Lower Limb Biomechanics in Walking, Running and Cycling: Implications for Overuse Injury

by

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Author’s Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners. I understand that my thesis may be made electronically available to the public.

Abstract

A common anecdotal theory among endurance athletes is that cycling results in fewer knee injuries compared to running. This is thought to be due in part to a lower impact ground reaction force in cycling, compared to running. Thus, as these endurance athletes age, there is a tendency to shift from running participation to cycling participation in order to avoid injury. However, the knee has been reported to be the most commonly injured region in both sports, with similar injury rates (Clarsen, Krosshaug, & Bahr, 2010; James, 1995). While it has been found that cycling does typically result in a lower peak ground reaction force compared to running (Gatti et al., 2017), it is unknown how other mechanisms (which could potentially lead to injuries such as iliotibial band syndrome and patellofemoral pain syndrome) differ between these exercise modalities. There has been an abundance of research conducted assessing the impacts of cycling on subsequent running performance (i.e. triathlon performance), specifically from a physiological point of view (Heiden & Burnett, 2003; Hue, Le Gallais, Chollet, Boussana, & Préfaut, 1998). No study to date, however, has explicitly compared the biomechanics of running and cycling. The purpose of the current study was to compare dynamic joint stiffness, co-contraction of the muscles surrounding the knee, segment coordination variability and iliotibial band impingement measures between walking, running and cycling in young, experienced runner/cyclists, in order to elucidate the risk of developing knee injury in one activity over the other.

Fifteen healthy, trained runner/cyclists (11M, 4F, age: 25.1 ± 4.7 years, height: 1.80 ± 0.1 m, mass: 72.1 ± 8.2 kg) were recruited. Muscle activity for 7 lower limb muscles were collected using wireless surface electromyography. External ground reaction forces were collected using force plates for the walking and running trials and an instrumented force pedal

for the cycling trials. 3D kinematics were collected using an active motion capture system. Participants performed 6 trials of walking at a self-selected pace, 6 trials of running at a pace equivalent to 70% of their maximal heart rate and 6 minutes of cycling at an intensity of 65% of their maximal heart rate. These intensities were selected to represent a typical, social weekend activity. A walking or running trial consisted of one progression overground on a 20m runway and a cycling trial consisted of 30 second efforts extracted from a continuous, steady state, 6-minute trial on the cycle ergometer. Walking was assumed to be a relatively low injury-risk activity and was performed to act as a baseline to which running and cycling could be compared. Kinematic, kinetic and electromyographical signals were analyzed during the stance phase of walking/running and the downstroke of cycling. These portions of the respective activities were chosen since they are the main propulsion producing phases of their respective activities.

Compared to walking and cycling, running generally had a larger dynamic joint stiffness and co-contraction index. For the entire stance/downstroke, and when it was broken into an initial and terminal phase, running had the largest DJS, followed by walking and then cycling (all $p<0.0001$). For stance/downstroke as well as the terminal phase, for all muscle groupings, running had a greater CCI compared to walking and cycling (all $p<0.05/4$), which were not different from each other. For the initial phase, for the VLLG and VMMG muscle groupings, running had a greater CCI compared to walking and cycling (both $p<0.05/4$), which were not different from each other. For the VLBF muscle grouping, running had a greater CCI compared to walking, which was greater than cycling ($p=0.002$) and for the VMST muscle grouping, walking and running were larger than cycling, but were not different from each other ($p=0.002$). The coordination variability was not different between walking, running or cycling for the sagittal thigh / sagittal shank or sagittal shank / sagittal foot segment coupling (both $p>0.05$). In terms of IT band impingement measures, running had a larger ITBEX compared to walking and cycling, which were not different from each other ($p=0.0001$). In terms of IT band impingement duration over an equivalent cumulative load, there were no differences between walking, running and cycling ($p=0.164$). The outcome measures that were different from walking (assumed to be a relatively low-injury risk activity) might be a contributor to injury for that activity. Of the outcomes analyzed in the current study, joint stiffness and co-contraction thus could potentially play a role in running injuries. None of the studied outcomes, however, when interpreted in isolation, likely play a role in cycling injuries. At the large knee flexion angles, such as those exhibited during the downstroke of cycling, the patellofemoral contact pressure, area and force have been found to be increased compared to full extension (Lewallen, Riegger, Myers, & Hayes, 1990). Thus, perhaps large flexion angles may be an important factor, either in isolation in combination with the outcome measures analyzed such as the co-contraction indices and coordination variability, to overuse injuries in cycling.

This was the first study to explicitly compare the biomechanics of running and cycling in the same study. When comparing walking, running, and cycling, significant differences were found in the dynamic joint stiffness and co-contraction index. Since injury rates between the two sports are very similar, these findings suggest that the same injuries could manifest from different injury mechanisms. Runners may opt to cross train in cycling but should be warned that due to large knee flexion angles under load, there are mechanisms of injury during cycling that may also result in overuse injury.

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It’s been a great couple of degrees at Waterloo, and I’m very excited for my next adventure!

Table of Contents

List of Figures

List of Tables

List of Abbreviations

ACSM	American College of Sports Medicine
BDC	Bottom Dead Center
BF	Biceps Femoris
CSEP	Canadian Society for Exercise Physiology
DJS	Dynamic Joint Stiffness
DoF	Degrees of Freedom
EMG	Electromyography
GM	Gluteus Maximus
HRm	Maximum Heart Rate
IE	Index of Effectiveness
IT Band	Iliotibial Band
ITBEX	Iliotibial Band EMG Exposure
ITBIZ	Iliotibial Band Impingement Zone
$ITBIZ(t)_{cumul\_load}$	Iliotibial Band Impingement Zone duration for equivalent cumulative loading
$ITBIZ(t)_{eq\_workout}$	Iliotibial Band Impingement Zone duration for comparisons to Farrell et al. (2003) of an equivalent workout
$ITBIZ(t)_{per\_rep}$	Iliotibial Band Impingement Zone duration per repetition (stride or pedal revolution)
ITBS	Iliotibial Band Syndrome
KAM	Knee Adduction Moment
KFM	Knee Flexion Moment
LG	Lateral Gastrocnemius

MG	Medial Gastrocnemius
OA	Osteoarthritis
PFPS	Patellofemoral Pain Syndrome
RPE	Rating of Perceived Exertion
ST	Semitendinosus
TDC	Top Dead Center
TFL	Tensor Fascia Latae
VL	Vastus Lateralis
VM	Vastus Medialis
VO_2max	Maximal Oxygen Uptake

Chapter 1 : Introduction

Among endurance athletes, it is common for aging runners to shift to cycling, with the aim of mitigating injuries perceived to be related to running. Specifically, a lower incidence of cycling injuries is thought to be a consequence of the lower impact ground reaction forces (GRF) experienced in cycling compared to running. Contrary to this belief, however, similar injury rates have been reported between running and cycling. The overall incidence of non-traumatic injury (also called overuse injury) is up to 92.4% in running (Van Gent et al., 2007) and up to 85% in cycling (Baskins, Koppel, Oliver, Stieber, & Johnston, 2016; Dettori & Norvell, 2006) over the course of one year. Of these injuries, 42% (Clement, Taunton, Smart, & Mcnicol, 1981; Taunton, 2003) and 36% (Clarsen et al., 2010) are at the knee for running and cycling, respectively. Of overuse injuries to the knee, in both modalities, iliotibial band syndrome (ITBS) and patellofemoral pain syndrome (PFPS) are the two most prevalent (Callaghan, 2005; Nielsen, Nohr, Rasmussen, & Sorensen, 2013). The close relationship between ITBS and PFPS injury may be explained due to the iliotibial band having an indirect insertion on the patella (Hudson & Darthuy, 2009; Mendonça et al., 2016).

In running, ITBS (also referred to as runner's knee) has an incidence of ~15% (Baker, Souza, & Fredericson, 2011; Ellis, Hing, & Reid, 2007; Noehren, Davis, & Hamill, 2007; Taunton, 2003) and is the main cause of lateral knee pain in runners (Aderem & Louw, 2015). Further, PFPS has an incidence of 25% in running (Neal, Barton, Gallie, O'Halloran, & Morrissey, 2016; Sprenkel, 2014; Taunton, 2003) and is a debilitating, chronic injury that can last multiple years, requiring adjustments to training for the majority of those affected (Stefanyshyn, Stergiou, Lun, Meeuwisse, & Worobets, 2006).

In cycling, ITBS accounts for up to 24% of all overuse injuries in the knee (Ellis et al., 2007; Holmes, Pruitt, & Whalen, 1993) and PFPS (also known as biker's knee) has been found to account for up to 36% of all overuse injuries in the knee (Clarsen et al., 2010; Holmes et al., 1993; Weiss, 1985). In professional cyclists, 36% of cyclists are troubled with PFPS, with PFPS being responsible for 57% off all injuries requiring time off training/racing (Clarsen et al., 2010).

In Canada, there were 350,000 reported finishers of running races in 2015, with many more estimated to run recreationally (Athletics Canada, 2015). In the US, 17 million runners finished over 30,000 races across all distances in 2016 (Running USA, 2016), with 47.4 million Americans estimated to have run for fitness in 2016 (Statista, 2016). Half of all finishers were between 25-44 years old (Running USA, 2016), with 31% of all finishers being older than 45. With participation rates dropping rapidly from the largest age group (35-44 years old) to just 3% in the oldest measured age group (65+ years old) (Figure 1.1), the most recent statistics support the theory that running participation is more prevalent in a younger population, as opposed to older populations. Running participation is currently at an all-time high, at the peak of a 300% increase from 1990-2013 (Running USA, 2016). Additionally, an estimated 33 million Americans rode their bikes for 1 hour an average of 6 times per month in 2016 (Baskins et al., 2016).

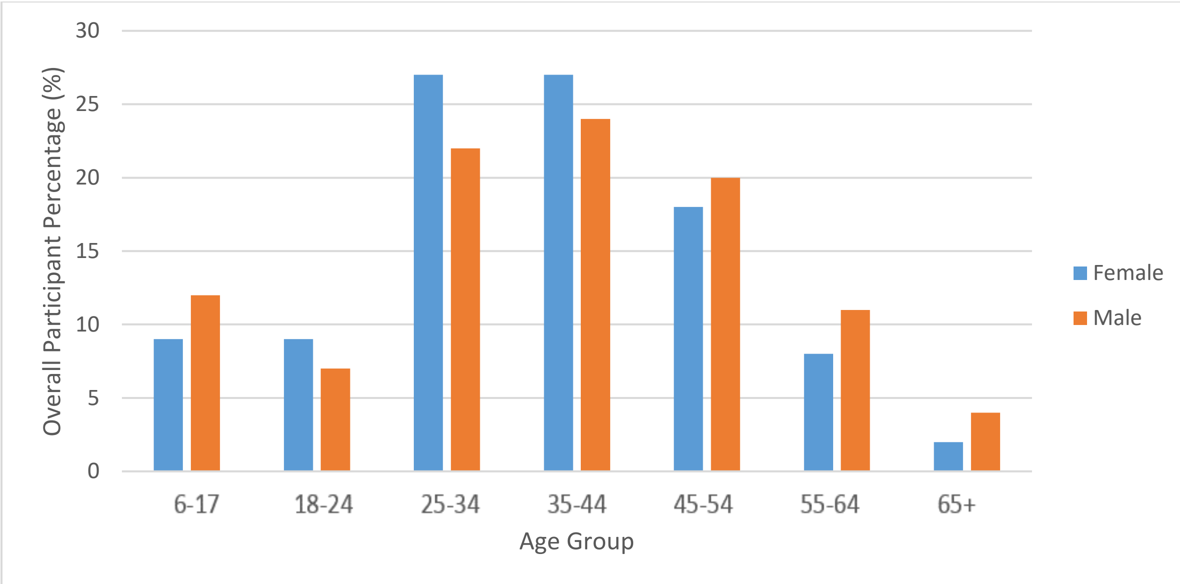


Figure 1.1: 2016 Running race participants in the U.S. by age and sex (data taken from the 2016 state of the Sport – U.S. Road Race Trends (Running USA, 2016)).

Clearly, a large proportion of individuals in Canada and the US participate in running and cycling as forms of physical activity in a society growing exponentially in terms of health conscientiousness. Endurance activities, such as running and cycling, improve cardiovascular and mental health, and decrease diabetes and mortality risks (Ghorbani et al., 2014; Lo et al., 2017; Petrovic-Oggiano, Damjanov, Gurinovic, & Glibetic, 2010; Williams, 2015). However, no study to date has performed an extensive biomechanical analysis explicitly comparing running and cycling. It is therefore unknown how participation in these exercise modalities affects the body from a biomechanics point of view in terms of injury mechanisms and knee health. Previous studies of running and cycling are mostly physiological, with emphasis on running performance after cycling, for triathlon applications (Bernard et al., 2003; Hausswirth et al., 2001; Heiden & Burnett, 2003; Hue et al., 1998; Millet & Vleck, 2000; Millet, Vleck, & Bentley, 2009; Vercruyssen et al., 2002). The current study fills this gap in the literature by comparing running and cycling to each other, as well as to walking, which was assumed to be a relatively low injury-risk activity.

A moderate intensity was the focus of the current thesis because this would be the intensity of a typical long run or long ride, which most athletes would perform for fitness or recreation. In order to quantify differences, dynamic joint stiffness, muscular co-contraction, segment coordination variability and IT band impingement measures were assessed. These outcome measures were selected since they are often investigated in walking, running and cycling literature and have previously been linked to overuse injury (Chmielewski, Hurd, Rudolph, Axe, & Snyder-Mackler, 2005; Farrell, Reisinger, & Tillman, 2003; Heiderscheit, Hamill, & Van Emmerik, 2002; Stefanyshyn & Nigg, 1998; Williams, Davis, Scholz, Hamill, & Buchanan, 2004). Understanding how these outcome measures compare between exercise modalities may help to identify similarities or differences in mechanisms of overuse injury in running and cycling, which may be of interest to endurance athletes, coaches and sport biomechanics researchers.

Chapter 2 : Objectives and Hypotheses

2.1 Objectives

The purpose of the current study was to compare lower limb kinetics, kinematics and electromyography (EMG) between walking, running and cycling in an active, experienced cohort aged 18-35, in order to investigate what could contribute to the similar injury risks between running and cycling. All outcome measures for running and cycling were compared to a walking task, which acted as a relatively low-risk comparison activity. The current study had four primary objectives:

1. *To determine how dynamic joint stiffness (DJS) of the knee differs between the different exercise modalities during physiologically equivalent bouts of exercise intensity.*

Physiologically equivalent bouts of exercise intensity were defined as exercise which elicits a "moderate intensity" heart rate (65%-71% heart rate max (HRm)) (Garber, Blissmer, Deschenes, Franklin, & Lamo, 2011), according to the American College of Sports Medicine (ACSM). Higher DJS has been associated with more advanced tibiofemoral OA (Zeni & Higginson, 2009) and was predictive of patellofemoral OA worsening (Chang et al., 2017) during walking. DJS has also been linked to a greater risk of tibial stress fracture (Milner, Hamill, & Davis, 2007), with a larger DJS being associated with an increased risk of bony injury and a lower DJS linked to a greater risk of soft tissue injury (Butler, Crowell, & Davis, 2003).

2. *To determine how muscular co-contraction for combinations of vastus lateralis (VL), vastus medialis (VM), gastrocnemius lateralis (LG), gastrocnemius medialis (MG) biceps femoris (BF) and semitendinosus (ST) differ between the exercise modalities.*

Greater co-contraction has been associated with increased joint contact forces (Hodge & Harris, 1986; Sasaki & Neptune, 2010; Winby, Lloyd, Besier, & Kirk, 2009), increased joint contact pressure (Li & Park, 2004) and increased energy expenditure in older adults (Hortobágyi et al., 2009; Hortobágyi, Finch, Solnik, Rider, & Devita, 2011). Increased co-contraction has also been retrospectively linked to PFPS in walking and running (Besier, Fredericson, Gold, Beaupré, & Scott, 2009) and has been identified as a coping mechanism for instability (Gantchev & Dimitrova, 1996; Hurd & Snyder-Mackler, 2007; Slijper & Latash, 2000),

3. *To determine how segment coordination variability differs between exercise modalities.*

Segment coordination represents segment control strategies adopted to perform a task. A lower segment coordination variability has been thought to be more detrimental to injury since forces may be applied more repeatedly to a localized point and less distributed about the surrounding tissues (Hamill, Palmer, & Van Emmerik, 2012).

4. *To determine how measures of IT band impingement differ between exercise modalities when considering multiple ways of defining exercise exposures.*

The EMG of gluteus maximus (GM) and cumulative ground reaction forces were used to determine measures of exercise exposures when comparing IT band impingement. The iliotibial band impingement zone (ITBIZ) (when the knee is flexed between 20° and 30°) has previously been defined as the range of motion where the IT band rubs against the lateral epicondyle of the femur and can cause irritation (Orchard, Fricker, Abud, & Mason, 1996). More time spent with the knee in the ITBIZ has been associated the onset of ITBS (Farrell et al., 2003; Orchard et al., 1996). A novel cumulative exposure metric is proposed in the current study, termed the IT band EMG exposure (ITBEX). It is the product of the number of times the knee passes through the ITBIZ in one minute, the angular velocity of knee flexion while it is in the ITBIZ and the sum of the integrated EMG of GM while the knee is in the ITBIZ. The ITBEX has never been investigated before, however, it is possible that when GM is activated, the IT band could become more tensioned, contributing to overuse injury. ITBIZ duration over an equivalent cumulative ground reaction force was calculated to investigate relationships between external reaction force and IT band impingement in walking, running and cycling.

In addition, two secondary objectives have been defined, prompted by findings in previous literature. The correlation between DJS and muscular co-contraction was investigated. It has previously been postulated that DJS in the knee may be explained by an increase in muscular co-contraction about the knee (McGinnis, Snyder-Mackler, Flowers, & Zeni, 2013). Co-contraction is often a strategy adopted to increase stiffness of the lower limb (McGinnis et al., 2013; Zeni & Higginson, 2009) and can increase knee compression forces (Hodge & Harris, 1986; Sasaki & Neptune, 2010).

The IT band impingement measures from the current study were also compared to those of a hypothetical equivalent workout. It has been postulated that more repetitions and total duration with the knee flexed in the ITBIZ has been associated with the onset of ITBS (Farrell et al., 2003). This comparison between running and cycling in an equivalent workout has previously been approximated (Farrell et al., 2003). An equivalent workout was suggested to be a 1 hour run and a 1.25-hour bike ride, during which, the knee was reported to spend 250 seconds in the ITBIZ over 4800 events in running, compared to 330 seconds in the ITBIZ over 6600 events in cycling. In this study, however, the running data were not experimentally measured so it is unknown how these values compare within the same individual, during exercise of equivalent intensities.

2.2 Hypotheses

The specific hypotheses of the primary objectives of the current study are as follow:

1. *The DJS will be greater in running compared to walking and cycling.*

DJS is the resistance of a joint's muscle and soft tissue to an external applied moment (Zeni & Higginson, 2009). It is reported as the slope of the linear regression of the external knee flexion moment (KFM) plotted against the knee flexion angle (Chang et al., 2017). Pedal reaction forces during cycling are expected to be lower than ground reaction forces during running (Gatti et al., 2017) and thus, lower knee flexion moments should be found in cycling. Because cycling is expected to have lower knee flexion moments and to have a larger range of motion, the slope of the regression line (DJS) is expected to be lower during cycling. In addition, walking is expected to have knee flexion moments similar to cycling and lower than running.

2. *Co-contraction will be greater in running compared to walking and greater for walking compared to cycling for all muscle groupings.*

Running is expected to have the largest co-contraction due to the more ballistic and high-velocity nature of the task, requiring more knee joint stabilization to perform the task and keep the body upright. While walking is a lower velocity task, it still requires the maintenance of upright posture through support at the foot-ground contact and thus is expected to have higher co-contraction compared to cycling. In cycling, support surfaces include pedals, handlebars and the seat, adding additional contact points to provide support.

3. *Coordination variability will be greater for walking compared to running and greater for running compared to cycling.*

Coordination variability is thought to explain how individuals perform the same task repetitively given redundancies in degrees of freedom of movements (further explained in section 3.2.2.4). A lower coordination variability is thought to be linked to injury, while a greater variability is related to healthy and high-performing individuals (Hamill et al., 2012). It has previously been reported that running has a trend of reduced variability compared to walking (Seay, Van Emmerik, & Hamill, 2011), although a statistical test was not run to determine significance in this study. Cycling with clipless pedals (where the forefoot portion of the cycling shoe is rigidly attached to the pedal) has reduced degrees of freedom (DoF) compared to walking and running, since the foot is restrained by the pedal system (reduced to 3DoF) and the pelvis is restrained by the saddle (reduced to 3DoF). This may affect the cyclist by limiting the range of motion and possible strategies available to perform the task. These constraints may result in lower coordination variability and have an effect on the knee joint kinematics. Running is hypothesized to have less coordination variability compared to walking because running is a more physically demanding (higher velocity) task. This higher demand is expected to limit the number of possible strategies available to the athlete to achieve the task. When landing from a jump or double float phase of running, certain constraints are imposed on the individual that must be performed to complete the task successfully while maintaining balance and control in preparation for the next stride. Walking is expected to have the largest coordination variability due to the low intensity and least amount of constraints imposed to perform the task.

4. *A) The ITBEX will be greatest for running compared cycling and greater for cycling compared to walking.*

The ITBEX is a cumulative measure, combining the number of times that the knee is flexed between 20° and 30° with the integrated EMG over the same interval. For each pedal revolution in cycling, Farrell et al. (2003) estimated that the knee would be flexed in this range for 38ms with the participant cycling at a power output of 280W and a cadence of 80-90 RPM. They also reported that in running, the knee would be flexed in that range for 75ms for each stride, based on previous data. In the previous running study, participants ran at a self-selected pace (between 2.78 and 3.89 m/s). In addition, the EMG during running is anticipated to be larger than that of walking or cycling. Thus, the ITBEX is expected to be largest for running.

4. B) The cumulative load ITBIZ duration will be greater for cycling compared to walking and running.

It has previously been reported that at a self-selected moderate intensity, 45 minutes of cycling results in the same external force cumulative load as a 15-minute run (Gatti et al., 2017). As mentioned in Hypothesis 4A, it has also been approximated that running has an ITBIZ duration per repetition of 75ms and cycling has 38ms per repetition (Farrell et al., 2003), with the cadence of running and cycling anticipated to be similar. Taking these results and calculating ITBIZ duration over an equivalent cumulative load, it is expected that cycling will have a larger ITBIZ duration compared to running. Walking and running are expected to be similar, since walking should have a lower cadence compared to running, but a larger ITBIZ duration per step due to longer stance durations.

Pertaining to the first of the secondary objectives, the DJS and muscular co-contraction is anticipated to be positively correlated for all muscle groupings for walking, running and cycling.

It has previously been hypothesized that DJS and co-contraction could be correlated (Chang et al., 2017; McGinnis et al., 2013). Though no strong correlations were previously found for walking (Chang et al., 2017; McGinnis et al., 2013), it has never been compared during running or cycling. DJS in running has been found to increase with increasing gait speed (Arampatzis, Brüggemann, & Metzler, 1999). It has been postulated that co-contraction acts to stabilize the joint via joint stiffening (Chmielewski et al., 2005).

Comparing IT band impingement measures over an equivalent workout, the current study aimed to validate results previously published. The number of ITBIZ events has previously been estimated by Farrell et al. (2003) over a workout of equivalent exposure (a 1 hour run vs. 1.25-hour bike ride) to be approximately 4800 ITBIZ events in running and 6600 ITBIZ events in cycling. These values for running were approximated from a separate study, under different testing conditions. Further, in addition to approximating the number of ITBIZ events, the total time the knee would be flexed in the ITBIZ over each specific workout was also determined. Using the assumption that over a pedal revolution, 38ms is spent in the ITBIZ for cycling and that for each stride, 75ms is spent in the ITBIZ for running, they found that running would result in a total of 330s in the ITBIZ and 250s for cycling (Farrell et al., 2003).

2.3 Hypothesis Summary Table

Table 2.1: Summary of hypotheses outlining outcome measures, expected result, signal over which each outcome measure is analyzed and statistical test used to determine differences.

	Outcome Measure	Hypothesis	Signal	Statistical Test
1.	DJS	Running > (Walking = Cycling)	Mean DJS per repetition	One-way repeated measures ANOVA
2.	CCI	Running > Walking > Cycling	Mean CCI per repetition	
3.	Coordination Variability	Walking > Running > Cycling	Intra-individual standard deviation of segment coordination	
4a)	ITBEX	Running > Cycling > Walking	Mean iEMG per minute in the ITBIZ	
4a)	$ITBIZ(t)_{cumul\_load}$	Cycling > (Walking = Running)	Mean ITBIZ duration over same cumulative load	

Chapter 3 : Literature Review

The objectives of this literature review are to establish a basic understanding of running and cycling mechanics (section 3.1), to broadly describe injuries common to both running and cycling (ITBS & PFPS) (section 3.2), to review previous literature comparing the biomechanics of running and cycling (section 3.3) and to explain the need for and the method for defining equivalent exercise exposures in the current study (section 3.4).

The benefits of endurance activities have widely been proven to offer numerous health benefits leading to increased cardiovascular health, mental health, and decreased mortality risk (de Hartog, Boogaard, Nijland, & Hoek, 2010; Ghorbani et al., 2014; Lo et al., 2017; Petrovic-Oggiano et al., 2010; Williams, 2015), among others. A six-week running program was found to increase mental health, decrease body mass index and improve VO_2max (maximal oxygen uptake; a performance metric quantifying the body's oxygen consumption during physical activity) (Ghorbani et al., 2014). On average, individuals who run live 3 years longer than those who do not run, have a 57% reduced risk of mortality by cardiovascular disease (Williams, 2015) and 30% reduced risk of mortality by all other diseases (Lo et al., 2017). Additionally, de Hartog et al. (2010) found that individuals 18-64 years old who switched to cycling to work from driving to work would gain 3-14 months of life expectancy, controlling for potential mortality effects (air pollution, accident rates). The Canadian Society for Exercise Physiology (CSEP) recommends that adults should perform at least 150 minutes per week of moderate – to vigorous-intensity aerobic physical activity, in bouts of 10 minutes or more (Tremblay et al., 2011). Further, the ACSM's physical activity recommendations state that individuals should perform between 150 minutes to 300 minutes of moderate-intensity, or 75 minutes to 150 minutes of vigorous-intensity aerobic physical activity per week (American College of Sports Medicine, 2018). As mentioned in the Canadian Physical Activity Guidelines provided for CSEP, running and cycling both qualify and are recommended as forms of physical activity.

Health practitioners often prescribe physical activity to patients in order to promote longevity and health. However, activities such as running and cycling present a risk of injury that may prohibit or discourage further physical activity, resulting in more sedentary behaviour.

3.1 General Background

Recreational running and cycling can both be classified as long distance endurance events (barring sprint running and cycling events, which are not the focus of the current study and involve different mechanics (Amen et al., 2011; Mann & Hagy, 1980)). Both involve repetitive motions at regular cadences that have distinct phases bilaterally. The following section will briefly outline the running stride and cycling pedal revolution.

3.1.1 Phases in the Gait Stride

Current gait analyses typically consist of two distinct phases; the stance phase and swing phase (Novacheck, 1998). Each phase is subdivided into more specific portions dependent on various biomechanical characteristics. One gait cycle consists of the events between when one foot first contacts the ground and when the same foot contacts the ground again. During human gait, the stages of a unilateral step consist of initial contact (0%-2% of the gait stride), loading response (2%-12%), midstance (12%-31%), terminal stance (31%-50%), pre-swing (50%-62%), initial swing (60%-75%), mid-swing (75%-87%) and terminal swing (87%-100%) (Rancho Los Amigos National Rehabilitation Center, 2001). These features occur temporally offset in the opposite leg to result in bipedal motion. Gait between walking and running share these common features, differentiated by altered temporal characteristics. During walking gait, the stance phase for each leg lasts approximately 62% of the gait cycle (Novacheck, 1998). Since stance phase is greater than 50% of the gait cycle, there is an interval of double-leg stance. As gait speed increases, stance time decreases, resulting in a double leg float above the ground. During running, typically ~40% of the gait cycle is in stance phase (in contact with the ground), depending on running pace (Novacheck, 1998). The remaining 60% is split between a double float phase where both feet are off the ground and a swing phase.

3.1.2 Phases in the Pedal Stroke

Compared to running, cycling biomechanics is a relatively new field of study, with the earliest recorded concepts of the bicycle being traced back to the mid-1500's (Lessing, 1997). The modern bicycle in its current form (2 wheels propelled by a crank and chain) began to develop early in the 19th century (Hamer, 2005). Today, modern recreational road cycling bikes are designed to optimize comfort, aerodynamics, and weight, often adopting technology used by professional cyclists. In recent years, there has been a heightened interest in research pertaining to cycling related injury (Dieter, McGowan, Stoll, & Vella, 2014; Farrell et al., 2003). Unlike running, which is an exercise modality in which form/technique is primarily led by an individual's anthropometrics and neuromuscular coordination, the kinematics of cycling is, in part, dependent on the characteristics of the bicycle and the interaction of the fit of the cyclist on the bicycle. Recreational and competitive cyclists are often "clipped in" to the pedals at their shoe, which reduces motions at the foot from 6 DoF to 3 DoF (lateral/medial translation, eversion/inversion and adduction/abduction are restricted or reduced). A bicycle has many adjustment options to best fit the rider in order to optimize performance, comfort and reduce injury such as saddle fore/aft, saddle height, handlebar reach, handlebar height, handlebar width, crank length and crank Q-factor, among others. Inappropriate adjustments can detrimentally alter the cyclist's movements, resulting in injury. No consensus has been agreed upon regarding an optimal bike fit (Fonda, Sarabon, & Li, 2014). Since no two individuals have the same anthropometric ratios, few guidelines have been accepted.

In order to optimize performance and reduce injury, historically, a knee angle of between 25°-30° is recommended at the bottom dead center (BDC) of the pedal stroke (Burke & Pruitt, 2003; Holmes et al., 1993), a torso angle of ~45° with the horizontal and an arm-torso angle of between 75°-90° is generally accepted for a standard road bike.

As with running, the cycling motion occurs in distinct phases. The cycle pedal revolution is broken into just two phases: the downstroke (often referred to as the power phase) and upstroke (often referred to as the recovery phase) (Fang, 2014). During the downstroke, force is applied downwards onto the pedal from the foot to apply a propulsive force to the bike. This occurs approximately between top dead center (TDC, when the crank arm is at the 12 o'clock position) and bottom dead center (BDC, when the crank arm is at the 6 o'clock position). At the same time, the opposite leg is in the upstroke phase, where it completes the final 180° from BDC to TDC (Bini, Hume, & Kilding, 2013). Ideally, the force applied to the pedal occurs with an index of effectiveness (IE) of 1.0 when the cyclist if clipped into the pedal. This would indicate that the resultant force (i.e. direction of force application) is perfectly perpendicular to the crank arm. In reality, however, the IE is not always 1.0, due to a variety of factors such as realistic pedal positions, skill level of the cyclist, and reluctance to "pull up" on the pedal during the upstroke phase of cycling. A large IE is found during the downstroke phase, when force is applied to propel the bike forward, however a negative IE is sometimes found during the recovery phase. This indicates that a cyclist may not pull up during the upstroke in order to maintain a high index of effectiveness throughout the entire cycle and actually applies a downward force on the pedal (Mornieux, Stapelfeldt, Gollhofer, Belli, & Etienne, 2008).

3.2 Common Injuries in Running and Cycling

As described in Chapter 1, overuse injury rates for running and cycling are very similar, with the knee being the most common injury site for both exercise modalities (Clarsen et al., 2010; Taunton, 2003). Of these injuries to the knee, ITBS and PFPS are the two most common (Callaghan, 2005; Nielsen et al., 2013). The following section will describe these injuries and investigate previous literature pertaining to these injuries.

3.2.1 Biomechanical Injury Risk Factors

Due to a lack of prospective studies, specific causes of injury are difficult to identify. Non-biomechanical risk factors thought to contribute to injury are training errors (such as suddenly increasing mileage or intensity) or improper equipment (Farrell et al. 2003). However, certain biomechanical outcome measures have previously been investigated as they relate to injury and have been associated with injured states. Specifically, DJS, CCI, coordination variability and IT band impingement measures have been identified as such outcome measures. In this section, DJS will be examined as it relates to general overuse injury. CCI and coordination variability will be reviewed in more depth in sections 3.2.2 and the IT band impingement measures will be reviewed in section 3.2.3, as they relate to PFPS and ITBS, respectively.

DJS defined as the slope of the linear regression of KFM plotted against knee flexion angle, quantifies the resistance of a joint to an external applied moment (Chang et al., 2017). When the slope is positive, such as the case of increasing flexion angle and increasing flexion moment, work is done by external forces and absorbed by the system. When the slope is negative, such as the case with decreasing flexion angle with decreasing flexion moment, work is done by internal forces and produced by the system (Frigo, Crenna, & Jensen, 1996). A greater DJS is the result of a larger ground reaction force and/or lower joint range of motion.

DJS is not a true measure of stiffness since it involves a biological joint with both active and passive structures (Latash & Zatsiorsky, 1993). “Quasi-stiffness” is a more accurate term to describe the moment-angle relationship, since stiffness refers to exclusively passive structures that have the ability to store and release energy (Latash & Zatsiorsky, 1993; Rouse, Gregg, Hargrove, & Sensinger, 2013). Despite this, among some biomechanists, the term DJS has been adopted (Baltich, Maurer, & Nigg, 2013; Chang et al., 2017; Lark, Buckley, Bennett, Jones, & Sargeant, 2003; Stefanyshyn & Nigg, 1998) and was the term used in the current study to refer to the slope of the regression line of KFM and knee flexion angle to remain consistent with the literature.

Typically used for gait studies, DJS has been retrospectively found to be twice as high in individuals with tibiofemoral OA compared to healthy controls (Zeni & Higginson, 2009). In addition, in a 2-year prospective study of 391 knees, Chang et al. (2017) found increased DJS to be associated with patellofemoral OA disease worsening, however no correlations to tibiofemoral OA were found. These discrepancies in tibiofemoral findings may be due to a greater baseline DJS, which has not been examined. It has also been reported that overuse injury can occur when DJS is excessive, or when it is not sufficient to support the task being performed (Butler et al., 2003). DJS has been investigated extensively in the ankles during walking and running under differing conditions (Frigo et al., 1996; Gabriel et al., 2008; Hamill, Gruber, & Derrick, 2014; Powell, Williams, Windsor, Butler, & Zhang, 2014; Stefanyshyn & Nigg, 1998) as well as for the knee (Frigo et al., 1996; Hamill et al., 2014; Kuitunen, Komi, & Kyröläinen, 2002; Williams et al., 2004).

The relationship between DJS and injury is made more complex by the fact that stiffness is also associated with performance; a larger DJS is associated with a greater gait velocity, jump height and economy of motion (Butler et al., 2003). Increased DJS been linked to an increase in risk for bony injuries such as tibial stress fractures (Milner et al., 2007), while a lower DJS being associated with an increase in soft tissue injuries (Butler et al., 2003). These postulations were supported by Williams et al. (2001), who studied the knee stiffness of high- and low-arch runners. It was revealed that high-arched runners ran with a greater leg stiffness compared to low-arched runners. It was believed that these differences contributed to the understanding of the phenomena that high-arched runners experienced more bony injuries and low-arched runners experienced more soft tissue injuries. It has been suggested that since large DJS is linked to bony injuries and low DJS has been linked to soft tissue injuries, there may be an optimal level of stiffness which balances the risk of injury and performance (Butler et al., 2003). Further, it believed that these stiffness values are of the most importance for injury risk during the early stance phase of gait (Milner et al., 2007; Powell et al., 2014).

Chang et al. (2016) did not collect EMG data during their prospective study showing the link between patellofemoral OA and DJS, due to participant burden (muscle pairs for each joint and each side must be collected) and the difficulty of analyzing surface EMG on their study cohort with a larger BMI (average BMI = 28.4 kg/m^2). DJS does not incorporate muscle activity, however it has been hypothesized that greater muscle co-activation creates a stiffer joint which in turn yields greater joint loading and potentially an increase in OA progression (Chang et al., 2017; McGinnis et al., 2013). Williams et al. (2004) found low co-activation values during running due to the relatively more active quadriceps muscle group compared to the hamstring group during the stance phase, however this was not compared to DJS. Co-contraction was not found to be correlated to DJS in individuals with ACL injury during walking (Gardinier, 2009) or in healthy individuals during walking (McGinnis et al., 2013). This has not been investigated in running or cycling, so it is unknown if there is an association between DJS and co-contraction indices of these more physically demanding activities.

3.2.2 Patellofemoral Pain Syndrome

PFPS is non-specific knee pain located between the posterior aspect of the patella and anterior surface of the distal femur (Dieter et al., 2014). PFPS is thought to be caused by maltracking of the patella and is often localized to the lateral aspect of the patella (Dieter et al., 2014; Wünschel, Leichtle, Obloh, Wülker, & Müller, 2011). Although specific etiology is unknown (van Zyl, Schwellnus, & Noakes, 2001), knee adduction and muscular imbalances are thought to be associated with PFPS.

At the proximal aspect of the patella, the quadriceps muscles insert as the quadriceps tendon (Wilson, Press, & Zhang, 2009). The patella is embedded within an arrangement of soft tissue, distal to the quadriceps tendon and proximal to the patellar ligament. The quadriceps act to extend the knee using the patella as a lever arm to increase mechanical advantage (Blackburn & Craig, 1980). Other muscles and tissues also have an indirect influence on the patella via attachments to the surrounding fascia (Blackburn & Craig, 1980). Temporal muscle imbalance has been thought to lead to the development and/or progression of PFPS (Dieter et al., 2014; Grabiner, Koh, & Draganich, 1994; Van Tiggelen, Cowan, & Coorevits, 2009). Dieter et al. (2014) concluded that temporal imbalance in the vasti might not affect kinematics, however it may still have an influence on pain, while Souza & Gross (1991) and Voight & Weider (1991) propose that knee extensor imbalances disturb the normal kinematics of the patellofemoral joint, leading to injury.

Given that PFPS and ITBS are both highly prevalent in both running and cycling, it is not out of the question that they could share a common injury mechanism. When the iliotibial band is tight, the patella is pulled laterally due to the iliotibial band's attachment to the lateral retinaculum of the patella, inducing PFPS tightness (Hudson & Darthuy, 2009). Therefore, factors that lead to an increased risk of ITBS could also lead to an increased risk of PFPS.

The following section will describe previous literature investigating PFPS in running and cycling specifically, as well as describe methods previously used to investigate PFPS in running and cycling.

