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THE CLINICAL USEFULNESS OF VECTOR CODING VARIABILITY IN FEMALE RUNNERS WITH AND WITHOUT PATELLOFEMORAL PAIN

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THE CLINICAL USEFULNESS OF VECTOR CODING VARIABILITY
IN FEMALE RUNNERS WITH AND WITHOUT PATELLOFEMORAL PAIN

DISSERTATION

A dissertation submitted in partial fulfillment of the
requirements for the degree of Doctor of Philosophy in the
College of Education
at the University of Kentucky

By

Tommy Joseph Cunningham

Lexington, Kentucky

Director: Dr. Timothy L. Uhl, Associate Professor of Rehabilitation Sciences

Lexington, Kentucky

2012

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ABSTRACT OF DISSERTATION

THE CLINICAL USEFULNESS OF VECTOR CODING VARIABILITY IN FEMALE RUNNERS WITH AND WITHOUT PATELLOFEMORAL PAIN

It has been suggested that Patellofemoral Pain (PFP) may be the result of a coordinate state which exhibits less joint coordination variability. The ability to relate joint coordination variability to PFP pathology could have many clinical uses; however, evidence to support clinical application is lacking. Vector coding's coupling angle variability (CAV) has been introduced as a possible analysis method to quantify joint coordination variability. The purpose of this study was to assess the clinical usefulness of CAV measures from a dynamical systems perspective. This involved establishing the precision limits of CAV measures when physiological conditions are held constant, altering control parameters of knee pain and population then determining if the observed changes in CAV were clinically meaningful.

20 female recreational runners with PFP and 21 healthy controls performed a treadmill acclimation protocol then ran at a self-selected pace for 15 minutes. 3-D kinematics, force plate kinetics, knee pain and perceived exertion were recorded each minute. CAV were calculated for six knee-ankle combinations for 2 sets of 5 non-consecutive stride cycles at each capture period. Data were selected for the PFP group at a high (≥ 3) and low ($\leq \text{high}-2$) pain level in a non-exhausted state (< 14). Healthy data were used from the 11th minute of the running. Levels of agreement were performed between the 2 sets of CAV measures for both populations, a paired t-test compared low to high pain CAV measures and independent t-tests compared populations at the high pain state.

Several CAV measures showed a significant increase in value with an increase in pain and were significantly greater for the PFP group. None of the observed changes exceeded the precision limits of all CAV measures investigated. These results do not agree with previous claims that less variability is indicative of pathology but rather the opposite. This suggests that there might be an optimal amount of variability to maintain a healthy coordinate state with deviations in any direction being detrimental. However; due to the volatile nature of CAV measures, the clinical use of CAV is not recommended using current

analysis methods since changes observed weren't considered clinically meaningful.

KEYWORDS: Coordination, Gait, Kinematics,
Dynamical Systems, Running

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May 9, 2012

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Chapter 1: Introduction

Background

Variability in movement control strategies (MCS) is advantageous for the learning and performance of gait. (Stergiou, et al. 2006) From a dynamical systems perspective of motor control a MCS is a function of the complex interactions of three multidimensional control parameters; environment (e.g. external conditions), organism (e.g. population, physiological state) and task (e.g. movement goal). (Newell, et al. 1993) Subtle changes to a control parameter during seemingly identical movements results in inherent fluctuations to a MCS, while a large change to a control parameter results in a substantial shift in MCS and may be classified as a different movement entirely. Variability in MCS should be evident in variances in joint coupling kinematics and the order parameters used for kinematic analysis of given movements. (Kelso, et al. 1991, Newell, et al. 1993, Turvey 1990) Inferences to dynamical systems theory based on indirect measures are limited by the mathematical tools used to quantify joint coordination kinematics and their subsequent variances. The technique of “vector coding” (Heiderscheit, et al. 2002, Tepavac and Field-Fote 2001) is one method that is implemented in various forms that is used to quantify joint and segment couplings, providing a continuous measure of coordination commonly referred to as a coupling angle (CA). Vector coding quantifies the angle-angle diagram of two joints or segments for each point during a movement cycle. The standard deviation among measured cycles, coupling angle variability (CAV), has been introduced as a useful measure to quantify the amount of variability during different portions of the gait cycle and is suggested as being indicative of the amount of variability within a MCS. (Heiderscheit, et al. 2002) Further, the mean CAV value over gait intervals have been calculated and used for comparisons of MCS responses to changes in control parameters. (Dierks and Davis 2007, Ferber, et al. 2005, Heidercheit 2000, Heiderscheit, et al. 2002, Maulder 2011, Miller, et al. 2010, Pollard, et al. 2005, Wilson, et al. 2008)

