it has been stated that range, speed, and duration of pronation may all contribute toward patellofemoral pain [36], [41], [42].

Pronation of the foot is a multi-axial motion, consisting of calcaneal eversion and talar internal rotation. Current understanding of the ankle joint complex is that during landing, the calcaneus everts, and the talus internally rotates which drives internal rotation of the tibia [44], [45]. A commonly accepted mechanism leading to patellofemoral pain is that excessive pronation causes supination in mid-stance to occur later, and therefore the tibia to be internally rotated later in stance [91]. During knee extension, the prolonged internal rotation of the tibia is compensated further up the kinetic chain via internal rotation of the femur. Although this compensation mechanism maintains consistent interactions of the tibiofemoral joint, it may create abnormal stresses on the surrounding soft tissues due to its effect on patellar kinematics [92].

Despite evidence for this theory [42], many studies have been conducted that have found no differences in pronation related variables between populations with and without patellofemoral pain. Epidemiological studies have compared pre-injury static lower limb alignment in recreational runners, and found foot pronation angle to be irrelevant to the development of patellofemoral pain over the following six months [11]. The same conclusions were found using both static and dynamic measures of pronation during walking in army recruits [82], [93]. Additionally, no differences were found in kinematic variables such as maximum pronation angle, maximum pronation velocity, and time to maximum pronation between injured and uninjured groups of recreational runners [6], [94]. Conversely, some studies have shown that less pronation during the first 10% of stance may increase this risk [41], as might a delay in peak rearfoot eversion [42].
While a link has been drawn between pronation variables and patellofemoral pain, evidence of this relationship is inconsistent. This may indicate that movement patterns alone cannot account for the development of patellofemoral pain, and that its cause is multifactorial. Additionally, while foot pronation has been linked to tibial internal rotation with some consistency \cite{36, 43, 44}, the underlying mechanics and bone interactions resulting in this relation have not been researched in depth.

This pilot study seeks to further our understanding of the ankle joint complex, and specifically to evaluate coupling of the internal rotation of the talus and tibia throughout stance. This will be accomplished using biplanar videoradiography (BVR), as it is the only measurement modality capable of capturing talus motion \textit{in vivo}. A BVR system consists of two x-ray generators and their accompanying tubes. Each x-ray system is connected to a highspeed digital camera, allowing the x-ray images to be recorded at 125Hz when converted to visible light. Using known positions of the x-ray generators as well as established three-dimensional volumes of the participant’s bones, the motion of each bone can be tracked in a non-invasive manner. A hopping motion will be used in place of a running motion, due to availability of data. Additionally, use of the hopping data set will allow for intra-participant analysis, as there are data for several hops from each participant. To better inform future research, we evaluated timing, magnitude, and excursion of tibial rotation in global space, as well as talus and calcaneus rotation about the long axis of the tibia. These variables were evaluated as a population, as well as compared on an inter- and intra-participant basis to evaluate the generalizability of any conclusions to a population.

3.2 Methods

3.2.1 Data Collection

In this pilot study, a convenience sample of three participants (2F/1M) between the ages of 18 and 50 were selected from a larger cohort of 9 subjects. All participants had no known neurological or cardiovascular diseases and had no history of lower-limb injury in the previous 12
months. Participants were recruited from Queen’s staff and students by word of mouth or response to a poster. The data collection was conducted with Queen’s HSREB approval. All participants gave informed consent and were aware that they were able to withdraw from the collection at any time.

Participants attended two separate data collections. The first collection session was to confirm that participants met the inclusion criteria and to collect a CT scan of the foot and ankle (Revolution HD, General Electric Medical Systems, Chicago, IL). During the second visit, the participants were asked to complete a hopping task. In the laboratory, participants were introduced to the task, and given time to practice and become comfortable with it, to reduce the risk of losing balance during data collection. Participants were asked to hop on their right foot at a frequency of 2Hz. To maintain the hopping cadence, participants were asked to match the speed of a digital metronome. Once it was visually determined that participants were keeping cadence with the metronome, data was collected for a total of five hops (2.5 seconds). X-ray data was collected using a BVR system at 125 Hz. This system consists of two x-ray generators, and two x-ray tubes. Each x-ray system is connected to a high-speed digital camera, allowing for x-ray images to be recorded on video, with a frame rate of 125 Hz. Additionally, ground reaction force data was captured using a force plate (AMTI, Watertown, MA) at 1125 Hz. All measurement systems were spatially and temporally synchronized. The BVR system was calibrated with a custom object and images were undistorted using a process described in the literature [95].

3.2.2 Data Processing

The CT scans were imported into Mimics Research 19.0 (Materialise NV, Leuven, Belgium) for processing. In Mimics, we created surface models and partial volumes of each bone of interest by segmenting its outline in each slice of the CT scan. We then smoothed the surface files using Geomagic Wrap (3D Systems Inc. Rock Hill, South Carolina). Finally, based on the smoothed surface files, the inertial axes of each bone were calculated in Matlab (MathWorks,
Natick, Massachusetts) using custom software [96]. The tibial coordinate system is based on the inertial axes of the bone, with the x-axis pointed in the lateral direction, y-axis pointed in the anterior direction, and z-axis pointed in the superior direction.