3.2.2.1 Patellofemoral Pain Syndrome in Running

PFPS accounts for up to 25% of all overuse injuries to the knee in runners (Neal et al., 2016; Sprenkel, 2014; Taunton, Ryan, Clement, & McKenzie, 2002) and can potentially affect an individual for years, requiring alterations to their exercise routines (Stefanyshyn et al., 2006). A study consisting of retrospective and prospective components compared individuals with PFPS individuals with matched controls (Stefanyshyn et al., 2006). The study demonstrated that, retrospectively, 20 runners who had previously developed PFPS had higher knee abduction moment peak (130 Nm) than those who had never been affected by PFPS (105 Nm) (Stefanyshyn et al., 2006). Prospectively, 6 participants who went on to develop PFPS also had a higher knee abduction moment peak (78 Nm) compared to matched controls who did not (38 Nm). Further, it was discovered that in the retrospective component that participants with PFPS displayed a higher KAM impulse (17.0 ± 8.5 Nm·s) compared to healthy controls (12.5 ± 5.5

Nm·s). In the prospective study, the trend was similar with those who developed PFPS having a higher KAM impulse (9.2 ± 3.7 Nm·s) compared to matched healthy controls (4.7 ± 3.5 Nm·s).

Due to the many muscles having insertions on the patella as well as the importance of its role in the quadriceps tendon moment arm, muscle activations have been thought to influence patellofemoral pain (Besier et al., 2009). Between individuals with patellofemoral pain and pain-free controls, co-contraction was found to be significantly larger during walking, with patellofemoral pain group values trending to be larger during running (Besier et al., 2009). It was concluded that the increased joint contact force due to increased co-contraction could be causing pain in individuals in the patellofemoral pain group. Additionally, during abnormal lower limb kinematics (such as increasing internal rotation) the quadriceps muscles, having a direct insertion onto the patella, have been found to produce increased lateral forces on the patella (Powers, 2003) and have an influence on PFPS.

In addition to studying PFPS injury in gait from a ‘traditional’ biomechanics approach (EMG, joint moments), PFPS has been investigated using segment coordination (Hamill, Van Emmerik, Heiderscheit, & Li, 1999; Heiderscheit et al., 2002). Heiderscheit et al. (2002) found reduced variability in individuals with PFPS. Eight participants with and eight participants without PFPS ran on a treadmill both at a fixed pace of 2.68 m/s and at a self-selected pace. Though no differences were found averaging coordination variability over the entire stride cycle, the PFPS group showed a reduced variability in the transverse thigh / transverse shank segment coupling at heel strike during the self-selected running trials. Individuals with PFPS might constrain the available coordination patterns to avoid pain, and thus show a reduced variability. It is unknown, however, whether this reduction in the cause for PFPS or influenced by it. Additionally, Hamill et al. (1999) investigated coordination variability in runners with and without PFPS. With participants running at 2.5m/s, coordination variability during late stance was lower in the PFPS group, compared to healthy controls. It was concluded that when individuals with PFPS repeated actions with a lower variability and, despite relatively low loads being applied, the repetition over a less distributed area, might localized damage to the tissue.

3.2.2.2 Patellofemoral Pain Syndrome in Cycling

Patellofemoral pain syndrome, also referred to as “Biker’s Knee” accounts for approximately 35% of all overuse injuries caused by cycling (Clarsen et al., 2010; Holmes et al., 1993; Weiss, 1985). Despite the large proportion of cyclists anticipated to experience this injury, there is limited research addressing the topic in relation to cycling biomechanics.

A study by Ericson & Nisell (1987) found that peak patellofemoral compressive forces could approach 900N for cycling at just 120W (for reference, professional cycling sprinters can reach a power output of 2000W+). The absolute compressive forces were found to be independent of body weight, a potential consequence of the seated posture, and associated knee positions, during cycling. They reported that patellofemoral joint forces were found to be higher with increasing workload and lower with lowered saddle height. Cadence had no effect on the forces. In addition, these compressive forces were lower than the forces experienced during some activities of daily living.

As previously stated, quadriceps imbalance is hypothesized to result in patellar maltracking and lead to PFPS. This was first investigated in cyclists by Dieter et al. (2014). In a case-control study of 10 healthy and 7 PFPS cyclist participants cycling for 10 minutes at a RPE of 14 on the Borg 6-20 scale, temporal characteristics of VM, VL, BF, and ST were compared (Dieter et al., 2014). Despite the well documented hypothesis that vasti onset times are associated with PFPS, onset activity of the vasti was not found to be correlated with altered kinematics in cyclists but may still have been a contributor to pain (Dieter et al., 2014). Dieter et al. (2014) concluded that at the point of offset imbalance, the knee is flexed to ~147° and at this large flexion angle, the patellofemoral joint is stable, so no lateral movement should occur. Instead, the lateral pull may induce injury to the lateral soft tissue as the source of pain.

3.2.2.3 Co-Contraction Indices

Co-contraction is the simultaneous activation of muscle groups surrounding a joint (Rudolph, Schmitt, & Lewek, 2007). Increased co-contraction has been linked to increased joint contact forces (Hodge & Harris, 1986; Sasaki & Neptune, 2010; Winby et al., 2009), joint contact pressure (Li & Park, 2004), energy expenditure in older adults (Hortobágyi et al., 2009, 2011) and patellofemoral pain (Besier et al., 2009). Further, evidence suggests that increased co-contraction is a coping mechanism for joint instability (Gantchev & Dimitrova, 1996; Hurd & Snyder-Mackler, 2007; Slijper & Latash, 2000). Via forward dynamic modelling of human gait, Winby et al. (2009) and Sasaki & Neptune (2010) have displayed muscle contributions to joint contact forces that are not accounted for during inverse dynamic analyses. Specifically, Winby et al. (2009) found that muscle contributions provided more than 50% of the tibiofemoral joint contact load calculated in their EMG-driven model of human gait. In vitro testing has shown that co-contraction increased patellofemoral joint contact pressures (Li & Park, 2004). While applying a quadriceps muscle force of 400N and a hamstring force of 200N to cadaveric knees at 0°, 30°, 60°, 90° and 120°, in contrast to quadriceps muscle activation in isolation, the addition of hamstring co-contraction increased patellofemoral contact pressures by up to 24%. Thus, an increased co-contraction may affect joint loading to the articulations about the knee (tibiofemoral and patellofemoral) in ways that are not accounted for through inverse dynamics analyses.

3.2.2.4 Segment Coordination and Coordination Variability

Segment coordination and segment coordination variability explain the adaptations that individuals perform in order to successfully complete a given task (Heiderscheit et al., 2002). Using a modified vector coding approach (explained in section 4.3.2) (Sparrow et al., 1987), segment coordination is defined by coupling angles, which are the angles between pairs of consecutive points of an angle-angle plot of two articulating segments (Figure 3.1). These coupling angles are calculated for each percent of an event of interest (i.e. stance phase of gait). Segment angles are calculated relative to the global coordinate system.

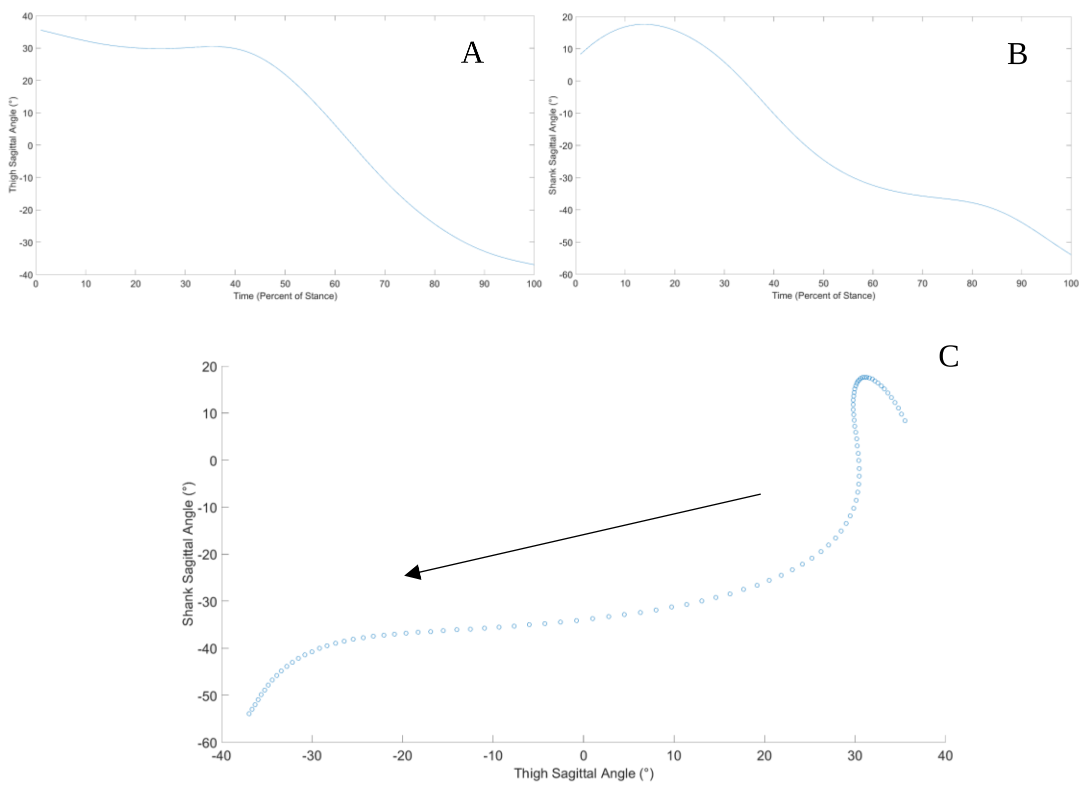


Figure 3.1: Segment angle plots for the A) sagittal thigh and B) sagittal shank for one stance phase of running. Segment angles are represented in the global coordinate system. The angle-angle plot of the data in A) and B) plotted against each other is depicted in C). The black arrow in C) indicates the direction of signal progression.

The coupling angle magnitude (0 - 360°, as measured form the right horizontal) represents the coordination strategy adopted to perform the task and can be interpreted by 4 distinct phases: anti-phase, in-phase, proximal phase and distal phase. The anti-phase pattern describes movement where the segments rotate in opposite directions (when coupling angles are 135° and 315°). The in-phase pattern describes movement where the segments rotate in the same direction (when coupling angles are 45° and 225°). The proximal phase pattern describes the movement where the proximal segment rotates, while the distal segment does not (when coupling angles are 0° and 180°). The distal phase pattern describes the movement where the distal segment rotates, while the proximal segment does not (when coupling angles are 90° and 270°). Since the coupling angles track a continuous movement, the segment angles rarely lie on these discrete values. Thus, 45° bins have been previously established to describe the primary coordination pattern of an action as one of the four phases (Figure 3.2) (Chang, Emmerik, & Hamill, 2008).

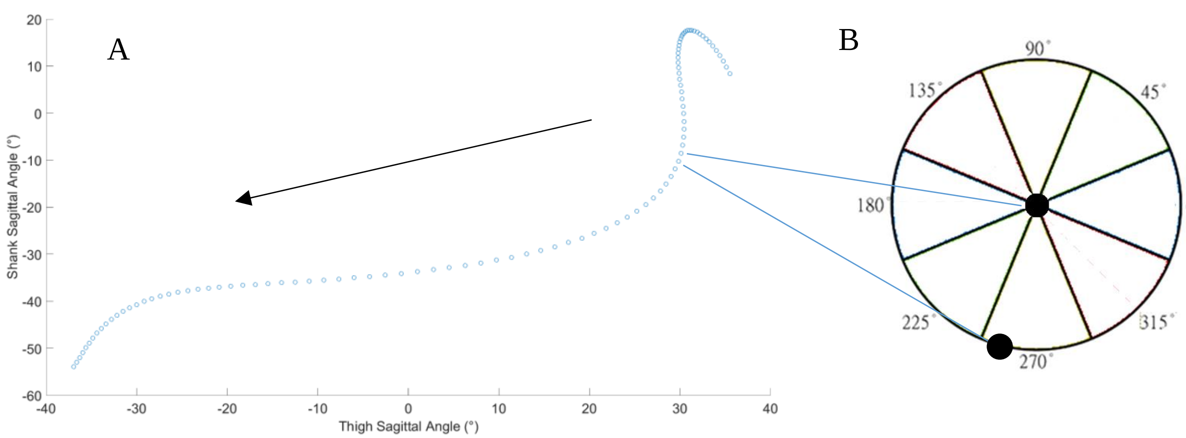


Figure 3.2: A) Angle-angle plot of the sagittal thigh and sagittal shank segment coupling, duplicate of Figure 3.1C. The black arrow indicates the direction of signal progression. B) Diagram for describing segment coordination patterns. When the centre black dot is positioned over a point in the angle-angle plot for segment couplings (expanded from A)), whichever section on the diagram that the next consecutive point falls into (the peripheral black dot), will determine the coupling angle (and the corresponding) segment coordination pattern at that percent of stance. In this instance, the coupling angle has a value of ~260° and would be classified as a proximal segment coordination pattern.

The coordination variability is the standard deviation of the coupling angle between trials at each percentage of the event being investigated. Healthy and high-performing individuals have been found to have greater coordination variability allowing them to adapt to changes in the environment (i.e. curbs, ice, uneven ground) by having more redundancies in degrees of freedom at their exposure (Hamill et al., 2012). Injured individuals with patellofemoral pain have shown to have lower segment coordination variability in the lower limb during running (Hamill et al., 1999). Increased coordination variability was thought to decrease the risk of injury by avoiding no specific tissue to be repeatedly engaged to the point of injury due to the range of variation of movements that could be recruited to perform the movement.

Using traditional biomechanics approaches (ground reaction forces, stride length, stride duration), increased variability has typically been thought to be detrimental and increase the risk of injury, such as risk of falls (Gabell, Simons, & Nayak, 2007; Hausdorff, Rios, & Edelberg, 2001). However, it may be helpful to distinguish ‘end-point’ variability from coordination variability. End-point variability is the variability of the final objective of a task. Arutyunyan et al. (1969) investigated the end-point variability and coordination variability of expert vs amateur marksmen. The expert marksmen, unsurprisingly, showed very low end-point variability (high accuracy) in a shooting task but their coordination variability was more varied compared to amateurs. In other words, there may be many ways to complete a task and experts can adapt to choose the most efficient solution. The same could be true for segment coordination of more complex tasks such as walking, running and cycling. The end-point variability of running at a constant pace with low variability in kinematics may be small (i.e. maintaining a constant pace) but the coordination variability required to do so may be large.

Individuals with little experience in a task perform by "freezing" degrees of freedom or limiting variation of a movement in order to maintain control. However, Hamill et al. (2012) have speculated that a very high coordination variability could also lead to injury, proposing that there is a happy middle ground where healthy, functioning athletes perform.

3.2.3 Iliotibial Band Syndrome

ITBS, first described by Renne (1975), is "rubbing" of the iliotibial band against the lateral femoral condyle. The iliotibial band originates from portions of the gluteus maximus and gluteus medius and tensor fascia latae muscles (Birnbaum et al., 2004; Muhle et al., 1999), while the distal insertion includes the lateral condyle of the tibia at Gerdy's tubercle (Orchard et al., 1996), the patella, head of the fibula and lateral intermuscular septum (Birnbaum et al., 2004). Its role is to primarily assist in hip adduction and to resist abduction. Further described by Orchard et al. (1996), ITBS was traditionally thought to be caused by friction caused by the iliotibial band crossing over the lateral femoral condyles at approximately 20°-30° of knee flexion, while gluteus maximus and tensor fascia latae contract, creating tension (Noble & Sa, 1979; Novacheck, 1998; Renne, 1975). This range of 20°-30° of knee flexion has been coined the "iliotibial band impingement zone" (ITBIZ) (Orchard et al., 1996). More recently, an alternative etiology for ITBS has been proposed, suggesting that the iliotibial band is held in place over the lateral aspect of the knee joint and as the knee is flexed into the ITBIZ, the femoral condyle moves *under* the iliotibial band, causing compression of the distal fibres and the layer of innervated fat over the lateral epicondyle (Fairclough et al., 2007; Falvey, Clark, Bryant, Briggs, & Mccrory, 2009). Regardless of the exact internal mechanism of ITBS, it is generally accepted that flexion of the knee through the ITBIZ is the primary cause of ITBS (Farrell et al., 2003).

Previous work in biomechanics has outlined possible causal factors leading to ITBS as well as movement alterations due to ITBS. In retrospective studies, greater knee flexion angles at heel strike (Miller, Lowry, Meardon, & Gillette, 2007), and weakness of hip external rotators, tight iliotibial band and altered neuromuscular control (Noehren, Schmitz, Hempel, Westlake, & Black, 2014) have been potential factors. In prospective studies, female runners who had developed ITBS demonstrated increased hip adduction, knee internal rotation and femur external rotation (Noehren et al., 2007), and male runners who had developed ITBS demonstrated greater hip internal rotation and knee adduction angles (Noehren et al., 2014).

Though the etiology of ITBS has not been agreed on, various impingement measures have been associated with ITBS. In a cadaveric study, Orchard et al. (1996) found that the posterior edge of the IT band is the most affected region of the IT band since, at full extension, the IT band was either overlying or just anterior to the lateral epicondyle. Thus, as the knee enters flexion, the posterior edge of the IT band rubs against the lateral epicondyle, which could lead to irritation. Both the duration with the knee spent in the ITBIZ as well as the number of times that the knee is flexed into the ITBIZ have previously been used to investigate the relationship between kinematics and ITBS (Farrell et al., 2003) and were investigated in the current study.

The following sections will investigate previous literature investigating ITBS in running and cycling specifically.

3.2.3.1 Iliotibial Band Syndrome in Running

ITBS is the most commonly reported knee condition in runners (Sprenkel, 2014) and the main cause of lateral knee pain in runners (Aderem & Louw, 2015). Its notoriety among endurance athletes has earned it the title of “Runner’s Knee” (Sprenkel, 2014). In subjects who previously had been diagnosed with ITBS, the average knee flexion angle at heel strike was found to be greater at 21.4° ± 4.3° (Orchard et al., 1996) and 20.6° ± 7.8° (Miller et al., 2007), in contrast to a heel strike angle of 15.3° in subjects who did not have a history of ITBS (Miller et al., 2007). It has been found that just after heel strike when GM and TFL (muscular attachments to the IT band) are most active (Mann & Hagy, 1980) and is when individuals with ITBS experience pain (Orchard et al., 1996).

There have also been sex differences related to ITBS development. In a prospective study of a cohort of female recreational runners followed for two-years, those who went on to develop ITBS presented increased hip adduction and knee internal rotation compared to matched controls (Noehren et al., 2007). The same group found that runners with ITBS had increased hip internal rotation and increased knee adduction, compared to controls.

3.2.3.2 Iliotibial Band Syndrome in Cycling

To date, only one experimental study has assessed biomechanics related to ITBS in cycling specifically (Farrell et al., 2003). Ten recreational cyclists were fit to a bicycle ergometer using a static bike fit with their knee angle between 25°-30° at BDC. Participants were asked to pedal for 5 minutes at 80-90 RPM at a workload of 280 W. Kinematics and kinetics were recorded. The results were a minimum knee flexion angle of 32.9° ± 7.2° with a pedal force of 230 N ± 64.8 N at that knee angle. The external reaction force was estimated to be 17%-19% of values previously reported for running in other studies. Further, duration in the ITBIZ was found to be 38ms for cycling (Farrell et al., 2003) and approximated to be 75ms for running, based on data previously published (Orchard et al., 1996). These were not equivalent intensities or calculated for the same individual. An individual’s cadence and step frequency inherently have a large influence on the calculation of duration in the ITBIZ. In cycling, the duration in the ITBIZ may be affected by factors such as the bicycle set up (i.e. saddle height) and/or cycling intensity (which may affect kinematic joint angles). The authors simulated a situation where they equated 4km cycled for every 1 km jogged and determined that cyclists would go through 30%-40% more ITBIZ events compared to running and that duration in the ITBIZ would be less for cycling (250s) than for running (330s). Speeds/pace and cadence/step frequencies were assumed for the scenario and not quantitatively equivalent in terms of intensity. It was concluded that since duration in the ITBIZ was less, reaction force was less, and number of ITBIZ events was greater for cycling compared to running, repetition is the primary factor leading to ITBS in cycling. The current study attempted to verify the assumptions and findings used to come to these conclusions.

3.3 Previous biomechanics work comparing running and cycling

The majority of studies investigating cycling and running focus on the physiology of the sports. Specifically, differences between VO_2, heart rate (Borg, Hassmén, & Lagerström, 1987; Faulkner, Roberts, Elk, & Conway, 1971), energy expenditure (Achten, Venables, & Jeukendrup, 2003) and run performance after cycling, for applications to triathlon performance (Hausswirth et al., 2001; Hue et al., 1998; Millet & Vleck, 2000; Millet, Vleck, & Bentley, 2009; Vercruyssen et al., 2002) are common themes.

As mentioned in Chapter 1, only one other study has measured ground reaction forces between running and cycling (Gatti et al., 2017). In this study, the main outcome measure was mean cartilage transverse relaxation times before and after bouts of running and cycling. Cumulative ground reaction force was calculated to represent a 15-minute run based on participant impulse ground reaction force from running multiplied by step cadence. The participants were then asked to ride on a bicycle ergometer instrumented with an instrumented force pedal, until the cumulative load of cycling reached the pre-calculated cumulative load equal to that of a 15-minute run. To produce an equivalent cumulative GRF, cycling duration was approximately three times that of the running duration. No kinematic or kinetic comparisons were made between the two exercise modalities and the intensity of the exercise bouts was self-selected. Speeds and power output for running and cycling varied (7.9-13kph and 85-200W, respectively). Cadence was regulated to 80RPM for cycling and step cadence for running was self-selected. Participants were screened for fitness level, but a very good cyclist who is not a trained runner, or vice versa, may have skewed the results as to how running compared to cycling.

Other studies (such as Farrell et al. (2003), described in section 3.2.3), have compared running to cycling using previously published data and hypothetical situations. No study to date has directly compared the biomechanics of running and cycling during regulated equivalent bouts of exercise intensity.

3.4 Equivalent Exercise Exposures

This study was the first to perform a direct kinematic analysis between running and cycling. In order to conduct such an analysis to relate the findings to an equivalent level of effort between running and cycling, both exercise modalities must be performed at an equal intensity. A measure of percent HRm is typically used when prescribing intensities for research studies, however two issues are present as they relate to the current study.

First, age-related equations are typically used to determine individual’s HRm. HRm is affected by training level and age (age equations can be inaccurate by up to 20 bpm in individuals over 40 years old) (Mollard et al., 2007; Nes, Janszky, Wisløff, Støylen, & Karlsen, 2013; Tanaka, Monahan, & Seals, 2001; Zavorsky, 2000). Since the current study was performed on highly trained athletes, this method may result in inaccurate results. Thus, a physiological test was conducted to directly determine HRm by having the participants perform a fatigue test to exhaustion. A maximum consumption of oxygen test (VO_2max test) was performed, since the protocol between a heart rate max test and VO_2max test are similar, in terms of intensity and time. The protocol for a VO_2max test can be conducted while performing a variety of exercise modalities. Of interest to this study, however, are the protocols for running and cycling. The main objective of the VO_2max test was to obtain a value for the participant's HRm. HRm has been found to differ between modalities in untrained individuals (Millet et al., 2009), but in trained triathletes, HRm has been found to be the same between modalities (Astrand & Saltin, 1961; Caputo, Mello, & Denadau, 2003; Fontana, Boutellier, & Knöpfli-Lenzin, 2009; Millet et al., 2009; Roecker, Striegel, & Dickhuth, 2003). Thus, only one test was needed, as opposed to a cycling maximal test and a running maximal test.

Second, heart rate is typically higher during running compared to cycling for equivalent rates of perceived exertion (RPE). Studies have found that for an equivalent RPE, percent lactate threshold or percent VO_2, the submaximal HR for running is approximately 10-20bpm or 5%-10% HRm higher than cycling (Fontana et al., 2009; Hetzler, Seip, Boutcher, & Pierce, 1991; Millet et al., 2009; Roecker et al., 2003). In many studies, non-significant differences in HR at the same RPE were ~7bpm, which is still considered clinically relevant (Millet et al., 2009; Mollard et al., 2007). In order to maintain the recommended intensity zone for "moderate intensity" typical of a recreational activity, running heart rate was prescribed at 70% HRm ± 5bpm and cycling heart rate was prescribed at 65% ± 5bpm. These values are both in-line with

ACSM recommendations for moderate intensity which has been found to simulate a long slow distance activity or a typical recreational outing (Stagno, Thatcher, & van Someren, 2007) and is an intensity at which lactate does not accumulate in the body (Hetzler et al., 1991).

3.5 Summary of Relevant Literature

To summarize, it remains unknown how factors such as dynamic joint stiffness, muscular co-contraction, segment coordination, and IT band impingement measures differ between running and cycling. Traditionally, runners have shifted to cycling participation in an attempt to preserve knee health and continue in endurance sport pain- and injury- free. However, overuse injury rates between running and cycling have been found to be similar (92.5% and 85%, respectively) with the knee being the most commonly injured region and the most common injuries being PFPS and ITBS for both sports.

DJS has previously been studied for walking and running and has been linked to OA (Chang et al., 2017) as well as bony injury and soft tissue injury (Butler et al., 2003). During walking and running, muscular co-contraction has previously been associated with patellofemoral pain (Besier et al., 2009) and has been reported as a coping mechanism for instability (Gantchev & Dimitrova, 1996; Hurd & Snyder-Mackler, 2007; Slijper & Latash, 2000). Further, co-contraction has been linked to increased knee joint contact forces (Hodge & Harris, 1986; Sasaki & Neptune, 2010; Winby et al., 2009), knee joint contact pressure (Li & Park, 2004) and energy expenditure in older adults (Hortobágyi et al., 2009, 2011). Segment coordination and coordination variability have been previously reported for walking and running and have implications for overuse injury risk. A lower coordination variability has been associated with an injured state (Hamill et al., 2012, 1999). ITBS risk has been previously studied by quantifying the duration with the knee flexed in the ITBIZ, number of repetitions and ground reaction force for both running and cycling (Farrell et al., 2003; Orchard et al., 1996). It was speculated that for cycling, number of repetitions played a bigger role in the development of ITBS compared to force or ITBIZ duration (Farrell et al., 2003)

No study to date has assessed the differences between running and cycling kinematics or kinetics. To compare the two sports, these variables had to be investigated during equivalent exposures to each other to provide meaningful comparisons. The current study compared these variables during an exercise bout of moderate intensity, similar to that of a recreational ride or run, to represent the activity habits of an individual performing these sports for leisure. A target heart rate of 65%-71% HRm was sought to conform to ACSM guidelines, however due to differences in heart rate at similar levels of perceived exertion (Hetzler et al., 1991), cycling was performed at a lower heart rate than running. Running trials were performed at a pace equivalent to 70% HRm ± 5bpm and cycling was performed at 65% HRm ± 5bpm.

Chapter 4 : Methods

4.1 Study Population

Fifteen participants between the ages of 18 and 35 were recruited (Table 4.1). Participants were required to have at least one year of self-reported experience in both running and cycling. Participants were pre-screened to verify that in the last year, on average, they performed at least one run and one bike ride per week (totaling a minimum of 30 minutes running and 60 minutes cycling, respectively) and that they have competed in at least one triathlon or duathlon in the past. Further, experience with clipless pedals was required since the instrumented pedals had a clipless design. The study population was recruited from local running and cycling clubs and University classes, teams and clubs. Additionally, a recruitment email was forwarded by Triathlon Ontario to members of every registered triathlon club in Ontario. In addition to the fifteen participants who were eligible and participated in the study, thirteen individuals responded to recruitment measures. Four respondents from various classes on campus were excluded due to not meeting the experience requirements. Three respondents from local training groups were excluded due to current lower limb injury. Six respondents from the Triathlon Ontario email list were excluded; one due to not meeting the experience requirements and five for exceeding the maximum age limit.

Table 4.1: Demographics of study participants.

Characteristic	
Sex (M/F)	11/4
Leg Dominance (L/R)	3/12
Age (years)	25.1 (4.7)
Height (cm)	1.80 (1.0)
Weight (kg)	72.1 (8.2)
BMI (kg/m^2)	22.2 (2.3)
Running Experience (years)	9.1 (4.8)
Cycling Experience (years)	7.4 (5.4)
Competition Experience (years)	5.3 (3.9)
Weekly Training (hours/week)	8.0 (3.1)

Note: Values are reported in mean (SD), where applicable.

Both running and cycling experience was sought because kinematics, kinetics, and EMG can differ between trained and untrained cyclists (Chapman et al., 2008) and intra- and inter-individual EMG variance is higher in untrained cyclists. As well, untrained athletes would be more likely to fatigue, which has been found to alter running and cycling kinematics (Derrick, Dereu, & McLean, 2002; Dingwell, Joubert, Diefenthaeler, & Trinity, 2008; Mizrahi, Verbitsky, Isakov, & Daily, 2000; Williams et al., 1991).

As previously described, individuals presenting with injury symptoms (PFPS, ITBS, etc.) have been shown to display altered kinetics and kinematics (Cowan, Hodges, & Bennell, 2001; Dieter et al., 2014; Patil, Dixon, White, Jones, & Hui, 2011; Wilson et al., 2009). Thus, individuals who were currently experiencing injury symptoms of the lower limb or within the past month were excluded. Further exclusion criteria were individuals with a history of previous injury requiring surgery, lower limb pathologies, or diagnosis of chronic disease, answering "yes" to any of the questions on the GAQ (Appendix A), less than 80% on the Lower Extremity Functional Scale (LEFS) (Appendix A) or those who had been instructed by a health care practitioner to not engage in physical activity. In addition, due to the importance of heart rate in the current study, individuals who were currently taking or had taken medication in the previous

14 days which acts to alter heart rate/circulatory properties were excluded. Examples include beta blockers, calcium channel blockers, digitalis glycosides, sodium channel blockers, potassium channel blockers, and blood thinners. As rubbing alcohol and adhesive strips were to be used for instrumentation, individuals with allergies to the aforementioned products were also excluded.

The study was approved by the University of Waterloo Office of Research Ethics and prior to participation in the study, all participants provided written, informed consent.

4.2 Experimental Design

The current study was a cross-sectional, within-subject study design with three conditions (walking, running and cycling) applied to all participants. The study consisted of two collection days. On the first day, participants completed screening questionnaires to validate their eligibility in the study and to confirm that there were no contraindications to participating. If all inclusion criteria were satisfied, a VO_2max test was performed in order to obtain a quantitative measure of the participant's physical fitness and to obtain a value of maximal heart rate.

Since overground running (on a 20m runway) was to be studied in this investigation, it was not expected that they would achieve the target heart rate (70% HRm ± 5bpm) during the overground trials. Thus, on the second day, the participants were asked to run on a treadmill in order to find a pace that would elicit 70% HRm ± 5bpm and then subsequent overground trials were performed at this pace. There was no trial to pre-determine cycling pace to elicit 65% HRm ± 5bpm because this heart rate could be achieved during the data collection trials on the ergometer.

After the treadmill run, the participants were asked to perform walking, running and cycling trials, in a quasi-randomized order. All trials within an activity type were performed consecutively. The order of activities always consisted of walking and running trials being performed back-to-back (i.e. never walking, cycling, running or in the reverse order) due to logistical constraints of re-digitizing the foot Optotrak markers (Appendix F). Walking trials consisted of 6 progressions down the runway at a self-selected normal walking pace, running trials consisted of 6 progressions down the runway at the pre-determined treadmill speed and the cycling trial consisted a 6-minute cycling bout at a heart rate of 65% HRm ± 5bpm. Activities were separated by 10 minutes of rest. While there have been no studies comparing walking, running and cycling biomechanics, and thus there is no standardized rest period between these activities, the 10-minute rest was assumed to allow adequate washout time for any biomechanical changes that may occur due to the previous exposure. Previous work has indicated that walking has no effect on subsequent running or cycling performance (Gardner, 2013; Gardner et al., 2015; Noehren et al., 2014), and that cycling can result in minimal lower limb angle changes during subsequent running (up to a 3.9° increase in ankle dorsiflexion and less than 1° in sagittal plane angles of the knee). These changes typically affect less experienced athletes and wash out within 5 minutes (Bonacci et al., 2010).

4.2.1 Experimental Protocol

On the first collection day (Figure 4.1), participants were greeted by the principal researcher in the lobby of the Centre for Community, Clinical and Applied Research Excellence (CCCARE), before being escorted to the cardiovascular assessment room (AHS TJB-1143). The participants read, and if they agreed to participate, signed a letter of informed consent. The participants then completed three forms. The first form was the participant screening questionnaire (Appendix A). This form consisted of questions about the participant's past and current health status to identify any contraindications to taking part in the current study. This form contained the Get Active Questionnaire (GAQ), which collects general health information to determine suitability for physical activity. No participants answered "yes" to one or more of the questions on the GAQ, the participant's collection, which would have indicated the need to consult a physician before physical activity. Next, a shortened version of the International Physical Activity Questionnaire (IPAQ) was completed to provide a quantitative assessment of physical activity and to ensure eligibility for participation (Appendix B). The last form the participants were asked to fill out is a participant information questionnaire (Appendix C). This questionnaire collects general information of the participant including sex, age, self-described running style, years of competitive experience, and a description of current equipment used (i.e. shoes, pedals, etc.).

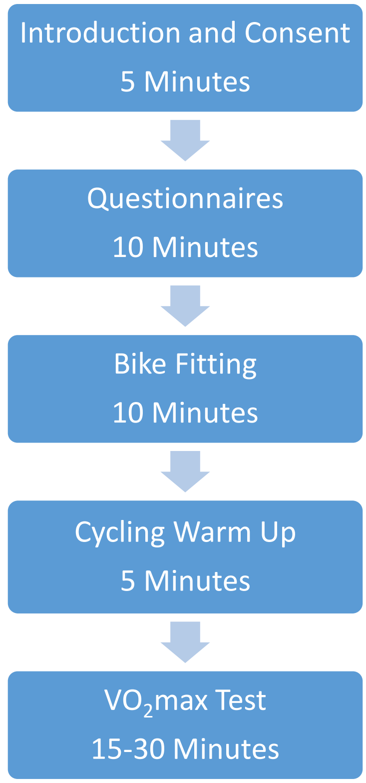


Figure 4.1: Day One – Experimental Protocol. Total duration was approximately 1 hour.

The participants were then asked to change into suitable athletic clothing (i.e. cycling shorts, t-shirt and athletic shoes). Participant height and weight was recorded. Participants performed a graded VO_2max test according to the ramp protocol on a cycle ergometer (Lucia, Hoyos, Santalla, Perez, & Chicharro, 2002). The participants sat on the cycle ergometer and it was adjusted to the participant's preference. VO_2 was measured using the VMAX Encore Metabolic Cart (Becton, Dickenson and Company, Franklin Lakes, New Jersey, USA) and heart rate was monitored using a 3-lead electrode placement. The cycle ergometer used was an Ergoline VIAsprint 150P (ergoline GmbH, Bitz, Germany), which is an electromagnetically resisted ergometer for which the resistance is controlled by the VMAX system and has a load range of 6-999 W. The ergometer was able to maintain a constant load as long as the participant's cadence was between 60 and 130 RPM.

The protocol began with a 5-minute warm up to prepare the participant for the test of increased intensity. The warm up was performed at a self-selected intensity and cadence. Following the warm up, the VO_2max test commenced. Starting at a power output of 25W, while maintaining a cadence of at least 60RPM, the resistance was increased by 25W every minute until failure or the participants were unable to continue (Lucia et al., 2002). The participants were fitted with a secured mouthpiece, through which the inhaled and exhaled air passed into the VMAX system for oxygen/carbon dioxide measurement. A nose clip was placed over the participant's nose to prevent air from bypassing the measurement system. As the participants breathed in and out during activity, the contents of the air was calculated by the VMAX system. Before each collection, the VMAX machine was calibrated for volume flow and oxygen/carbon dioxide gas proportion using built-in calibration tools. The VO_2max test results provided the participant's HRm, which was used to calculate the target heart rate at which running and cycling bouts were performed during the second testing day. These target heart rates were determined to elicit an exercise intensity of "moderate intensity" as defined by the ACSM, which is defined as 65%-71% HRm (Stagno et al., 2007). This intensity was expected not to induce any fatigue in the participant, as fatigue has been shown to result in alterations to joint kinematics while cycling (Dingwell et al., 2008) and running (Derrick et al., 2002; Mizrahi et al., 2000; Williams et al., 1991). Following the VO_2max test, the participant was given as long as they needed to cool down. The total collection time for the first day was approximately 1 hour.

On the second collection day, the participants arrived at the Biomechanics of Human Mobility Laboratory (BMH 1405) in Burt Matthews Hall. The participants were reminded of the study protocol, given the results from their VO_2max test, and informed of the previously calculated heart rate at which they were required to perform the running and cycling bouts. Due to known discrepancies between submaximal heart rate during running and cycling at equivalent perceived intensities, running heart rate was prescribed at 70% HRm ± 5bpm and cycling was prescribed as 65% HRm ± 5bpm, as described in section 3.4. Participants completed a secondary participant screening questionnaire (Appendix D). This questionnaire was an abbreviated version of the questionnaires completed on the first day to check that no medical changes had occurred between visits and to record physical activity since the last visit. Participants changed into running clothes. Participant height and weight were recorded.

The participants were asked to sit on the cycle ergometer and it was adjusted according to the protocol advocated by Bini and colleagues (2016). Bini et al. (2016) compared static and dynamic knee flexion angles from bike fits and found that statically measuring the sagittal plane angles with the pedal at the 3 o'clock position resulted in "trivial" differences between measured static and dynamic sagittal joint angles. While the participant was seated on the bike with their feet on the pedals placed at the 3 o'clock position, the saddle was adjusted to elicit a knee angle of 60° ± 5° with the knee inline vertically with the pedal spindle axle (plumb line). The handlebars were positioned such that the torso and the horizontal elicit a 40°-50° angle and the arms and the torso elicit a 75°-90° angle (Figure 4.2). Due to the experience level of the participants, it was expected that they would have a preferred bike fit. A 25mm saddle height adjustment range around the set saddle height was allowed to accommodate the participant's preference. Despite the literature reporting a 4%-7% change in saddle height can affect knee joint angles by up to 25% (Sanderson & Amoroso, 2009), it was expected that the participants would move in their seat to get to a comfortable position if the saddle were not adjusted to their comfort. The position of the saddle, handlebars and crank center of rotation were recorded for each participant.