One perspective on musculoskeletal injuries of the lower extremity from a dynamical systems perspective has been suggested by Hamill et al. (Hamill, et al. 1999) They suggest that a person's inability to exhibit variations in their joint coordination patterns can increase the frequency of loading of soft tissue and eventually lead to an overuse condition and pathological state. (Hamill, et al. 1999) This theory has yielded many investigations that have examined CAV measures to describe changes in MCS variability in relation to differences in population (Dierks and Davis 2007, Ferber, et al. 2005, Heiderscheit, et al. 2002, Maulder 2011), knee pain (Heidercheit 2000), sex (Pollard, et al. 2005), skill level (Wilson, et al. 2009, Wilson, et al. 2008) and speed. (Miller, et al. 2010) Despite CAV measures growing use, little evidence of its effectiveness to support the dynamical systems perspective to lower extremity injury has been presented. Heterogeneous sample populations (Ferber, et al. 2005), small sample sizes (DeLeo, et al. 2004, Wilson, et al. 2008) inconsistent analysis procedures and differing dependent measures (Mullineaux, et al. 2008) are some of the many reasons evidence to support a dynamical systems perspective to lower extremity injury is lacking. Preliminary data suggests that CAV measures may behave according to dynamical systems theory. (Miller, et al. 2010) A feature of this preliminary investigation showed a predictable response of CAV measures to a theoretical control parameter, the Lorentz Attractor (Lorenz 1963), and to changes in speed. Small changes of speed has routinely been identified as having a large effect on running gait parameters such as kinematics, stride parameters (Mann and Hagy 1980, Williams 1985) and variability (Li, et al. 1999) and may suggest speed is a possible control factor that may affect CAV measures and should be controlled in studies utilizing CAV as an output measure. Other factors should be considered. When measuring changes in CAV measures corresponding to a physiological control parameter during a violent movement such as running, inability to control for extraneous changes in a physiological system may lead to error in measures and misinterpretation of results. It is necessary to adequately identify and limit all possible sources of

error and study changes in the physiological system in a controlled manner to interpret MCS variability.

To be clinically useful, an analysis tool must be reliable (measures are repeatable) and valid under a physiological construct being studied (responds predictably to a physiological change). The clinical usefulness of CAV measures has yet to be established. Camera resolution and sampling frequency have improved with motion capture technology progression and are possibly more accurate in detecting variation in movement than when CAV was introduced as a measure. (Mullineaux, et al. 2008) Various gait cycle processing techniques such as normalization procedures and gait cycle event definitions can have possible effects on CAV measures which can mislead interpretation of results. (Mullineaux, et al. 2008) Inconsistent methods in calculation of CAV, joint and segments being compared and intervals in which mean CAV values are composed makes comparison among studies difficult and thus establishment of clinical validity difficult. CAV measures of interest calculated using refined measurement and analysis techniques should be investigated and the reliability assessed to discern measurement error and physiological variation when control parameters and MCS have remained constant. It has been suggested that test-retest reliability should be performed on any clinical measure prior to its clinical interpretation and is a necessary procedure to establish measurement error. (McGinley, et al. 2008) This procedure has largely been ignored for CAV measures during gait. (Maulder 2011) The precision of any measurement method or analysis tool has associated error. CAV only has a full scale range of 0 to 82°. (Batschelet 1981) Previous literature has only observed changes in CAV of upwards as to 10°, approximately 12% of full scale range. The reliability of CAV measures should demonstrate consistent values below 10° in order to interpret these observed changes to be considered clinically meaningful. Accurately assessing CAV measures is achievable with current motion capture technology and analysis methods. Reliability assessment is necessary to understand the scope in which clinical meaning can be inferred from CAV measures within a dynamical systems context.

Identifying changes in MCS supports one construct that variability can measure meaningful physical changes in gait and is a valid clinical measure of a physiological change. In order to test this construct, one approach is to change a single control parameter while attempting to keep all other control parameters relatively constant. One such example of this change could be found in runners with Patellofemoral Pain (PFP), where the onset of knee pain may indicate an important change in an organism control parameter. (Heidercheit 2000) This simple change in the organism can be used to evaluate the validity of vector coding in distinguishing between MCS within a population of injured runners (e.g. pre/post pain onset) or between populations (e.g. healthy/injured). Knee pain as a control parameter during treadmill running has been evaluated in the past when assessing CAV measures. (Heidercheit 2000) Small changes in pain and a low magnitude of pain is cited as a limitation when assessing the effects of a reduction in pain (Heidercheit 2000) and population differences. (Heiderscheit, et al. 2002) The average change in pain magnitude was less than 2 on a visual analogue scale and may not have been clinically meaningful. (Crossley, et al. 2004, Piva, et al. 2009) Reliability values were not reported and it is speculated that the limited findings might also be the results of confounding factors such as sex, state of fatigue, running speed, environment (overground vs. treadmill) and injury.