The partial volumes were imported into Autoscorer [56] (Brown University, Providence, RI). Digitally reconstructed radiographs based on the partial volumes of the tibia, talus, and calcaneus were aligned with the x-ray images across the trial using a semi-automated approach. Using the calibrated locations of the x-ray sources during collection, this alignment in two views results in a known location and orientation of each bone. Rotation matrices were generated based on the known locations and orientations and were exported for further analysis.

3.2.3 Data Analysis

The rotation matrices for the bones throughout the trial, as well as the inertial axes were imported into Matlab. We used this information to calculate the orientation of the calcaneus and the talus with respect to the tibial coordinate system (Figure 1.1). The talus and calcaneus orientations were calculated using a ZYX Euler sequence, that is, with the final rotation being around the z-axis when rotating from the calcaneus or talus orientation to the tibial coordinate system (Figure 3.2). This Euler sequence was chosen for its ability to convey the rotations about the long axis of the tibia more distinctly. Additionally, the axial rotation of the tibia in the global reference frame was calculated using an XYZ Euler sequence, so that in any orientation of the tibia, the first rotation from the tibial frame toward the global reference frame is about the long axis of the tibia.
Figure 3.1: Inertial axes of the right tibia from the anterior (left) and lateral (right) views. The X, Y, and Z axes are represented by the red, green, and blue axes, respectively. The Euler sequence used to determine the rotations from the talus and calcaneus axes (not shown) to the tibial axes first rotated the X and Y axes into the correct plane, then rotated around the Z-axis last.
Figure 3.2: Inertial axes of the tibia, talus, and calcaneus of the right foot from the anterior (left) and lateral (right) views. The X, Y, and Z axes are represented by the red, green, and blue axes, respectively.

Synchronized ground reaction force data was also imported into Matlab. Using the force data, all angle vs time curves were temporally cropped and normalized to 100% of ground contact time for each hop, as determined with a threshold of 25N of vertical ground reaction force. All angles were represented as a change in their respective angle from the moment of contact.

To evaluate the relationships between the angles of interest on a sample-population level, the means and standard deviations of tibial rotation in the global reference frame, as well as talus and calcaneus rotation in the tibial reference frame were calculated across the sample. Additionally, to identify inter-participant variations, these same measures were calculated for each participant independently. Finally, to evaluate intra-participant consistency from one hop to the next, the angles during each jump were evaluated on a trial by trial basis. The quantitative values compared
between these curves were the timing of maximum internal rotation of the tibia in the first half of
stance in the global reference frame, timing of maximum internal and external rotation of the talus
in the tibia reference frame, and talus rotation excursion in the tibia reference frame. Timing of all
events was expressed as a percent of stance, as determined with synchronized force plate data. Due
to the small sample size, t-tests were not conducted.

3.3 Results

Of the data collected from three participants, there were two hops from subject 1, five hops
from subject 2, and three hops from subject 3 that could be analyzed. This inconsistency is due in
part to errant timing of imaging, and in part due to the participants moving partially outside of the
capture volume during collection. It should be noted that to keep the population measures
representative of all participants, means and standard deviations were calculated using only the first
2 hops from each participant.

We first visualized the data on a population-level (Figure 3.3). Averaged across all
participants, the maximum internal rotation of the tibia in the global frame occurred at 27% of
contact time, at 9.5° more internally rotated than at initial contact. Additionally, maximum talus
internal and external rotation with respect to the tibia occurs at 18% and 60% of contact time,
respectively, with values of 2.9° of internal rotation, and 6.0° of external rotation compared to the
position at contact time. This gives an internal/external rotation excursion of 8.9° over 42% of
contact time.
Figure 3.3: Average internal rotation of the calcaneus (blue) and talus (red) with respect to the tibia, and internal rotation of the tibia in the global reference frame (green) across all participants. Data points of interest highlighted with dotted lines are: maximum internal and external rotation of the talus with respect to the tibia, and maximum internal rotation of the tibia in the global frame.

Similarly, we visualized the global rotation of the tibia, as well as the relative rotation of the talus and calcaneus averaged for each participant (Figure 3.4). All hops for each participant were used, to represent participant motion most accurately. Additionally, each individual hop was evaluated, and the minimum and maximum of each variable of interest was recorded for each participant (Table 3.1). This allows for comparisons of these variables both within and between subjects. When comparing data between subjects, subject 2 had noticeably less internal rotation of the tibia in the global reference frame, as well as less internal rotation of the talus with respect to the tibia. In addition, peak internal rotation of both of these signals occurs earliest in subject 2. Despite this, the rotation of the calcaneus with respect to the tibia is similar across all subjects with respect to maximum external rotation, internal/external rotation excursion, and maximum external rotation timing (Figure 3.4). Other differences of note are the comparatively large maximum external rotation of the talus for subject 2 and the comparatively late maximum external rotation of
the talus for subject 3. Although statistical significance was not proven due to the small sample size, the differences noted were outside of the variability of the measurement system. As such, the differences cannot be due to measurement error, and are taken to be representative of the true motions of the participant.