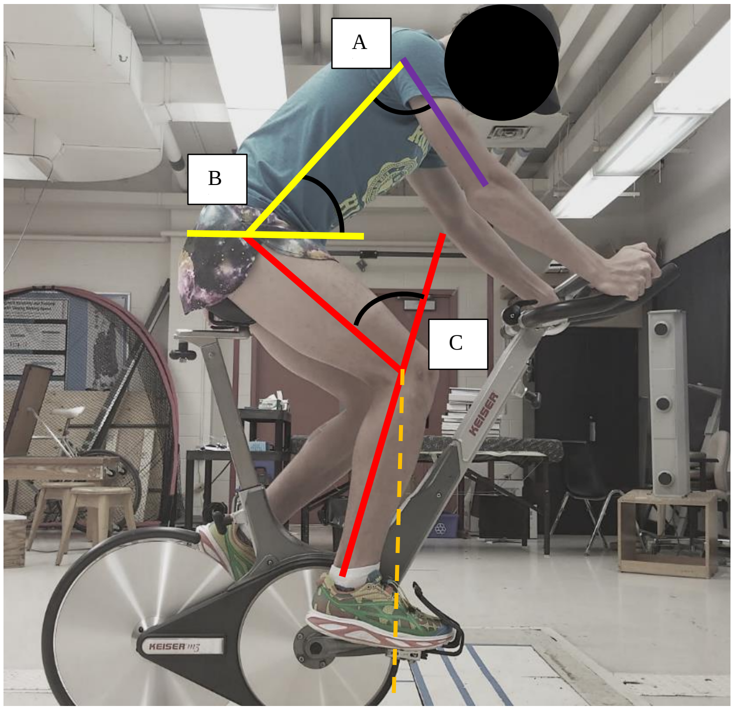


Figure 4.2: Cycle Ergometer Set-Up. A) Arm/Torso Angle (75°), B) Torso/Horizontal Angle (40°), C) Knee Angle (60°) with Pedal at 3 o'clock, the orange dotted line is the plumb line.

Participants were instrumented unilaterally on the right leg with EMG surface electrodes over VL, VM, LG, MG, BF, ST and GM, (Figure 4.3). Before application of the EMG electrodes, the area was shaved of hair and wiped clean with alcohol. Three trials of maximal sprint runs were performed over a 30-meter long runway in order to normalize the EMG data. This method has previously been shown to produce appropriate normalization values for studies investigating walking, running and cycling EMG (Chuang & Acker, 2019). Participants were given time to warm up before completing the maximal sprints by walking/jogging on a treadmill at a self-selected pace for 5 minutes.

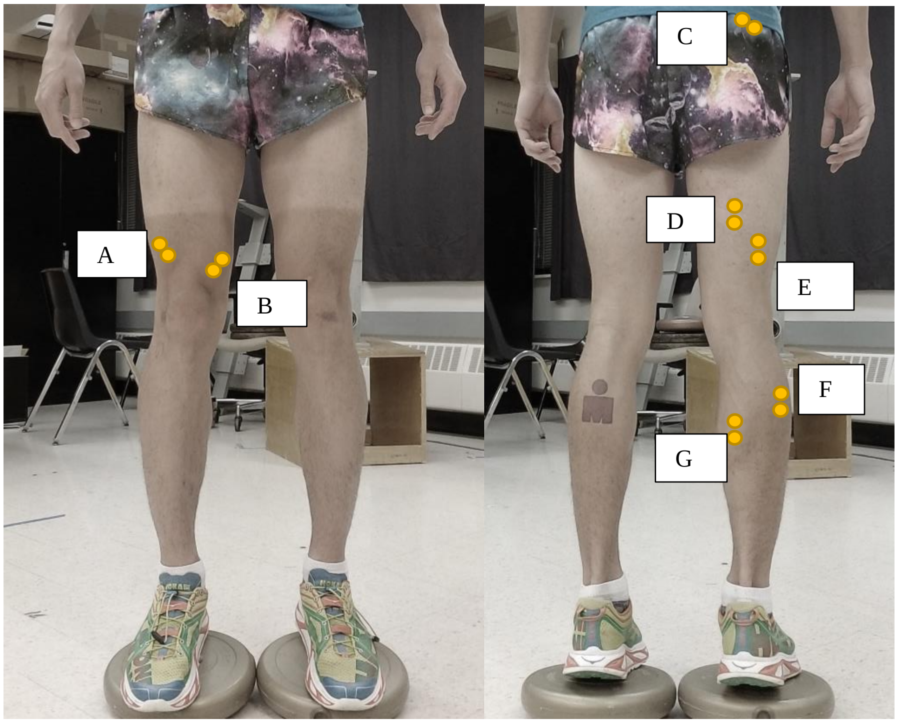


Figure 4.3: Unilateral Electromyography Placement for the Right Leg. A) Vastus Lateralis, B) Vastus Medialis, C) Gluteus Maximus, D) Biceps Femoris, E) Semitendinosus F) Gastrocnemius Lateralis, G) Gastrocnemius Medialis

After the maximal sprints, the participants performed the treadmill run. Participants performed a 5-10 minute run warm up at a self-selected pace followed by an increase to a speed that elicits 70% HRm ± 5bpm over the following 5 minutes. The treadmill was set to a 1% incline, since this has been found to replicate the metabolic expenditure of that of outdoor running at speeds between 2.92m/s and 5.0m/s (speeds that were expected in the current study) (Jones & Doust, 1996). The 5-10 minute warm up was in accordance to recommendations by the ACSM. Once 70% HRm ± 5bpm was achieved, the run was maintained for an additional 1 minute to confirm the pace and to allow heart rate to stabilize. The investigator stopped the test in the event the participant indicated any distress.

After the treadmill run, the participants were instrumented with motion capture marker clusters unilaterally on the right foot, shank, thigh and on the lower back (Figure 4.4). Motion capture calibration was performed to define bony landmarks and joint centers. A standing calibration trial, a knee flexion/extension trial and a hip range of motion trial was recorded in order to determine functional joint centers during data processing. Bony landmarks are listed in Table 4.2 (section 4.3.2).

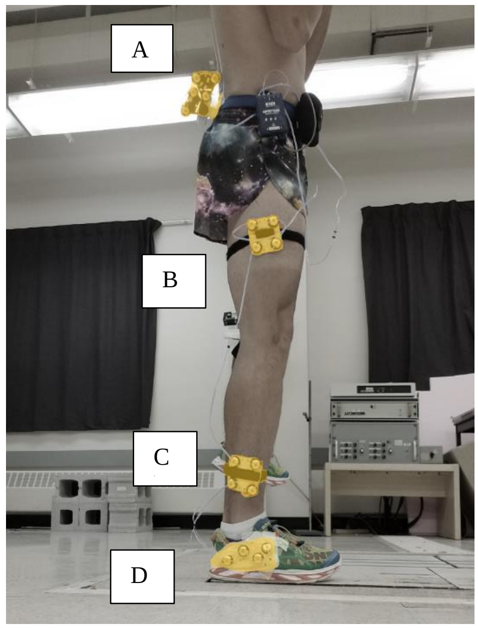


Figure 4.4: Kinematic Marker Placement, Unilaterally on the Right Leg. Rigid bodies with IREDs are located on the A) pelvis, B) right thigh, C) right shank and D) right foot.

Next, depending on the pre-selected block assigned to the participant, the following was performed as A) then B) or B) then A).

A) The participant was given 10 minutes rest. In a random order, the participants performed either walking at a self-selected normal pace or running at the pre-determined pace meant to elicit 70% HRm. If this block occurred second, the participant's foot was re-digitized after putting on running shoes.

Gait trials were performed down a 20-meter long laboratory runway (Figure 4.5) over the force plates until 6 acceptable trials were collected, always starting on the same mark and always starting with the same foot, unless otherwise indicated by the researcher. Acceptable trials included those where the right foot was isolated on a single force plate during stance, the motion tracking markers were visible to the collection cameras, EMG data showed no signs of abnormalities (such as signal spikes, dropouts or clipping) and the pace was within the set constraints. The walking and running pace were monitored using timing gates. The walking and running orders were randomized by participant. For the first few walking trials, the participants were asked to proceed at a self-selected comfortable pace. During these trials, the participant was instructed by the principal investigator to start at varying distances from the force plates, until their stance was isolated to one force plate. Over these walking 'familiarization' trials, walking pace was determined and the average was set as their preferred walking pace. For consecutive walking trials, their walking speed was monitored to within 10% of this preferred walking pace.

For the running trials, the first few trials were meant to familiarize the participant with running at the pre-determined pace meant to elicit 70% HRm ± 10%. Once they were comfortable consistently running at the set pace, the participant was instructed by the principal investigator to start at varying distances from the force plates until their stance was isolated to one force plate. Between trials, participants were given 1-minute recovery including a walk or light jog back to the starting mark. After the trial, if done after B) the participant was given as long as necessary to cool down.

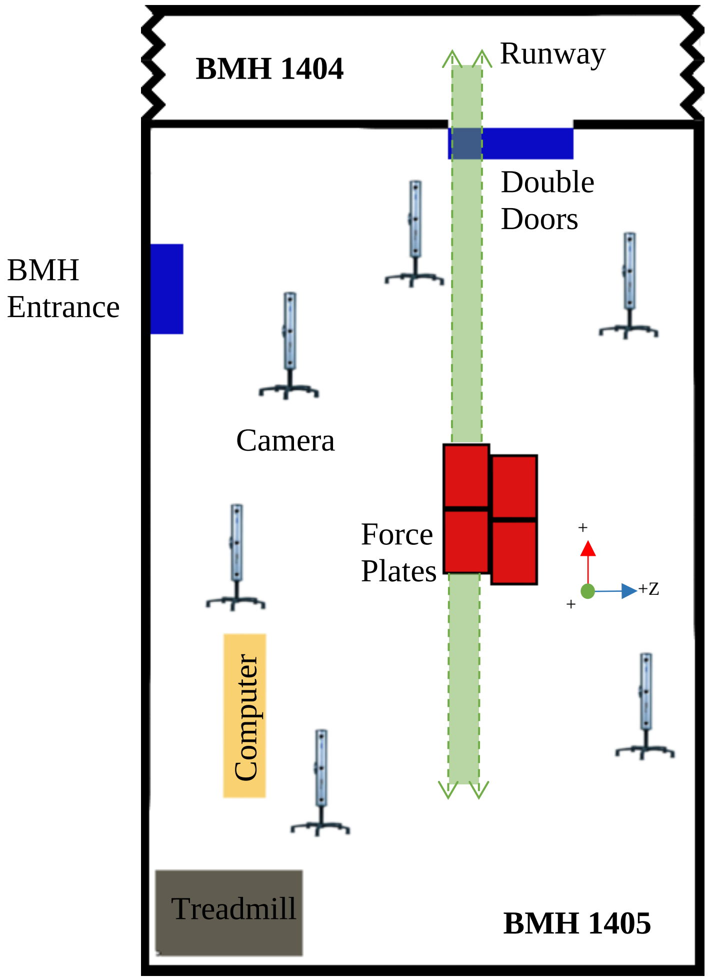


Figure 4.5: Laboratory Runway Set-up

B) The cycle ergometer was moved to the center of the collection area (Figure 4.6). 10-minutes of rest was given to the participant. If this block occurred second, the participant was re-digitized to track the bike in the global space, track the instrumented pedal with respect to the right foot and locate the center of pressure of the right foot (section 4.3.3). Participants then begin the 11-minute cycling trial. The first 5 minutes was a cycling warm up while the resistance was gradually increased to elicit 65% HRm ± 5bpm at a self-selected cadence. The final 6-minutes were performed at 65% HRm ± 5bpm at a self-selected cadence. The data were collected for 30 seconds, on every minute of the 6-minute steady state exposure. After the trial, if done after A) the participant was given as long as necessary to cool down.

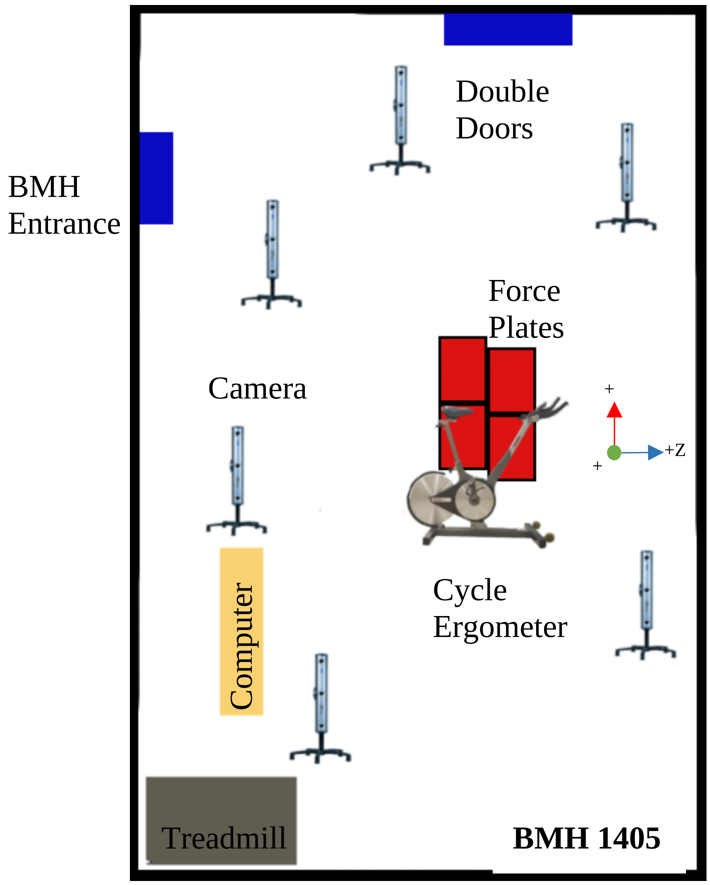


Figure 4.6: Cycle Ergometer Laboratory Set-up.

After all trials were complete, all equipment was removed from the participant and any re-usable laboratory equipment (marker clusters, surface electromyography signal boxes) were cleaned and disinfected with an alcohol-based solution. Any clothing provided to the participant was collected and washed before its next use. The total collection time for the second day was approximately 2 hours and 40 minutes (Figure 4.7).

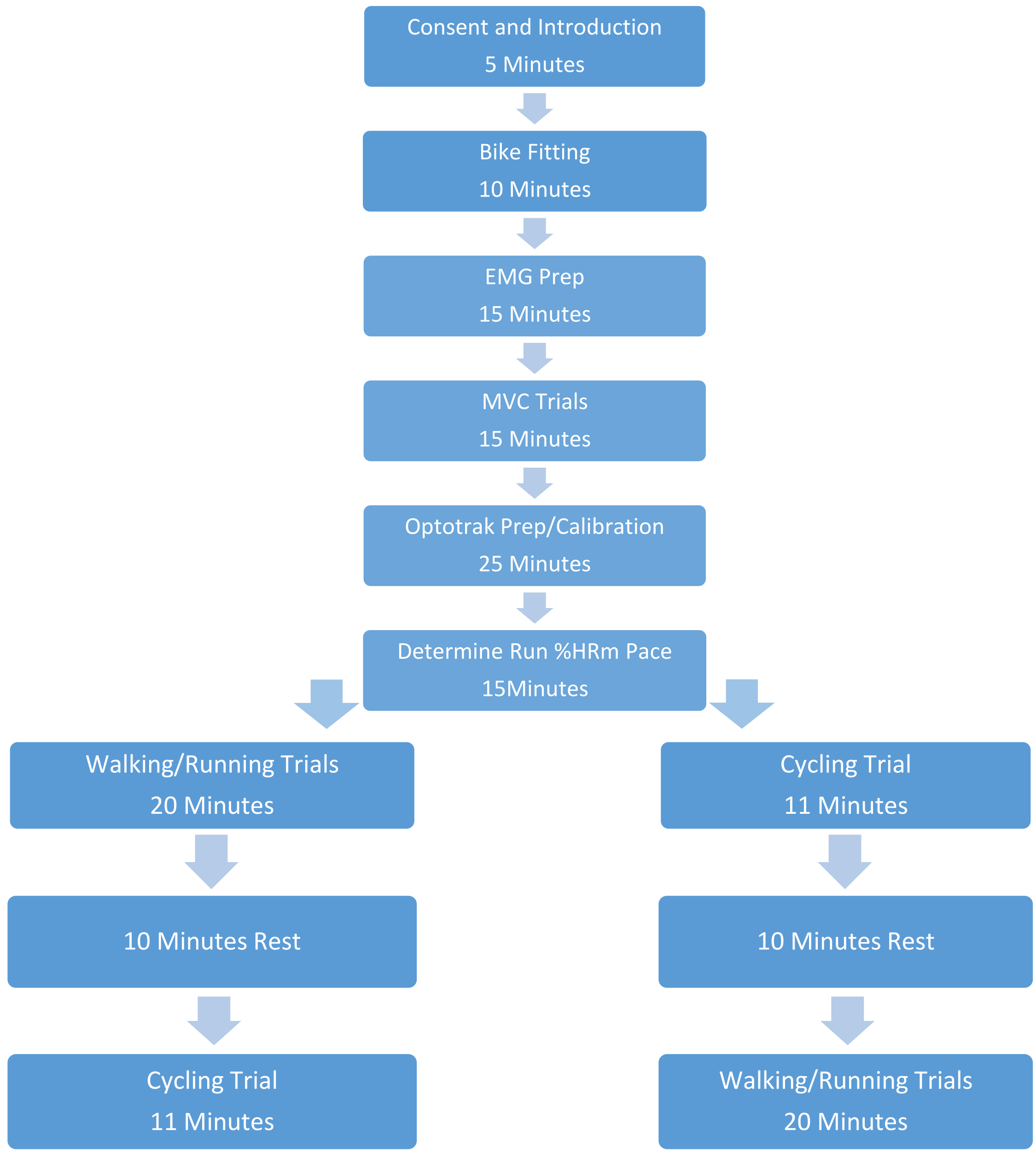


Figure 4.7: Day Two - Experimental Protocol. Total approximate duration was 2 hours and 40 minutes.

4.3 Data Analysis

Inherently, walking, running and cycling are very different activities and produce vastly different signals for the same outcome variables. Previous research investigating each activity in isolation has typically used characteristics specific to each activity in order to partition the data for further analysis. Such examples include heel strike, toe off and swing phases for gait (Novacheck, 1998) and TDC/BDC for cycling (Dieter et al., 2014; Farrell et al., 2003). Even walking and running (both forms of bipedal gait) have unique characteristics that make comparisons difficult (i.e. double stance phase of walking or double float phase of running (Novacheck, 1998). Thus, in the current study, phases within each activity were defined in order to keep comparisons as consistent as possible.

Holistically, the stance phase of walking and running and the downstroke of the pedal revolution of cycling will be used in the following analysis. These phases were chosen since they are the primary power-producing phases for each respective activity (i.e. the swing phase of walking/running and the up-stroke of cycling do not produce as much propulsive power relatively, if at all). Before the data were segmented into the stance phase or down-stroke, the raw data were filtered such that there were an adequate number of padding points (minimum 20) both before and after the signal of interest to ensure that filtering end effects would be removed when trials/phases were cropped to appropriate lengths (Smith, 1989).

The stance phase in walking and running will be defined as from heel strike (when a force of 20N is registered on the force plate) and toe off (when the force decreases below 20N) (Powell, Andrews, Stickley, & Williams, 2016).

The downstroke of cycling will be defined as from pedal force initiation (when the normal force of the pedal begins to increase) to when the foot is at bottom dead center. This definition was used in an attempt to capture the power-producing phase of cycling. Due to the cyclic nature of cycling, pedal force initiation occurs at every instance when the normal force of the pedal is at a local minimum and begins to rise (solid green line, bottom panel Figure 4.8). Typically, in cycling literature, the downstroke is defined as from TDC (when the pedal is at 12 o'clock) to BDC (when the pedal is at 6 o'clock) (Fonda & Sarabon, 2010). A post hoc analysis of the data revealed that in some participants, a magnitude of force was already being applied to the pedal before TDC (dashed green line, top panel Figure 4.8) and in others, force began to rise in the pedal after TDC. Since the cycling data were being compared to gait, where the beginning of the stance phase was defined as when a force was applied to the foot (and not defined by a kinematic parameter), the beginning of the downstroke of cycling was defined as when the pedal force started to rise at each pedal revolution.

Unlike gait, however, the end of the downstroke was not defined as when there was a cessation of force applied to the foot. Due to the attachment of the foot to the pedals via the clipless shoe mechanism, there was never a time when a force was not acting on the foot (bottom pane, Figure 4.8), which prohibits the same end of signal definition. Thus, for cycling, BDC was used to define the end of the downstroke since that is when it was assumed that the power-producing phase of cycling ends.

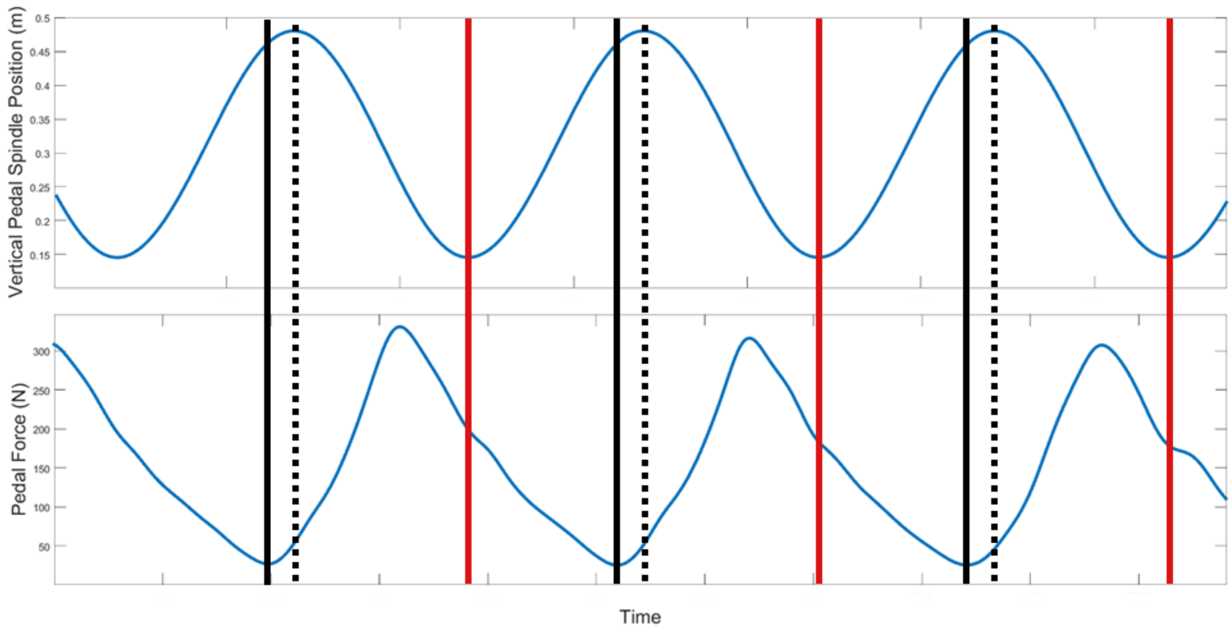


Figure 4.8: Pedal position (top) and normal pedal reaction forces (bottom) during cycling. The dashed black lines indicate where TDC occurred. The solid black lines indicate where pedal loading initiation occurred. The solid red line indicates where BDC occurred.

For all outcome measures, the data were first processed and analyzed over the interval of the stance of each walking/running stride and downstroke of each cycling pedal revolution. For DJS and CCI, the signal was further partitioned into an *initia*l and *terminal* phase, separated by peak KFM. This separation was meant to be representative of loading and unloading of the knee joint and has been previously used to partition signals of DJS (Frigo et al., 1996). The signal for segment coordination and coordination variability was partitioned into thirds, to represent early, mid and late phases (Hafer & Boyer, 2018; Hafer, Silvernail, Hillstrom, & Katherine, 2016). The signals for the IT band impingement measures were assessed over the stance phase of walking and running and for the downstroke of cycling.

4.3.1. Physiological Measurements

The VO_2max was measured using the VMAX Encore Metabolic. The VMAX system measures the percentage of oxygen (O_2) consumed and exhaled during usage to produce a measurement of VO_2 in relative (mL/kg/min) or absolute (L/min) terms. During the VO_2max test on the cycle ergometer, VO_2max was defined as the 10-second averaged VO_2 at which failure occurs or the value at which VO_2 stops increasing with increasing workload.

Maximal heart rate was measured using a 3-lead electrode placement. HRm was defined as the highest 10-second average heart rate achieved during the VO_2max test.

4.3.2 Kinematics

4.3.2.1 Raw Data Processing

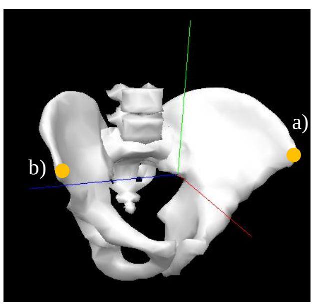
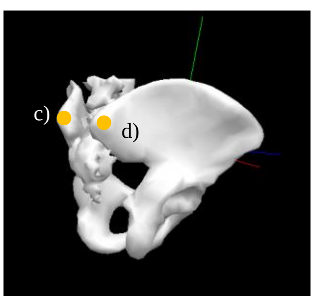
The 3D kinematic data were recorded using an Optotrak camera system (NDI Inc., Waterloo, Ontario, Canada). After collection space calibration and alignment (global coordinate system Y+ up, X+ in the direction of travel, Figure 4.5/Figure 4.6) (Wu & Cavanagh, 1995), rigid bodies with infrared diodes (IREDs) were placed on the lower back to represent the pelvis, and unilaterally on the right thigh, shank and foot of the participant. Bony landmarks, which were digitized during a standing reference trial, assisted in assigning a local coordinate system (LCS) to each segment according to the Visual3D convention (Table 4.2). For the standing reference trial, participants were instructed to stand upright, still and naturally for a duration of 6 seconds. This enabled a V3D model to be applied to the digitized landmarks for further processing. The LCS was assigned such that the Y+ axis was defined as a vector from the origin along the long axis of the segment, the X+ axis was perpendicular and pointing in an anterior direction, with the Z+ axis being the cross product of the X+/Y+ axes, according to the right-hand rule. The dot products of the segment axis vectors were calculated and Euler angles were extracted using the ZXY transformation matrix (Equation 1), using the proximal segment as the reference (fixed) frame. The angles of the Z axis of the right limb was multiplied by -1 to produce a positive knee flexion angles and external moments.

$$R = Z(\alpha)X(\beta)Y(\gamma)$$
$$= \begin{bmatrix} \cos(\alpha)\cos(\gamma) - sin(\alpha)\,sin(\beta)\,sin(\gamma) & \cos(\beta)\sin(\alpha) & \cos(\alpha)\sin(\gamma) - cos(\gamma)\,sin(\alpha)\,sin(\beta) \\ \cos(\gamma)\sin(\alpha) + cos(\alpha)\,sin(\beta)\,sin(\gamma) & \cos(\alpha)\cos(\beta) & \sin(\alpha)\sin(\gamma) - cos(\alpha)\,cos(\gamma)\,sin(\beta) \\ -\,cos(\beta)\,sin(\gamma) & sin(\beta) & \cos(\beta)\cos(\gamma) \end{bmatrix}$$

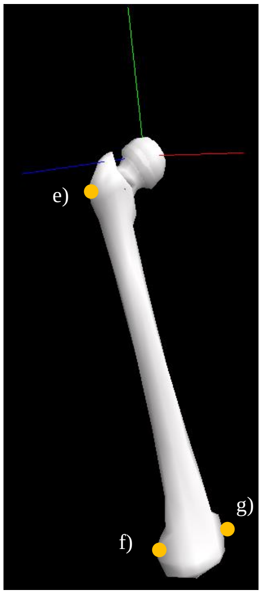
Equation 1: ZXY Rotation Matrix for calculating segment and joint angles.

Where: R is the ZXY rotation matrix, α is the angle (in rad) of the first elemental rotation (Z), β is the angle (in rad) of the second elemental rotation (X) and γ is the angle (in rad) of the third elemental rotation (Y).

Table 4.2: Coordinate System and Landmarks for the Segments of the Body. Local coordinate systems are located at the segment origin, and bony landmarks are indicated by a yellow dot with alphabetical referencing.

<u>Pelvis</u>   	Origin: The origin is the midpoint between the left and right ASIS (a, b) Z-Axis: A line connecting the left ASIS and right ASIS, pointing to the right ASIS X-Axis: A line in the plane defined by the two ASIS and midpoint of the two PSIS, pointing anteriorly and perpendicular to the Z-Axis. Y-Axis: A line orthogonal to the X-Z plane, to create a right-handed coordinate system *Bony Landmarks:* *a. Left Anterior Superior Iliac Spine* *b. Right Anterior Superior Iliac Spine* *c. Left Posterior Superior Iliac Spine* d. *Right Posterior Superior Iliac Spine*

Thigh



Origin:

The origin is the center of the functional hip joint center (Schwartz & Rozumalski, 2005), defined during kinematic calibration

Y-Axis:

The line connecting the midpoint of the lateral and medial femoral epicondyles (f, g) and the origin, pointing proximally

X-Axis

The line perpendicular to the plane defined by the origin, greater trochanter (A) and the midpoint of the lateral and medial femoral epicondyles (f, g), pointing anteriorly

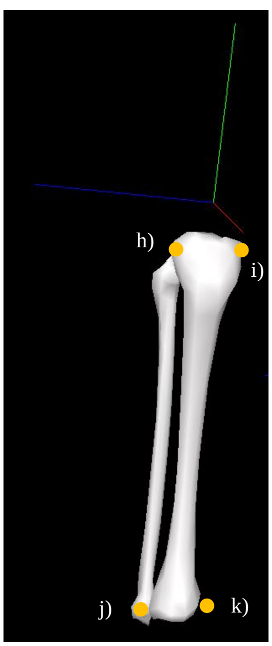
Z-Axis:

The line orthogonal to the X-Y plane, to create a right-handed coordinate system

Bony Landmarks:

e. *Greater Trochanter*
f. *Lateral Femoral Epicondyle*
g. *Medial Femoral Epicondyle*

Shank



Origin:

The origin is the functional knee joint center (Jensen, Lugade, Crenshaw, Miller, & Kaufman, 2016), defined during kinematic calibration

Y-Axis:

The line connecting the midpoint of the lateral and medial malleoli (j, k) and the midpoint of the lateral and medial tibial condyles (h, i), pointing proximally

X-Axis

The line perpendicular to the plane defined by the origin, the lateral tibial condyle (h) and the midpoint of the lateral and medial malleoli (j, k), pointing anteriorly

Z-Axis:

The line orthogonal to the X-Y plane, to create a right-handed coordinate system

Bony Landmarks:

a. *Lateral Tibial Condyle*
b. *Medial Tibial Condyle*
c. *Lateral Malleolus*
d. *Medial Malleolus*

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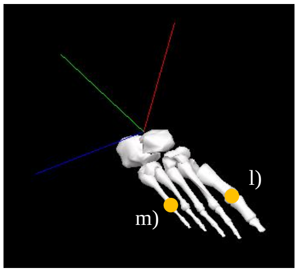
<u>Foot</u>



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Origin:

The origin is the midpoint of the lateral and medial malleoli (j, k)

Y-Axis:

The line connecting the midpoint of the 1st metatarsal head and the 5th metatarsal head (l, m) and the midpoint of the lateral and medial malleoli (j, k), pointing proximally

X-Axis:

The line perpendicular to the plane defined by the origin, the lateral malleolus (j) and the midpoint of the 1st metatarsal head and the 5th metatarsal head (l, m), pointing anteriorly

Z-Axis:

The line orthogonal to the X-Y plane, according to the right-hand rule

Bony Landmarks:

j. *Lateral Malleolus*
k. *Medial Malleolus*
l. *1st Metatarsal Head*
m. *5th Metatarsal Head*

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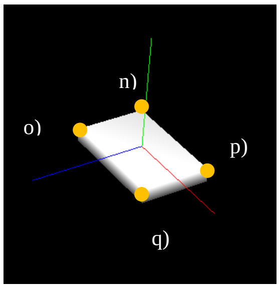
<u>Pedal</u>



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Origin:

The origin is the center of the pedal

Z-Axis:

The line passing through the midpoint of the anterior medial corner and posterior medial corner (n, p) and the midpoint of the anterior lateral corner and posterior lateral corner (o, q), pointing laterally

X-Axis:

The line passing through the midpoint of the anterior medial corner and anterior lateral corner (p, q) and the midpoint of the posterior medial corner and posterior lateral corner (n, o), pointing laterally

Y-Axis:

A line orthogonal to the X-Z plane, according to the right-hand rule

Bony Landmarks:

n. *Posterior Medial Corner*
o. *Posterior Lateral Corner*
p. *Anterior Medial Corner*
q. *Anterior Lateral Corner*

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The kinematic data were recorded at the highest binary multiple frequency possible using 19 IREDs, which was 128Hz. The data were processed using a custom Visual 3D pipeline (v6.01.07, C-Motion, Germantown, MD, USA) and Matlab script (Version R2015a, Mathworks Inc., Natick, MA). All kinematic data were processed using a 4th order Butterworth low-pass filters. Due to the different nature expected from the gait and cycling signals, different cutoff frequencies were applied to the filters for the gait and cycling data. Cycling is a more rhythmic motion with few transient impact signals. Within the literature, a variety of cutoff frequencies and filtering methods have previously been used to process kinematic data for running and cycling. The most common filtering method used is a 4th order Butterworth filter with a cutoff frequency of between 4 Hz to 25 Hz for cycling (Bini & Diefenthaeler, 2010; Bini, Tamborindeguy, & Mota, 2010; Bini, Diefenthaeler, & Mota, 2010; Dingwell et al., 2008; Fang, 2014; Korff, Fletcher, Brown, & Romer, 2011; Marsh, Martin, & Sanderson, 2000; Shen, 2015) and a cutoff frequency of between 4 Hz and 15 Hz for running (Bus, 2003; Fellin, Manal, & Davis, 2010; Hunter, Marshall, & McNair, 2005; Quigley & Richards, 1996; Santos-Concejero et al., 2017). Bini and colleagues (2016) stated that cycling kinematics have a natural frequency of 1.5 Hz while cycling at 90 RPM. It has been recommended that in order to filter cycling kinematic data, a cutoff frequency of at least 5Hz should be used since the natural frequency of cycling is approximately 1.5 Hz (Bini et al., 2016). 6 Hz was used to filter cycling kinematic data (Fang, 2014; Gardner et al., 2015; Shen, 2015). To process gait kinematic data, a cutoff frequency of 12Hz was used. This frequency has been found to contain 95% of the signal content while runners ran overground at 3.35 m/s (Fellin et al., 2010), which is similar to the pace that was expected in the current study. Walking was filtered at the same cutoff frequency as running

(Keller et al., 1996; Li et al., 1999; Seay et al., 2001) since it was assumed all necessary frequencies would be contained within the same frequencies as running.

4.3.2.2 Data Reduction

4.3.2.2.1 Coordination Variability

Segment coordination and coordination variability was calculated for sagittal shank / sagittal foot (Boyer, Freedman Silvernail, & Hamill, 2016) and sagittal thigh / sagittal shank segment couplings (Heiderscheit et al., 2002). These segment couplings have been previously investigated for walking and running and have been thought to be related to increased risk of injury. A modified vector coding technique was used to calculate segment coordination using equation 2 (Sparrow, Donovan, Van Emmerik, & Barry, 1987).

$$\theta_{i,j} = tan^{-1}\left[\frac{y_{i,j+1} - y_{i,j}}{x_{i,j+1} - x_{i,j}}\right]$$

Equation 2: Segment coordination coupling angle formula.

Where $0 \leq \Theta \leq 360°$, j is the percent of stance/downstroke for the ith repetition. And x and y are the distal and proximal segment angles for each coupling, respectively, as measured in the global coordinate system. This modified vector coding approach calculates the coupling angle between 2 consecutive time points on an angle-angle plot of two segments, with respect to the right horizontal (Figure 4.9). The result is a plot of coupling angle at each percent of stance/downstroke (Figure 4.10). These coupling angles represent the segment coordination and describe how segments move with respect to one another (i.e. anti-phase, in-phase, proximal phase or distal phase). Coordination variability is calculated by finding the intra-individual standard deviation of the coupling angle for each percent of the data between strides or pedal revolutions (Figure 4.11).

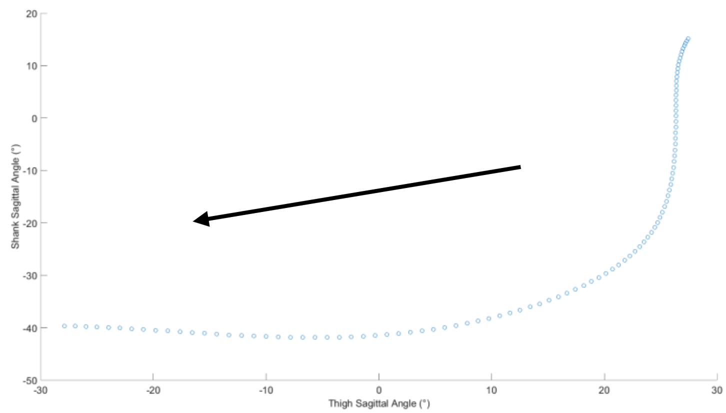

Figure 4.9: Sagittal thigh/sagittal shank segment coupling (angle-angle plot) for the stance phase of running for one trial of a single participant. The black arrow indicates the direction of progression.

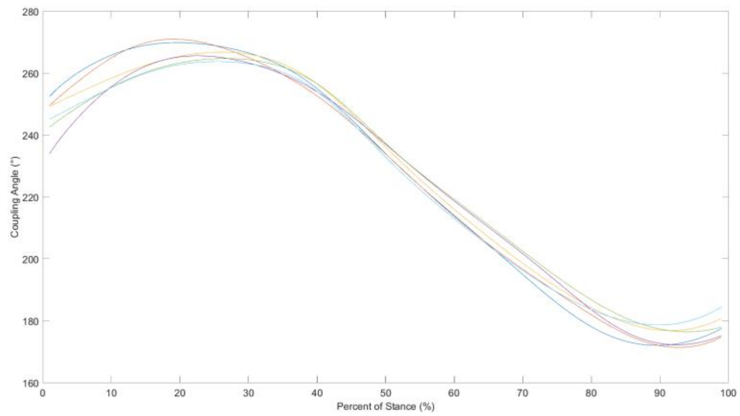

Figure 4.10: Plot of coupling angles for all trials from a single participant during overground running at each percent of the stance phase of the sagittal thigh/sagittal shank segment coupling.