A majority of reported PFP cases are female. (Taunton, et al. 2002) Females have been shown to have distinct differences in their running joint kinematics (Csintalan, et al. 2002) and variability measures (Pollard, et al. 2005) possibly subjecting them to more excessive lateral patellofemoral joint forces (Lee, et al. 1994, Lee, et al. 2001, Mizuno, et al. 2001) than males would experience. (Powers 2003) Similarly, joint kinematic changes have been observed after exhaustive treadmill runs (Derrick, et al. 2002, Dierks, et al. 2008) and speed of locomotion. (Li, et al. 2005) These observations suggest that a cohort may exist within the PFP population whereby females exhibit specific kinematic patterns different from other PFP groups (Powers 2003) and large changes in fatigue or speed can alter running kinematics all together. Previous

studies have not adequately controlled all of these factors and possible control parameters which may have affected CAV results.

The reliability of vector coding techniques when all physiological control parameters remain constant has never been assessed. The reliability of any CAV measures has not been investigated in female recreational athletes or an injured population during treadmill running. Also, the validity of CAV measures as a clinically useful measure from a dynamical systems context to orthopaedic injuries has not been established. A study is needed to address these limitations. The expected result of this study will provide information on the validity of a CAV measures as a useful clinical tool. This will be determined by first; reporting the reliability of CAV measures when no physiological changes have occurred and then; by evaluating observed changes in CAV measures when physiological changes have occurred in the context of the precision of CAV measures like any valid biomechanical tool.

Statement of the Problem

The intra-subject reliability of MCS as measured using CAV derived from vector coding analysis has not been established for healthy runners and runners with PFP. Further, it is unknown whether a transition of MCS resulting from a single change in an organism control parameter is detectable using CAV measures. It is unknown if CAV measures can delineate a possible change in MCS when a clinically significant change in knee pain occurs in a PFP population. It is also unknown if CAV measures can detect a different MCS between a PFP population in a painful state from healthy controls. CAV measures used in the literature have been insensitive to many changes in control parameters in most previous lower-extremity gait analyses. Therefore, more sensitive CAV measures need to be investigated to establish their clinical utility when interpreted from a dynamical systems perspective.

Purpose

The purposes of the present study are to:

1. Determine the intra-subject reliability of CAV measures when control parameters remain constant for a PFP and healthy population of runners.
2. Determine the clinical validity of CAV measures when a physiological state control parameter (knee pain) is altered for runners with PFP.
3. Determine the clinical validity of CAV measures to distinguish when the control parameter of population differs. (PFP runners vs. healthy runners)

Research Hypotheses

Aim 1: Asses the test-retest reliability of vector coding order parameters for healthy runners and runners with PFP between a first set of five non-consecutive gait cycles and a second set of five non-consecutive gait cycles.

Hypothesis 1: Changes to CAV measures in both healthy and PFP populations with physiological variables held constant will be less than 10% of the 81° full scale range capable of CAV measures (8.1°).

Aim 2: Asses the validity of vector coding variability measures in response to a clinically significant increase in knee pain (pain change ≥ 2).

Hypothesis 2: There will be a statistically significant and clinically meaningful decrease in vector coding variability values with an increase of pain during a treadmill run for runners with PFP.

Aim 3: Asses the validity of vector coding variability measures in distinguishing between healthy runners and runners with PFP.

Hypothesis 3: Vector coding variability values will be significantly greater and clinically meaningful in healthy runners than runners with PFP.

Significance of the Study

The use of vector coding variability measures has recently been introduced to the literature as a possible tool to distinguish between physiological conditions and populations. Although mathematically reliable and valid, vector coding variability measures have not been empirically proven to be clinically useful from a dynamical systems perspective. As with any biomechanical measure, it must be reliable and valid within a specific clinical construct to be justified as clinically useful. The reliability of CAV measures needs to be established to understand the measurement error limits associated with CAV measures. These values will be expressed as the level of agreement with a 95% confidence interval. This approach defines the boundary limits that need to be exceeded for a change in CAV to be clinically meaningful. If a change in CAV due to a parameter change is found to exceed the 95% confidence interval, the investigated parameter would indicate that there is reasonable confidence that the change in physiological state is beyond measurement error and has clinical validity as a true change affecting the CAV. If a difference in CAV is found to exceed the established reliability limits when comparing a PFP and healthy population, it will mean that CAV is clinically valid in the construct that it can detect a change in population.

Assumptions

For this study it will be assumed that physiological variations are inherent during movement and manifest in joint coordination variability measures. Concerning manipulation and measurement of parameters, it will be assumed that only a single control parameter will be manipulated (knee pain) while others (fatigue, sex, preferred speed, environment) will remain constant. Moreover, the measured pain and perceived exertion (fatigue) values reflect a physiological state of the MCS further defining a construct for female recreational runners at a preferred running speed.