In comparing intra-participant consistency, subject 1 had the most consistent data across all measures; however, only 2 hops were analyzed. The magnitudes of the ranges are very similar for participants 2 and 3; however, participant 3 has a comparatively broad ranges of timing of maximum external rotation of the talus, and of total excursion time.
Figure 3.4: Average internal rotation of the calcaneus (blue) and talus (red) with respect to the tibia, and internal rotation of the tibia in the global reference frame (green) for participants 1-3 (a-c, respectively). Data points of interest highlighted with dotted lines are: maximum internal and external rotation of the talus with respect to the tibia, and maximum internal rotation of the tibia in the global frame.
Table 3.1: Summary of angles of interest across subjects.

<table>
<thead>
<tr>
<th></th>
<th>Subject 1</th>
<th></th>
<th>Subject 2</th>
<th></th>
<th>Subject 3</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Average</td>
<td>Range</td>
<td>Average</td>
<td>Range</td>
<td>Average</td>
<td>Range</td>
</tr>
<tr>
<td>Maximum Tibia Internal Rotation Angle</td>
<td>13.0°</td>
<td>11.4° - 14.8°</td>
<td>0.9°</td>
<td>1.0° - 2.6°</td>
<td>14.3°</td>
<td>14.0° - 15.9°</td>
</tr>
<tr>
<td>Maximum Tibia Internal Rotation Timing</td>
<td>26%</td>
<td>25% - 27%</td>
<td>13%</td>
<td>12% - 41%</td>
<td>27%</td>
<td>26% - 28%</td>
</tr>
<tr>
<td>Maximum Talus Internal Rotation Angle</td>
<td>5.5°</td>
<td>5.2° - 5.9°</td>
<td>2.3°</td>
<td>0.3° - 7.0°</td>
<td>4.6°</td>
<td>2.7° - 8.8°</td>
</tr>
<tr>
<td>Maximum Talus Internal Rotation Timing</td>
<td>19%</td>
<td>18% - 19%</td>
<td>12%</td>
<td>4% - 14%</td>
<td>21%</td>
<td>20% - 26%</td>
</tr>
<tr>
<td>Maximum Talus External Rotation Angle</td>
<td>1.9°</td>
<td>1.6° - 2.8°</td>
<td>11.6°</td>
<td>7.5° - 14.3°</td>
<td>4.8°</td>
<td>-1.4° - 11.6°</td>
</tr>
<tr>
<td>Maximum Talus External Rotation Timing</td>
<td>55%</td>
<td>51% - 56%</td>
<td>49%</td>
<td>45% - 56%</td>
<td>63%</td>
<td>57% - 69%</td>
</tr>
<tr>
<td>Talus Internal / External Excursion Range</td>
<td>7.4°</td>
<td>6.8° - 8.7°</td>
<td>13.9°</td>
<td>12.9° - 15.1°</td>
<td>9.4°</td>
<td>7.3° - 14.6°</td>
</tr>
<tr>
<td>Talus Internal / External Excursion Time</td>
<td>36%</td>
<td>33% - 37%</td>
<td>37%</td>
<td>32% - 49%</td>
<td>41%</td>
<td>31% - 49%</td>
</tr>
</tbody>
</table>

3.4 Discussion

In this pilot study we have analyzed the motion of the tibia, and relative motion of the talus during a hopping activity and compared results at both an inter- and intra- participant level. The patterns and findings from this pilot may serve to direct future studies toward the level generalisation at which data should be analyzed and reported. Kinetic data is often represented as means and standard deviations across a population [8], [36]; however, in the study of overuse injuries the participants that are farther from the mean curve, constantly to one side of the curve, or
otherwise deviating from the average motion of the population may contribute the most meaningful
data.

Coupling of the internal rotation of the tibia and talus bones is not seen in our data; however, there are similarities in their motion. Coupling would be represented by a prolonged constant angle of the talus with respect to the tibia. Averaged across all participants, a pattern emerges that during the first 20% of contact, the tibia is internally rotating in space, while the talus is also internally rotating with respect to the tibia. This is in agreement with previous research, which used eversion of the foot as a proxy-measure for internal rotation of the talus [43]. Based on our data, the internal rotation of the talus is leading the tibial rotation during early stance. Further, global internal rotation of the tibia peaks shortly after peak internal rotation of the talus, which is again in line with available literature [43]. We do not have the data here to suggest which bone ‘drives’ the motion of the other; however, literature states it is the internal rotation of the talus which drives that of the tibia [44], [45]. Our data suggests that this could be a plausible mechanism for the transfer of internal rotation up the kinetic chain.

Additionally, after peak internal rotation of both the tibia and talus, rotation of the tibia is nearly constant. During this period, the calcaneus rotates externally with respect to the tibia to a range of approximately 20-25° of external rotation. Throughout this time, the talus also switches to external rotation about the tibia, although to a lesser extent than the calcaneus. We speculate that the talus may act as a buffer, allowing independence of the rearfoot and shank with respect to internal rotation. This is not in line with literature, which has previously found the internal rotation of the subtalar joint to be an order of magnitude higher than that of the talocrural joint [97]. The deviation from previous knowledge may be due to a difference of motions (hopping vs. walking) or collection methods (continuous vs. discrete points). During this time, coupling between the tibia and talus is also not observed. If coupling between the talus and tibia were to occur this late in stance, the talus would need to articulate significantly more with respect to the calcaneus. For
example, if the talus is rotated 5° external to the tibia, that 5° difference would have to be accounted for at the subtalar joint, resulting in a subtalar joint angle of approximately 20° of internal rotation. If the morphology of the subtalar joint prohibits such a large joint angle, the rearfoot and tibia would be unable to rotate independently of each other at this point in stance.