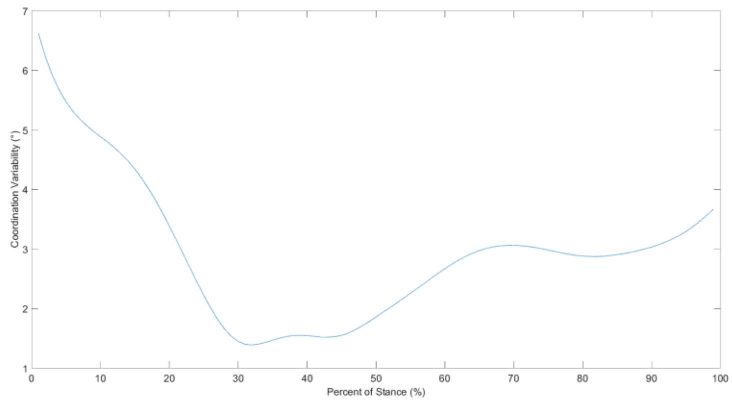

Figure 4.11: Intra-individual coordination variability for all running trials during overground running at each percent of the stance phase of the sagittal thigh/sagittal shank segment coupling for one participant.

For walking and running, each trial consisted of one stance phase available for analysis. For each participant, the coordination variability was calculated as the intra-individual standard deviation of the coupling angles between strides. The grand mean coordination variability for walking and running were calculated by averaging the coordination variability across participants.

As for cycling, each 30-second trial consisted of 30-40 pedal revolutions available for analysis. To account for the greater number of pedal revolutions compared to gait strides, coordination variability was first found for each trial. For each cycling trial, the intra-trial coordination variability between pedal revolutions within each trial was found and then this trial coordination variability was averaged across trials to find the coordination variability for each participant. Essentially, coordination variability was reduced to one value per cycling trial to align with the 6 walking and running trials before being averaged for each participant. The grand mean coordination variability for cycling was calculated by averaging the coordination variability across participants.

The grand mean coordination variability curves were then partitioned into thirds to represent early, mid and late phases (Hafer & Boyer, 2018; Hafer et al., 2016).

4.3.2.2.2 IT Band Impingement Measures

The other kinematic outcome investigated was IT band impingement. As previously stated in section 3.2.3, the ITBIZ is when the IT band is in contact with the lateral femoral condyle between 20° and 30° of knee flexion. For the walking and running trials, the amount of time that the knee was in the ITBIZ for each stance phase was calculated and averaged across all trials for each participant (Figure 4.12/Figure 4.13). For the cycling trials, the amount of time that the knee was in the ITBIZ for each pedal revolution was calculated and the mean ITBIZ duration per pedal revolution was found (Figure 4.14). Collectively, ITBIZ duration for each stance phase and pedal revolution are referred to as $ITBIZ(t)_{per\_rep}$. Only the stance phase of walking and running were analyzed due to collection space constraints: the subsequent heel strike often occurred outside the collection volume. Further, since the stance phase is the only part of the walking or running stride where force is being applied to the foot, the stance phase was thought to be most detrimental and thus was the focus in the current study. Thus, the downstroke of the cycling pedal revolution was compared to the stance phase of walking and running. The kinematic collection frequency for the collection data was 128Hz. The number of data frames with the knee flexed between 20° and 30°, inclusively, was found and divided by 128 to yield the $ITBIZ(t)_{per\_rep}$, in seconds.

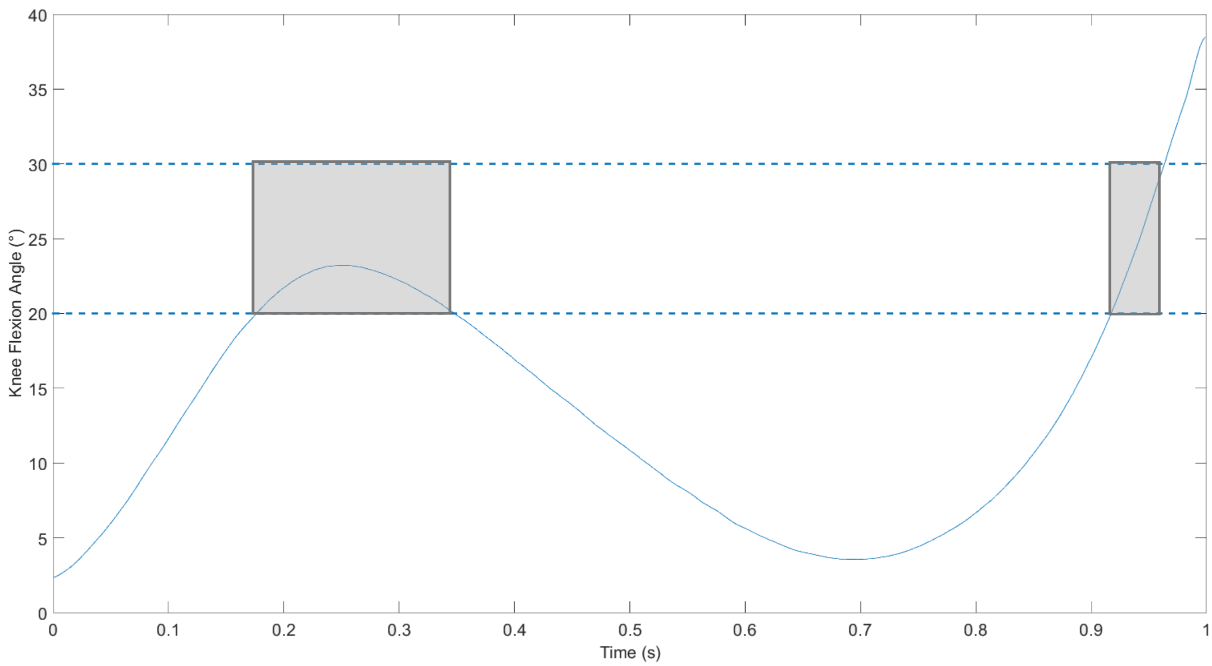

Figure 4.12: Knee flexion angles for one stance phase of walking for a single participant. The shaded area between the blue bars represents the interval in which the knee was flexed in the ITBIZ.

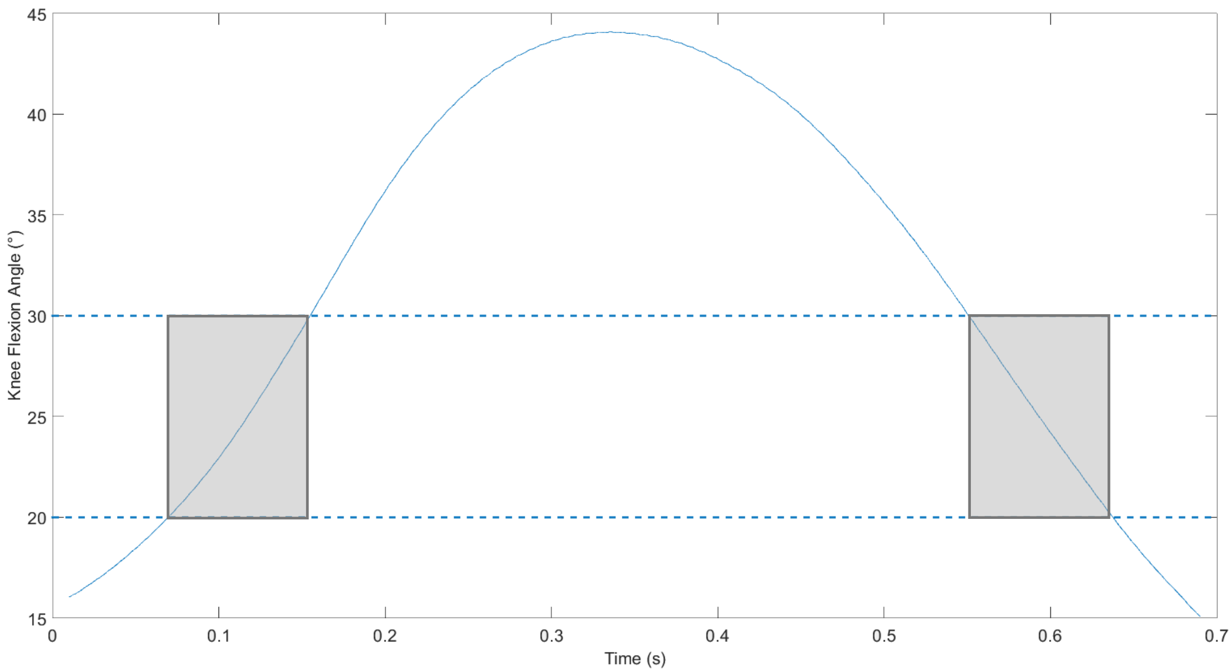

Figure 4.13: Knee flexion angles for one stance phase of running for a single participant. The shaded area between the blue bars represents the interval in which the knee was flexed in the ITBIZ.

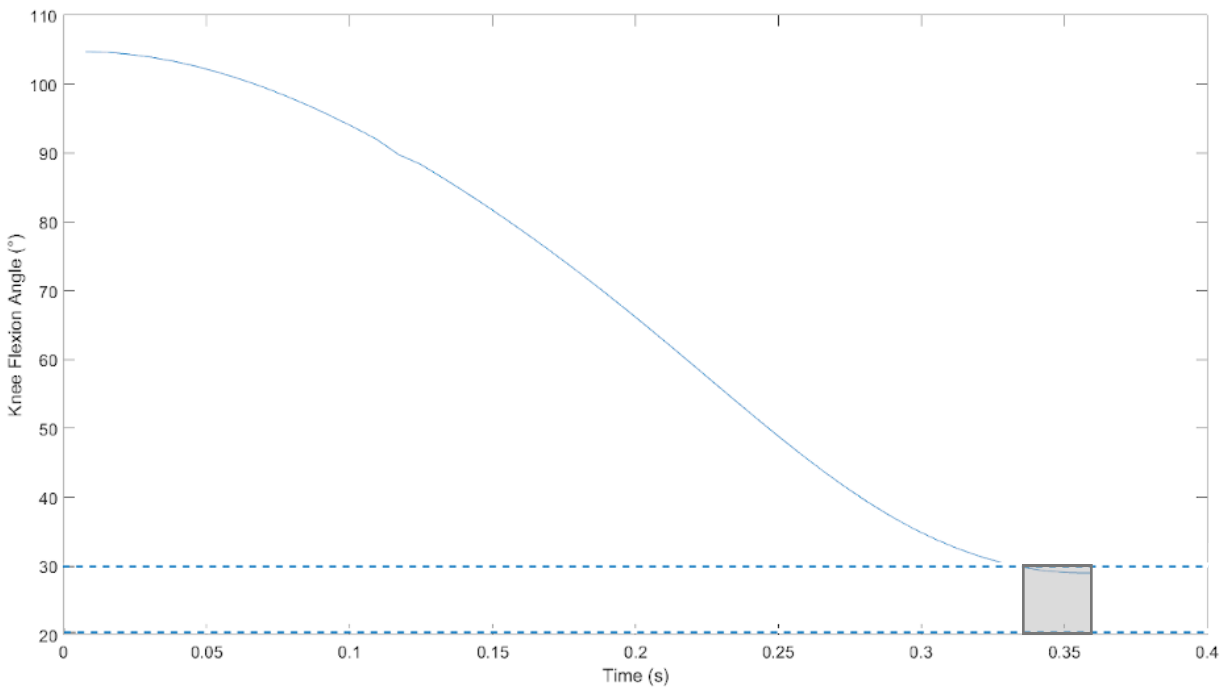

Figure 4.14: Knee flexion angles for one downstroke during cycling. The shaded area between the blue bars represents the interval in which the knee was flexed in the ITBIZ.

Typically, a static bike fit requires the bike to be set-up such that the participant's leg is flexed between 20° and 30° at BDC. As stated in section 4.2.1, in order to mitigate biasing the data, the bike was set-up according to a protocol established by Bini and colleagues (2016). While the participant was seated on the bike with feet on the pedals placed at the 3 o'clock position, the saddle was adjusted to elicit a knee angle of 60° ± 5°. This was done in an attempt to avoid forcing the participant to cycle with their knee flexed in the ITBIZ. As a result, it was anticipated that participants may not always achieve a knee flexion of 20° and 30° during a pedal revolution. Some participants may cycle with their knee angle entering the ITBIZ for every pedal revolution and some may cycle with their knee angle never entering the ITBIZ, except for a few outlier cases. The participants were thus separated into two groups: ITBIZ and non-ITBIZ cyclists. The criteria for ITBIZ cyclists were for at least three of the six collected 30-second trials, at least half of the pedal revolutions contained ITBIZ events.

In order to compare ITBIZ duration between walking, running and cycling, three comparisons were made. The purpose of these comparisons was twofold: to assess ITBIZ duration on a more tangible, meaningful scale and to compare data in the current study to results previously reported.

The first comparison extrapolated the $ITBIZ(t)_{per\_rep}$ to represent a 60-minute activity. When exercising, 60-minutes is fairly representative of a workout duration (i.e. spin class or ~10km run). The $ITBIZ(t)_{per\_rep}$ for each activity was multiplied by the average cadence for each participant and by 60 minutes. The result is the ITBIZ duration experienced per hour and will be represented as $ITBIZ_{60}$ (units are seconds). For this analysis, $ITBIZ_{60}$ was compared over stance phase of walking and running and the downstroke of cycling.

The second comparison will attempt to verify data presented by Farrell et al. (2003) (see section 3.2.3.2, where they assumed that running distance to cycling distance ratio of 4:1 was a good approximation of an equivalent workout). Values for this comparison will be referred to as $ITBIZ(t)_{eq\_workout}$. It was found that ITBIZ duration was 250s and 330s for cycling and running, respectively, when comparing a 1.25-hour bike ride to a 10km run. In order to support these findings, the same calculations by Farrell et al. (2003) was performed, substituting the assumptions made (cadence, $ITBIZ(t)_{per\_rep}$, and run pace) for measured data. Since cycling distance cannot be accurately calculated (due to factors such as rolling resistance, drag, etc.), the assumption of a 1.25-hour ride was used in the current calculation (which was assumed by Farrell et al. to be representative of a 40km ride). For running, the participants run paces were used to approximate their 10km running time. The $ITBIZ(t)_{per\_rep}$ for running were multiplied by their cadence and 10km time (calculated based on their run pace), to produce the ITBIZ duration over a 10km run. For cycling, the $ITBIZ(t)_{per\_rep}$ was multiplied by their cadence and 1.25 hours to produce the ITBIZ duration over a 40km ride. Since there was no equivalent for walking, it was excluded from this analysis. The stance phase of running was compared to the full pedal revolution of cycling, to maintain consistency with the methods for the previous study.

The final comparison made compared ITBIZ duration over an equivalent cumulative load and will be referred to as $ITBIZ(t)_{cumul\_load}$. This will be further described in section 4.3.3.2.

In summary, three comparisons will be measured for ITBIZ duration. The first will be a comparison of a 60-minute activity ($ITBIZ_{60}$), the second will compare the modalities similar to Farrell et al., (2003) ($ITBIZ(t)_{eq\_workout}$) and the third will compare modalities up to an equivalent cumulative load ($ITBIZ(t)_{cumul\_load}$).

4.3.3. Kinetics

4.3.3.1 Raw Data Processing

The force data were collected from AMTI OR6-7 force plates (AMTI, Watertown, MA, USA) and a 2D force transducer pedal (Figure 4.15) (Novatech, UK) for gait and cycling trials, respectively. The force transducer pedal was equipped with Shimano 105 clipless pedals (Shimano, Inc., Sakai, Japan) in order to allow the participants to interface with the cycle ergometer similarly to their own personal bike. Shimano SPD-SL zero-degree float cleats (Shimano, Inc., Sakai, Japan) were installed on the bottom of the participant's cycling shoes to limit movement between the foot and pedal. The contralateral pedal was a dummy pedal with the same dimensions, mass and modified clipless pedal system as the instrumented pedal in order to allow for symmetry between legs. All force plate data were pre-processed through the amplifiers with a built in 2nd order low-pass critically damped filter with a cutoff frequency of 1050 Hz. All kinetic data were collected at a sampling frequency of 2048 Hz.

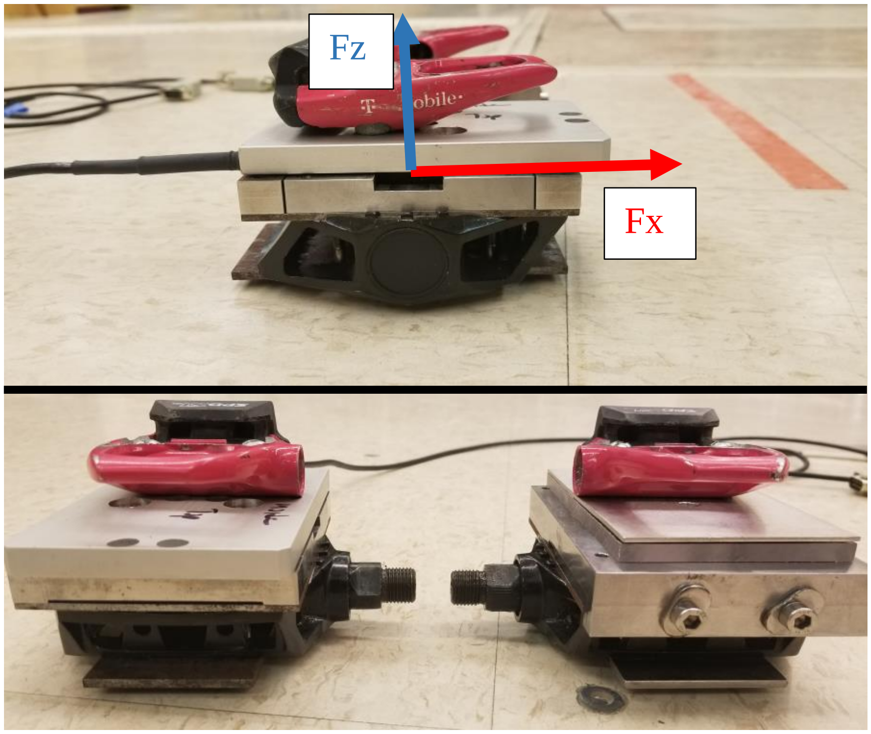

Figure 4.15: The experimental set-up for the instrumented pedal. A Shimano clipless pedal system was attached securely to the force transducer pedal. A dummy pedal was fabricated to match the stack height, positioning of the clipless pedal with respect to the spindle and mass. Top: Lateral view of the instrumented pedal with coordinate system. Bottom: Frontal view comparing the height of each pedal.

Similar to kinematics, to reflect the differences between running and cycling signals, the kinetics were processed using different cutoff frequencies. Both sets of data were filtered through a 4th order Butterworth low-pass filter. Common cutoff frequencies are between 15 Hz and 96 Hz for running (Blackmore, Willy, & Creaby, 2016; Gatti et al., 2017; Hunter et al., 2005; Keller, Weisberger, Ray, & Hasan, 1996; Quigley & Richards, 1996; Santos-Concejero et al., 2017) and between 6 Hz and 25 Hz for cycling (Bini & Diefenthaeler, 2010; Bini et al., 2010; Broker & Gregor, 1990; Fang, 2014; Korff et al., 2011; Marsh et al., 2000; Mornieux et al., 2008; Shen, 2015).

A cutoff frequency of 20 Hz was used for the gait kinetic data and has been previously used by Gatti and colleagues (2017). A cutoff frequency of 10 Hz was used for the cycling kinetic data. This cutoff frequency had been previously used by multiple other authors assessing cycling kinetics (Bini & Diefenthaeler, 2010; Bini et al., 2010; Broker & Gregor, 1990; Marsh et al., 2000) and was also used by Gatti et al. (2017).

Due to the limitations of the 2D force transducer pedal, center of pressure could not be calculated at each time point. Thus, in order to apply the force signal to the model in Visual3D, a center of pressure location was estimated for each participant. During cycling digitization, an F-Scan 3000E Tekscan sensor (Tekscan, Inc., Boston, Massachusetts) was placed inside of the participant's cycling shoe (Figure 4.16).

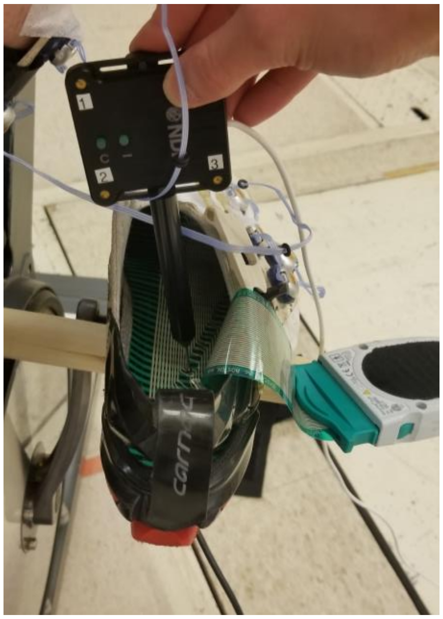

Figure 4.16: F-Scan 3000E sensor inside of a cycling shoe. Three points were digitized within the shoe and on the Tekscan sensor simultaneously in order to map a center of pressure location to the foot.

Three points (A, B and C) were digitized on the Tekscan F-Scan 3000E in the shoe in order to map the Tekscan sensor to the rigid body cluster which was attached firmly to the cycling shoe (Figure 4.17a). The participant then performed 5-10 pedal revolutions in order to identify where center of pressure occurred on the Tekscan sensor (Figure 4.17b). Center of pressure was defined as the center of pressure during the 5-10 pedal revolutions in the frame that had the largest summed forces registered. The center of pressure from the Tekscan sensor was then mapped into the foot coordinate system (Figure 4.17c). In Visual3D, a landmark for center of pressure was created by mapping the center of pressure with respect to points A, B and C. A vector from point B to point A was defined (length 1), in which the endpoint was the orthogonal intersection of the vector from the center of pressure, in the plane defined by points A, B and C (length 2). Force signals from the pedal were then transformed into the foot coordinate system and applied to the foot at the center of pressure (Figure 4.18).

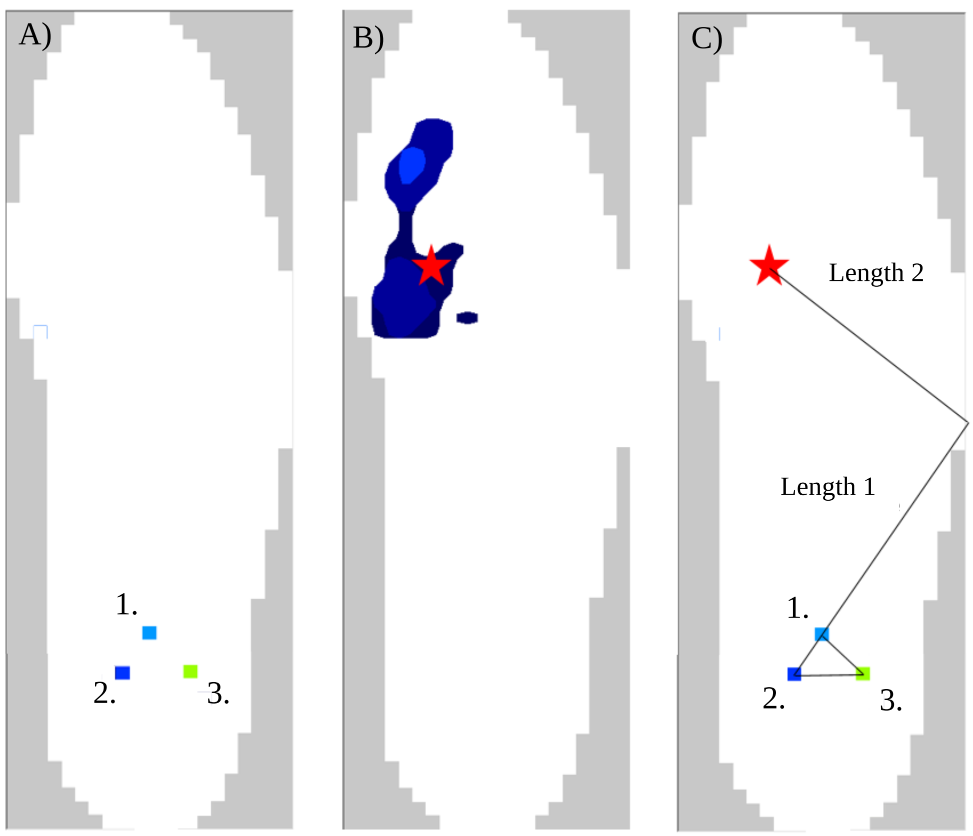

Figure 4.17: Tekscan sensor foot center of pressure location. A) Three reference points recorded on the Tekscan sensor that were also digitized with respect to the foot cluster that was attached to the shoe/pedal interface. B) Tekscan recording of the foot during the frame with the largest sum of force. The red star indicates where the center of pressure was calculated. C) Deduction of center of pressure location with respect to the three reference points.

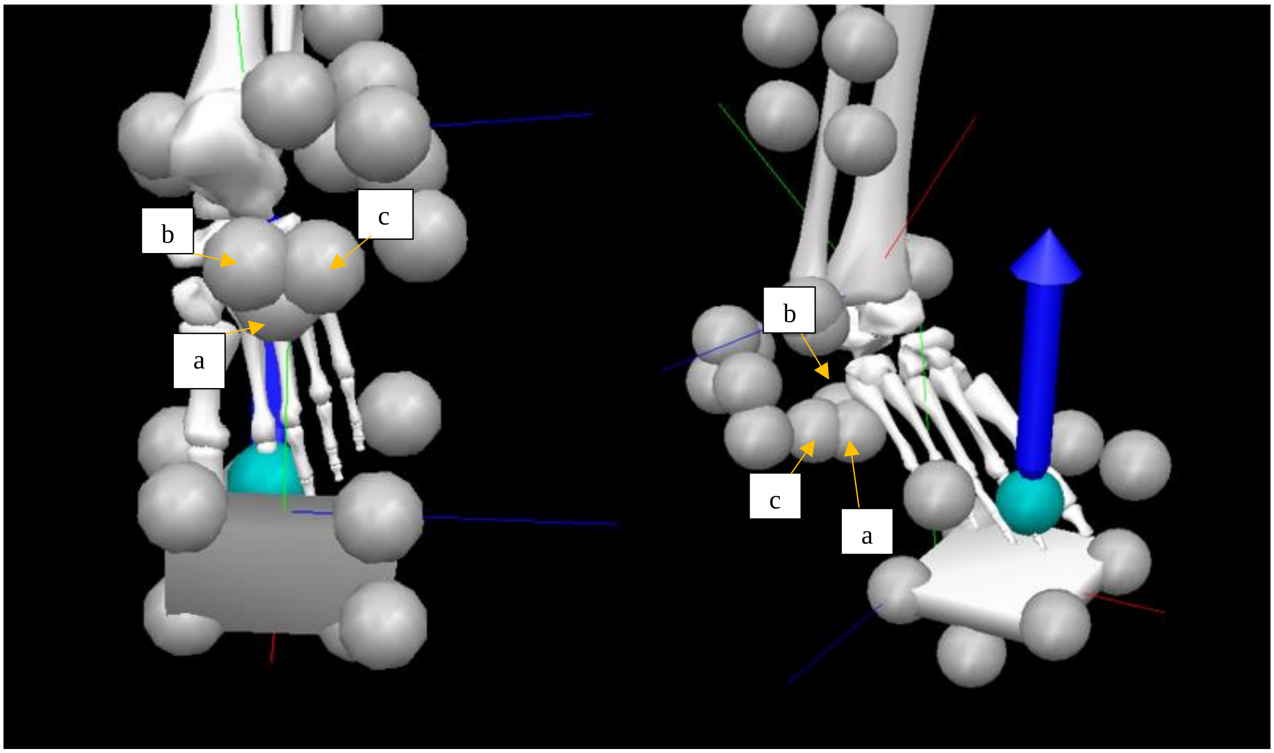

Figure 4.18: Force signals applied to the foot model at the center of pressure location during cycling. a, b, and c are the three reference points as recorded with Tekscan and digitized with respect to the foot cluster. The blue sphere is the center of pressure location, as depicted in Figure 4.17c.

4.3.3.2 Data Reduction

4.3.3.2.1 Dynamic Joint Stiffness

Dynamic joint stiffness of the knee, previously described in section 3.2.1, is the slope of the regression line of a plot of knee flexion moment versus knee flexion angle. For each activity, the knee flexion moment and knee flexion angle were calculated for stance and downstroke and plotted against each other for each stride/pedal revolution.

Due to the cyclic and non-linear nature of moment/angle plots for gait, analysis of dynamic joint stiffness is often partitioned into different quasi-linear phases bounded by inflection points in the plots (Frigo et al., 1996). In the current study, the stance and downstroke were partitioned in to two phases (the initial and terminal phases). The initial phase for running and cycling was defined from heel strike/initiation of pedal loading, respectively, to the maximum knee flexion moment (Figure 4.19). The terminal phase for running and cycling was defined from the maximum knee flexion moment to toe off/BDC, respectively (Figure 4.19). In walking, the DJS is more complex from heel strike to toe off, due to the double stance phase (Figure 4.19a) and has often been partitioned in up to 4 phases (Frigo et al., 1996). In order to reduce the data to be able to compare it to the two phases previously described for running and cycling, the DJS of walking was analyzed during single leg stance only. The terminal phase was defined as maximum knee flexion moment to minimum knee flexion angle, following the inflection points of the DJS plot. It was important to isolate the DJS terminal phase to single leg stance only because once the swing leg initiates heel strike, it has been assumed to have an influence of the stiffness of the contralateral leg. This is evident by the near horizontal portion at the right of the walking DJS figure (first column, Figure 4.19a). The data extending from the red star to the right side of the figure represent the double support phase of walking, where the slope of the DJS approaches zero. This portion of the data was not included in any analyses.

A linear regression line was fit to the plotted data from each of the trials for the entire stance/downstroke, the initial phase and the terminal phase (Figure 4.19). The mean slope of the regression line, in Nm/kg/deg, for each activity for each participant was calculated and used for statistical analysis.

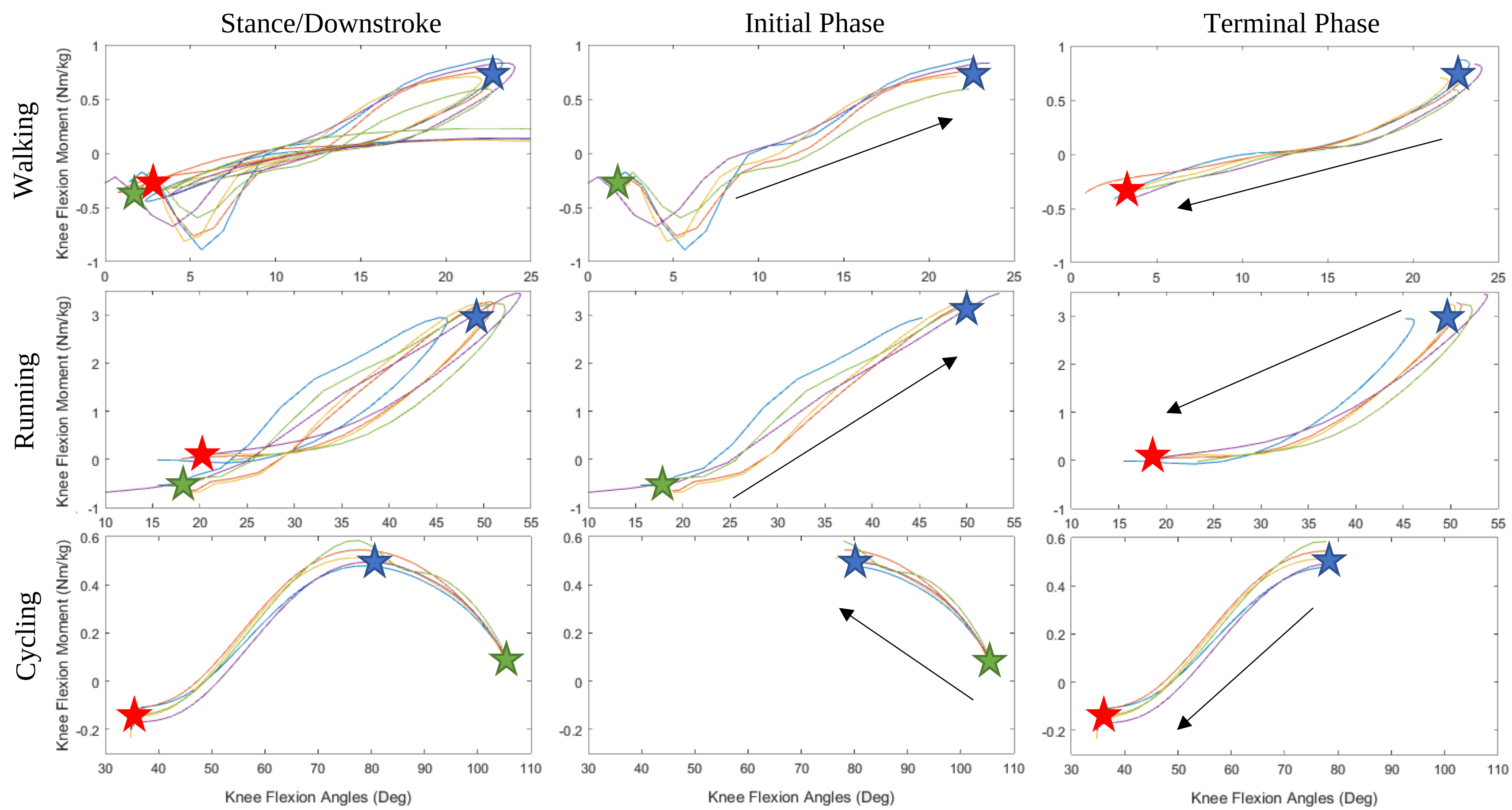

Figure 4.19: DJS plots for all trials of a single participant for walking, running and cycling for the Stance/Downstroke, which was then partitioned into: the loading phase and the terminal phase. The black arrows indicate the direction of progression. The green star represents heel strike (for walking and running) or initiation of pedal force (for cycling). The blue represents max knee flexion moment (the end of the loading phase and the start of the terminal phase). The red star represents contralateral heel strike (for walking), toe off (for running) or BDC (for cycling).

4.3.3.2.2 IT Band Impingement Measures

The cumulative vertical ground reaction force loading was calculated for a representative 15-minute run (Gatti et al., 2017). The value of this cumulative load was used to find an equivalent cycling duration which elicits the same cumulative load for the vertical pedal reaction force. The time for cycling to reach an equivalent cumulative load as a 15-minute run was expected to be approximately 45-minutes (Gatti et al., 2017). For running, the $ITBIZ(t)_{per\_rep}$ was multiplied by the run cadence and by 15 minutes to produce the ITBIZ duration over a 15-minute run. For walking and cycling, cumulative loads were calculated for each participant and the $ITBIZ(t)_{per\_rep}$ were multiplied by the average cadence and multiplied by a time duration that would produce equivalent cumulative loads to the 15-minute run.

The equations used to calculate the cumulative loading times were adopted from Gatti et al. (2017) (Equations 3 & 4).

$$CL = R * I * t_{run}$$

Equation 3: Cumulative loading equation.

Where: CL is the cumulative load for the run, in newtons. R is the number of running repetitions (strides) per minute, I is the integral of the vertical ground reaction force per repetition (N), and t_{run} is the duration of the run, in minutes.

$$t_{activity} = \frac{CL}{R*I}$$

Equation 4: Cumulative loading equation solving for time.

Where: $t_{activity}$ is the walking or cycling duration equivalent to the 15-minute run, CL is the cumulative load of the 15-minute run from Equation 3, R is the number of repetitions (walking steps or pedal revolutions) per minute and I is the integral of the vertical reaction forces (during walking or cycling). In order to get a true measure of cumulative vertical reaction forces for cycling, this value was taken from the entire pedal revolution, and not just the downstroke.

Using $t_{activity}$, $ITBIZ(t)_{cumul\_load}$ was calculated using Equation 5 to determine the duration with the knee flexed in the ITBIZ during walking, running and cycling of a cumulative load.

$$ITBIZ(t)_{cumul\_load} = t_{activity} * \ cadence \ * ITBIZ(t)_{per_{rep}}$$

Equation 5: ITBIZ duration equation for comparing cumulative loads.

Where: $ITBIZ(t)_{cumul\_load}$ is the duration with the knee flexed in the ITBIZ during an activity with a cumulative load equal to that of a 15-minute run, $t_{activity}$ is the walking or cycling duration equivalent to the 15-minute run, cadence is the number of repetitions for each respective activity and $ITBIZ(t)_{per\_rep}$ is the duration with the knee flexed in the ITBIZ during each stance phase of walking and running and downstroke of cycling.

4.3.4. Electromyography

4.3.4.1 Raw Data Processing

EMG was collected for seven muscles, unilaterally on the right side of the body. The muscles being collected were VL, VM, LG, MG, BF, ST and GM (Figure 4.3). The purpose of collecting EMG was so that co-contraction could be calculated and to assess tensioning of the IT band.

EMG was recorded using the Cometa Wave Plus EMG system (Cometa, Cisiano, Italy) at a sampling frequency of 2048 Hz. The EMG sensors had a fixed signal amplifier with a magnitude of 1000 and filter the raw analog signal with a bandpass filter of 10-500Hz. Data had its bias removed then was full wave rectified and low-pass filtered with a 4th order 6Hz Butterworth filter (Hubley-Kozey, Deluzio, Landry, McNutt, & Stanish, 2006).

For all EMG locations, two Ag-AgCl surface electrodes (Ambu Blue Sensor N, Denmark) with an inter-electrode distance of 2 cm were placed parallel to the fibres of the muscle, also according to SENIAM guidelines. The EMG placement for the VL muscle was 2/3 of the distance between the anterior superior iliac spine (ASIS) and the lateral edge of the patella or over the bulk of the muscle bulge during resisted knee extension. VM EMG was placed 80% of the distance between the ASIS and anterior joint space of the medial knee or over the bulk of the muscle bulge during resisted knee extension. LG EMG was placed 1/3 of the distance between the head of the fibula and the heel. MG EMG was placed 1/3 of the distance between the most prominent bulge of the muscle belly. BF EMG was placed halfway between the ischial tuberosity and the lateral epicondyle of the tibia or over the bulk of the muscle bulge during resisted knee flexion. ST EMG was placed halfway between the ischial tuberosity and the medial epicondyle of the tibia or over the bulk of the muscle bulge during resisted knee flexion. GM EMG was placed halfway between the sacrum and the greater trochanter or over the bulk of the muscle bulge during resisted hip extension in a prone position.