Limitations

- Worn bearings were repaired on one of the belts of the treadmill during the study. This repair required 15 subjects to use the treadmill's opposite belt and force plate for data collection. Before the bearings were fixed, it is possible that low frequency noise generated from the rotating bearings could have affected force data.
- Electrical noise was intermittently introduced to the force plate signals from other laboratory equipment. This noise was typically greater than 80 Hz.
- Markers adhered to the feet would occasionally fall off or become loose. This essentially altered the rigid body assumption of the foot if all markers were to be used. Visual screening of foot markers was performed and markers were chosen to represent the foot that routinely remained attached to the foot.
- Onset of pain had already occurred prior to reaching a preferred running speed in some individuals not allowing comparison between a fresh state and painful state at the preferred running speed.
- RPE values changed more than 1 from a fresh state to the maximum pain reached in most individuals. This did not allow comparison between a fresh state to the maximum state of pain while controlling for perceived fatigue.

Delimitations

- The population of this study was delimited to 20 female recreational runners diagnosed with PFP and 21 otherwise healthy female recreational runners aged 18-45 years of age.
- The treadmill speed and the order of the running in which subjects performed was delimited to their preferred walking speed for a time period of 3 minutes, followed by a 3.3 m/s run for 2 minutes, followed by a 25

minute run at their preferred running pace and finishing at their initial preferred walking pace until recovered.

- Measurements of physiological state are delimited to a Numeric Pain Rating Scale (NPRS) for knee pain and a perceived rated exertion scale for fatigue. NPSR measurements were restricted to integer increments.
- All physiological measurements were only recorded every minute.
- Equipment used was delimited to a Bertec dual force gauge treadmill (TM-09-PBertec, Columbus, OH) for the collection of ground reaction forces (1200Hz), a combination of 15 Eagle and Eagle4 cameras (Motion Analysis Corporation, Santa Rosa CA), accuracy of <1mm and collection frequency of 300Hz.
- Data reductions were delimited to the selection and normalization of gait phases, calculation of joint angles and CAV measures. Analyses of joint coupling combinations were delimited to knee flexion/extension, valgus/varus and internal/external rotation; and ankle plantar/dorsi flexion and inversion/eversion.

Operational Definitions

- Clinically Useful Measure: A measure that is sufficiently reliable to observe real changes and responds predictably to a physiological change.
- Patellofemoral Pain (PFP): Subjects who report retro or peri-patellar pain at after exclusion of some knee conditions as determined by a certified physical therapist or athletic trainer. Exclusion criteria included: intra-articular pathology, peripatellar tendinitis and bursitis, plica syndromes, Sinding Larsen's disease, Osgood Schlatter's disease, neuromas and a history of surgery or traumatic injury to the knee.(Thomee, et al. 1999) Diagnosis was confirmed if a minimum pain level of 3 out of 10 on a numeric pain scale during the course of the study protocol.
- Healthy Recreational Runner: Person who runs a minimum of 10 miles per week (6 miles per week if reduced due to symptoms).

- PFP Recreational Runner: Person who ran a minimum of 10 miles per week or 6 miles per week if reduced due to symptoms.
- Movement Control Strategy (MCS): The lower limb neuromuscular response to the control parameters of environment, organism and task to physically perform a stride cycle.
- Joint Coupling: Simultaneous coordination between two joints throughout a movement cycle (e.g. stride cycle).
- Coordination: The relative timing and magnitude of kinematic variables describing between two or more adjacent or non-adjacent segments.
- Order Parameter: State of a MCS as defined by the dependent measures which respond to changes in a control parameter.
- Control Parameter: Independent variable of one of three categories; environment, organism or task, any of which can be manipulated to alter the Movement Control Strategy.
- Coupling Angle (CA): Vector coding output measure assessing coordination between two joints or segments over a stride cycle. Units are in degrees.
- Coupling Angle Variability (CAV): Vector coding output measure assessing coordination variability between two joint or segments over several stride cycles. Is the circular standard deviation the mean coupling angle.
- Coupling Angle Variability Mean (CAV_{Mean}): Mean coupling angle variability over a selected interval of a normalized gait cycle.
- Coupling Angle Variability Local Maximum (CAV_{Max}): Maximum coupling angle variability value within a specified interval of stride.
- Location of Coupling Angle Variability Local Maximum (CAV_{MaxLoc}): Location of a CAV_{Max} value within a specified interval of stride. Units are in percentage of stride from heel strike.
- Knee Joint: Articulations between the thigh and shank.
- Ankle Joint: Articulations between the shank and rear-foot.