As shown in this study, variations in bone kinematics are likely to exist in a participant pool. In studies of overuse injury, these variations must be captured to accurately shed light on which factors contribute to injury. While the above observations can be drawn from data averaged across the entire population, they do not appear to be consistent across all participants. For example, very little internal rotation of the tibia was observed from the motion of subject 2. Additionally, in subject 2 the talus and calcaneus follow an almost identical rotation pattern after approximately 20% of stance. Based on these observations, the above hypotheses of tibial internal rotation being driven by talus rotation and of the talus acting as a buffer during mid-stance cannot hold true. While we have not done adequate analysis to determine the cause of the difference of subject 2 from the other subjects, it is possible that this is due to a different morphological structure in any of the ankle bones. This inter-participant variation also serves to prove that analysis of population-averaged data may be problematic in injury studies, as it may not adequately represent the motions of each participant, but a middle-ground between several different motions.

At an intra-participant level, different amounts of variation were observed between the participants. For example, looking at the talus internal / external excursion time, while all three subjects have a similar fastest excursion time, the most variation is seen from subject 3 (excursion times ranging from 33%-74% of total stance time), with subject 1 showing the least variation (34%-39% of stance time). Step-to-step variations or consistencies within a subject is often lost during motion capture based biomechanical injury studies due to averaging over an entire population; however, studies have shown that coordinative variability may be an important factor in injury development [98].
The main limitation of this pilot study is the quantity of data. While the data presented has highlighted the importance of a participant-specific approach to gait analysis, the conclusions would be more certain with additional trials from each participant, as well as more participants, and repetitive visits to assess consistency of participant motion over time. Additionally, the variation between the number of valid hops from each participant presents a limitation to the comparison of both intra- and inter-participant variability. Statistically, more variation will be found in the participant who has conducted the most trials, and from our data, this is the case for many of the variables analyzed. Despite these limitations, this pilot study was able to illustrate the importance of a participant-specific approach to analysis of bone kinematics through investigation of the talocrural joint.

The use of a hopping data set for a proof of concept study into running related injuries is also a limitation. Although the overall motion was different than would be found in running, there are likely similarities between the bone motions within the ankle joint complex between hopping and a forefoot strike run, as the arch is loaded and force transferred to the leg in a similar manner. In the future, a running motion should be used to confirm this hypothesis. Additionally, both forefoot and rearfoot strike patterns should be evaluated for tibiotalar coupling.

Based on the information presented in this pilot study, researchers must take care to accurately represent data of bone kinematics and interactions. As shown in comparison of population-averaged data to participant-averaged data, representation of an entire population may be misleading due to variations within a seemingly homogenous group of healthy, active, adult participants. This is particularly of importance in the study of overuse injuries. While differences may not be evident when averaged across entire populations of injured and uninjured runners, there may be variations between participants that have been washed out by the level at which the data was averaged. As such, a participant-specific approach should be used when possible for evaluation of ankle bone kinematics. Additionally, moving forward, future studies should work toward
determining factors, such as specific morphology, which can affect ankle joint kinematics. If ankle joint kinematics were to be explained by specific, measurable morphological factors, it would assist in the generalisation of findings across a population. For example, if the amount of narrowing in the talar dome is found to correlate with internal/external rotational excursion during gait, the relation of these two factors may assist researchers to identify the root cause of injuries related to tibial internal rotation. Additionally, a link between morphology and specific movement patterns could help to link two of the three pillars which can lead to overuse injury – dosage, specific motion, and morphological structure. Furthering the breadth of this knowledge, especially as it relates to motions implicated in overuse injury will aid in better understanding the kinematics of the lower limb, as well as preventing overuse injuries into the future.
Chapter 4

Discussion and Conclusions

Overuse injuries are diverse, and can affect several joints, bones, and soft tissues. As such, they are also varied and multifactorial in nature. Common gait measurement techniques, including optical motion capture, inertial measurement units, and biplanar videoradiography can contribute to our collective knowledge of overuse injury causation in an integrated manner. A study design that combines measurement techniques to include analysis of bone structure, training variables, and specific movement patterns will grant greater insight to the multi-factorial nature of such injuries. We conducted two studies using all three measurement systems with the objectives to lay the groundwork for future studies in the wild, and to couple these measurements with detailed biomechanical analyses.

First, we used IMUs to monitor gait and to detect several common gait features: strike pattern, stride length, stance width, a bouncing gait, and a genu-valgus gait. Using the robust resultant angular velocity signals from the shanks and feet, we found that statistical parametric mapping was an effective way to detect gait features. All features tested were able to be detected using this method except for stance width, and areas of common significance were determined by either the foot-mounted or shank-mounted IMUs. Using these areas of common significance, IMUs have the potential to be used in the wild to gain a greater understanding of a person’s running gait style. Additionally, by combining the results from both the shank-mounted and foot-mounted IMUs, the statistical certainty that a change in gait style has occurred could be increased. In this study we also verified the use of IMUs in measuring resultant angular velocity of the lower limbs by comparing IMU-based measurements to optical motion capture based measurements. The level of agreement between the systems showed that IMUs were an accurate modality for measuring resultant angular velocities during running. Finally, we related the gait changes detected by IMUs
to changes in joint angles based on optical motion capture data. This relationship between IMU data and more traditional biomechanics measures will give researchers a more detailed representation of a person’s gait than gait features alone when measured outside of the lab environment, as their joint angles can be statistically inferred. The outcomes of this study have shown us that not only are IMU-based gyroscopic readings a plausible modality for monitoring gait in the wild, but that changes in joint angles, traditionally measured using laboratory-based systems, can be inferred without the use of traditional lab-based measurement systems. With further testing and validation, this ability could be extended to data collected outside of the lab, in a more ecologically valid setting. This could serve as a plausible method for collecting high-level data in future studies of overuse injuries in the wild.