Normalization is the process in which EMG data are adjusted to be representative of a meaningful reference value. Typically, this is performed via a maximal voluntary isometric contraction (MVC). However, normalizing to MVC's for dynamic activities such as running and cycling has been found to be inaccurate and not always reliable. In a review by Ball & Scurr (2013), the need for EMG normalization of running and cycling during sprint activities was emphasized. They concluded that typical normalization techniques such as MVC may be inappropriate for high-velocity muscle contractions such as those presented when running and cycling. These exercise modalities have unique muscle activation characteristics (such as muscle length changes), which are not accurately reflected by the specificity of the conditions of an MVC (Sinclair et al., 2015). Though no standardized method of normalization has been agreed upon for cycling or running specifically, it is universally accepted that the traditional MVC method is inadequate (Albertus-Kajee, Tucker, Derman, Lamberts, & Lambert, 2011; Ball & Scurr, 2010, 2013; Suydam, Manal, & Buchanan, 2016). Further, the method of EMG normalization has been found to affect the interpretation of the EMG signal (Sinclair, Brooks, Edmundson, & Hobbs, 2012). MVC methods have been found to result in low levels of reliability during cycling (Sinclair et al., 2015).

In a running study, Kyrolainen et al. (2005) found much greater max EMG values during sprint running compared to MVC values. Kyrolainen et al. (2005) found greater max EMG values for BF, GM, VL, and gastrocnemius during sprint running. In cycling, Rouffet & Hautier (2008) also found greater max EMG values during sprint cycling compared to MVC values for GM, soleus, BF and VL. Albertus-Kajee and colleagues (2010 & 2011) analyzed EMG normalization techniques in cycling and running in separate studies. Both studies concluded that dynamic measures of maximum EMG should be taken for these activities and included VL, VM, rectus femoris, BF, MG and LG. Further, in both studies, the sprint method of normalization was recommended since it was sensitive to exercise intensity changes, resulted in good reproducibility and had low intra-subject variation compared to tradition MVC methods, specifically for studies taking place on one day and where absolute EMG amplitude is desired. In the current study a 30m maximal sprint effort on a runway was performed to quantify the maximum EMG signal for all muscles. This method has been shown to produce the greatest EMG normalization values, however, with lower repeatability compared to traditional MVC methods or during sprint cycling (Chuang & Acker, 2019). Three sprint run normalization trials were performed and the greatest EMG value for each muscle across trials was defined as the normalization value to be used when normalizing EMG data according to the following equation:

$$Normalized\ EMG\ (i) = \frac{EMG\ Signal\ of\ Interest\ (i)}{Max\ EMG\ (V)}$$

Equation 6: Formula for EMG normalization.

Where i is the ith datum point of the signal of interest for a specific muscle, and max EMG is the maximum voltage obtained for the specific muscle during the sprint run.

4.3.4.2 Data Reduction

4.3.4.2.1 Co-Contraction Indices

As with all other signals, the EMG signals for each trial were cropped to only include the stance of gait and the downstroke of cycling as the signal of interest.

EMG co-contraction for VL-LG, VM-MG, VL-BF and VM-ST muscle pairs were calculated using the equation 7:

$$\sum_{n}^{i=1} \mathrm{CCI(n)} = \left(\frac{LM(n)}{MM(n)}\right)\left(LM(n) + MM(n)\right)$$

Equation 7: Formula for co-contraction indices, adopted from Rudolph et al. (2001).

Where LM is the EMG value of the less activated muscle and MM is the EMG value of the more activated muscle. The co-contraction index (CCI) is unitless and represents a magnitude for the amount of co-contraction between antagonistic muscles. The CCI was measured at every 1% and summed to provide a CCI over the stance/downstroke (Figure 4.20). The CCI was also calculated for the initial phase and terminal phase so that its relationship to the associated DJS value could be explored. As with DJS, the initial and terminal phases were separated by the peak KFM.

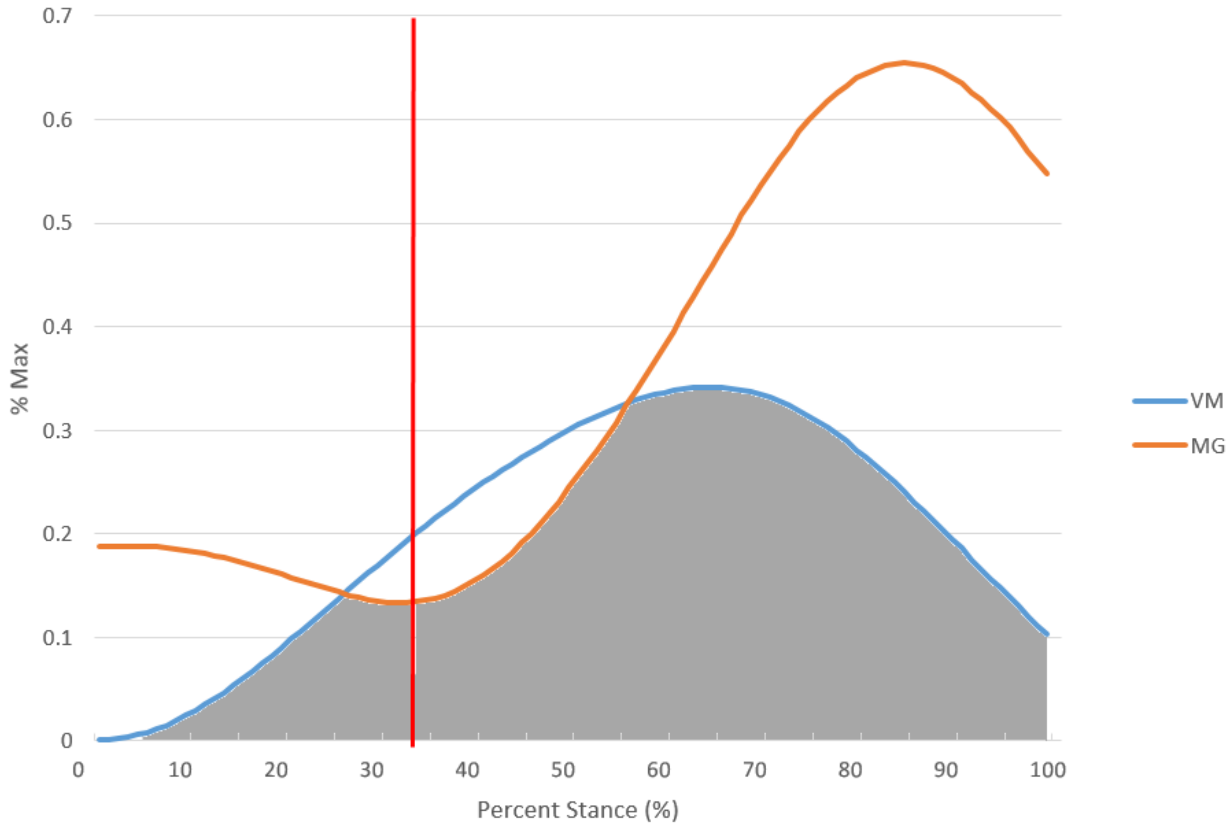

Figure 4.20: Plot of the CCI for the VMMG muscle coupling for one stride of running for one participant, from heel strike to toe off. The shaded region represents the signal common to both muscles, represented by the CCI. The red line is coincident with the peak knee flexion moment which was used to separate the signal for analyses.

4.3.4.2.2 IT Band EMG Exposure

The EMG of GM, the angular velocity of the knee joint and the ITBIZ(t)$_{per\_rep}$ were used to establish a novel exposure metric to compare potential injury risk of the IT band. As previously described, the IT band rubs against the lateral femoral epicondyle when the knee is flexed between 20° and 30°. However, when GM and TFL are activated, their muscle bellies shorten, creating increased tension on the IT band due to its attachment. It is possible that this additional tensioning of the IT band when it is in the impingement zone may be a possible explanation for IT band injury. In the current analysis, only the EMG of GM was considered because it is a much larger muscle compared to TFL and it is anticipated that it would have more influence over the tensioning of the IT band. Thus, the IT band EMG exposure (ITBEX) for a single trial is defined by Equation 8 for walking, running and cycling:

$$ITBEX = cadence \times \sum_{i=1}^{n} [\overline{\|\omega\|}_i * iEMG_{GM(i)}]$$

Equation 8: IT band EMG exposure formula for a given trial.

Where: n is the number of times the knee was flexed in the ITBIZ during each trial (this was typically three times for walking, twice for running and once for cycling), $\overline{\|\omega\|}_i$ is the mean of the absolute instantaneous angular velocity (rad/s) of knee flexion across ITBIZ event i, $iEMG_{GM(i)}$ is the integrated EMG of GM (%MVC·s/rep) during ITBIZ event i, and cadence is the repetitions per minute measured for the specific activity. The ITBEX value has a unit of %MVC/min. For each repetition of each activity, each time the knee was flexed in the ITBIZ, the knee flexion angle was differentiated using 2-point differentiation to produce the angular velocity. The absolute value of the angular knee velocity was taken since this analysis was performed irrespective of whether the knee was flexing or extending. ITBEX was averaged across trials. In this analysis, only the stance phase of walking and running and the downstroke of cycling was included.

4.4 Statistical Analysis

Statistical analysis was performed using a custom SPSS syntax (IBM Corp., Armonk, NY) and a custom Matlab code. Student's paired t-tests were performed to determine whether there was a significant difference between the participant's predicted HRm and actual HRm.

Significance was determined using an alpha level of 0.05. One-way ANOVAs were performed to determine whether there was an effect of activity on cadence, peak external foot reaction force, knee flexion moment, knee range of motion, maximum knee flexion angle and minimum knee flexion angle. A Bonferroni correction was used to correct for multiple pairwise comparisons. Significance was determined using an alpha level of 0.05.

Hypotheses 1-4 were tested using 1-way repeated measures ANOVAs to determine if there was difference in DJS, co-contraction indices, coordination variability and IT band impingement measures between activities, respectively. A Bonferroni correction was used to correct for multiple comparisons and were specific for each dependent variable. The alpha level was divided by *n*, where *n* was the number of datasets analyzed for each outcome variable. Thus, the alpha level for co-contraction indices was divided by 4 (to account for the 4 CCIs: VLLG, VMMG, VLBF and VMST muscle groupings), the alpha level for coordination variability was divided by 2 (to account for 2 coordination variability outcome measures sagittal foot/sagittal shank and sagittal thigh sagittal shank segment couplings) and alpha level for IT band impingement measures was divided by 6 (to account for the six comparisons made using $ITBIZ(t)_{per\_rep}$ as a factor).

Pearson Product Moment Correlations were calculated to find relationships between DJS and the CCI for each muscle pairing, for each phase. Significant correlations ($p < 0.05$) were identified as high correlation if $r > 0.7$, moderate correlation if $0.3 < r < 0.7$ and low correlation if $r < 0.3$ (Ratner, 2009). Student's paired t-tests were performed to compare both the number of ITBIZ events and $ITBIZ(t)_{eq\_workout}$ over an equivalent workout between running and cycling. Student's paired t-tests were performed since a criterion for walking had not been previously defined and the purpose of this assessment was to compare against previous work. Significance was determined using an alpha level of 0.05.

Prior to performing statistical analysis, the data were checked for normality, common variance, mean of zero and independence. In cases where Mauchly's test indicated that sphericity was violated, Greenhouse-Geisser correction was applied to the p-values.

Chapter 5 : Results

All 15 participants completed the protocol as planned. Running and cycling trials were all performed at their prescribed HRm. Two participants data had to be excluded due issues with collection files. The participants' physiological measurements (VO_2max results, predicted and actual HRM) are summarized in Table 5.1. Participants had a significantly lower actual HRm, compared to the predicted HRm ($t(14) = 2.815$, $p = 0.014$) by an average of 6.2 BPM.

Table 5.1: Physiological outcome measures.

Outcome Measure	Value	Maximum	Minimum
VO_2max (mL/kg/min)	61.5 (8.1)	74.2	41.4
VO_2max (L/min)	4.5 (0.7)	5.7	3.0
Predicted HRm (BPM)	194.9 (4.7)	202	185
Actual HRm (BPM)	188.7 (9.3)	204	166

Note: Values are presented as mean (SD), where applicable.

The participants kinematic and kinetic characteristics are summarized in Table 5.2. There was a significant effect of activity on cadence ($F_{2,24} = 88.099$, $p < 0.0001$), peak external foot reaction force ($F_{2,24} = 317.713$, $p < 0.0001$) and peak knee flexion moment ($F_{2,24} = 235.736$, $p < 0.0001$). Post-hoc pairwise comparisons revealed the following: Walking had a lower cadence compared to running and cycling (all $p < 0.0001$), however running and cycling were not different from each other ($p = 0.098$). Running had a larger peak external foot reaction force than walking ($p < 0.0001$), which was larger than cycling ($p < 0.0001$). Running had a larger peak knee flexion moment than walking ($p < 0.0001$), which was larger than cycling ($p = 0.004$).

Table 5.2: Kinematic and kinetic outcome measures during the stance phase of walking and running and the downstroke of cycling.

	Walking	Running	Cycling
Speed (m/s)	0.9 (0.12)	2.4 (0.33)	
Resistance (gear)			11.6 (1.5)
Cadence (RPM)	58.8 (3.1)	85.2 (4.6)^{A}	79.9 (7.0)^{A}
Peak External Foot Reaction Force (N)	863.0 (174.5)	1917.4 (234.0)	345.7 (106.2)
Peak Knee Flexion Moment (Nm/kg)	0.9 (0.3)	2.9 (0.5)	0.5 (0.1)

Note: Values presented as mean (SD). Values with the same superscript are not significantly different from each other ($p<0.05$).

5.1 Dynamic Joint Stiffness

The dynamic joint stiffness curves for walking and running were similar, with cycling appearing drastically different (Figure 5.1). Walking and running produced dynamic joint stiffness plots progressing clockwise, and cycling produced a dynamic joint stiffness plot progressing in a counterclockwise direction, including the loading phase having in a negative slope.

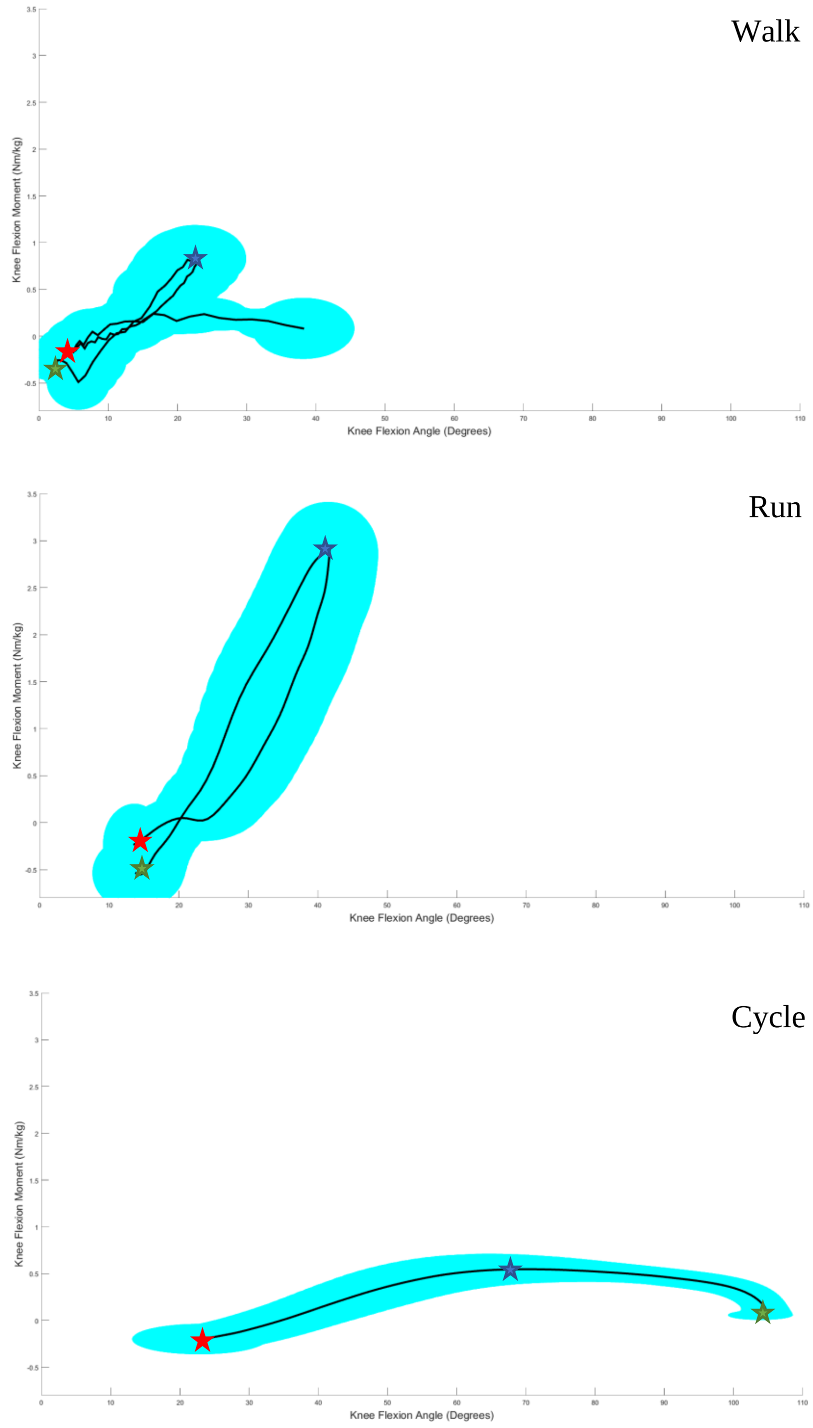

Figure 5.1: Mean DJS plots for walking, running and cycling, for all participants. The green star represents heel strike (for walking and running) or initiation of pedal force (for cycling). The blue star represents max knee flexion moment (the end of the loading phase and the start of the terminal phase). The red star represents contralateral heel strike (for walking), toe off (for running) or BDC (for cycling).

Dynamic joint stiffness, the slope of the linear regression of the moment-angle plots, are displayed in Figure 5.2, for stance/downstroke and for the initial and terminal phases. For all phases, there was a main effect for activity (Stance/Downstroke: $F_{1.182,14.1785} = 107.944$, $p < 0.0001$; Initial: $F_{1.182,14.179} = 64.192$, $p < 0.0001$; Terminal: $F_{2,24} = 37.021$, $p < 0.0001$), with post hoc analyses revealing significant differences occurred between all activities (all $p < 0.05$), except between walking and running during the terminal phase ($p = 0.186$). For the stance/downstroke, running had the largest DJS (0.108 ± 0.030 Nm/kg/deg), followed by walking (0.032 ± 0.011 Nm/kg/deg) and then cycling (0.005 ± 0.002 Nm/kg/deg). Over the initial phase, running had the largest DJS (0.129 ± 0.041 Nm/kg/deg), followed by walking (0.065 ± 0.015 Nm/kg/deg) and then cycling (-0.015 ± 0.005 Nm/kg/deg), which had a negative slope. Over the terminal phase, running had the largest DJS (0.107 ± 0.035 Nm/kg/deg), followed by walking (0.081 ± 0.035 Nm/kg/deg) and then cycling (0.018 ± 0.004 Nm/kg/deg).

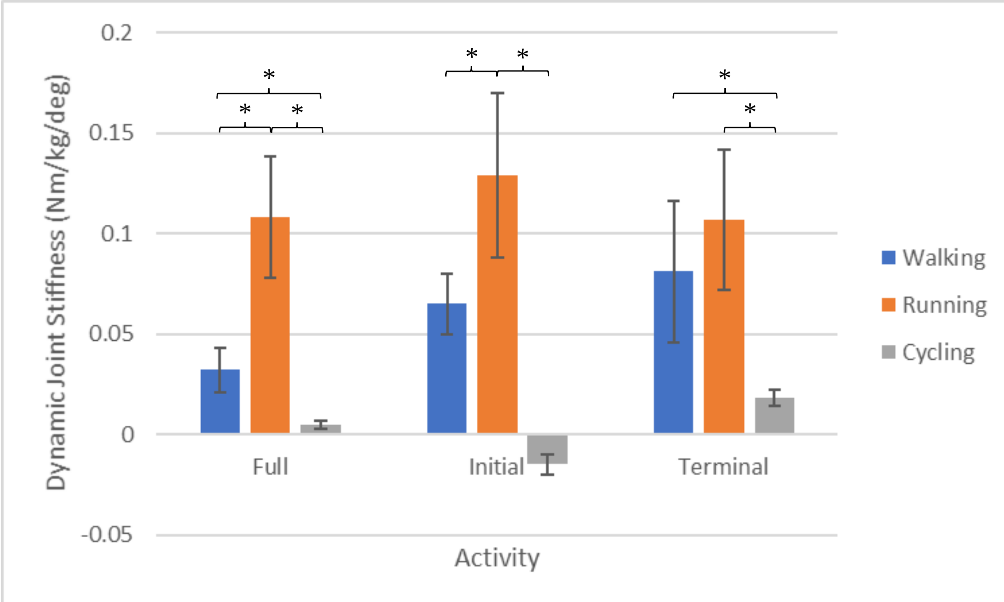

Figure 5.2: Mean dynamic joint stiffness values for the knee across all participants during each phase analyzed. Vertical error bars represent standard deviations.

5.2 Co-Contraction Index

The average co-contraction index was determined for stance/downstroke (Figure 5.3), initial phase (Figure 5.4) and terminal phase (Figure 5.5), for each activity. Over stance/downstroke, there were significant effects of activity for all muscle groupings (all $\alpha < 0.05/4$) (VLLG: $F_{1.327,15.929} = 61.194$, $p < 0.0001$; VMMG: $F_{1.176,11.756} = 37.952$, $p < 0.0001$; VLBF: $F_{1.192,13.112} = 24.941$, $p < 0.0001$; VMST: $F_{1.032,12.386} = 20.230$, $p = 0.001$)(Figure 5.3). Post hoc tests revealed that running had a larger CCI than both walking and cycling for all muscle groupings (all $p < 0.05$) and that walking and cycling were not different from each other.

Over the initial phase, there were significant effects of activity for all muscle groupings (all $\alpha < 0.05/4$) (VLLG: $F_{2,24} = 16.053$, $p < 0.0001$; VMMG: $F_{1.185,11.848} = 21.207$, $p < 0.0001$; VLBF: $F_{1.322,14.545} = 12.6720$, $p = 0.002$; VMST: $F_{1.262,15.139} = 12.890$, $p = 0.002$) (Figure 5.4). Post-hoc tests reveal running had the largest CCI for VLLG and VMMG groupings (both $p < 0.05$) and that walking and cycling were not different from each other (all $p > 0.05$). In VLBF, running had a larger CCI than walking, which was larger than cycling (all $p < 0.05$). In VMST and medial knee, walking and running were larger than cycling, but not different from each other (all $p < 0.05$).

For the terminal phase, there were significant effects of activity for all muscle groupings (all $\alpha < 0.05/4$)(VLLG: $F_{2,24} = 59.908$, $p < 0.0001$; VMMG: $F_{1.334,14.673} = 29.849$, $p < 0.0001$; VLBF: $F_{1.225,13.472} = 19.588$, $p < 0.0001$; VMST: $F_{1.050,12.598} = 16.903$, $p = 0.001$) (Figure 5.5). Post-hoc tests reveal running had the largest CCI for all muscle groupings and that walking and cycling were not different from each other (all $p < 0.05$).

Of the 12 muscle grouping comparisons (4 groupings x 3 phases each), walking only produced a greater CCI compared to cycling for the initial phase of VLBF and VMST. Cycling never produced a larger CCI compared to walking and neither walking nor cycling produced a greater CCI compared to running.

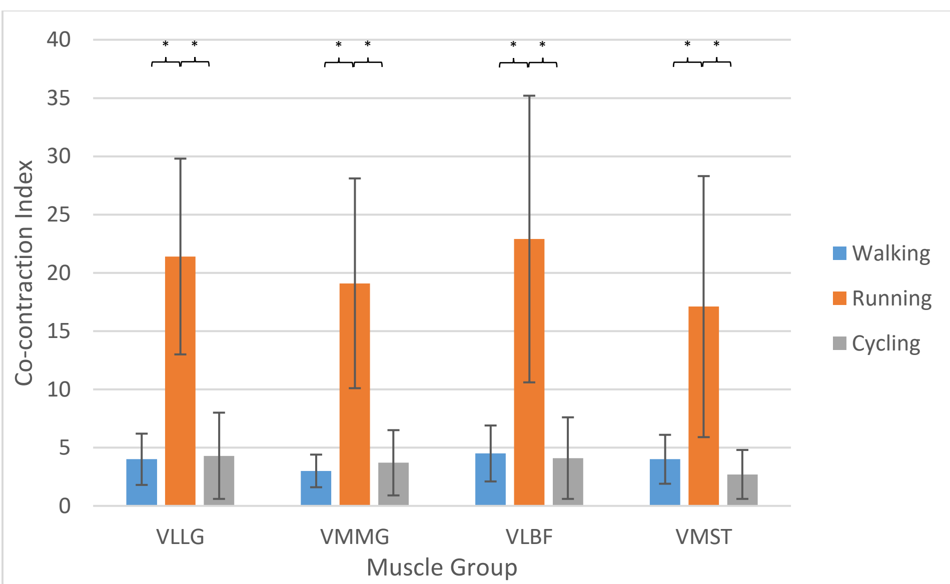

Figure 5.3: Mean co-contraction indices across all participants for muscle groupings about the knee for walking, running and cycling during the stance/downstroke of each activity. Vertical error bars represent standard deviations.

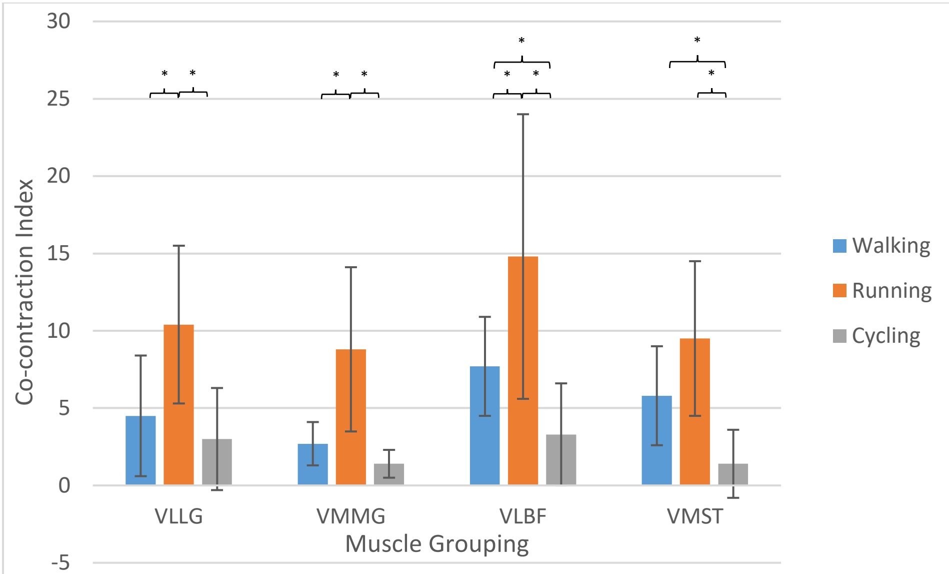

Figure 5.4: Mean co-contraction indices across all participants for muscle groupings about the knee for walking, running and cycling during the initial phase of stance/downstroke of each activity. Vertical error bars represent standard deviations.

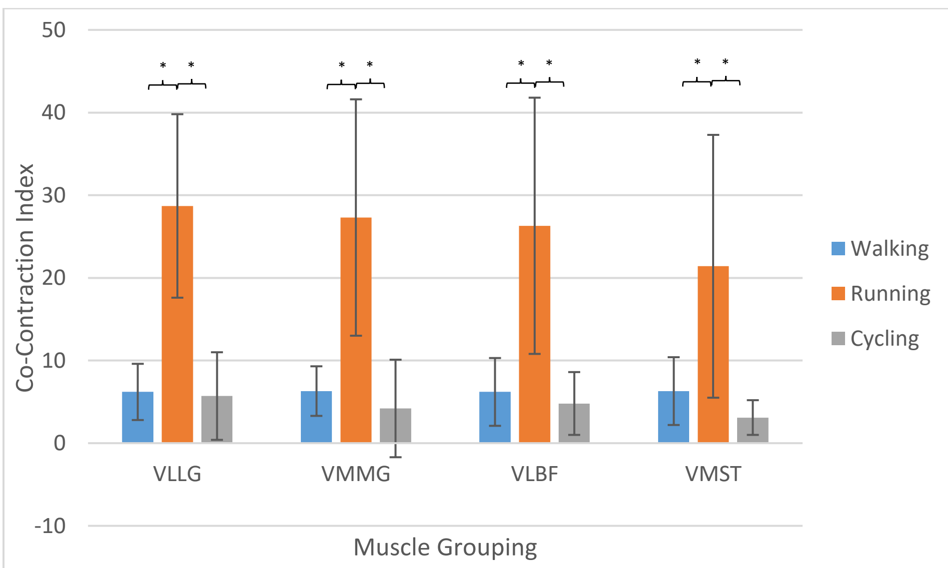

Figure 5.5: Mean co-contraction indices for muscle groupings about the knee for walking, running and cycling during the terminal phase of stance/downstroke each activity. Vertical error bars represent standard deviations.

5.3 Dynamic Joint Stiffness / Co-Contraction Index Correlation

Pearson correlations were calculated to determine the relationship between DJS and CCI for each activity for stance/downstroke and the initial phase and terminal phase (Table 5.3).

There were no strong correlations with DJS for any muscle pairing for any activity. The CCI for walking had at least a moderate correlation for at least one phase for every muscle grouping. The CCI for running had no moderate correlations to DJS and the CCI for cycling had a moderately negative correlation to DJS for the full pedal revolution of the VMMG muscle grouping.

Table 5.3: Pearson Moment Correlations between DJS and CCIs of each muscle grouping.

Activity	Phase	VLLG	VMMG	VLBF	VMST
Walking	Full	0.309*	0.398*	0.255*	0.032
	Early	0.072	-0.347*	-0.201	-0.308*
	Late	0.213	-0.125	0.379*	0.063
Running	Full	-0.022	0.178	0.066	-0.089
	Early	-0.005	-0.032	0.040	-0.190
	Late	0.025	-0.017	0.115	0.137
Cycling	Full	-0.139	-0.425*	0.029	0.025
	Early	0.256*	-0.125	-0.021	-0.157
	Late	0.041	-0.276*	0.039	0.060

Note: Significant correlations ($p < 0.05$) were identified as high correlation if $r > 0.7$, moderate correlation if $0.3 < r < 0.7$ and low correlation if $r < 0.3$ (Ratner, 2009). Values indicated by a * are significant ($p<0.05$).

5.4 Segment Coordination & Coordination Variability

Segment coordination and coordination variability over stance/downstroke was split into thirds to represent early, mid and late phases of the activity). Table 5.4 displays the segment coordination for each segment coupling averaged over each phase. Despite being forms of gait, walking and running did not share the same segment coordination patterns for all phases of both couplings (Table 5.4) with the largest differences occurring during the late phases.

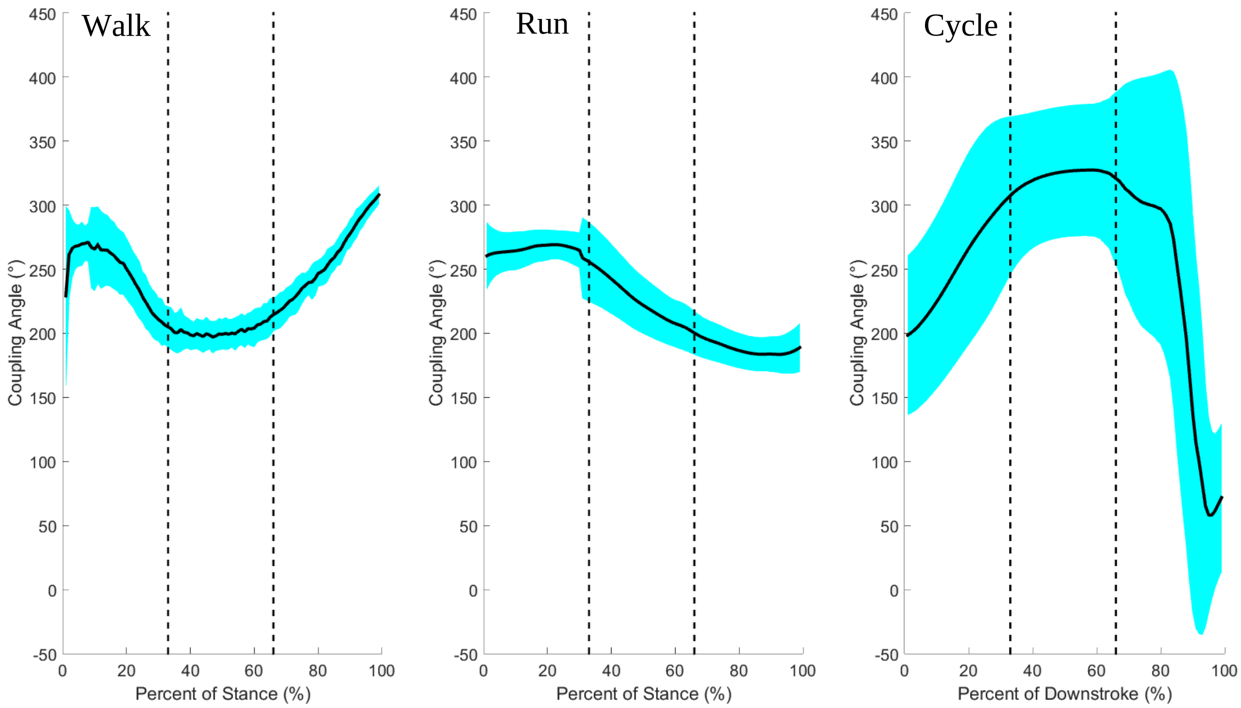

Figure 5.6: Mean segment coordination coupling angles for the sagittal thigh/sagittal shank segment coupling for A) walking stance phase, B) running stance phase and C) cycling downstroke for all participants. The dashed lines separate the signals into thirds, to represent the early, mid and late stages.

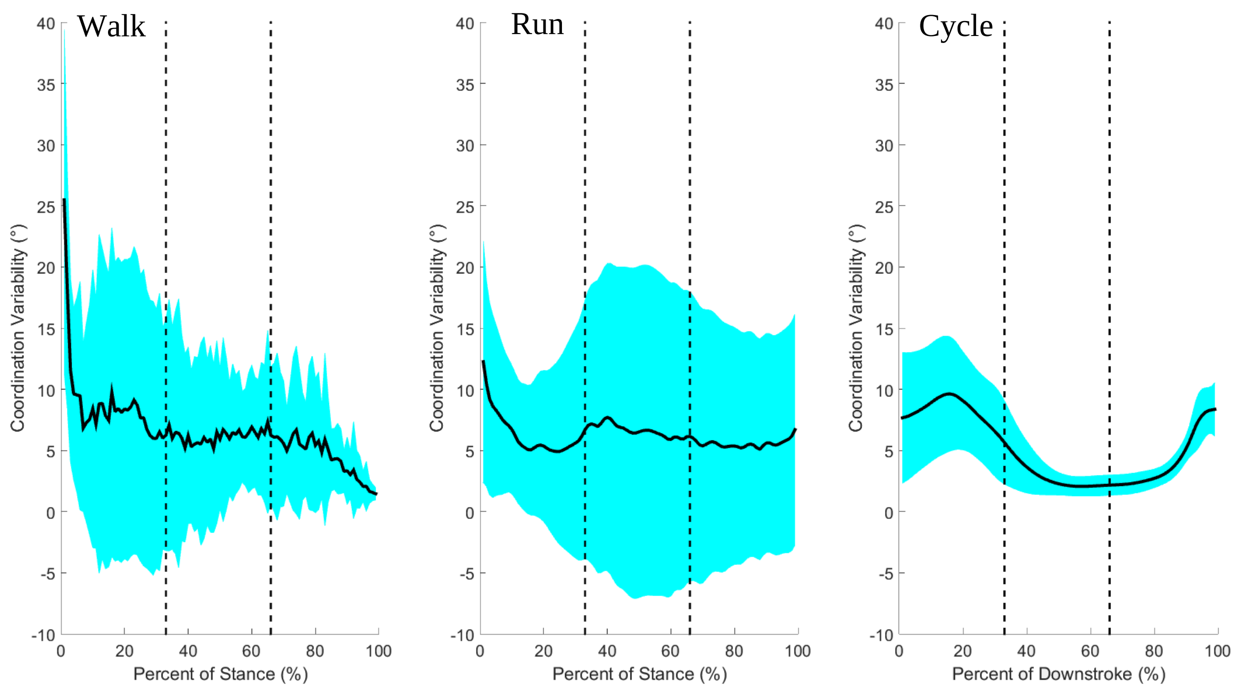

Figure 5.7: Mean Coordination variability for the sagittal thigh/sagittal shank segment coupling for A) walking stance phase, B) running stance phase and C) cycling downstroke for all participants. The dashed lines separate the signals into thirds, to represent the early, mid and late stages.

There was no significant effect of activity (p > 0.05/6) on coordination variability for either phase of the stance phase/downstroke for either the sagittal shank / sagittal foot segment coupling (Early: $F_{2,24} = 0.534$, p = 0.593; Mid: $F_{2,24} = 0.598$, p = 0.558; Late: $F_{2,24} = 1.385$, p = 0.270) or the sagittal thigh / sagittal shank segment coupling (Early: $F_{2,24} = 0.369$, p = 0.695; Mid: $F_{1.346,16.156} = 0.814$, p = 0.415; Late: $F_{1.190,14.276} = 0.190$, p = 0.0.712).

Table 5.4: Segment coordination values for the corresponding segment couplings during walking, running and cycling.

Segment Pairing	Activity Phase	Walking	Running	Cycling
Sagittal Shank v. Sagittal Foot	Early	214.3 (4.2)	220.5 (10.0)	221.0 (32.9)
	Mid	210.2 (13.8)	207.5 (9.0)	221.0 (34.6)
	Late	231.4 (4.6)	258.8 (8.2)	117.4 (35.1)
Sagittal Thigh v. Sagittal Shank	Early	248.0 (6.0)	266.0 (7.7)	274.2 (33.6)
	Mid	202.2 (6.5)	226.9 (4.5)	339.7 (32.1)
	Late	257.8 (6.1)	188.3 (6.4)	238.6 (42.9)

Note: Values are presented in degrees (SD).

Table 5.5: Coordination variability values for the corresponding segment couplings during walking, running and cycling.