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Chapter 2: Review of Literature

The purpose of this study is to evaluate the clinical reliability and validity of a mathematical technique termed “vector coding” when used to describe joint coordination variability of the lower extremity during running gait. Vector coding has been introduced to gait analysis under the premise that it is a tool used to quantify inherent variation in joint coupling interactions present in gait. Accordingly, this review of literature is intended to present the argument for use of vector coding as a legitimate clinical gait analysis tool and discuss the additional requirements needed for assessment of reliability and validation of vector coding variability measures. This chapter is organized to first describe movement variability in the context of a dynamical systems perspective. Secondly, the progression of measurement methods to quantify joint coordination variability is presented beginning with continuous relative phase variability and progressing to vector coding variability. Benefits and limitations of each respective tool used are discussed. The use of vector coding variability measures is further reviewed in the context of clinical gait analysis and orthopaedic injuries from a dynamical systems perspective with limitations of previous literature summarized and critiqued for improvements. Lastly, the overuse condition of Patellofemoral Pain is discussed as a valid construct to study vector coding variability measures.

Movement Variability from a Dynamical Systems perspective

Theoretical Background

Quantification of movement variability is a necessary aspect of understanding how people navigate and respond to their environment via their motor control system. (Davids, et al. 2004, Glazier, et al. 2006, Hamill, et al. 2006, Stergiou, et al. 2006, Wheat and Glazier 2006) Previously viewed as detrimental to a control system or measurement noise, movement variability is now thought to be a necessary functional characteristic inherent within a healthy motor control system. (Glazier, et al. 2006, Hamill, et al. 2006, Stergiou, et al. 2006) A motor control system is incredibly complex when considering even just a fraction of the total amount of degrees of freedom (DOF) in a body (10^2 joints, 10^3 muscles, 10^3 cell types and 10^4 neurons) that must be considered to perform a task. (Kelso 1995, Wheat, et al. 2002) Generating a trajectory of a limb that is satisfactorily repeatable seems like an insurmountable task and variability within a control system is expected.

The dynamical systems theory of motor control was introduced by Bernstein. (Bernstein 1967) Recognizing the overwhelming complexity and timeliness required to perform a task, Bernstein proposed that the DOF within a given control system behave according to a non-linear dynamical system and its complexity is a product of the number of elements in the system and the dimensionality of the system. The DOF can be reduced by the number of equations of constraint that can describe the system embodied by coordinative structures. (Fitch, et al. 1982, Tuller, et al. 1982) Coordinative structures essentially reduce the DOF by using groups of muscles that often span several joints and act as a single unit. Coordinative structures are autonomous of each other and are independently tuned through spontaneous adjustments to control parameters which dictate response. (Fitch, et al. 1982, Turvey 1990) The autonomous nature of coordinative structures enables the ability of a combination of coordinative structures to perform complicated tasks with relatively simple adjustment of input parameters. (Fitch, et al. 1982)

Coordinative structure response to changes in control parameters and interaction among structures is dependent on the demands required of the task being performed, the external conditions of environment in which the task is being performed and the internal conditions of the organism performing the task. (Davids, et al. 2003) The resulting response of coordinative structures in the performance of a task gravitates towards a preferred attractor state otherwise termed a coordinate state. (Turvey 1990) A coordinate state will exhibit inherent variability as a result of the actions of coordinative structures and is categorized as anatomical, mechanical or physiological. (Turvey, et al. 1982) This type of variability is specific to the context defined by the aforementioned combination of task, environment and organism (Turvey, et al. 1982) and switching between coordinate states requires a functional variability which enables utilization of differing coordinative structures. (Kelso 1995) Switching between coordination states behaves according to non-equilibrium systems with the resulting coordinate state and its variability described by an identifiable and measureable order parameter. (Turvey 1990) Adherence to all of these characteristics must be apparent to be considered a plausible theory from a dynamical systems perspective of motor control.

Application

The landmark work of Kelso (Kelso 1984, Kelso 1995, Kelso, et al. 1991, Kelso, et al. 1979) presented a valid construct of application of dynamical systems to joint coordination. Kelso was able to identify a control parameter and measure responses to an order parameter that behaved according to dynamical systems theory. Summarizing, observing the inter-segmental coordination of oscillating fingers, Kelso identified frequency of movement as a control parameter and the relative phase between the two oscillating segments as a method of measurement to assess coordination. With adjustment of frequency, Kelso was able to observe variability patterns that exhibited predictable coordinate state shifts in accordance with all the characteristics defining a system in non-equilibrium. Several key observations were made. The order parameter

of relative phase was bi-modal with values either in-phase or out of phase. Also, coordination patterns displayed a hysteretic effect and switched to in-phase with increasing frequency but did not transition to out of phase with decreasing frequency. The transition to in-phase was preceded with a variability increase which allowed spontaneous and non-linear switching of coordinate states. All of these findings coincide with a dynamical systems framework and provide a good example of what is required of an application using dynamical systems as a theoretical perspective of study.