In the second study, we used biplanar videoradiography to analyze talus motion in vivo. In overuse injury literature, most in vivo biomechanics studies are conducted using optical motion capture, which cannot capture talus motion. Using optical motion capture, researchers can measure shank and calcaneus motions using skin-based landmarks; however, no such landmarks exist for talus motion due to its location in the ankle. Due to this gap in the literature, the mechanisms linking internal and external rotation of the rearfoot to internal and external rotation of the tibia are not yet fully understood. The primary objective of this pilot study was to gain further understanding of the link between talus motion and tibial rotation. We found that internal rotation of the talus during early stance leads the internal rotation of the tibia. While our pilot did not include the necessary information to determine which bone ‘drives’ the motion of the other, the theory that tibial internal rotation is driven by rotation of the talus beneath it [42], [44], [45] may be plausible based on our data. Further, our data shows that the tibia and talus are not coupled after approximately 25% of stance. At this point in stance, the talus externally rotates with respect to the tibia. Due to the angles observed of external rotation of the talus and calcaneus both with respect to the tibia, it is
hypothesized that the talus acts as a buffer between these two bones during mid- and late-stance to allow for independence of the rearfoot and shank with respect to their internal rotation.

Additionally, this study served to investigate the necessity of subject-specific data analysis vs population averaged data. While observations regarding the motion of the talus were drawn from analysis of data at the population level, inconsistencies were found at both the intra-and inter-subject level. Between subjects, it was noted that one of the 3 subjects presented notably different motion of both the tibia and the talus, as compared with the sample mean. While at the surface-level, this pilot used a homogenous participant pool of healthy, active, young adults, this result highlights the need for subject-specific analysis and modelling of functional biomechanics. Additionally, this may serve as motivation toward future studies to further develop an understanding of the cause of such variations in bone kinematics between participants.

While interpreting our results and their implications, we must acknowledge the limitations of the studies themselves. In the first study, an experimental paradigm was that to validate the IMU modality, data had to also be captured using another method, namely optical motion capture. Due to the necessity of optical motion capture data, the study was conducted in a laboratory using an instrumented treadmill. This is paradoxical given the motivation of the study to validate the use of IMUs for studies outside of the laboratory environment. Each participant’s data was collected at a constant speed, and on level ground. Outside of the laboratory environment, these variables would not be able to be controlled, and this would affect the results of the gait feature detection technique presented. Additionally, in this study, the gait changes were intentionally contrived by the runners, and as such are likely exaggerated compared to the natural changes in any of these features that may be seen in recreational runners. As a result of these limitations, further validation of this method of gait feature detection would be required prior to its adoption as a measurement modality in a large-scale study.
In the second study, limitations were predominantly due to the scale of the study. With more subjects, the results could be further refined and confirmed with a larger data collection, both in terms of the number of participants, and the number of trials from each participant. Statistically, this would allow researchers to draw more meaningful conclusions, especially with regards to inter- and intra-participant variations. Additionally, due to the nature of the available data, although the motivation of this study was based on running-related injuries, the action of participants during the study was a landing and hopping motion. This motion is likely similar to a forefoot strike motion in running, but the bone kinematics may vary from those of a rearfoot strike action.

The variation shown between participants in both studies is of importance to inform future studies of overuse injury development. In the IMU study, the decision was made to determine areas of the gait cycle which could be used to detect gait features on an individual basis, and then to compile the locations of common interest across the population. Prior to this approach, the same method was attempted using population-averaged data. With this method, not all areas identified as significant to the detection of gait features were applicable across the population. In sections of the gait curve where the population-averaged SPM result showed a significant difference between two opposing gait features, this was often inconsistent when observed on a participant-specific basis. For example, in comparison of the bouncing gait feature to the regular gait, the population averaged SPM detected significant differences from 30%-40% of the gait cycle, as well as 62%-72%, and 78%-92%. When analyzed on a participant-specific basis, less than 80% of participants showed a detectable difference in these areas, except for 34%-36% of gait stance (Table 3.1). This illustrates that at a body-segment level, there are important variations between participants. As such, care must be taken to analyze data in a way which does not dilute or play down the differences between individuals. Especially in research of overuse injuries, these seemingly small variations among participants may hold valuable information into the development of injury.
Similarly, in the second study, the focus was on inter-participant variability. This study proved that the same principal of subject-specific analysis hold true when transitioning from analysis at the segmental level to analysis at the bone level. Again, using population-averaged data, conclusions were drawn regarding motion of the talus with respect to the tibia, and motion of the tibia in global space. While these observations remained consistent for two of the participants when looked at individually, the third participant showed drastically different motions of both the talus and the tibia. Given that the differences between subjects were larger than the error of the measurement modality, as well as larger than the standard deviation of the participants’ motion, the inter-subject variations cannot be ignored. As such, we propose that in analysis of the ankle joint complex, data should be analyzed on a subject-specific basis prior to drawing any overarching conclusions about the population, as means and standard deviations have proven to not adequately present the motion across the population.