Segment Pairing	Activity Phase	Walking	Running	Cycling
Sagittal Shank v. Sagittal Foot	Early	9.2 (11.7) ^{AB}	6.0 (5.2) ^{AC}	6.7 (5.6) ^{BC}
	Mid	11.4 (9.3) ^{AB}	7.1 (12.6) ^{AC}	9.3 (5.9) ^{BC}
	Late	4.0 (4.4) ^{AB}	7.9 (13.6) ^{AC}	9.6 (5.6) ^{BC}
Sagittal Thigh v. Sagittal Shank	Early	8.8 (10.0) ^{AB}	6.4 (6.7) ^{AC}	8.2 (3.8) ^{BC}
	Mid	6.2 (6.2) ^{AB}	6.6 (12.6) ^{AC}	2.8 (1.2) ^{BC}
	Late	4.4 (3.4) ^{AB}	5.5 (9.7) ^{AC}	4.2 (0.8) ^{BC}

Note: Values are presented in degrees (SD). Values with the same superscript are not significantly different from each other ($p<0.05$).

5.5 Iliotibial Band EMG Exposure

The mean ITBIZ$(t)_{per_rep}$ was calculated for all three activities, averaged for each stance phase in walking and running and for each downstroke in cycling. For walking and running (blue shading and red shading, respectively, Figure 5.8), every stance contained two events of ITBIZ during stance, typically once during the loading response, and once during the pre-swing. In cycling (yellow shading, Figure 5.8), some participants entered the ITBIZ near the end of the downstroke, when the pedal was near BDC. Eight out of fifteen participants consistently presented ITBIZ events and were thus classified as ITBIZ cyclists (section 4.3.2). Of these eight ITBIZ classified cyclists, three were excluded due to contaminated EMG signals from GM. The data for ITBIZ impingement measures are displayed in Table 5.6.

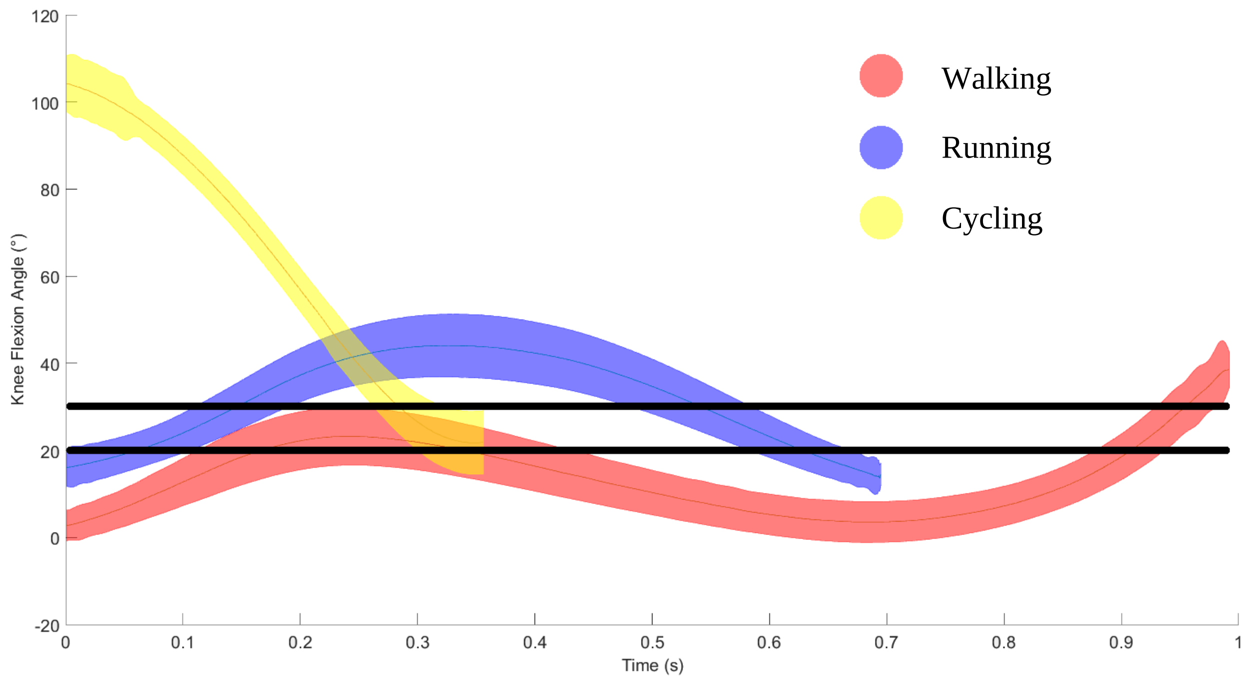

Figure 5.8: Knee flexion angles for one stance phase of walking (red) and running (blue), and one downstroke of cycling (yellow), averaged across all participants. The area between the black bars represents the interval in which the knee was flexed in the ITBIZ.

There was no significant effect of activity on ITBIZ(t)$_{per\_rep}$ ($p > 0.05/6$) ($F_{1.077,7.540}$ = 7.293, p = 0.0527), ITBIZ(t)$_{60}$ ($F_{1.164,8.150}$ = 3.917, p = 0.079) or ITBIZ(t)$_{cumul\_load}$ ($F_{2,14}$ = 2.062, p = 0.164). There was a significant effect of activity on ITBEX ($F_{2,10}$ = 20.950, p < 0.0001) (Table 5.6).

For the ITBEX, running (161.8 ± 70.3 %MVC/min) was larger than both walking (25.4 ± 27.3 %MVC/min) and cycling (5.6 ± 3.8 %MVC/min), which were not different from each other.

Comparing equivalent workouts, there was a difference in the number of ITBIZ events between running and cycling (t(7) = -4.654, p=0.002 but no difference in the duration with the knee flexed in the ITBIZ between running and cycling (t(7) = 2.567, p=0.037).

Table 5.6: ITBIZ durations and ITBEX values for walking, running and cycling.

	Walking	Running	Cycling
ITBIZ Per Rep (ms)	115.9 (69.4)^{AB}	53.8 (8.1)^{AC}	41.5 (16.6)^{BC}
ITBIZ over 60 minutes (sec)	403.4 (239.3)^{AB}	274.8 (59.7)^{AC}	195.8 (73.0)^{BC}
ITBIZ Events Over Equivalent Workout	-	6907.4 (2848.3)	11709.8 (1415.6)
ITBIZ Equivalent Workout (sec)	-	319.8 (96.8)^{C}	497.8 (189.7)^{C}
ITBIZ Cumulative Load (sec)	102.1 (59.7)^{AB}	68.7 (14.9)^{AC}	111.5 (45.3)^{BC}
ITBEX (%MVC/min)	25.4 (27.3)^{B}	161.8 (70.3)	5.6 (3.8)^{B}

Values are presented as mean (SD). Values with the same superscript are not significantly different from each other ($p<0.05$).

5.6. Results Summary Tables

Table 5.7: Summary table for the main hypotheses.

Question	Hypotheses	Conclusion	Supporting Evidence
How does stiffness at the knee differ when walking, running and cycling?	The DJS will be greater in running compared to walking and cycling during equivalent bouts of walking, running and cycling.	Rejected	DJS was greatest for running when considering the entire stance/downstroke and in the initial phase. In the terminal phase, walking and running were not different, but were larger than cycling. Cycling had the lowest DJS for all phases
How does the co-contraction of the muscles around the knee differ between walking, running and cycling?	The co-contraction will be greater in running compared to walking and greater for walking compared to cycling for all muscle groupings during equivalent bouts of walking, running and cycling.	Rejected	CCI was greatest for running for all phases and all muscle groupings, except for the VMST groupings, where it was not different from walking in the initial phase. Walking had a greater CCI compared to cycling for VLBF and VMST muscle groupings in the initial phase.
How does coordination variability of the lower limb differ between walking, running and cycling?	Coordination variability will be greater for walking compared to running and greater for running compared to cycling.	Rejected	There were no differences in the coordination variability between walking, running and cycling for any segment couplings.
How do measures of exercise intensity affect duration of the knee flexed in the ITBIZ between walking, running and cycling?	The ITBEX will be greatest for running compared cycling, which will be greater compared to walking.	Rejected	Running was larger than both walking and cycling, which were not different from each other.
	The ITBIZ(t)$_{cumul\_load}$ will be greater for cycling compared to walking and running.	Rejected	There were no differences in the ITBIZ(t)$_{cumul\_load}$ between walking, running and cycling

Chapter 6 : Discussion

The primary objective of the current study was to assess the biomechanical differences between walking, running and cycling, in order to provide insight to how various outcome measures may contribute to overuse injury. To accomplish this, participants were asked to perform 6 trials each of moderate intensity running and cycling, as well as 6 trials of walking at a self-selected pace. Walking was included to act as a presumed low-risk control to compare running and cycling against, which allowed for a practical comparison that is not often taken into account when interpreting findings. A moderate intensity of activity was selected to represent a typical recreational/social run or bike ride. From the data collected, outcome variables were grouped into four categories (dynamic joint stiffness, co-contraction, segment coordination, and IT band impingement measures). These outcomes were investigated due to their association with overuse injury (Hamill et al., 1999; Heiderscheit et al., 2002) and long-term knee health (Chang et al., 2017; Hafer et al., 2016; McGinnis et al., 2013). Greater DJS has been associated with more severe knee osteoarthritis (Chang et al., 2017; Zeni & Higginson, 2009), greater risk of tibial stress fracture (Milner et al., 2007) as well as bony injury and soft tissue injury (Butler et al., 2003). Greater muscular co-contraction has been associated with increased energy expenditure (Hortobágyi et al., 2009, 2011), greater joint contact forces (Hodge & Harris, 1986; Sasaki & Neptune, 2010; Winby et al., 2009) as well as increased joint contact pressure (Li & Park, 2004). Additionally, co-contraction has been found to be a coping mechanism strategy for instability (Gantchev & Dimitrova, 1996; Hurd & Snyder-Mackler, 2007; Slijper & Latash, 2000) and linked to PFPS in walking and running (Besier et al., 2009). Lower coordination variability has been associated with greater risk of overuse injury (Hamill et al., 2012). The number of ITBIZ events has been associated with ITBS (Farrell et al., 2003). It has previously been postulated that repetition was more important compared to reaction forces as a contributing factor for ITBS in cyclists, in addition to anatomical differences and differing training habits. To advance this work, a novel exposure metric was developed involving knee angular velocity and EMG, as well as a cumulative loading analysis, to attempt to investigate why ITBS is still a risk to cyclists.

Generally, DJS was greater in running compared to walking and cycling, indicating that increased knee stiffness may contribute to injury in running. CCI was greatest in running compared to walking and cycling, indicating that increased muscle activation may contribute to injury in running. Coordination variability was not different between walking, running and cycling. Lastly, the ITBEX was greatest in running compared to walking and cycling but the $ITBIZ(t)_{cumul\_load}$ was not different between walking, running and cycling. Both the knee angular velocity and magnitude of muscle activation of GM might contribute to overuse injury in running. It is unlikely that coordination variability or time spent in the ITBIZ, given an equivalent cumulative load play a role in contributing to overuse injury.

The following discussion sections will elaborate on the results found in the current study, compare and contrast them to findings from previous literature and explain any implications with regards to the hypotheses and objectives of the study.

6.1 Dynamic Joint Stiffness

There is a complex relationship between performance and injury risk in DJS. A larger DJS is linked to bony injury such as tibial stress fracture and knee OA (Chang et al., 2017; Milner et al., 2007; Zeni & Higginson, 2009). Thus, it would seem logical to attempt to reduce DJS by means of altering ground or shoe properties. However, this may not be feasible at the cost of performance and in addition, a lower DJS at the knee could lead to more soft tissue injuries (Butler et al., 2003; Granata, Padua, & Wilson, 2002) or increase the risk of injury to other joints (Butler et al., 2003). It is thought that there is an optimal level of stiffness to balance injury risk and performance (Butler et al., 2003). In the current study, the DJS was generally larger for running compared to walking at a self-selected pace and cycling at an equivalent intensity to running. It was hypothesized that the DJS would be greater in running compared to walking and cycling, which was rejected due to walking and running having similar DJS during the terminal phase. DJS may contribute to overuse injury in running due to its greater magnitude during the initial phase, where weight acceptance and work absorption occurs.

When the knee is stiff during walking and running, energy is lost by the system to the tissues of the body, as eccentric contraction of the knee extensors occurs (Frigo et al., 1996), which might cause injury (Warren et al., 1999). Since the knee joint is not a perfectly elastic structure, this energy is either lost or partially applied to the hip via bi-articular muscles (Frigo et al., 1996). Though there is no threshold for injury for the DJS of the knee, it is typically accepted that a larger DJS is associated with a greater risk for bony injury compared to a lower DJS (Butler et al., 2003).

Though DJS has been calculated for walking and running, it has never been found for cycling. The results of this study show cycling as a vastly different activity in terms of DJS from gait. The moment-angle curve progresses counterclockwise, with the initial phase comprised of an increasing knee moment and decreasing knee angle, while the terminal phase involves a decreasing knee moment and a decreasing knee angle. As described in Table 6.1, this indicates that work is always produced (the joint moment never opposes the angular velocity), and never absorbed. There is never a state of knee ‘stiffness’ in cycling (work absorption); the knee instead has compliance. In walking and running, upon heel strike, the knee extensors apply an extensor force about the knee, while knee flexion occurs. This acts to decelerate the body and control the landing to stabilize the joint during early stance. In cycling, no such requirement exists. Due to the cyclic nature of the pedal revolution, inducing a dynamically stiff knee is not relevant to the task. Not having a stiff knee is notable, in that in cycling, eccentric contraction of the knee extensors never occurs, which might be associated with injury (Warren et al., 1999).

Cycling was shown to have a positive DJS value in the terminal phase, which was lower than walking in the terminal phase, which is when the primary thrust is developed in walking. Cycling is often recommended as a safe alternative for physical activity to reduce stresses on the knee for individuals after TKA, rehabilitation after injury to as a low stress activity. A lower DJS can be interpreted as a lower loading of tissues in the body (Blatich et al., 2013). Though cycling has a lower DJS (which has been previously associated with an increase in soft tissue injury (Butler et al., 2003), it is unlikely to contribute to overuse soft tissue injury, since it appears that stiffness is not a requirement to successfully complete the task. The interpretations of DJS in the current study supports the notion that cycling is a safe alternative for physical activity for these populations.

Table 6.1: Summary of interpretations of DJS slopes.

	Positive Slope	Negative Slope
Progressing Left to Right	Work Absorbed	Work Produced
Progressing Right to Left	Work Produced	Work Absorbed

For the stance/downstroke and initial phase, running resulted in the greatest DJS, followed by walking, then cycling (Figure 5.2). In the terminal phase, walking and running were larger compared to cycling, but not different from each other. These findings can be explained by breaking down the DJS into its components: the knee flexion moment and knee flexion angle. Running had the largest ground reaction force (1917.4 ± 224.8 N), and the largest peak knee flexion moment (2.9 ±0.5 Nm/kg). Walking had the second largest ground reaction force (863.0 ±167.7 N), and the second largest peak knee flexion moment (0.9 ± 0.3 Nm/kg). Cycling had the lowest peak ground reaction force (345.7 ± 102.1 N), and the lowest peak knee flexion moment (0.5 ± 0.1 Nm/kg). This information, coupled with cycling having the largest knee range of motion (79.4 ± 11.4°), walking having the second largest range of motion (35.6 ± 9.3 °) and running having the least range of motion (31.7 ± 6.5°) over the signal analyzed, contributed to the differences found in DJS components (moment-angle curve slope). Even though running had a lower range of motion (denominator of the slope equation) compared to walking, the much larger knee flexion moment (numerator) of running was a more important factor. Cycling had the lowest knee flexion moment and largest range of motion, resulting in the lowest slope (smallest DJS).

The DJS plots for walking and running consist of work absorption and work production sections (Figure 5.1). As weight is accepted following heel strike, the knee flexion moment increases as the knee flexion angle increases (an increasing positive slope, Figure 5.21). This is indicative of work absorbed by the joint and characterizes the initial phase of walking and running (Frigo et al., 1996). After weight acceptance during the initial phase, as the stride progresses from mid to late stance, both the knee flexion moment and the knee flexion angle decreases (a decreasing positive slope, Figure 5.1). This represents work produced by the joint (Frigo et al., 1996). In cycling, DJS has never been investigated, to the knowledge of the author. The initial phase has an increasing negative slope and the terminal phase has a decreasing positive slope, both representative of work production (Figure 5.1). Like cycling, walking and running had a decreasing, positive slope in the terminal phase, where work was produced and the joint is compliant (i.e. the joint is moving in the direction of the knee flexion moment and thus produces no resistance to the applied force.) The DJS has been determined in previous studies (Milner et al., 2007; Powell & Williams, 2018). Results from the initial phase of these studies are presented in Table 6.2. Milner et al. (2007) measured the DJS for individuals with and without previous tibial stress fractures to investigate correlations between DJS and impact shock between groups. The results revealed that DJS was positively correlated to impact shock and that the group with previous injury ran with a greater DJS compared to the control group. For their control group, the DJS was found to be lower than that in the current study. The DJS in the previous study was calculated from heel strike to the first vertical ground reaction peak (the interval which they referred to as initial contact) whereas DJS in the current study was calculated from heel strike to the peak knee flexion moment, which occurred more towards midstance. In addition, Milner and colleagues only studied female runners, and it has been shown that females have a lower leg stiffness compared to matched males in a hopping task (Granata et al., 2002), which could explain the difference between findings. When Powell & Williams (2018) measured DJS in younger and older runners, the DJS was found to be similar to that found in the current study (Table 6.2). It was found that older runners (63.6 ± 3.6 years) ran with lower knee stiffness, suggesting that running biomechanics change with age, perhaps to compensate for decreased capacity or to preserve joint function.

Table 6.2: DJS comparison during running between the current study, Milner et al. (2007) and Powell & Williams (2018).

	Chuang (2019)	Milner et al. (2007)	Powell & Williams (2018)
Initial Phase DJS (Nm/kg/deg)	0.129 (0.041)	-	0.119 (0.02)
Initial Phase DJS (Nm/kg-m/deg)	0.117 (0.02)	0.030 (0.015)	-
Age (years)	25.1 (4.7)	24 (9)	33.6 (5.2)
BMI (kg/m^2)	22.2 (2.3)	22.0	22.9
Running Pace (m/s)	2.4 (0.33)	3.7	3.35
Notes	Self-Selected Pace	Young, healthy female runners	Young, healthy runners (sex unspecified)

Note: Values are presented in mean (SD), where applicable.

DJS of the knee has also previously been found for walking, comparing total knee arthroplasty participants to controls (McGinnis et al., 2013) and for individuals with moderate OA, severe OA and controls (Zeni & Higginson, 2009). Results for the initial phase of these studies are compared to those of the current study in Table 6.3. The findings of McGinnis and colleagues (2013) were greater than that found in the current study over the initial phase, which may be due to the faster walking pace than in the current study. However, the previous study did have an older cohort than the current study (62.7 ± 6.6 years vs. 25.1 ± 4.7 years, respectively). However, Zeni & Higgins (2009) present a DJS that very closely matches the DJS found in the current study over the initial phase. The similar values may be attributed to the similar parameters between the studies (pace, cadence, BMI), despite Zeni & Higgins (2009) also having an older cohort. In contrast to running, where DJS was found to decrease with age, the current study combined with this previous study shows that, the DJS of walking may be less affected by age and more influenced by gait parameters.

Table 6.3: DJS comparison for walking between the current study, McGinnis et al. (2013) and Zeni & Higginson (2009).

	Chuang (2019)	McGinnis et al. (2013)	Zeni & Higgins (2009)
Initial Phase DJS (Nm/kg/deg)	0.065 (0.015)	-	0.066
Initial Phase DJS (Nm/kg-m/deg)	0.019 (0.006)	0.042 (0.015)	-
Age (years)	25.1 (4.7)	62.7 (6.6)	58.9 (11.4)
BMI (kg/m^2)	22.2 (2.3)	29.0 (4.8)	24.9 (3.7)
Walking Pace (m/s)	0.9 (0.12)	1.44 (0.15)	1.0
Cadence (steps/min)	58.8 (3.1)	-	54.6 (4.0)
Notes	Self-Selected Pace	Self-Selected Pace	Controlled Pace

Note: Values are presented in mean (SD), where applicable.

One method to attempt to reduce overuse injury at the knee might be to adapt methods to alter the DJS in running to some optimal level. However, in addition to relevancy to injury, DJS has also been associated with performance. As a task’s demand increases (increasing running speed, jump height, for example) the DJS has been found to increase (Arampatzis et al., 1999; Farley, Houdijk, Van Strien, & Louie, 1998; Granata et al., 2002; Kuitunen et al., 2002; Stefanyshyn & Nigg, 1998). This is explained by greater external forces being applied to the body, resulting in a greater stiffness being produced to resist the increased applied forces in order to perform the task with increasing effort (Butler et al., 2003).

There has been research investigating modifications to stiffness and it has been found that after training volleyball players to 'land softer' after jumping, injury rates were lower over the course of a season, compared to a control group. One method of altering leg stiffness during running might be to alter footfall landing pattern (Arampatzis et al., 1999; Butler et al., 2003; Seyfarth, Geyer, Gunther, & Blickhan, 2002) or alter the foot/ground contact properties (Farley et al., 1998; Ferris, Louie, & Farley, 1998; Smith & Watanatada, 2002).

6.2 Co-Contraction Index

For all phases of all muscle groupings, except for the loading phase of the VMMG muscle grouping, running had the largest CCI compared to walking and cycling. Walking and cycling were only different during the loading phase for VLBF. Neither walking nor cycling produced larger CCI compared to running. This is unsurprising, since the CCI calculation used in the current study incorporates not only the ratio of muscle activity between antagonistic muscles, but also the magnitude of the muscle activation (Rudolph et al., 2001). Running is a ballistic, high-velocity activity that requires a lot of muscle activation in order to gain control after heel strike and propel the body forward near toe off (Besier et al., 2009; Novacheck, 1998).

An increase in CCI has been found to increase joint contact forces (Hodge & Harris, 1986; Sasaki & Neptune, 2010; Winby et al., 2009), joint contact pressures (Li & Park, 2004) and energy expenditure in older adults (Hortobágyi et al., 2009, 2011). Moreover, CCI is also a coping mechanism strategy for instability (Gantchev & Dimitrova, 1996; Hurd & Snyder-Mackler, 2007; Slijper & Latash, 2000) and has been linked to PFPS in walking and running (Besier et al., 2009). Previous research investigating CCI has primarily focused on differences between injured and uninjured participants (Chmielewski et al., 2005), moderate/severe/control OA participants (Hubley-Kozey, Hill, Rutherford, Dunbar, & Stanish, 2009; Rudolph et al., 2001; Schmitt & Rudolph, 2008) and age-related changes (Lo et al., 2017). Relevant to the current study are the behavior of the control/healthy cohorts in these studies.

For healthy individuals during walking, the values found in the current study are generally of smaller magnitude to those found in previous literature (Table 6.4). In these studies, the CCI was measured from heel strike (or 100 ms prior) to an inflection in the knee joint moment.

Table 6.4: Comparison of co-contraction indices of the current study to previous literature.

	Chuang (2019)	Childs et al. (2004)	Chmielewski et al. (2005)	Lewek et al., (2004)	Hubley-Kozey (2009)	Rudolph et al. (2007)
VLLG	4.5 (3.9)	-	-	15 (8)	10 (5)	7.8 (2.2)
VMMG	2.7 (1.4)	-	40.0 (9)	11 (8)	9 (5)	8.1 (2.8)
VLBF	7.7 (3.2)	15 (9)	30.4 (11.9)	19 (10)	16 (6)	13.2 (3.2)
VMST	5.8 (3.2)	-	-	16 (10)	19 (10)	12.3 (2.9)
Walking Speed (m/s)	0.9 (0.12)	1.12 - 1.34	Self-Selected Pace	1.5 (0.2)	1.37 (0.18)	1.39 (0.08)
Age (years)	25.1 (4.7)	62 (10)	Not Reported	49.5 (6.1)	49.2 (9.7)	20.6
Status	Healthy	Healthy	Healthy	Healthy	Asymptomatic	Healthy

Note: Values are presented as degrees (SD), where applicable.

The larger CCI values found by Chmielewski et al. (2005) may be explained by their use of a movable platform. The platform was in a 'locked' condition, but the act of walking over an unfamiliar apparatus may have invoked some anxiety/instability causing an increase in co-contraction. The findings of Childs (2004), Lewek et al. (2004) and Hubley-Kozey (2009) were also larger than those found in the current study. All participants in these studies were older than that of the current study, and age has been shown to be linked to an increased co-activation of antagonistic muscles (Hortobágyi et al., 2009; Hortobágyi, Fernandez, & Rothwell, 2006). The findings of Rudolph and colleagues (2007) were closest to the co-contraction values found in the current study. Though both studies had young participant cohorts, the faster walking pace in the previous study may have resulted in the larger CCI values. It is interesting to note that the quadriceps/hamstring groupings tend to be larger than their respective medial/lateral quadriceps/gastrocnemius groupings. Further, physical activity levels were not reported in the previous studies. The highly active, younger cohort of the current study may have an increased efficiency which could explain the lower CCI values found.

6.3 Dynamic Joint Stiffness / Co-Contraction Index Correlation

It has previously been proposed that DJS and CCI could be correlated, since a higher muscular co-contraction is thought to prevent instability, potentially creating a stiffer joint (McGinnis et al., 2013). In the current study, moderate significant correlations were found during some phases of walking, but none for running, and moderate negative correlations for cycling for the VLLG and VMMG muscle groupings. McGinnis et al. (2013) found no significant relationship between DJS and CCI about the ankle at a self-selected walking pace (1.44 ± 0.15 m/s).

In the current study, it was hypothesized that there would be at least a moderate correlation for all phases for all muscle groupings. This hypothesis was rejected and a possible explanation for these findings could be that the relationship (or lack thereof) between DJS and CCI in trained individuals is influenced more strongly by experience and efficiency, rather than a need to address instability. Being able to produce sufficient joint stiffness to perform a task could be related to personal style and technique. Given that the participants in the current study were highly trained athletes (competition experience: 5.3 ± 3.9 years, weekly training: 8.0 ± 3.1 hours/week), the conditions in the study were not novel and the CCI may have been lower as an adaptation to reduce energy expenditure. More moderate correlations may have been found in walking since it is a less demanding task that every participant performed at a fairly uniform pace and cadence. The more demanding tasks of running and cycling may invoke more varied muscular coordination patterns between participants, revealing less correlations to DJS.

6.4 Segment Coordination & Coordination Variability

Segment coordination explains the possible strategies that an individual may use to perform a given task (Hafer & Boyer, 2018; Hamill et al., 1999). By plotting segment angles (expressed with respect to the global coordinate system) against each other, the coupling angles between consecutive data points were calculated using a modified vector technique. Coordination variability describes the variance of the segment coordination stride to stride or pedal revolution to pedal revolution. In contrast to end-point variability, where an increased variability is unfavourable and associated with less experience, an increase in coordination variability is beneficial and associated with healthy, higher performing individuals (Arutyunyan et al., 1969, Hamill et al., 2012).

Segment coordination has previously been compared between walking and running (Boyer et al., 2016; Hafer & Boyer, 2018) and it was found that walking and running had different coordination patterns. Further, Hafer et al. (2018) compared the segment coordination and coordination variability between young, active older and less active older adults. A comparison of segment coordination is displayed in Table 6.5, comparing the results of the current study to those of the young cohort from Hafer et al. (2018). In general, the values are consistent with previous literature.

Table 6.5: Comparison of segment coordination for walking between the current study and Hafer & Boyer (2018). Green values represent “In-Phase” coordination, red values represent “Anti-Phase” coordination, blue values represent “Proximal Segment” coordination and yellow values represent “Distal Segment” coordination.

Segment Coupling	Phase	Current Study	Hafer & Boyer (2018)
Sagittal Shank v. Sagittal Foot	Early	214.3 (4.2)	209.2 (3.5)
	Mid	210.2 (13.8)	202.1 (7.0)
	Late	231.4 (4.6)	224.6 (4.3)
Sagittal Thigh v. Sagittal Shank	Early	248.0 (6.0)	240.2 (5.5)
	Mid	202.2 (6.5)	206.7 (7.8)
	Late	257.8 (6.1)	268.9 (7.4)

Note: Values are presented in degrees, mean (SD).

The Hafer group has also determined the coordination variability for walking and running in a number of different studies (Boyer et al., 2016; Hafer & Boyer, 2017, 2018; Hafer et al., 2016). These data are displayed and compared in Table 6.6 and Table 6.7, for walking and running, respectively. For walking, the coordination variability values are similar, with values from the current study being slightly lower in general than those previously reported for the mid and late phases. In the previous work, participants walked at a faster pace of 1.38 ± 0.14 m/s, compared to an average of 0.9 ± 0.12 m/s in the current study.

Table 6.6: Comparison of coordination variability for walking between the current study, Hafer & Boyer (2017) and Hafer & Boyer (2018).

Segment Coupling	Phase	Current Study	Hafer & Boyer (2017)	Hafer & Boyer (2018)
Sagittal Shank v. Sagittal Foot	Early	9.2 (11.7)	-	8.1 (4.4)
	Mid	11.4 (9.3)	-	11.1 (5.9)
	Late	4.0 (4.4)	-	4.8 (4.2)
Sagittal Thigh v. Sagittal Shank	Early	8.8 (10.0)	6.0 (1.2)	7.6 (3.9)
	Mid	6.2 (6.2)	4.0 (1.0)	5.1 (2.7)
	Late	4.4 (3.4)	4.4 (0.8)	4.9 (3.2)

Note: Values are presented in degrees, mean (SD).

Coordination variability differences between the current study and previous studies were larger for running than for walking. All studies recruited experienced runners, so it is unlikely that these differences were due to a difference in experience/ability. Running pace was not reported by Hafer et al. (2016) and Boyer et al. (2016), however participants were asked to run at a self-selected pace and cadence typical of a comfortable run (similar to the instructions of the current study). Hafer & Boyer (2017) had participants run at a self-selected moderate pace, which was 3.2 ± 0.4 m/s.

Table 6.7: Comparison of coordination variability for running between the current study, Hafer et al. (2016), Hafer & Boyer (2017), Boyer et al. (2016).

Segment Coupling	Phase	Current Study	Hafer et al. (2016)	Hafer & Boyer (2017)
Sagittal Thigh v. Sagittal Shank	Early	6.4 (6.7)	3.8 (1.7)	4.0 (0.8)
	Mid	6.6 (12.6)	3.1 (0.4)	3.0 (0.8)
	Late	5.5 (9.7)	5.8 (3.5)	7.0 (3.6)

Note: Values are presented in degrees, mean (SD).

A key difference between the studies conducted by Hafer & Boyer (2017) and Hafer & Boyer (2018) and the current study was the environment in which data were collected. The previous studies had participants walk/run on a treadmill whereas in the current study participants ran overground on a runway. In the current study, during the running trials, participants may have still been accelerating when this data was collected. Further, the greater variability may have come from some participants altering their stride to land on the force plate, despite the instruction to look forward while running and the researchers flagging any trial where the participant appeared to alter their gait to land on the force plate. Treadmill running could have been more comfortable and may have been collected while the participant was running at a steady state.

Despite having different segment coordination magnitudes, coordination variability was not significantly different between any activities for any segment coupling in the current study. These results are similar to a previous study who found similar variances between phases for walking and running for 8 participants (Li et al., 1999). In this study variation was calculated using continuous relative phase, as opposed to the modified vector coding technique used in the current study. Both studies were performed at self-selected walking pace and an average run pace of 2.24 m/s, though Li et al. (1999) constrained that speed for the participants, while in the current study it was tailored for each participant.

Segment coordination and coordination variability has never been calculated for cycling; this dynamical systems approach offers a novel comparison between walking, running and cycling. It was hypothesized that, due to the limited degrees of freedom in cycling, coordination variability would be the lowest, however, this was not found to be the case. For all segment couplings, there were no differences between activity, for any of the three phases (early, mid or late). A lower coordination variability has been associated with an injured state, lower level performance and inexperience (Arutyunyan et al., 1969; Hamill et al., 2012). If walking can be assumed to be a low-risk control activity, in which all participants perform with a high level of experience/competency, the similar coordination values found for running and cycling could help support the justification that these athletes were highly trained in both running and cycling. It has been theorized that in addition to a low coordination variability being associated with injury, when coordination variability is excessive, there may also be an association to injury as well (Hamill et al., 2012). Thus, there is thought to be an optimal range for coordination variability where normal, healthy function occurs (Hamill et al., 2012). Since the values for running and cycling found in the current study were not difference from walking, these data may provide an indication of this optimal range of coordination variability.

6.5 Iliotibial Band Impingement Measures

The ITBIZ was defined as when the knee is flexed between 20° and 30°, where impingement of the IT band can occur, leading to ITBS. ITBS has been attributed to increased repetition, anatomical differences (leg length discrepancy, varus knee alignment or excessive joint rotation), and improper training (i.e. too much volume/intensity) (Farrell et al., 2003). In the current study, the stance phase of walking and running were compared to the downstroke of the cycling pedal revolution. $ITBIZ(t)_{per\_rep}$ was found to be 115.9 ± 69.4 ms during each stance phase of walking, 53.8 ± 8.1 ms during each stance phase of running and 41.5 ± 16.6 ms per downstroke of cycling. Extrapolated over a one-hour activity, $ITBIZ(t)_{60}$ was found to be 6.7 ± 4.0 minutes for walking, 4.6 ± 1.0 minutes for running and 3.3 ± 1.2 minutes for cycling, taking into account participant's cadence for each activity. Finding the ITBIZ duration for an equivalent cumulative load, $ITBIZ(t)_{cumul\_load}$ was found to be 102.1 ± 59.7s during walking, 68.7 ± 14.9s during running, and 111.5 ± 45.3s during cycling. A novel exposure metric (ITBEX) was proposed, which incorporated the knee angular velocity and iEMG of GM to provide an indication of whether the IT band was being tensioned by musculature while it was in the ITBIZ. The ITBEX was 25.4 ± 27.3 %MVC/min for walking, 161.8 ± 70.3 %MVC/min for running and 5.6 ± 3.8 %MVC/min for cycling. Walking was included in the analysis of the current study to act as a baseline, since walking has no reported increase in ITBS risk. Walking, in fact, had a trend of having a larger ITBIZ duration than that of running and cycling in the current study. This is primarily due to the fact that over a typical walking gait stride, the knee is in the ITBIZ for a larger percentage. Thus, the notion that ITBIZ duration or repetition alone is a prominent factor for ITBS (Farrell et al., 2003) is not supported by the data in the current study.

ITBIZ has been previously investigated in a cycling-oriented study by Farrell et al. (2003). The minimum knee flexion angle was 32 ± 7.2° and it was reported that ITBIZ per cycle revolution was 38ms for the whole pedal revolution at an average cadence of 88 RPM at an intensity of 280 W. That minimum flexion angle was slightly larger than that found in the current study, which was 27.99 ± 8.04° for athletes who were classified as ITBIZ cyclists (section 5.5). ITBIZ per cycle revolution for cycling in the current study (79.6 ± 37 ms) was more than double that of the value previously published. This discrepancy can be attributed to differences in bike set up. Farrell et al. (2003) fit the participants to the cycle ergometer by statically placing their knee angle at 25-30 degrees with the pedal at bottom dead center. As previously shown (Bini & Hume, 2016), this method of static bike fit often does not result in the same measurements taken dynamically. The bike fit method from the current study was selected since it has been shown to produce optimal power, which happened to result in the knee often being flexed in the ITBIZ for some participants.

Farrell et al. (2003) also addressed the issue that ITBS had a similar prevalence in running and cycling, despite cycling having a much lower GRF. Citing a study by Orchard et al. (1996) for running data, it was concluded that for a workout of equivalent exposure (10km run vs. 40km bike ride), running would have a total time of 330s with the knee flexed in the ITBIZ, with cycling having 250s with the knee flexed in the ITBIZ. The same calculations were performed in the current study and the opposite relationship was found. Running had an $ITBIZ(t)_{eq\_workout}$ duration of 320 ± 97s over a 10km run and cycling had an ITBIZ duration of 498 ± 190s for a simulated 40km bike ride. Farrell et al. concluded that since cycling had a decreased impingement duration compared to running, other factors such as anatomical differences, repetition, improper bike set-up and training errors are the most likely influencers of ITBS. However, this was a general observation with no statistical analysis. Comparing ITBIZ duration in the current study, there was no significant difference between running and cycling duration when the comparison was based on the hypothetical equivalent workout or when extrapolating the data for a one-hour activity. It is important to note that in this analysis, the stance phase of running was compared to the entire downstroke of cycling, consistent with methods previously published (Farrell et al., 2003)

Farrell et al. (2003) found repetition to play a more influential role in the onset of ITBS compared to measures involving external foot forces, since it was found that cycling involved more impingement events, less external foot forces and less duration with the knee flexed in the ITBIZ compared to running. For an athlete running at 4.5 m/s with a 1 m stride length, Farrell et al. approximated that 4800 impingement zone events will take place, assuming 1 impingement event per stride. The results of the current study show that for each walking and running stride, the knee is flexed into the ITBIZ twice per stride – once just after heel strike and once just before toe-off (Figure 5.8), which should have doubled the 4800 used in the previous analysis to 9600. In the current study, the cadence for walking, running and cycling was 59 ± 3.1 RPM, 85 ± 4.6 RPM and 80 ± 7.0 RPM, respectively. Running and cycling were found to have a significantly larger cadence compared to walking but were not statistically different from each other. For an average run or ride of 1 hour, these could equate to ~5100 steps or ~4800 pedal revolutions, respectively. Although this is greater than the ~3540 steps for one hour of walking, many people take more than 3540 steps per day, with no reported increase of ITBS. Modern fitness devices, in fact, recommend an arbitrary 10,000 steps per day as a goal. These data further indicate that repetition alone may not be as influential to the onset of ITBS as previously thought.

It was hypothesized that task intensity may play a role in ITBS development. Running and cycling are typically done at a greater intensity than walking, which may help to explain the similar injury risk. Thus, the novel ITBEX was proposed to incorporate not only the ITBIZ duration, but the knee flexion velocity as well as the muscle activation of GM over that interval. The results revealed that running was greater than both walking and cycling, but walking and cycling were not different from each other (Table 5.6). Thus, the ITBEX might provide insight as to why ITBS is so prevalent in runners. Running tended to have more EMG activation compared to walking or cycling, and cycling tended to have the fastest angular knee velocity compared to walking and running with the knee flexed in the ITBIZ. Thus, muscle activity could potentially be more of an influence for ITBS in running, while knee angular velocity could potentially be more of an influence in cycling.