An inability to transition between coordinate states has been suggested to be indicative of a pathological locomotion pattern during walking. (Hamill, et al. 1999, Van Emmerik, et al. 1999) Less joint coordination variability has been observed between the thorax and pelvis in patients with Parkinson's disease when compared to healthy controls while walking. (Van Emmerik, et al. 1999) The lower amount of observed variability is thought to inhibit coordinative state changes resulting in a pathological condition. In this construct, walking has been suggested as a plausible construct to apply dynamical systems with velocity as control parameter and relative phase variability between the thorax and pelvis as order parameters. Velocity has been identified as a likely control parameter in walking gait (Schoner, et al. 1990, van Emmerik and Wagenaar 1996) and the transition from walk to run. (Diedrich and Warren 1995) Relative phase has also been identified as a plausible order parameter in the assessment of joint coordination as previously discussed.

An inability to transition between coordinate states has also been suggested to be indicative of a pathological lower extremity coordination pattern which can lead to overuse injuries of the knee during running. (Hamill, et al. 1999) In presentation of this theory, the order parameter of continuous relative phase variability was suggested to measure response from a control parameter not identified but related to population. The two populations investigated were individuals with patellofemoral pain and healthy individuals. Since the introduction of this theory, continuous relative phase has been shown to only be a valid assessment of joint coordination for movements that oscillate at a 1:1

ratio and are sinusoidal. (Peters, et al. 2003) This is applicable for movements of the pelvis and thorax but not of segments of the lower extremity during running. (van Emmerik, et al. 2004) Further, identification of a control parameter is required to methodically manipulate to determine if this theory coincides with dynamical systems theory. Taking these limitations into consideration, stride length, frequency and knee pain were assessed as possible control parameters with vector coding variability introduced as a possible order parameter to detect change. (Heidercheit 2000, Heiderscheit, et al. 2002) Results to support this theoretical application of dynamical systems were minimal. Details concerning these limitations of these studies that may have led to inconclusive results are discussed in more detail later in this chapter. The limitations of identifying a control parameter and measuring predictable change of an identified order parameter still needs to be done to consider this a plausible theory from dynamical systems perspective.

Summary & Conclusion

Joint coupling and coordination is necessary to perform a task (Arutyunyan, et al. 1968) and inherent variation in the coordinative structures are necessary. (Amazeen, et al. 1998, Turvey 1990) Variability of a coordinative structure is thought to manifest in joint coordination variability; (Turvey 1990) therefore joint coordination variability has become a topic of study in the application of dynamical systems theory. Constructs necessary for a study to coincide with dynamical systems theory include an identifiable control parameter and measurable order parameter. The order parameter must predictably respond consistent with a non-linear system in non-equilibrium. (Turvey 1990) Kelso provided an example of manipulation of a control parameter (frequency) and a predictable response of an order parameter (relative phase) when describing the control of joint coordination between oscillating fingers. (Kelso 1995) These results have provided the framework for applying dynamical systems theory to other movements (Haken, et al. 1985) including locomotion. (Li, et al. 1999, Schoner, et al. 1990, van Emmerik and Wagenaar 1996) It has

been suggested that less variability inhibits switching of coordinate states and thus can be indicative of a pathological condition. (Van Emmerik, et al. 1999) Further, it has been suggested that less variability in lower extremity joint coordination can produce damage to soft tissue by not allowing switching among coordinate states thus creating an overuse pathology in the knee. (Hamill, et al. 1999) This theory still requires identification of a control parameter and predictable response of an order parameter to be considered a valid application of dynamical systems theory.

Development of Vector Coding

Analysis of coordination patterns from a dynamical systems perspective requires quantifying joint or segment movement patterns and the pattern variation across a number of cycles.(Hamill, et al. 1999) Continuous relative phase is a common method that has been employed to investigate movement patterns in a dynamical systems context; however, its validity as a measurement tool is highly questionable. A gait analysis method commonly referred to as “vector coding” has emerged as a potentially clinically useful gait analysis tool to assess the inherent variability exhibited in lower extremity coordination patterns. (Heiderscheit, et al. 2002, Tepavac and Field-Fote 2001) Simply, vector coding (VC) and similar but different methods preceding its development, all involve quantification of an angle-angle diagram (Grieve 1968) and statistical analysis of the respective output variable and its variation. This variation has been interpreted as indicative of inherent variability in movement control strategies. “Vector Coding” is an inclusive term that refers to many processes. To avoid confusion and for reasons which will be detailed in this review, proper referencing of “Vector Coding” should involve four separate method processes to reference; 1. quantifying the angle-angle diagram, 2. deriving continuous joint coordination measures, 3. deriving continuous joint coordination variability measures and 4. statistical analysis.