Through these studies, we have laid the groundwork for future research into overuse injury etiology using a variety of different measurement techniques. We have shown that there is promise for the future use of IMUs for studies in the wild in a way which they have not been used previously. Many previous studies using IMUs have focused on integration of the measured signals back to joint angles [71]–[73], but we have presented a methodology of using directly measured IMU signals to monitor several features of running gait which may be important to the development of overuse injuries. Additionally, we have provided initial insight into the motion of the talus with respect to the tibia, and the interactions of these bones. The interactions of these bones within the ankle joint complex have previously been shown to be of importance to the development of patellofemoral pain [41], [42], although a detailed explanation of the mechanism at play has not been presented.

Combining the methods used in both studies could provide the framework for a study into the development of overuse injuries which is unparalleled in its detail. Many studies thus far focus
on only one of the three factors contributing to injury – dosage, structure, and motions. Using IMUs for consistent gait monitoring in the natural environment can serve to monitor any measure of dosage (distance, time, number of strides, pace, etc.), as well as monitor motions and gait features in an ecologically valid way. Further, the use of biplanar videoradiography can grant an understanding of bone structure and bone interactions that cannot be measured with any other modality. This can also serve to bridge the gap between segmental-level motions, and subject-specific bone structure in an attempt to explain variations in running styles across a population. Finally, we have shown that analysis on a subject-specific basis, followed by the identification of similarities and differences across a population can present results not seen using population-averaged techniques. As such, this method of analysis should be considered at every step.

One such approach to the integration of these measurement modalities is to use IMUs to monitor a large number of runners in the wild over a period of months. Using this data, different running styles could be determined with the gait feature detection system. Following this period of the study, several representatives of each running style could be brought into the lab to participate in a study using the biplanar videoradiography system. Use of this system will allow for analysis of not only the underlying bone interactions across a variety of running styles, but also analysis of the morphological factors across a range of participants.

Some aspects of this research have led to the development of new research questions. From the first study, there are several research questions that must be addressed prior to adoption of this gait monitoring technique in a large-scale study. The most pressing question presented is if the method of gait feature detection will be robust outside of the laboratory environment. Since the motivation of the study was to develop a system for monitoring gait in an ecologically valid environment, the system must be robust to ecological variability. This is a multifactorial question, that will require validation of the measurement system during changes in running speed, slope, and terrain, among other variables. Further, while the system in its current configuration was designed
to be robust to errors and variation in sensor placement, these claims have not been tested. As such, prior to allowing participants to affix the sensors to themselves for use, expected errors in placement and orientation of the sensor should be intentionally made in a controlled environment, and the system validated under such conditions. These errors may include orientation of the shank sensor, location of the sensor around the shank (i.e. on the lateral side instead of medial), and placement within the shoe (i.e. at the top, middle, and bottom of the laces).

The second study presented also gave rise to several new research questions. First and foremost, this brief study into talus motion, and coupling between the talus and tibia has given new insight into the interactions of these two bones during a hopping motion. These observations need to be confirmed by a larger scale study. Additionally, due to the potential link to overuse injuries in running, this experiment should be repeated with a running motion with both a rearfoot striking population and a forefoot striking population. This would present a greater understanding regarding the interactions of the talocrural joint in running and give insight into the cause of tibial internal rotation during running. Additionally, a study should be designed with the objective of determining if the internal rotation of the tibia is driven by that of the talus, or vice-versa. Perhaps most importantly, an attempt should be made to explain the inter-participant variation. As this study showed, there were noticeably different movement patterns of the bones seen across only three subjects. With a larger cohort, researchers may be able to draw connections between the different motions observed and morphological details of the runners. Moving forward with this work, confirmation of the findings regarding the tibiotalar interactions observed will be addressed first. One approach may be to analyze participants on a subject-specific basis, as previously discussed, and determine if there are several common patterns of motion across the population. With this knowledge, morphological factors, such as the wedge angle of the talar dome will be analyzed across the population in an attempt to correlate form to function.
In conclusion, we have used three measurement modalities to develop an understanding of what is possible in future overuse injury studies. Our first study determined a method for using IMUs independently of any other measurement modality to statistically detect changes and variations in eight running gait features. The second study gave insight into the interactions of the tibia and talus during stance and allows for the plausibility of the theory that tibial internal rotation is driven by internal rotation of the talus. Both of these studies have highlighted the necessity of subject-specific analysis and modelling in running due to inter-athlete variations. Together, these studies have formed the basis for future large-scale studies into the development of overuse injury etiology. The combination of inertial measurement units and biplanar videoradiography will allow for future research to look at overuse injury using a comprehensive approach, encompassing all three major factors in injury development – dosage, structure, and movement patterns.
References


Appendix A

IRB Approval for Data Collection in Chapter 2

QUEEN'S UNIVERSITY HEALTH SCIENCES & AFFILIATED TEACHING HOSPITALS RESEARCH ETHICS BOARD (HSREB)