Another approach to investigating the relationship between ITBS and intensity was to determine a cumulative load for each activity. In this instance, cumulative loading is defined as the summation of the external forces being applied to the foot over a given interval. Repetitive loading of tissues may decrease their tolerance over time and reduce their capacity to function without injury (Marra et al., 2014). Gatti et al. (2017) determined the cumulative load of the external ground reaction forces of healthy men (25.8 ± 4.2 years) running at a self-selected moderate pace for the equivalent of a 15-minute run. With the same participants cycling at a moderate intensity, it was found that a 15-minute run had the same cumulative load of a 46-minute cycling duration. In other words, it took approximately 3 times as long in cycling to accumulate the same cumulative load as running. In the current study, using the same analysis methods as Gatti et al. (2017), a 15-minute run had the same cumulative load as a 33.8 ± 1.9-minute cycling duration. Thus, the current study found cycling to require, on average, 2.2 times the duration to that of running to get to the same cumulative load. This difference in cumulative load time may be due to the more specifically trained cohort as well as the regulation of exercise intensity with heart rate in the current study. The study population of the previous study had an average weekly activity history of 6570.6 ± 4158.7 MET/week, which is equivalent to approximately a 40 ± 27 minute run every day, or a little over 4.5 ± 3.2 hours of running per week (Kyu et al., 2016), as an example. For comparison, the participants of the current study averaged 8.03 ± 3.05 hours per week. When ITBIZ duration was compared over the same cumulative load of a 15-minute run, there was no difference between walking (102.1 ± 59.7s), cycling (111.5 ± 45.3s) or running (67.8 ± 14.9s).

Farrell and colleagues (2003) suggested that the number of repetitions of impingement zone events is the major contributor to ITBS in cycling, with the aforementioned force and ITBIZ duration being less important. The rationale was that the reaction force in cycling on the foot is one-fifth of that found during running and since the duration spent in the ITBIZ over a given workout was approximated to be less in running compared to cycling (330s and 250s, respectively), the greater number of repetitions must have been important. The results of the current study found evidence to the contrary. During walking and running, the knee was flexed into the ITBIZ twice per stance phase. This doubles the previous approximation of ITBIZ events for walking and running revealing that the number of ITBIZ events is actually less during cycling. Further, in contrast to the same workout as calculated by Farrell and colleagues (a simulated 10 km run and 40km bike), the time spent in the ITBIZ was larger for cycling (497.8 ±189.7s), compared to running (319.9 ± 96.8s). The running ITBIZ(t)$_{per\_rep}$ was similar to that previously published (previously found to be 75ms (Farrell et al., 2002)), but the cycling ITBIZ(t)$_{per\_rep}$ was nearly double the duration when considering the entire pedal revolution when comparing to data previously published (previously found to be 38ms (Farrell et al., 2002)). When only including the downstroke of the pedal revolution, ITBIZ(t)$_{per\_rep}$ was similar between running and cycling (53.8 ± 8.1s vs. 41.3 ± 16.6s, respectively). Lastly, despite the force being approximately one-fifth in cycling compared to that of running, when compared to the same cumulative load, there were no differences between activities. Due to the similarities between ITBIZ(t)$_{per\_rep}$ between the activities, repetition alone might not be enough to explain ITBS risk in running and cycling.

6.6 Rationale of Cycling Injuries

Running and cycling injury rates are very similar (Dettori & Norvell, 2006; Van Gent et al., 2007), however the outcome measures in the current study cannot definitively explain why. Since the DJS and CCI in running were so much greater compared to walking and cycling, different mechanisms might be at play as they relate to overuse injuries. Considering all the outcome measures, cycling was not significantly different compared to walking – which was assumed to be a low injury risk activity. Based on these findings, cycling should not have a higher incidence of injury compared to walking. This might be interpreted in two ways: that the outcome measures in the current study do not contribute to overuse cycling injury or that there are other factors at play. A possible explanation of these findings might be the increased knee flexion angle exhibited during cycling, specifically near TDC, when the lower limb first begins to apply pressure to the pedal. At TDC during cycling, the knee is flexed to greater than 100° (Figure 4.14). This is well above the typical knee flexion angle of either walking or running during the stance phase, where force is also being applied to the foot in those activities. Around this flexion angle, the cyclist begins to apply force to the pedal rapidly. With the knee in so much higher flexion, a greater joint contact force created by co-contraction could be detrimental, as it is at these flexion angles that tibiofemoral joint contact area decreases as much as 25% (Yao et al., 2008). A decreased contact area could mean that instead of the contact forces being distributed over a larger area, forces are instead localized, increasing the stresses at a region of interest. Additionally, *in vitro* studies have shown that a knee flexed to 90°, compared to a more extended knee (tested at 30° of knee flexion), patellofemoral contact characteristics are altered (Lewallen et al., 1990). When applying a moment of 35 Nm (similar to that found during cycling) contact area increased by 80%, contact pressure increased by 110% and contact force increased by 230%. Modelling of the knee during walking, running and cycling has also revealed that maximum patellofemoral contact force is ~950N for walking (Shelburne, Torry, & Pandy, 2005), ~2300N for running (Flynn & Soutas-Little, 1995) and ~980N for cycling (Ericson & Nisell, 1987). Despite peak external reaction forces in cycling being only ~15% of that found in running and 40% of that found in walking, the peak patellofemoral contact forces are estimated to be nearly 40% that of running and similar to that of walking. This suggests that at these large knee flexion angle postures, a greater proportion of external reaction forces is transferred to the patellofemoral joint.

Even with a lower CCI in cycling compared to running, altered patellofemoral joint contact forces, contact pressures and contact areas could potentially contribute more to overuse injury in cycling due to these changes to joint kinematics found at a large knee flexion angle.

These altered patellofemoral characteristics may also have implications for the findings of coordination variability. For both segment couplings, there was no difference between walking, running or cycling for any phase analyzed. As Hamill et al. (2012) stated, overuse injury is a multifactorial problem and is probably caused by an interaction of many variables. The dynamical systems approach to segment coordination and coordination variability is thought to explain the influence of many of these variables. With low coordination variability, the repeated use of a particular region of tissue about a joint may lead to an increased risk of injury. With greater coordination variability, forces are more distributed over a larger proportion of the tissue and not concentrated at localized point. The changes to tibiofemoral joint contact area, contact area, contact pressure and contact force may also interact with the coordination variability to result in greater influence to overuse injury. As previously mentioned, peak patellofemoral contact force has been estimated to be ~950N (Shelburne et al., 2005), ~2300N

(Flynn & Soutas-Little, 1995) and ~980N (Ericson & Nisell, 1987) for walking, running and cycling, respectively. Though the peak patellofemoral forces are similar between walking and cycling, the coordination variability of cycling, specifically in these larger flexion knee postures which are not experienced during the stance phase of walking, may explain why patellofemoral pain develops during cycling and is often not at an elevated risk during walking.

With respect to ITBS, impingement of the IT band only occurs when the knee is flexed between 20-30 degrees. This impingement zone is coincident with the late phase of the cycling pedal revolution and depending on bike fit, is only applicable for some people (see section 5.5). In running, this is coincident with the early phase and late phases of the stance phase. Considering the insertion of the IT band at Gerdy's tubercle of the tibia (Orchard et al., 1996), sagittal rotation of the shank with respect to the thigh could influence how the IT band moves over the lateral femoral epicondyle. There were no differences, however, between walking, running or cycling during the early or late phase of the sagittal thigh / sagittal shank coupling (when the knee is flexed in the ITBIZ). This coordination variance in walking would also suggest that there is a risk for ITBS in walking, which has not been reported. The cadence of running and cycling would typically eclipse those in walking, leading to more instances of impingement per minute and the low variation could be more detrimental for these tasks due to repetition. It is thus unlikely that coordination variability plays a role in ITBS development.

Chapter 7 : Limitations

There were a few identified limitations to the current study regarding the experimental setup and the processing and interpretation of the outcome variables. With respect to the experimental setup, there were inherent differences to analyzing running and cycling in a laboratory compared to studying these activities in a more realistic, outdoor setting. In the current study, the running trials were limited to a 20-meter runway and ground reaction forces from only one stance phase could be collected for each trial. Due to the volume of the collection space, the swing phase for each participant was often not captured and thus, was excluded from analyses. The short runway and force plate set-up prohibited running at the target heart rate and measuring data for successive strides. The solution to determining the run speed was to have the participants run on a treadmill elevated to a 1% grade to determine their target heart rate and then have the participants run at the same speed on the runway.

In terms of cycling, a fixed, stationary ergometer did not allow for the lateral side to side sway that typically occurs during outdoor riding and the ergometer setup (i.e. saddle position, handlebar position) may have made joint kinematics less natural. The limited lateral sway of the bike may have reduced the coordination variability found during cycling. It is possible that on a mobile bicycle that there may have been greater coordination variability due to the motion of the bicycle beneath the rider. In the literature, there is little consensus on how to optimally fit a bike to an individual (Fonda et al., 2014). However, dynamic bike fits have been recommended to ensure similar kinematics between testing and bike set up (Fonda et al., 2014). Unfortunately, these dynamic bike fits are time consuming and can bias the knee flexion angle to fall within the impingement zone. Thus, an alternate bike fit method that has been shown to produce similar hip, knee and ankle angles to a dynamic bike fit was used (Bini & Hume, 2016), as described in section 4.2.1.

In addition to the ergometer set up parameters, the instrumented pedals used may have been unfamiliar to participants. Though athletes with experience with clipless pedals were recruited, the stack height of the instrumented pedals (the distance from the pedal spindle to the interface at the bottom of the shoe) was approximately 5cm higher than a typical commercially available pedal to make room for the force transducer. Anecdotally, participants reported an unusual feel to the pedals initially, that was gone after the first few minutes of the warm-up. This can be corroborated by the fact that the coordination variances for cycling were relatively low and similar to that of walking and running, which could indicate that the instrumented pedal had a small to negligible impact on the participant's kinematics.

As no study has yet to quantitatively compare the biomechanics of cycling compared to running, there are few variables which have been used to compare both modalities. ITBIZ duration has been previously studied for running/gait (Orchard et al., 1996) and cycling (Farrell et al., 2003). However, DJS and joint coordination have only previously been determined for walking and running. DJS is often analyzed for gait during the stance phase or portions of stance phase (Chang et al., 2017; Günther & Blickhan, 2002; Stefanyshyn & Nigg, 1998; Zeni & Higginson, 2009). In order to define a cycling equivalent, DJS for cycling was determined during pedal loading as an analog to stance phase. To take into account the different characteristics of the signals (Figures 4.15), DJS was first defined over stance/downstroke, and then broken down into an initial phase and terminal phase. Partitioning the DJS has previously been done by Frigo et al. (1996) who divided gait strides into quasi-linear phases bounded by inflection points in the DJS plot.

Similarly, segment coordination is often analyzed during the stance phase of gait. It has never been measured in cycling. Segment coordination data collection is often done on a treadmill and it has been shown that 10 and 8 strides best represents reliable coordination analyses for walking and running, respectively (Hafer & Boyer, 2017). These values were established by taking the lowest stride count that produced a mean within 10% of a 15 stride mean. 5-6 strides were used in the current study, which was approximately within 15% of a 15 stride mean, according to data from Hafer & Boyer (2017). Given the small standard deviations for most coordination variances in the current study, this number of strides was deemed acceptable for this analysis.

Lastly, due to limitations of a two-axis custom force pedal, center of pressure could not be calculated in real time during data collections. To account for this, center of pressure of each participant was estimated using a Tekscan F-Scan 3000E pressure sensor during 5-10 pedal revolutions, before data collection began. A limited number of pedal revolutions were captured using the Tekscan sensor due to the capacity of each sensor (1-7 'uses' each) and the limitations of the Visual3D software used to compute the joint kinetics. Each Tekscan sensor is recommended for less than 50 steps/repetitions, which would equate to approximately 45 seconds of a cycling trial, making it ineffective for the 6-minute cycling protocol. Further, Visual3D requires fixed values to create new landmarks (center of pressure in this case), meaning that a static COP position location relative to the foot tracking markers were required. During the downstroke, across all pedal revolutions recorded with the Tekscan sensor, the medial-lateral center of pressure moved by an average of 3.57 ± 1.94 mm and the anterior-posterior center of pressure moved by an average of 7.43 ± 4.95 mm. Further, when isolating center of pressure movement within a single downstroke, the medial-lateral center of pressure moved by an average of 2.09 ± 1.10 mm and the anterior-posterior center of pressure moved by an average of 4.87 ± 4.48 mm. The use of a fixed center of pressure is an acceptable estimate, since it is assumed that these changes in foot center of pressure within the cycling shoe would not significantly affect the kinetics of the lower limb.

Chapter 8 : Future Directions & Contributions

Future research should expand on the findings of the current study to an older cohort. The current study was performed on young, healthy, well-trained adults, which lays the groundwork for future research. The motivation for the current study was that as runners age, there is a shift from running participation to cycling participation in order to avoid injury caused by the larger impact forces experienced while running. While the results of the current study corroborate this rationale (running had a larger DJS and CCI, in part due to larger ground reaction forces), future research should investigate how these results compare to those of an older cohort. In addition, prospective studies are required to determine causation of overuse injury in running and cycling. According to Tanaka & Seal (2003), endurance performance peaks at around 35 years of age, followed by decreases over the next 15-25 years before a drastic drop off is experienced. This trend is similar for athletes of all ability levels (Joyner, 1993) and decline in performance is more drastic in women compared to men (Donato et al., 2003; Joyner, 1993; Tanaka & Seals, 1997). The decrease in performance is the result of a decreased capacity to maintain training intensity and volume, as well as decreased cardiovascular properties such as stroke volume, maximal heart rate and VO_2max (Tanaka & Seals, 2008).

The results of the current study indicate that CCI was much lower in cycling compared to running and thus, co-contraction may not be a primary contributor to injury in cycling. However, a lower CCI isn't necessarily beneficial if there are additional changes to joint loading parameters. As mentioned in section 6.2, at large knee flexion angles, such as those during the initial phase of cycling, contact areas, pressures and forces are altered from a more extended knee. In the current study, these effects of large knee flexion angles were not measured and should be investigated to further the understanding of the effects that contribute to overuse injury in running and cycling.

As the ITBEX was a novel metric proposed in the current study, it has never been previously investigated and future work should investigate its relationship to IT band injury. A longitudinal study should be performed to explore whether there is an association to injury risk between iEMG of GM and cadence of each activity.

As mentioned in section 4.3.2.2.2, due to the size of the collection space, only the stance phase of the running stride could consistently be recorded for each participant. Thus, the stance phase of walking and running was compared to the downstroke of cycling. There were no differences in $ITBIZ(t)_{per\_rep}$ between walking, running or cycling. Thus, including the swing phase of walking and running and the upstroke of cycling into this analysis may help to parse out differences, since the IT band would pass through the ITBIZ additionally during the swing phase. In future studies, the full gait stride for walking and running should be included to provide a more complete analysis.

Both males and females were welcome to participate in the study. Although there has been some research revealing differences in the kinematics and kinetics between the sexes (Ferber et al., 2003; Phinyomark et al., 2015; Sinclair & Selfe, 2015; Willson et al., 2012) and it has previously been advocated that results should not be collapsed across sex (Schache et al., 2003), due to the small sample size in the current study, it was not feasible to compare sexes. With larger Q-angles, women may have different lower limb frontal plane alignment compared to men (Emami et al., 2007), which may result in different findings, specifically pertaining to the DJS and coordination variability. With an increased Q-angle, the sagittal plane moment of the knee may differ, affecting the DJS calculations. Additionally, an increased Q-angle may also affect the segment coordination and coordination variability findings, since it breaks down segment angles into the frontal, transverse and sagittal planes. Future work should be done to identify if there is an effect of sex on the outcome variables.

Furthermore, the activities performed in the current study were done at a moderate intensity, and future research should consider these activities at a higher intensity. The goal of many endurance athletes is to apply the fitness and form gained during training to a more competitive, racing setting. Analysis of varying intensities could offer insight into how overuse injury might be affected by faster run paces/increasing cycling resistances, since it is unlikely that an individual maintains a precise level of moderate intensity throughout an entire workout.

Lastly, future research should investigate frontal plane kinetics and kinematics. There is limited data available for the frontal plane of cycling, specifically in middle-aged and older adults (Fang et al., 2016). PFPS has been associated with frontal plane moments (Stefanyshyn et al., 2006; Myer et al., 2010; Myer et al., 2014; Myer et al., 2015). In addition, a frontal plane analysis may supplement the understanding of ITBS. It has been accepted that when the knee is between 20° and 30° of knee flexion, the IT band is impinged by the lateral epicondyle of the femur. It is possible that either frontal knee angle or frontal knee moments, while the knee is flexed in the ITBIZ, may provide more information on the mechanism of ITBS injury.

The current study contributed to the field of knee and sport biomechanics by being the first to explicitly compare the kinematics, kinetics and electromyography of walking, running and cycling. Many studies have investigated the biomechanics of running and cycling in separate studies, or compared the influence of one sport on successive performance of the other (Bernard et al., 2003; Hausswirth et al., 2001; Heiden & Burnett, 2003; Hue et al., 1998; Millet & Vleck, 2000; Millet, Vleck, & Bentley, 2009; Vercruyssen et al., 2002). In addition, the inclusion of a comparison to walking (a presumed low-risk activity of daily living) allowed for a practical comparison that is not often taken into account when interpreting findings. Further, no study had previously investigated DJS or segment coordination for cycling, providing novel data to the research community to build upon.

The current study provided contributions to the understanding of potential injury mechanisms in running and cycling. The rates of overuse injuries (such as ITBS and PFPS) are quite similar in running and cycling (Baskins et al., 2016; Dettori & Norvell, 2006; Neptune, Wright, & Van Den Bogert, 2000; Van Gent et al., 2007). However, the factors considered in this work, which are known to be related to overuse injuries, were, in many cases, significantly different between the two activities. This apparent discrepancy is resolved if, in fact, the mechanisms for developing the same overuse injuries are different between running and cycling.

To address the common perception that cycling participation reduces overuse injury compared to running, some of the data from the current study appear to corroborate this idea. The DJS and CCI values found during running are greater than walking and cycling due in part to the larger ground reaction forces applied to the body. A greater ground reaction force would increase the knee joint moments (resulting in a larger DJS), as well as greater muscle activation (CCI) in order to maintain stability and control over the motion. These outcome variables are associated with greater risk of bony overuse injury (Butler et al., 2003), increased joint contact force (Hodge et al., 1986; Lu et al., 1997; Winby et al., 2009; Sasaki & Neptune, 2010) and retrospectively linked to PFPS (Besier et al., 2009), among others.

There were no differences between running or cycling and walking for the coordination variability or $ITBIZ(t)_{cumul\_load.}$ The novel ITBEX was greater for running, compared to walking and cycling, which could indicate that the EMG magnitude and knee angular velocity while the knee is flexed within the ITBIZ could contribute to ITBS in running. The ITBEX has never been investigated before, however an increase in activity of the musculature acting on the IT band while it is in the ITBIZ is suggested to increase tension on the IT band. A larger ITBEX could indicate either that muscle activity in the impingement zone tensions the IT band or that the IT band moves over the condyles at a faster rate, possibly producing more irritation, or a combination of both.

Despite the lower ground reaction forces experienced during cycling, overuse injury prevalence is still similar to that of running. The outcome measures in the current study could not adequately explain this phenomenon. As described in section 6.6, perhaps these outcome measures, in combination with the larger knee flexion angles achieved while cycling may contribute to overuse injury.

Running overuse injury has previously been thought to be linked to impact forces, which has been supported by findings of the current study. Cycling does appear to be less detrimental with respect to impact-related variables (DJS, CCI). However, the injury rates in cycling have been found to be comparable to those of running, and findings of the current study suggest that in addition to the outcome variables investigated, repetitive loading in large knee flexion angles might also help to explain overuse injury, rather than impact force. Those suffering from running-related impact injuries could consider cycling as a form of cross training/rehabilitation but should be cautioned that a risk of injury is still present.

Chapter 9 : References

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Chapter 10 : Appendices

Appendix A: Participant Screening Questionnaire

This questionnaire asks some questions about your health status. This information is used to guide us with your entry into the study.

Contradictions to participation in this study include:

- ☐ Any previous history of knee pain that required medical intervention or time off from work or training longer than three days, within the past month
- ☐ Previous knee surgery
- ☐ Inability to participate in continuous light exercise for 30 minutes
- ☐ Experienced bouts of dizziness and/or fainting
- ☐ Allergy or sensitivity to alcohol

<u>Past Relevant Health History (Check all that apply):</u>

Musculoskeletal pain/disorders

	Hip/Thigh Injury, please specify:_________________________________
	Knee/Shank Injury, please specify:________________________________
	Ankle/Foot Injury, please specify:________________________________

Cardiovascular Disorders

	Acute myocardial infraction		Cardiac arrhythmias
	High risk unstable angina		Active endocarditis
	Symptomatic severe aortic stenosis		Decompensated symptomatic heart failure
	Acute pulmonary embolus or pulmonary infarction		Acute myocarditis or pericarditis
	Left main coronary stenosis		Moderate stenotic valvular heart disease
	Tachyarrhythmias or bradyarrhythmias		Atrial fibrillation
	Hypertrophic cardiomyopathy		A/V Block

Current Relevant Health History (Check all that apply):

- ☐ Irregular Heartbeat
- ☐ Chest Pain
- ☐ Leg Pain/injury
- ☐ Back pain/injury
- ☐ Illness requiring medication within the past week (e.g. flu/cold)
- ☐ Fatigue
- ☐ Persistent Coughing
- ☐ Dizziness
- ☐ Fainting

Allergies

- ☐ Rubbing Alcohol
- ☐ Adhesives

This next section, entitled “Get Active Questionnaire”, is an established self-screening questionnaire designed to assess your readiness to participate in physical activity and to address any potential risks related with exercising.

Get Active Questionnaire

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CANADIAN SOCIETY FOR EXERCISE PHYSIOLOGY –
PHYSICAL ACTIVITY TRAINING FOR HEALTH (CSEP-PATH®)

Physical activity improves your physical and mental health. Even small amounts of physical activity are good, and more is better.

For almost everyone, the benefits of physical activity far outweigh any risks. For some individuals, specific advice from a Qualified Exercise Professional (QEP – has post-secondary education in exercise sciences and an advanced certification in the area – see csep.ca/certifications) or health care provider is advisable. This questionnaire is intended for all ages – to help move you along the path to becoming more physically active.

- ☐ I am completing this questionnaire for myself.
- ☐ I am completing this questionnaire for my child/dependent as parent/guardian.

PREPARE TO BECOME MORE ACTIVE

The following questions will help to ensure that you have a safe physical activity experience. Please answer **YES** or **NO** to each question before you become more physically active. If you are unsure about any question, answer **YES**.

YES	NO	
		1 Have you experienced **ANY** of the following (A to F) **within the past six months**?
○	○	**A** A diagnosis of/treatment for heart disease or stroke, or pain/discomfort/pressure in your chest during activities of daily living or during physical activity?
○	○	**B** A diagnosis of/treatment for high blood pressure (BP), or a resting BP of 160/90 mmHg or higher?
○	○	**C** Dizziness or lightheadedness during physical activity?
○	○	**D** Shortness of breath at rest?
○	○	**E** Loss of consciousness/fainting for any reason?
○	○	**F** Concussion?
○	○	**2** Do you currently have pain or swelling in any part of your body (such as from an injury, acute flare-up of arthritis, or back pain) that affects your ability to be physically active?
○	○	**3** Has a health care provider told you that you should avoid or modify certain types of physical activity?
○	○	**4** Do you have any other medical or physical condition (such as diabetes, cancer, osteoporosis, asthma, spinal cord injury) that may affect your ability to be physically active?

NO to all questions: go to Page 2 – ASSESS YOUR CURRENT PHYSICAL ACTIVITY

YES to any question: go to Reference Document – ADVICE ON WHAT TO DO IF YOU HAVE A YES RESPONSE

Get Active Questionnaire

CSEP | SCPE — THE GOLD STANDARD IN EXERCISE SCIENCE AND PERSONAL TRAINING

ASSESS YOUR CURRENT PHYSICAL ACTIVITY

Answer the following questions to assess how active you are now.

1. During a typical week, on how many days do you do moderate- to vigorous-intensity aerobic physical activity (such as brisk walking, cycling or jogging)? ☐ DAYS/WEEK
2. On days that you do at least moderate-intensity aerobic physical activity (e.g., brisk walking), for how many minutes do you do this activity? ☐ MINUTES/DAY

For adults, please multiply your average number of days/week by the average number of minutes/day: ☐ MINUTES/WEEK

Canadian Physical Activity Guidelines recommend that adults accumulate at least 150 minutes of moderate- to vigorous-intensity physical activity per week. For children and youth, at least 60 minutes daily is recommended. Strengthening muscles and bones at least two times per week for adults, and three times per week for children and youth, is also recommended (see csep.ca/guidelines).

GENERAL ADVICE FOR BECOMING MORE ACTIVE

Increase your physical activity gradually so that you have a positive experience. Build physical activities that you enjoy into your day (e.g., take a walk with a friend, ride your bike to school or work) and reduce your sedentary behaviour (e.g., prolonged sitting).

If you want to do **vigorous-intensity physical activity** (i.e., physical activity at an intensity that makes it hard to carry on a conversation), and you do not meet minimum physical activity recommendations noted above, consult a Qualified Exercise Professional (QEP) beforehand. This can help ensure that your physical activity is safe and suitable for your circumstances.

Physical activity is also an important part of a healthy pregnancy.

Delay becoming more active if you are not feeling well because of a temporary illness.

DECLARATION

To the best of my knowledge, all of the information I have supplied on this questionnaire is correct.
If my health changes, I will complete this questionnaire again.

I answered NO to all questions on Page 1

Sign and date the Declaration below

I answered YES to any question on Page 1

Check the box below that applies to you:

- ☐ I have consulted a health care provider or Qualified Exercise Professional (QEP) who has recommended that I become more physically active.
- ☐ I am comfortable with becoming more physically active on my own without consulting a health care provider or QEP.

Name (+ Name of Parent/Guardian if applicable) [Please print] | Signature (or Signature of Parent/Guardian if applicable) | Date of Birth

Date | Email (optional) | Telephone (optional)

With planning and support you can enjoy the benefits of becoming more physically active. A QEP can help.

☐ Check this box if you would like to consult a QEP about becoming more physically active.
(This completed questionnaire will help the QEP get to know you and understand your needs.)

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Get Active Questionnaire – Reference Document

ADVICE ON WHAT TO DO IF YOU HAVE A YES RESPONSE

Use this reference document if you answered YES to any question and you have not consulted a health care provider or Qualified Exercise Professional (QEP) about becoming more physically active.

1 Have you experienced ANY of the following (A to F) within the past six months?	
A A diagnosis of/treatment for heart disease or stroke, or pain/ discomfort/pressure in your chest during activities of daily living or during physical activity? ☐ YES	Physical activity is likely to be beneficial. If you have been treated for heart disease but have not completed a cardiac rehabilitation program within the past 6 months, consult a doctor – a supervised cardiac rehabilitation program is strongly recommended. If you are resuming physical activity after more than 6 months of inactivity, begin slowly with light- to moderate-intensity physical activity. If you have pain/discomfort/pressure in your chest and it is new for you, talk to a doctor. Describe the symptom and what activities bring it on.
B A diagnosis of/treatment for high blood pressure (BP), or a resting BP of 160/90 mmHg or higher? ☐ YES	Physical activity is likely to be beneficial if you have been diagnosed and treated for high blood pressure (BP). If you are unsure of your resting BP, consult a health care provider or a Qualified Exercise Professional (QEP) to have it measured. If you are taking BP medication and your BP is under good control, regular physical activity is recommended as it may help to lower your BP. Your doctor should be aware of your physical activity level so your medication needs can be monitored. If your BP is 160/90 or higher, you should receive medical clearance and consult a QEP about safe and appropriate physical activity.
C Dizziness or lightheadedness during physical activity ☐ YES	There are several possible reasons for feeling this way and many are not worrisome. Before becoming more active, consult a health care provider to identify reasons and minimize risk. Until then, refrain from increasing the intensity of your physical activity.
D Shortness of breath at rest ☐ YES	If you have asthma and this is relieved with medication, light to moderate physical activity is safe. If your shortness of breath is not relieved with medication, consult a doctor.
E Loss of consciousness/ fainting for any reason ☐ YES	Before becoming more active, consult a doctor to identify reasons and minimize risk. Once you are medically cleared, consult a Qualified Exercise Professional (QEP) about types of physical activity suitable for your condition.
F Concussion ☐ YES	A concussion is an injury to the brain that requires time to recover. Increasing physical activity while still experiencing symptoms may worsen your symptoms, lengthen your recovery, and increase your risk for another concussion. A health care provider will let you know when you can start becoming more physically active, and a Qualified Exercise Professional (QEP) can help get you started.

After reading the ADVICE for your YES response, go to Page 2 of the *Get Active Questionnaire* – ASSESS YOUR CURRENT PHYSICAL ACTIVITY

Get Active Questionnaire – Reference Document

ADVICE ON WHAT TO DO IF YOU HAVE A **YES** RESPONSE

Use this reference document if you answered YES to any question and you have not consulted a health care provider or Qualified Exercise Professional (QEP) about becoming more physically active.

2 Do you currently have pain or swelling in any part of your body (such as from an injury, acute flare-up of arthritis, or back pain) that affects your ability to be physically active?	☐ YES
If this swelling or pain is new, consult a health care provider. Otherwise, keep joints healthy and reduce pain by moving your joints slowly and gently through the entire pain-free range of motion. If you have hip, knee or ankle pain, choose low-impact activities such as swimming or cycling. As the pain subsides, gradually resume your normal physical activities starting at a level lower than before the flare-up. Consult a Qualified Exercise Professional (QEP) in follow-up to help you become more active and prevent or minimize future pain.	
3 Has a health care provider told you that you should avoid or modify certain types of physical activity?	☐ YES
Listen to the advice of your health care provider. A Qualified Exercise Professional (QEP) will ask you about any considerations and provide specific advice for physical activity that is safe and that takes your lifestyle and health care provider's advice into account.	
4 Do you have any other medical or physical condition (such as diabetes, cancer, osteoporosis, asthma, spinal cord injury) that may affect your ability to be physically active?	☐ YES
Some people may worry if they have a medical or physical condition that physical activity might be unsafe. In fact, regular physical activity can help to manage and improve many conditions. Physical activity can also reduce the risk of complications. A Qualified Exercise Professional (QEP) can help with specific advice for physical activity that is safe and that takes your medical history and lifestyle into account.	

After reading the ADVICE for your YES response, go to Page 2 of the *Get Active Questionnaire* – ASSESS YOUR CURRENT PHYSICAL ACTIVITY

WANT ADDITIONAL INFORMATION ON BECOMING MORE PHYSICALLY ACTIVE?

- **csep.ca/certifications**
  CSEP Certified members can help you with your physical activity goals.
- **csep.ca/guidelines**
  Canadian Physical Activity Guidelines for all ages.

Appendix B: Participant Information Questionnaire

The following questionnaire provides us with general information about yourself and equipment that you currently use. **You may choose to not answer any question.** If you have any questions, please ask the research assistant.

1. What is your sex?	☐ Male ☐ Female
2. How old are you?	______________________
3. Which leg do you kick a ball with?	☐ Right ☐ Left
4. Would you consider yourself to be in a 'good' mood today?	☐ Yes ☐ No
5. What brand of running shoes are you wearing today? (e.g. Saucony, Nike)	______________________
6. What model of running shoe are you wearing today? (if known)	______________________
7. How old are the running shoes you are wearing today?	________ years ________ months
8. What brand of cycling shoes are you wearing today? (e.g. Shimano, Sidi)	______________________
9. What model of cycling shoe are you wearing today? (if known)	______________________
10. How old are the cycling shoes you are wearing today?	________ years ________ months
11. What is your preferred pedal system? (i.e. which pedal system do you use on your primary bike?)	______________________

Appendix C: Participant Physical Activity Questionnaire

This questionnaire asks some questions about your physical activity participation so that we can get a measure of your past training and racing experience as well as your current training volume, duration and intensity.

<table>
<tr><td colspan="2">Years of experience running</td><td colspan="2">_____years</td></tr>
<tr><td colspan="2">Years of experience cycling</td><td colspan="2">_____years</td></tr>
<tr><td colspan="2">Years of competition (if any)</td><td colspan="2">_____years</td></tr>
<tr><td colspan="2">On average, how long do you spend training per week?</td><td colspan="2">_____hours _____ minutes</td></tr>
<tr><td rowspan="3">Please specify your running/cycling competency level as (novice, intermediate, elite, professional)</td><td></td><td>Running per week</td><td>Cycling per week</td></tr>
<tr><td>Time</td><td>_____hours
_____ minutes</td><td>_____hours
_____ minutes</td></tr>
<tr><td>Distance</td><td>_____ km</td><td>____ km</td></tr>
<tr><td colspan="2">Personal best 10k run time (if known)</td><td colspan="2"></td></tr>
<tr><td colspan="2">Personal best FTP (if known)</td><td colspan="2"></td></tr>
</table>

The following section was adapted from the Adult Physical Activity Questionnaire (Booth, 2000) to record information on physical activity over the past 2 weeks.

In the past 2 weeks, have you done any of the following exercises, sports, or physically active hobbies?	Yes	No	How many times in the past 2 weeks did you go/do this activity?	On average, how many minutes did you spend doing each activity each time?	As a percent, how was the rate of perceived effort of these activities split? Light (Easy)	Moderate (Tempo)	Hard (Threshold/Max)
Walking for exercise							
Gardening or yard work							
Stretching exercises							
Weightlifting/Gym							
Jogging or running							
Aerobics/aerobic dance							
Riding a bicycle							
Stair climbing for exercise							
Swimming for exercise							
Tennis							
Golf							
Bowling							
Base/softball							
Hand/racquetball/squash							
Skiing							
Downhill							
Cross-Country							
Water							
Basketball							
Volleyball							
Soccer							
Football							
Other: (please specify)							

Appendix D: Secondary Participant Screening Questionnaire

This questionnaire asks some questions about your health status since your last visit to the lab.

Current Relevant Health History (Check all that apply):

☐	Irregular Heartbeat	☐	Fatigue
☐	Chest Pain	☐	Persistent Coughing
☐	Leg Pain/injury	☐	Dizziness
☐	Back pain/injury	☐	Fainting
☐	Illness requiring medication within the past week (e.g. flu/cold)		

The following section was adapted from the Adult Physical Activity Questionnaire (Booth, 2000) to record information on physical activity since your last visit.

In the past 2 weeks, have you done any of the following exercises, sports, or physically active hobbies?			How many times in the past 2 weeks did you go/do this activity?	On average, how many minutes did you spend doing each activity each time?	As a percent, how was the rate of perceived effort of these activities split?		
	Yes	No			Light (Easy)	Moderate (Tempo)	Hard (Threshold/Max)
Walking for exercise							
Gardening or yard work							
Stretching exercises							
Weightlifting/Gym							
Jogging or running							
Aerobics/aerobic dance							
Riding a bicycle							
Stair climbing for exercise							
Swimming for exercise							
Tennis							
Golf							
Bowling							
Base/softball							
Hand/racquetball/squash							
Skiing							
Downhill							
Cross-Country							
Water							
Basketball							
Volleyball							
Soccer							
Football							
Other: (please specify)							

Appendix E: Knee Anatomy

The knee joint consists of four bones – the tibia, fibula, femur and patella. The fibula is located lateral to the tibia and, together, make up the shank of the lower limb, positioned distal to the femur and articulating with the lateral and medial femoral condyles, respectively. The patella sits superiorly to the patellar surface on the anterior distal aspect of the femur. Articular cartilage is present on the articular surfaces of the tibia and femur to distribute forces and reduce friction during movement and is surrounded by an articular capsule filled with synovial fluid (Goldblatt & Richmond, 2003).

Intrinsically, the knee is supported by two ligaments: the anterior cruciate ligament and the posterior cruciate ligament. The anterior cruciate ligament runs from the lateral condyle of the femur, anteromedially to the anterior intercondylar area of the tibia. The anterior cruciate ligament resists anterior tibial translation relative to the femur and internal rotation of the tibia under the femur (Goldblatt & Richmond, 2003). The posterior cruciate ligament runs from the medial condyle of the femur to the posterior intercondylar area of the tibia. The posterior cruciate ligament resists posterior tibial translation relative to the femur and external tibial rotation under the femur (Goldblatt & Richmond, 2003). Extrinsically, the knee is supported by the lateral collateral (fibular) ligament and the medial (tibial) collateral ligament. Laterally, the femur and head of the fibula are supported by the lateral cruciate ligament, which resists varus forces. Medially, the femur and tibia are supported by the larger medial cruciate ligament, which resists valgus forces and is also attached to the medial meniscus (Goldblatt & Richmond, 2003).

The basic musculature of the knee joint consists of the thigh knee flexors and extensors (primarily the hamstrings and quadriceps, respectively), thigh adductors (gracilis) and the shank knee flexors (lateral and medial gastrocnemius) (Blackburn & Craig, 1980).

Of interest to this proposed study are the quadriceps, hamstring, and triceps surae muscle groups, in addition to gluteus maximus and tensor fascia latae. The quadriceps muscle group consists of the vastus lateralis, vastus medialis, vastus intermedius and rectus femoris (Blackburn & Craig, 1980). Collectively, they function to extend the knee and produce an extension moment (i.e. during stance phase of walking/running and during the down-stroke of cycling). The four quadriceps muscles taper together distally to form the quadriceps tendon (Blackburn & Craig, 1980). The quadriceps tendon inserts onto the superior aspect of the patella, and inferiorly, the patellar tendon attaches the apex of the patella to the tibial tuberosity. The three hamstring muscles are the biceps femoris, semimembranosus and semitendinosus. Collectively, they perform knee flexion (i.e. during the swing phase of running and upstroke of cycling) (Blackburn & Craig, 1980). Due to the complex structure of the knee anatomy, the hamstring muscles also have indirect attachments to the patella via the fascia. The purpose of the patella complex is to increase the lever arm of the quadriceps muscles to aid in knee extension (Blackburn & Craig, 1980). The triceps surae muscle group consists of gastrocnemius lateralis, gastrocnemius medialis, and soleus. The main function of the triceps surae are to plantarflex the ankle joint, however, due to the origin of the gastrocnemius proximal to the knee, it also has a small influence on knee flexion (Moore et al., 2009). Gluteus maximus originates at the medial pelvis and inserts onto the IT band and gluteal tuberosity of the femur (Martini et al., 2018). It functions to extend and externally rotate the thigh, while stabilizing the knee during full extension and maintaining hip abduction via its attachment to the IT band (Martini et al., 2018). Lastly, tensor fascia latae originates on the lateral iliac crest of the pelvis and also inserts onto the IT band. Its primary action provides abduction of the thigh, extension of the knee and lateral rotation of the shank (Martini et al., 2018).