Continuous Relative Phase

Other analysis methods have been developed to investigate movement patterns that do not involve angle-angle diagrams (Hamill and van Emmerik 2000), the most common being continuous relative phase (CRP) (Kelso 1995). CRP is described in detail elsewhere (Kelso 1995, Peters, et al. 2003, van Emmerik, et al. 2005, Wheat and Glazier 2006) but a brief description of CRP in the context of development of vector coding is warranted. CRP is useful as a measure of coordination in that it allows many cycles to be compared, maintains temporal (velocity) and spatial (angular) characteristics of segment data and gives a continuous measure of coordination throughout the entire movement cycle. (van Emmerik, et al. 2004) This has made CRP a popular analysis method in motor control literature in accessing control parameter relationships with order parameters defined by the specific coordination patterns and their variability. Changes in coordinative states are theorized to be accompanied or preceded by abrupt changes in coordination variability and can be measured using the standard deviation of the coordination measure. (Haken, et al. 1985, Kelso 1995, Turvey 1990) CRP-variability (CRPV) has been observed to change during transitions between attractor states during performance of simple bimanual tasks (Kelso 1995, Turvey 1990) and other more complicated tasks such as juggling (Post, et al. 2000), wrist movement (Amazeen, et al. 1998), trunk-pelvis rigidity in Parkinson's disease (Van Emmerik, et al. 1999) and gait transition speed. (Diedrich and Warren 1995, van Emmerik and Wagenaar 1996)

CRPV has been introduced as a possible clinical measure in identifying lower extremity movement patterns in orthopaedic injuries where less coordination variability

Table 2.1 Common abbreviations and definitions used within the text for joint and segment coupling relationships of the lower extremity.

Abbreviation	Definition
Joints	
HR	Hip Internal/External Rotation
HF	Hip Flexion/Extension
HA	Hip Ab/Adduction
KR	Knee Internal/External Rotation
KF	Knee Flexion/Extension
KV	Knee Valgus/Varus
AI	Ankle Inversion/Eversion
AF	Ankle Plantar/Dorsi Flexion
AA	Ankle Ab/Adduction
Segments	
TR	Thigh Internal/External Rotation
TF	Thigh Flexion/Extension
TA	Thigh Ab/Adduction
SR	Shank Internal/External Rotation
SF	Shank Flexion/Extension
SA	Shank Ab/Adduction
RI	Rearfoot Inversion/Eversion
RF	Rearfoot Flexion/Extension
RA	Rearfoot Ab/Adduction
FI	Forefoot Inversion/Eversion
FF	Forefoot Flexion/Extension
FA	Forefoot Ab/Adduction

is indicative of a constrained state and thus susceptible to injury. (Hamill, et al. 1999) In preliminary investigation of this theory, Hamill et al. (Hamill, et al. 1999) compared CRPV measures for all three thigh (TR,TF,TA)-SR segment couplings (proximal segment rotation-distal segment rotation; Table 2.1) and SR-RI between healthy individuals and those with Patellofemoral Pain Syndrome (PFP) during running. Mean CRP values were calculated over intervals of swing, stance, stride and four functional intervals of stance. Authors found greater CRPV values in 14 of 28 measures in healthy individuals but reported no standard deviations of the mean CRPV or performed any statistics comparing groups. Surprisingly, the largest difference found was near mid-stance for TA-SR where PFP CRPV values were more than 4 times that of healthy ($\sim 50^\circ$ vs. 12°) but were not mentioned in the discussion. Despite there being inconclusive distinction between groups, the authors concluded that their data demonstrated support for their theory and suggested these methods for further investigation of injuries. The clinical relevance of these findings remains unclear. (DeLeo, et al. 2004)

Following introduction of CRPV to the literature, various other lower extremity tasks, populations and injuries have been investigated using CRPV in the dynamical systems context; (Dierks 2005, Dierks and Davis 2004, Dierks and Davis 2007, Gittoes and Wilson 2010, Heiderscheit, et al. 1999, Li 2000, Miller, et al. 2010, Miller, et al. 2008, Wheat, et al. 2002) however, limitations to CRP have been presented suggesting CRP and its subsequent variability may be invalid measures of many lower extremity movements. Use of CRP requires violation of several assumptions if to be used in gait analysis. (Peters, et al. 2003, Wheat and Glazier 2006) CRP has been suggested to only be used in analysis of segments that oscillate in a sinusoidal fashion (Diedrich and Warren 1995) and at a 1:1 ratio, (Peters, et al. 2003) typical of motions observed by Kelso (Kelso 1995) and Van Emmerik et al. (van Emmerik and Wagenaar 1996, Van Emmerik, et al. 1999). Excluding hip motions in the sagittal plane, this assumption is not met for lower extremity joint movements. (Heidercheit 2000, Heiderscheit, et al. 2002) Additionally, normalization procedures are required to CRP values that