HSREB Initial Ethics Clearance

November 26, 2018

Dr. Rainbow
Department of Mechanical and Materials Engineering
Queen's University

ROMEOT/TRAQ #: 6025306
Department Code: MECH-066-18
Study Title: "An Affordable and Intelligent Wearable System for Overuse Injury Prediction and Prevention."
Co-Investigators: Dr. Andrew Dickinson, Miss Laura Hutchinson
Review Type: Delegated
Date Ethics Clearance Issued: November 26, 2018
Ethics Clearance Expiry Date: November 26, 2019

Dear Dr. Rainbow,

The Queen's University Health Sciences & Affiliated Teaching Hospitals Research Ethics Board (HSREB) has reviewed the application and granted ethics clearance for the documents listed below. Ethics clearance is granted until the expiration date noted above.

- Protocol v.2018OCT31
- Consent Form (version 3) v.2018NOV20
- Data Collection Form v.2018NOV09
- Poster v.2018NOV09

Documents Acknowledged:

- CORE training certificate L. Rogers 2017OCT23
- CORE training certificate A. Etemad 2018AUG15
- CORE training certificate J. Selinger 2018NOV08
- CV- Dr. M. Rainbow 2016OCT

Amendments: No deviations from, or changes to the protocol should be initiated without prior written clearance of an appropriate amendment from the HSREB, except when necessary to eliminate immediate hazard(s) to study participants or when the change(s) involve(s) only administrative or logistical aspects of the trial.

Renewals: Prior to the expiration of your ethics clearance you will be reminded to submit your renewal report through ROMEO. Any lapses in ethical clearance will be documented on the renewal form.

Completion/Termination: The HSREB must be notified of the completion or termination of this study through the completion of a renewal report in ROMEO.
**Reporting of Serious Adverse Events:** Any unexpected serious adverse event occurring locally must be reported within 2 working days or earlier if required by the study sponsor. All other serious adverse events must be reported within 15 days after study team members have become aware of the information.

**Reporting of Complaints:** Any complaints made by participants or persons acting on behalf of participants must be reported to the Research Ethics Board within 7 days of study team members becoming aware of the complaint.

**Note:** All documents supplied to participants must include the contact information for the Research Ethics Board. Investigators: please note that if your trial is registered by the sponsor, you must take responsibility to ensure that the registration information is accurate and complete.

Yours sincerely,

Albert F. Clark, PhD
Chair, Queen's University Health Sciences and Affiliated Teaching Hospitals Research Ethics Board

The HSREB operates in compliance with, and is constituted in accordance with, the requirements of the Tri-Council Policy Statement: Ethical Conduct for Research Involving Humans (TCPS 2); the International Conference on Harmonisation Good Clinical Practice Consolidated Guideline (ICH GCP); Part C, Division 5 of the Food and Drug Regulations; Part 4 of the Natural Health Products Regulations; Part 3 of the Medical Devices Regulations, and the provisions of the Ontario Personal Health Information Protection Act (PHIPA 2004) and its applicable regulations. The HSREB is qualified through the CTO REB Qualification Program and is registered with the U.S. Department of Health and Human Services (DHHS) Office for Human Research Protection (OHRP). Federalwide Assurance Number: FWA#: 00004184, IRB#: 00001173

HSREB members involved in the research project do not participate in the review, discussion or decision.
Appendix B
IRB Approval for Data Collection in Chapter 3

QUEEN'S UNIVERSITY HEALTH SCIENCES & AFFILIATED TEACHING HOSPITALS
RESEARCH ETHICS BOARD (HSREB)

HSREB Initial Ethics Clearance

January 22, 2018

Dr. Michael Rainbow
Department of Mechanical and Materials Engineering
Queen's University

ROMEOTRAQ: #6022292
Department Code: MECH-061-17
Study Title: Control of foot and ankle mechanics during hopping: deciphering the role of the plantar fascia and intrinsic foot muscles.
Co-Investigators: Dr. A. Dickinson, Dr. D. Borschneck, Miss L. Hutchinson, Dr. A. McGuire, Miss S. Kessler
Review Type: Full Board
Date of Full Board Meeting: December 11, 2017
Date Ethics Clearance Issued: January 22, 2018
Ethics Clearance Expiry Date: January 22, 2019

Dear Dr. Rainbow,

The Queen's University Health Sciences & Affiliated Teaching Hospitals Research Ethics Board (HSREB) has reviewed the application and granted ethics clearance for the documents listed below. Ethics clearance is granted until the expiration date noted above.

- Peer Reviews
- Budget – Version 2
- Recruitment Poster – Version 2 - 11/17
- Screening Form – Version 2
- Information/Consent Form – Version: Jan. 15, 2018

Documents Acknowledged:

- CV- M. Rainbow
- CORE – L. Hutchinson

Amendments: No deviations from, or changes to the protocol should be initiated without prior written clearance of an appropriate amendment from the HSREB, except when necessary to eliminate immediate hazard(s) to study participants or when the change(s) involves only administrative or logistical aspects of the trial.

Renewals: Prior to the expiration of your ethics clearance you will be reminded to submit your renewal report through ROMEO. Any lapses in ethical clearance will be documented on the renewal form.