The close relationship of knee structures and complexity of the knee anatomy makes it susceptible to injury, with many similarities and overlapping mechanisms of injury.

Appendix F: Protocol Order

Due to logistical constraints of participant instrumentation, the protocol for walking, running and cycling trials on day 2 were quasi-randomized. Walking and running trials were always performed consecutively, i.e. cycling was always performed first or last. Figure 10.3 displays a breakdown of the protocol variations, with the number of times each variation was performed. Randomizations were determined using a macro in Google Sheets (Alphabet Inc., Mountain View, Ca).

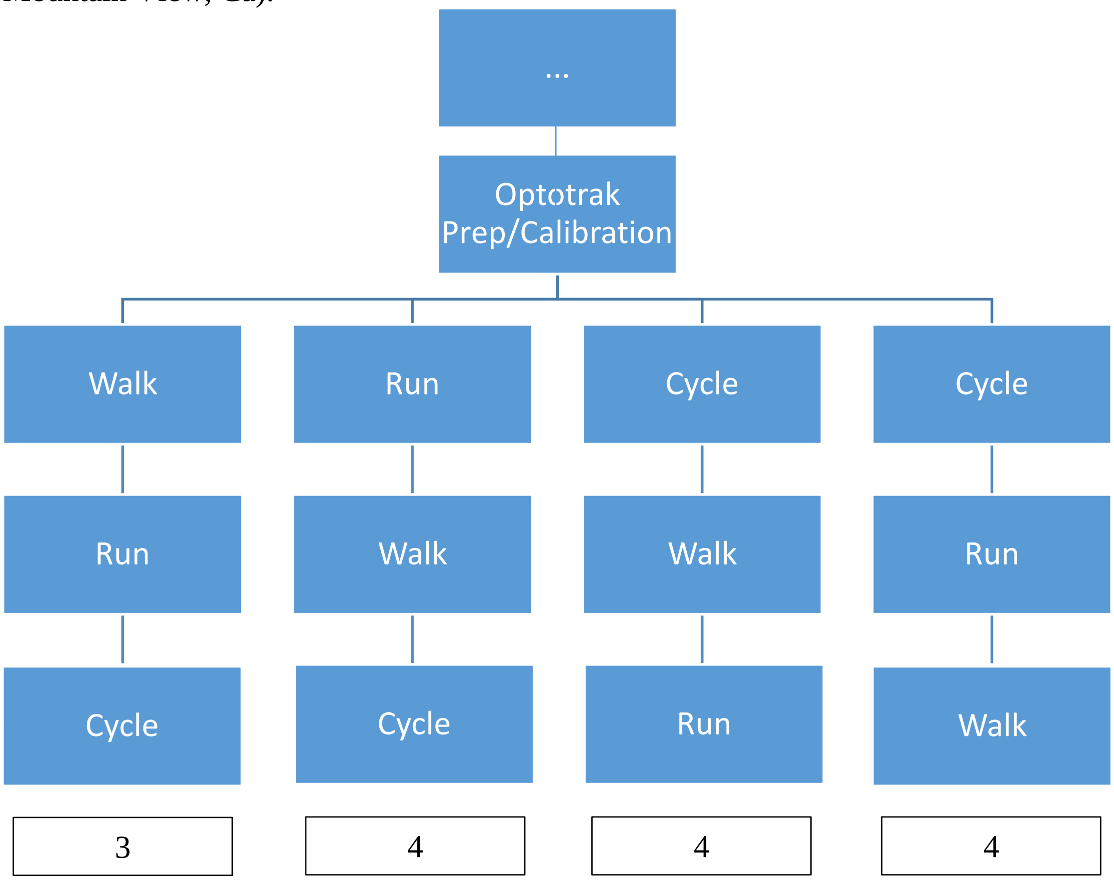

Figure 10.1: Breakdown of potential quasi-randomization protocols, with the number of instances each protocol occurred.

Appendix G: Bicycle Ergometer Set-Up

Methods of fitting the bicycle ergometer for each participant is crucial in assuring any differences found can attributed to true differences between activities, and confounded by differences in how the ergometer was set up. Lieberman (2007) stated that a 5% change in saddle height drastically alters kinematics by up to 35% and moments by up to 16%. It is therefore vital that a repeatable and reliable method be used to fit the ergometer to the participants to reduce inter-individual variance as much as possible. de Vey Mestdagh (1998) provides guidelines for setting handlebar reach and height based on anthropometric data to obtain correct upper body posture that have previously been used to fit ergometers (Bini et al., 2016). Several other methods to determine saddle positioning have been debated including the Lemond method, Thomas and Hamley method, heel method and Holmes method (Fonda et al., 2014). These methods use ratios of leg segment lengths (which can vary in efficacy between individuals) to determine saddle height and are performed during a static measurement (Bini et al., 2011; Fonda et al., 2014).

Static measurements of optimal saddle height have recently been called into question. The Holmes method (adjusting the static knee angle between 25° to 30°) and the Thomas and Hamley method (adjusting saddle height to 109% of inseam length) resulted in different saddle heights for the same person (Peveler et al., 2005; Peveler et al., 2008; Peveler et al., 2011). Further, these static set ups have produced knee angles out of the recommended 25° to 30° knee angle during dynamic cycling (Peveler et al. 2005). Thus, a dynamic bike fit has been advocated in order to optimally and accurately adjust saddle height (Fonda et al., 2014).

In a recent study by Bini et al. (2016), these concepts were confirmed and a novel bike fitting method was proposed. Compared to static measurement of knee angle at BDC, dynamic measurements were significantly smaller at BDC. However, at the 3 o'clock position, hip, knee and ankle angles were similar. It was concluded that to fit a bike statically, with the accuracy of a dynamic fit, a knee angle 60° should be set with the pedal at the 3 o'clock position during a static bike fit. Further, the saddle fore-aft position and position of the knee over the pedal axle can be adjusted concurrently. This method was thus, implemented in the current study. A 25mm saddle adjustment range for all measurements were allowed to accommodate the cyclist's preference, since it was expected that the participants would move in their seat to get to a comfortable position if the saddle were not adjusted to their comfort. Results from the bike fit from the current study are displayed in Figure 10.4 and Table 10.1.

Figure 10.2: Bicycle ergometer set up – participant knee, hip and arm angles measured during static bike fitting, adopted from Bini et al., 2016.

Table 10.1: Mean static joint angles during the bike set up. Measurements were taken while the crank arm was positioned at the 3 o'clock position.

Measurement	Angle (°)
Knee Angle	62.9 (3.8)
Hip Angle	87.5 (5.7)
Arm-Torso Angle	49.1 (4.9)

Note: Values are presented as mean (SD).

Appendix H: Instrumented Pedal

The instrumented pedal utilized in the current study was custom designed and manufactured by Novatech Measurements Ltd (Figure 10.5). The instrumented pedal has a capacity of ±1kN in the x-direction and ±3kN in the y-direction. In the current study, forces were below 150N in the x-direction and 400N in the y-direction. Calibration values are displayed in Table 10.2, comparing the results of the calibrations performed in the current study, to those previously experimentally tested for other investigations. The pedals were calibrated in the Y and X directions (Figure 10.6/Figure 10.9), tested for hysteresis (Figure 10.7) and drift (Figure 10.8)

Table 10.2: Instrumented Pedal Calibration Results

Parameter		Current Study – Post Collection	Gatti (2016)
Y-Direction	Force Cal	321.37 N/V	320.41 N/V
	Hysteresis	0.0002%	0%
	Drift	-2×10^{10} N/s over 10 minutes	0.0002 N/s over 90 minutes
X-Direction	Force Cal	107.0 N/V	102.40 N/V

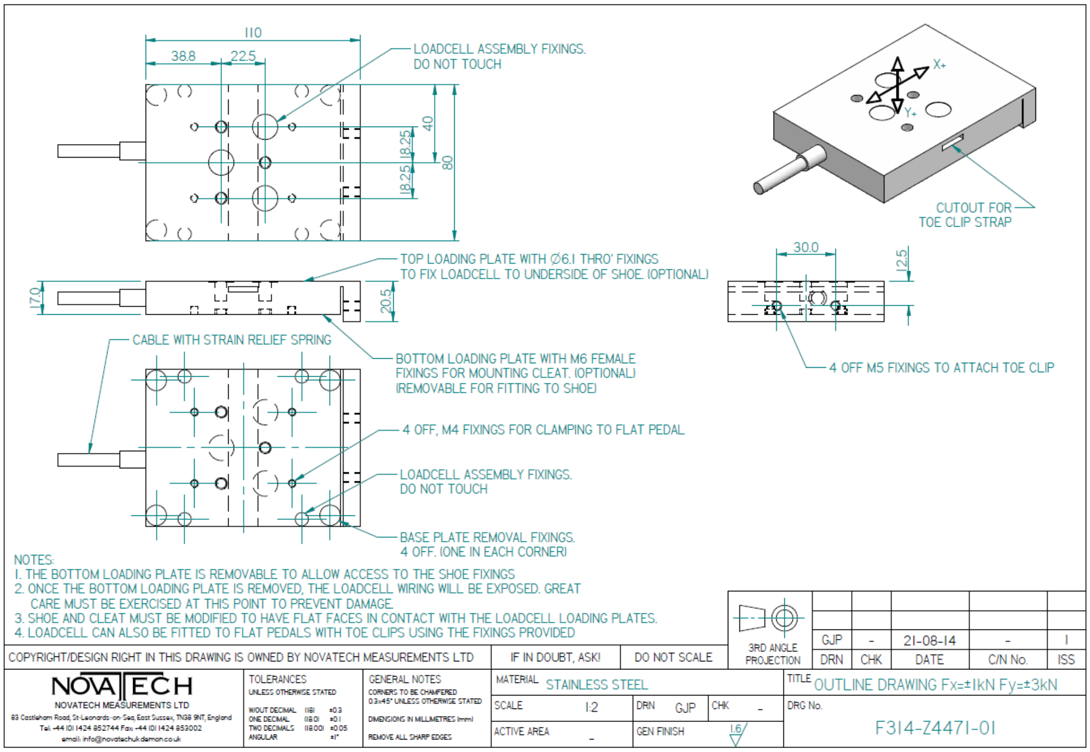

Figure 10.3: Schematic diagram for the instrumented pedal, provided by the manufacturer.

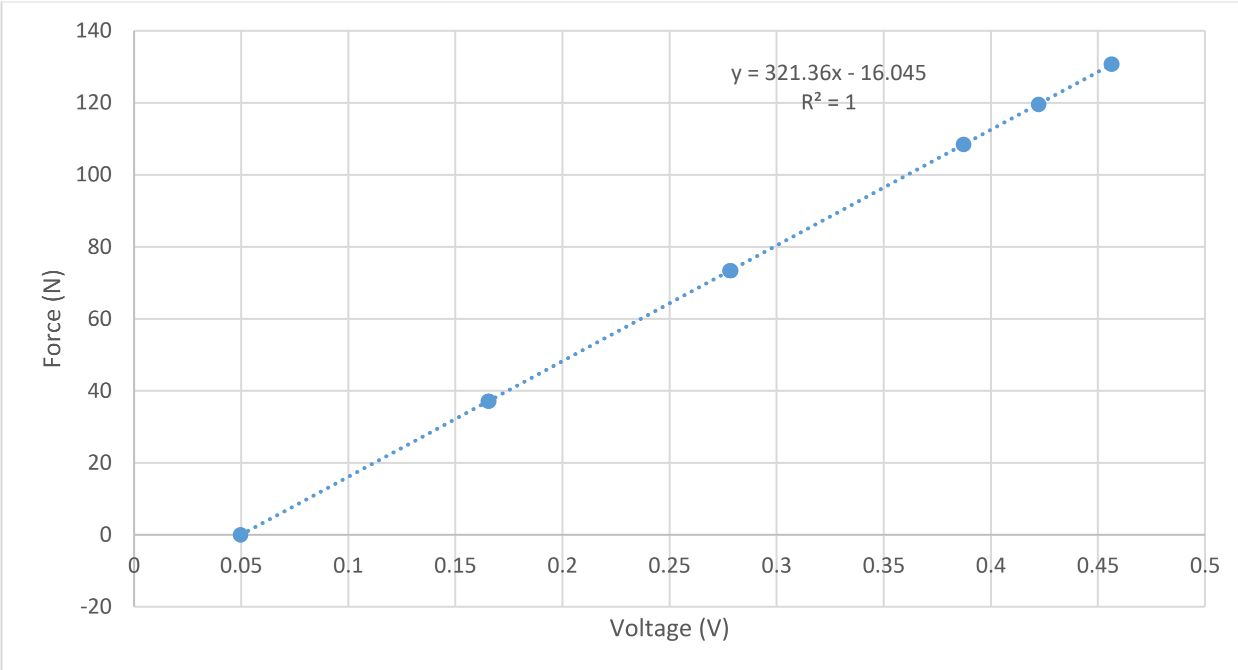

Figure 10.4: Force calibration curve for the y-direction of the instrumented pedal.

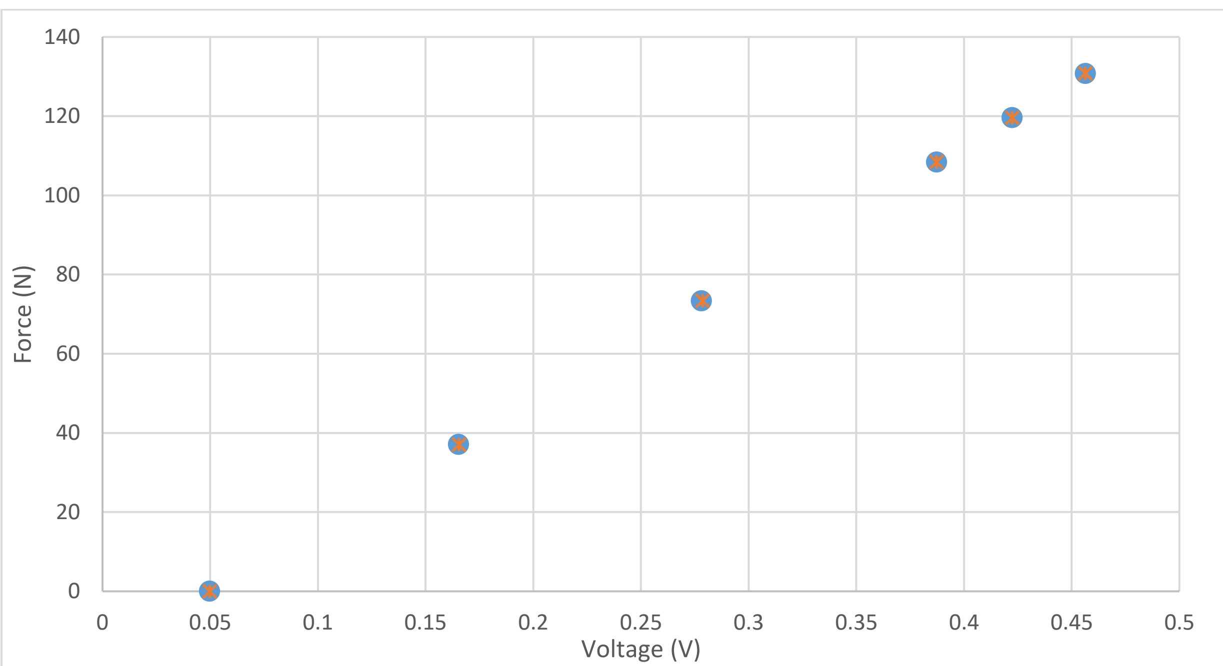

Figure 10.5: Hysteresis Curve for the y-direction of the instrumented pedal.

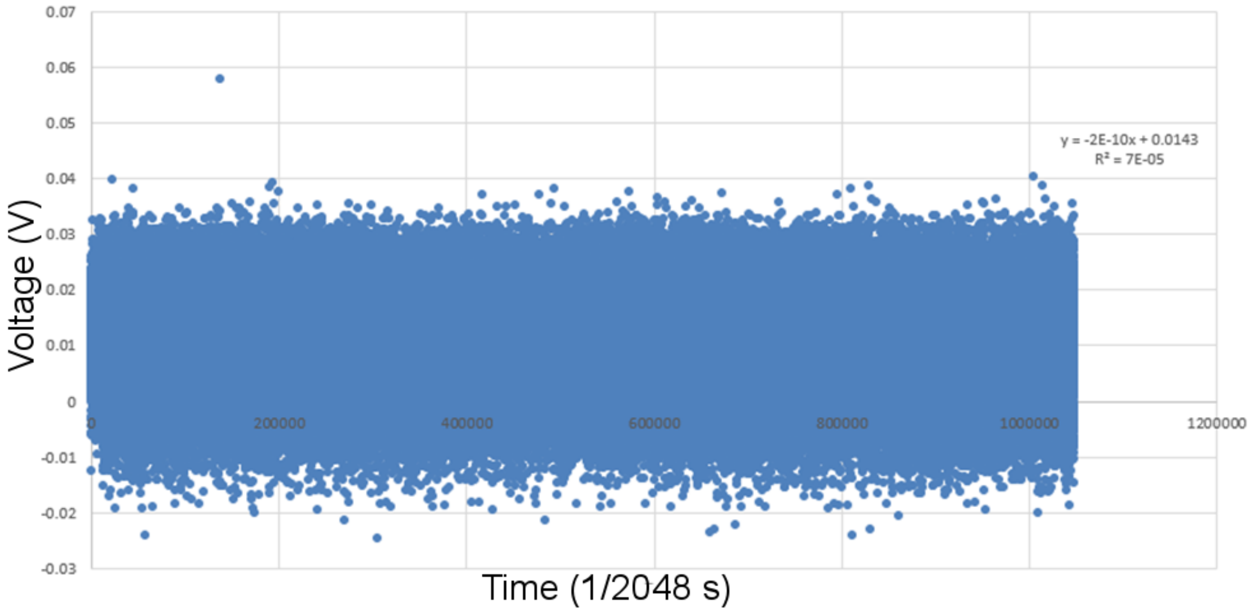

Figure 10.6: Drift Trial for the y-direction of the instrumented pedal.

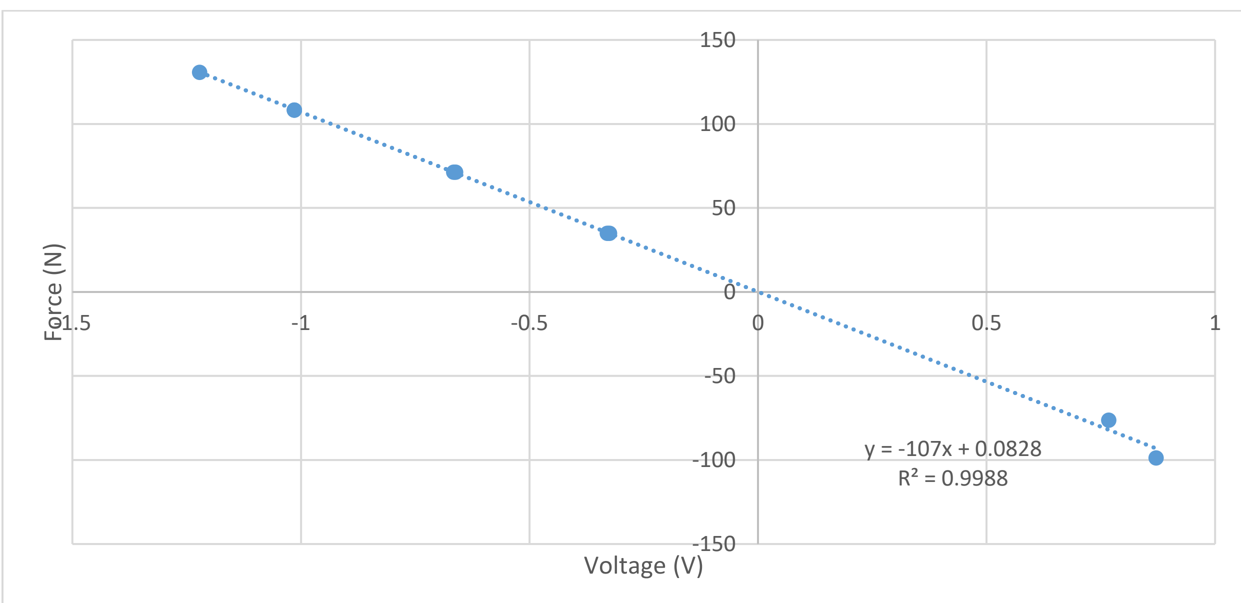

Figure 10.7: Force calibration curve for the x-direction of the instrumented pedal.

Appendix I: Co-Contraction Indices

Table 10.3: Co-contraction indices for each muscle grouping, for each activity over stance/downstroke.

Activity	VLLG	VMMG	VLBF	VMST
Walking	4.0 (2.2)	3.0 (1.4)	4.5 (2.4)	4.0 (2.1)
Running	21.4 (8.4)	19.1 (9.0)	22.9 (12.3)	17.1 (11.2)
Cycling	4.3 (3.7)	3.7 (2.8)	4.1 (3.5)	2.7 (2.1)

Note: Data is presented as mean (SD).

Table 10.4: Co-contraction indices for each muscle grouping, for each activity over the initial phase.

Activity	VLLG	VMMG	VLBF	VMST
Walking	4.5 (3.9)	2.7 (1.4)	7.7 (3.2)	5.8 (3.2)
Running	10.4 (5.1)	8.8 (5.3)	14.8 (9.2)	9.5 (5.0)
Cycling	3.0 (3.3)	1.4 (0.9)	3.3 (3.3)	1.4 (2.2)

Note: Data is presented as mean (SD).

Table 10.5: Co-contraction indices for each muscle grouping, for each activity over the terminal phase.

Activity	VLLG	VMMG	VLBF	VMST
Walking	6.2 (3.4)	6.3 (3.0)	6.2 (4.1)	6.3 (4.1)
Running	28.7 (11.1)	27.3 (14.3)	26.3 (15.5)	21.4 (15.9)
Cycling	5.7 (5.3)	4.2 (5.9)	4.8 (3.8)	3.1 (2.1)

Note: Data is presented as mean (SD).

Appendix J: Segment Coordination and Coordination Variation Plots

Figure 10.10 and Figure 10.11 display the segment coordination plots for all three segment couplings analyzed. Figure 10.12 and Figure 10.13 display the coordination variability plots for all three segment couplings analyzed. It is important to note that the coordination variability plots do not represent the variability of the segment coordination plots but instead the intra-individual variability that is masked by averaging across all participants.

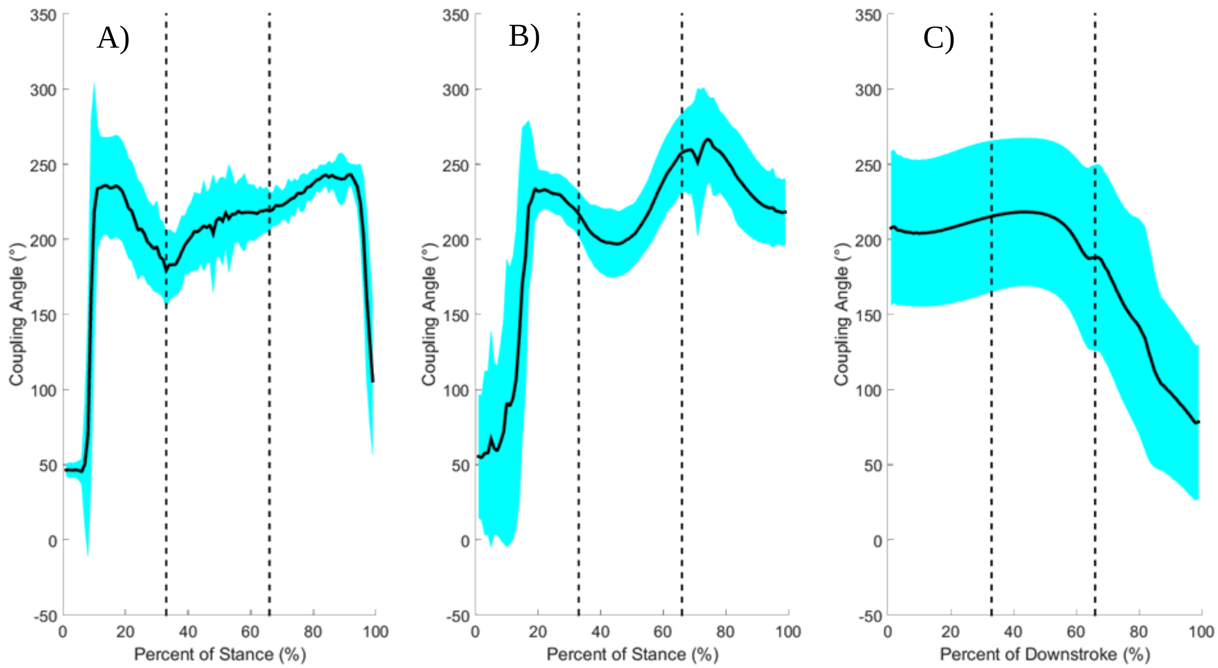

Figure 10.8: Segment coordination coupling angles for the sagittal shank/sagittal foot segment coupling for A) walking, B) running and C) cycling. The dashed lines separate the signals into thirds, to represent the early, mid and late stages.

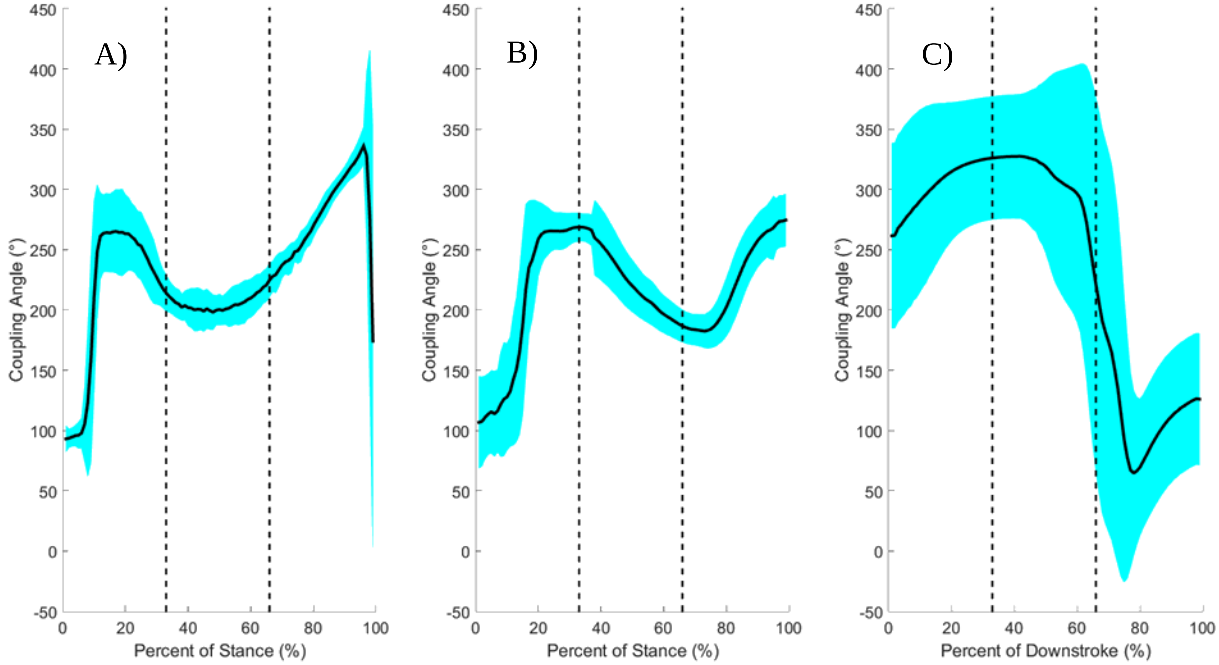

Figure 10.9: Segment coordination coupling angles for the sagittal thigh/sagittal shank segment coupling for A) walking, B) running and C) cycling. The dashed lines separate the signals into thirds, to represent the early, mid and late stages.

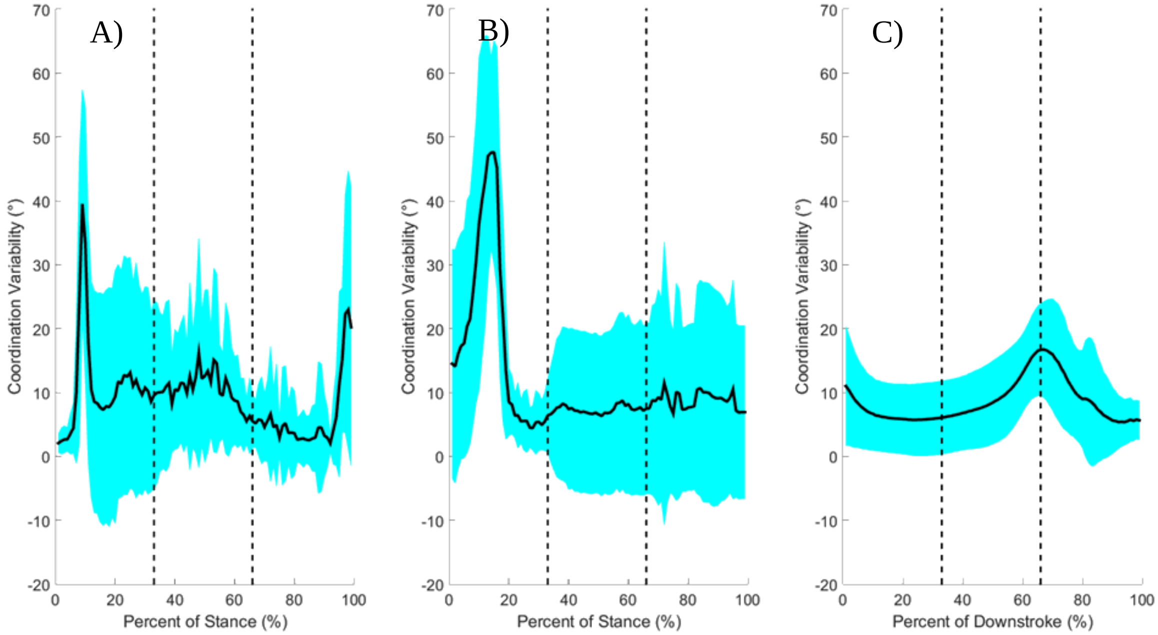

Figure 10.10: Coordination variability for the sagittal shank/sagittal foot segment coupling for A) walking, B) running and C) cycling. The dashed lines separate the signals into thirds, to represent the early, mid and late stages.

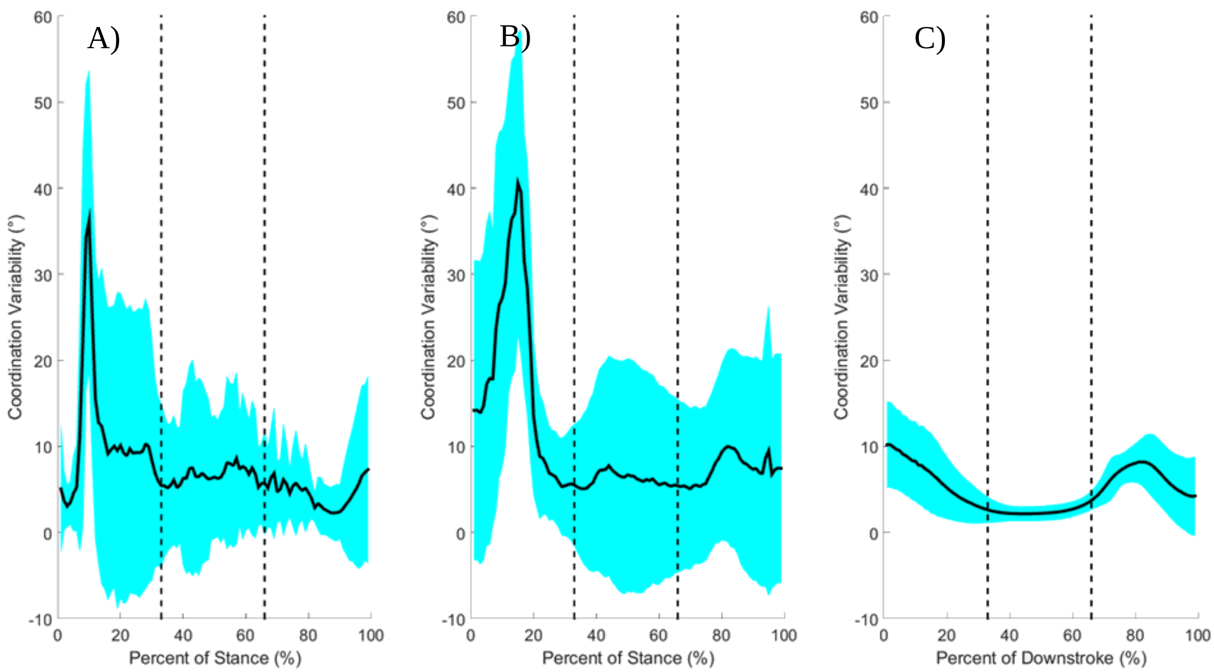

Figure 10.11: Coordination variability for the sagittal thigh/sagittal shank segment coupling for A) walking, B) running and C) cycling. The dashed lines separate the signals into thirds, to represent the early, mid and late stages.

Appendix K: Full ANOVA Results

K.1 Overall ANOVA Results

Table 10.6: ANOVA Results for Kinematic and Kinetic Analysis

Signal	DoF	Sum Square	Mean Square	F Value	Sig	G-G Correction	Partial Eta^2
Peak GRF	2	16682555.17	834277.584	317.713	0.0001	No	0.964
Peak KFM	2	42.300	21.150	235.763	0.0001	No	0.952
ROM	2	18214.985	10157.501	106.614	0.0001	No	0.899
Max Knee Angle	2	30185.296	15092.648	218.638	0.0001	No	0.948
Min Knee Angle	1.219	4900.639	4021.136	104.655	0.0001	Yes	0.897

Note: Significance was determined with an alpha level of 0.05 and are identified by an underlined p=value.

K.2 Dynamic Joint Stiffness ANOVA Results

Table 10.7: ANOVA Results for Dynamic Joint Stiffness

Signal	DoF	Sum Square	Mean Square	F Value	Sig	G-G Correction	Partial Eta^2
Stance/Downstroke	1.182	0.074	0.062	107.944	0.0001	Yes	0.900
Initial	1.182	0.084	0.071	64.192	0.0001	Yes	0.843
Terminal	2	0.054	0.027	37.021	0.0001	No	0.755

Note: Significance was determined with an alpha level of 0.05 and are identified by an underlined p=value.

K.3 Co-Contraction Index ANOVA Results

Table 10.8: ANOVA Results for Vastus Lateralis – Gastrocnemius Lateralis Co-Contraction

Signal	DoF	Sum Square	Mean Square	F Value	Sig	G-G Correction	Partial Eta^2
Stance/Downstroke	1.327	0.257	0.194	61.194	<u>0.0001</u>	Yes	0.836
Initial	2	0.040	0.020	16.053	<u>0.0001</u>	No	0.572
Terminal	2	0.447	0.223	59.908	<u>0.0001</u>	No	0.833

Note: Significance was determined with a Bonferroni correction (p = α/4) and are identified by an underlined p=value.

Table 10.9: ANOVA Results for Vastus Medialis – Gastrocnemius Medialis Co-Contraction

Signal	DoF	Sum Square	Mean Square	F Value	Sig	G-G Correction	Partial Eta^2
Stance/Downstroke	1.176	0.181	0.091	37.952	<u>0.0001</u>	Yes	0.791
Initial	1.185	0.035	0.029	21.207	<u>0.0001</u>	Yes	0.680
Terminal	1.334	0.392	0.294	29.849	<u>0.0001</u>	Yes	0.731

Note: Significance was determined with a Bonferroni correction (p = α/4) and are identified by an underlined p=value.

Table 10.10: ANOVA Results for Vastus Lateralis – Biceps Femoris Co-Contraction

Signal	DoF	Sum Square	Mean Square	F Value	Sig	G-G Correction	Partial Eta^2
Stance/Downstroke	1.192	0.276	0.231	24.941	<u>0.0001</u>	Yes	0.694
Initial	1.322	0.080	0.060	12.720	<u>0.002</u>	Yes	0.536
Terminal	1.225	0.347	0.284	19.588	<u>0.0001</u>	Yes	0.640

Note: Significance was determined with a Bonferroni correction (p = α/4) and are identified by an underlined p=value.

Table 10.11: ANOVA Results for Vastus Medialis – Semitendinosus Co-Contraction

Signal	DoF	Sum Square	Mean Square	F Value	Sig	G-G Correction	Partial Eta^2
Stance/Downstroke	1.032	0.163	0.158	20.230	0.001	Yes	0.628
Initial	1.262	0.033	0.026	12.890	0.002	Yes	0.518
Terminal	1.050	0.247	0.236	16.903	0.001	Yes	0.585

Note: Significance was determined with a Bonferroni correction (p = α/4) and are identified by an underlined p=value.

K.4 Coordination Variability ANOVA Results

Table 10.12: ANOVA Results for Sagittal Foot – Sagittal Shank Coordination Variability

Signal	DoF	Sum Square	Mean Square	F Value	Sig	G-G Correction	Partial Eta^2
Early	2	74.600	37.300	0.534	0.593	No	0.076
Mid	2	119.934	59.967	0.598	0.558	No	0.069
Late	2	215.939	107.969	1.385	0.270	No	0.366

Note: Significance was determined with a Bonferroni correction (p = α/2) and are identified by an underlined p=value.

Table 10.13: ANOVA Results for Sagittal Thigh – Sagittal Shank Coordination Variability

Signal	DoF	Sum Square	Mean Square	F Value	Sig	G-G Correction	Partial Eta^2
Early	2	42.432	21.216	0.369	0.695	No	0.070
Mid	1.346	113.824	84.544	0.814	0.415	Yes	0.0267
Late	1.190	13.726	11.538	0.190	0.712	Yes	0.026

Note: Significance was determined with a Bonferroni correction (p = α/2) and are identified by an underlined p=value.

K.5 IT Band Impingement Measures ANOVA Results

Table 10.14: ANOVA Results for IT Band Impingement Measures

Signal	DoF	Sum Square	Mean Square	F Value	Sig	G-G Correction	Partial Eta^2
ITBIZ(t)$_{per\_rep}$	1.077	0.025	0.024	7.293	0.027	Yes	0.648
ITBIZ(t)$_{60}$	1.164	175661.5	150876.4	3.917	0.079	Yes	0.650
ITBIZ(t)$_{cumul\_load}$	2	8103.1	4051.6	2.062	0.164	No	0.228
ITBEX	2	86760.3	43380.1	20.950	0.0001	No	0.807

Note: Significance was determined with a Bonferroni correction (p = α/6) and are identified by an underlined p=value.