Completion/Termination: The HSREB must be notified of the completion or termination of this study

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through the completion of a renewal report in ROMEO.

**Reporting of Serious Adverse Events:** Any unexpected serious adverse event occurring locally must be reported within 2 working days or earlier if required by the study sponsor. All other serious adverse events must be reported within 15 days after becoming aware of the information.

**Reporting of Complaints:** Any complaints made by participants or persons acting on behalf of participants must be reported to the Research Ethics Board within 7 days of becoming aware of the complaint.

*Note:* All documents supplied to participants must have the contact information for the Research Ethics Board.

Investigators please note that if your trial is registered by the sponsor, you must take responsibility to ensure that the registration information is accurate and complete.

Yours sincerely,

[Signature]

Chair, Health Sciences Research Ethics Board

The HSREB operates in compliance with and is constituted in accordance with the requirements of the Tri-Council Policy Statement: Ethical Conduct for Research Involving Humans (TCPS 2); the International Conference on Harmonisation Good Clinical Practice Consolidated Guideline (ICH GCP); Part C, Division 5 of the Food and Drug Regulations; Part 4 of the Natural Health Products Regulations; Part 3 of the Medical Devices Regulations, Canadian General Standards Board, and the provisions of the Ontario Personal Health Information Protection Act (PHIPA 2004) and its applicable regulations. The HSREB is qualified through the CTO REB Qualification Program and is registered with the U.S. Department of Health and Human Services (DHHS) Office for Human Research Protection (OHRP). Federalwide Assurance Number: FWA#: 00004184, IRB#: 00001173

HSREB members involved in the research project do not participate in the review, discussion or decision.
Appendix C
Supplementary Data for Chapter 2

Regions of common significance of the gait cycle for detecting changes between several gaits were summarized in Figure 2.2 (c) and (d) for the foot and shank IMUs, respectively. An example of the derivation of the areas of common significance was given in Figure 2.2 (a). Derivations of the areas of common significance for all gait comparisons are given below.

![Figure C.1: Derivation of areas of common significance between rearfoot strike and bouncing gait using IMU based resultant angular velocities of the feet. Resultant angular velocities are shown in the upper plot, and SPM results in the lower. Population-based SPM is shown as a black line, while individual SPM results are shown in dotted blue. Level of common significance is indicated at the bottom of the lower plot.]
Figure C.2: Derivation of areas of common significance between rearfoot strike and genu-valgus gait using IMU based resultant angular velocities of the feet. Resultant angular velocities are shown in the upper plot, and SPM results in the lower. Population-based SPM is shown as a black line, while individual SPM results are shown in dotted blue. Level of common significance is indicated at the bottom of the lower plot.

Figure C.3: Derivation of areas of common significance between rearfoot strike and forefoot striking using IMU based resultant angular velocities of the feet. Resultant angular velocities are shown in the upper plot, and SPM results in the lower. Population-based SPM is shown as a black line, while individual SPM results are shown in dotted blue. Level of common significance is indicated at the bottom of the lower plot.
Figure C.4: Derivation of areas of common significance between narrow and wide stance using IMU based resultant angular velocities of the feet. Resultant angular velocities are shown in the upper plot, and SPM results in the lower. Population-based SPM is shown as a black line, while individual SPM results are shown in dotted blue. Level of common significance is indicated at the bottom of the lower plot.

Figure C.5: Derivation of areas of common significance between short and long strides using IMU based resultant angular velocities of the feet. Resultant angular velocities are shown in the upper plot, and SPM results in the lower. Population-based SPM is shown as a black line, while individual SPM results are shown in dotted blue. Level of common significance is indicated at the bottom of the lower plot.
Figure C.6: Derivation of areas of common significance between rearfoot strike and bouncing gait using IMU based resultant angular velocities of the shanks. Resultant angular velocities are shown in the upper plot, and SPM results in the lower. Population-based SPM is shown as a black line, while individual SPM results are shown in dotted blue. Level of common significance is indicated at the bottom of the lower plot.

Figure C.7: Derivation of areas of common significance between rearfoot strike and genu-valgus gait using IMU based resultant angular velocities of the shanks. Resultant angular velocities are shown in the upper plot, and SPM results in the lower. Population-based SPM is shown as a black line, while individual SPM results are shown in dotted blue. Level of common significance is indicated at the bottom of the lower plot.
Figure C.8: Derivation of areas of common significance between rearfoot strike and forefoot striking using IMU based resultant angular velocities of the shanks. Resultant angular velocities are shown in the upper plot, and SPM results in the lower. Population-based SPM is shown as a black line, while individual SPM results are shown in dotted blue. Level of common significance is indicated at the bottom of the lower plot.

Figure C.9: Derivation of areas of common significance between narrow and wide stance using IMU based resultant angular velocities of the shanks. Resultant angular velocities are shown in the upper plot, and SPM results in the lower. Population-based SPM is shown as a black line, while individual SPM results are shown in dotted blue. Level of common significance is indicated at the bottom of the lower plot.
Figure C.10: Derivation of areas of common significance between short and long strides using IMU based resultant angular velocities of the shanks. Resultant angular velocities are shown in the upper plot, and SPM results in the lower. Population-based SPM is shown as a black line, while individual SPM results are shown in dotted blue. Level of common significance is indicated at the bottom of the lower plot.