can effect results (Peters, et al. 2003, Wheat and Glazier 2006, Wheat, et al. 2002) and lead to variability measures that can be erroneous and propagated with errors that can be misinterpreted as variability (Wheat and Glazier 2006, Wheat, et al. 2002). Also, the output measure of CRPV can be difficult for clinicians to interpret and relate to conceptually. (Kurz and Stergiou 2002, Peters, et al. 2003, Tepavac and Field-Fote 2001, Wheat and Glazier 2006)

Considering the non-sinusoidal movements often analyzed in gait and the need for clinicians to interpret results, the clinical usefulness of CRPV as a valid indicator of movement variability in gait is questionable. To account for these limitations, two very similar VC techniques were introduced to investigate variability in lower extremity movement patterns as a useful clinical measure. (Hamill, et al. 2000, Heidercheit 2000, Heiderscheit, et al. 2002, Tepavac and Field-Fote 2001). The main distinction between the two VC techniques is the output measure of variability used for comparison. In one technique, the standard deviation of a coupling pattern is found over a portion of a movement cycle and is used to evaluate the variability of the movement system. (Heiderscheit 2000) In the 2nd technique, the same coupling variability pattern is found but the variability is further normalized through several procedures to account for the amount of cycles and magnitude differences between cycles. (Tepavac and Field-Fote 2001) Development of these analysis techniques is further discussed.

Quantifying Angle-Angle Diagrams

VC's origins reside in quantification of relative motion diagrams. Relative motion diagrams, or angle-angle diagrams, are a useful tool in qualitatively describing coordination patterns between two body segments or joints. First introduced by Grieve (Grieve 1968) as a simple method to interpret movement patterns in the lower-limb during gait, efforts quickly concentrated on quantifying these diagrams for comparison between movements. Regardless of the specific methods used to quantify and compare angle-angle diagrams, two processes remain consistent. First, the angle-angle diagrams are constructed between two oscillators (segments or joints) and digitized to a curvilinear path, and secondly;

discrete variables are obtained from the digitized curve for statistical analysis of variations between different curves from the same person.

A novel method was presented by Freeman (Freeman 1961) to digitize geometric shapes using a chain encoding technique. This chain encoding technique superimposes a grid over a curvilinear line and encodes each successive point on the line using an 8 point scale with numbers ranging from 0 to 7, each representing increments of 45° as possible movements from one point on the grid to another. An entire curvilinear line would be described by a sequence of digits (Figure 2.1).

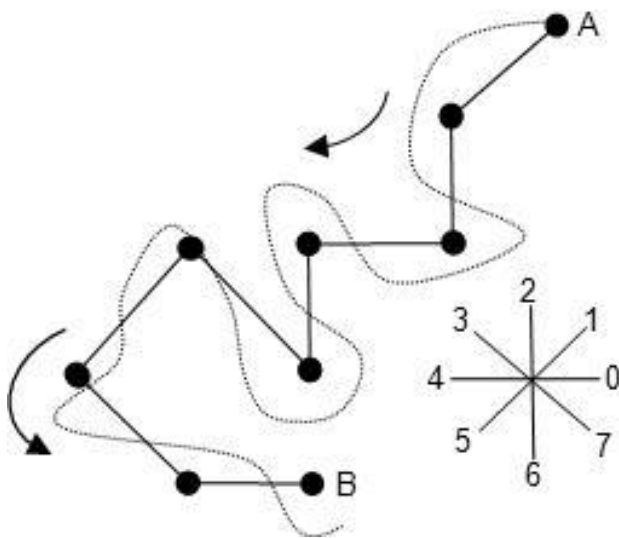


Figure 2.1 Example of an 8-element chain encoded curve starting at point A and proceeding to point B. The encoded curve is expressed as curve AB=56463570.

To compare similarities between two encoded curves, Freeman (Freeman 1961) introduced a discrete measure, the cross correlation coefficient ($R_{a,b}$), to compare the degree of similarity between two movement curves, **a** and **b** (Eq. 2.1).

$$R_{a,b} = \frac{1}{N} \sum_{i=1}^N a_i * b_{i+j} \quad j = 0, 1 \dots N - 1 \quad \text{Eq. 2.1}$$

Whiting and Zernicke (Whiting and Zernicke 1982) applied the cross-correlation technique to gait analysis of experienced male runners. In their study, angle-angle diagrams of the knee and hip joint during three treadmill activities (slow walk (.83 m/s), fast walk (1.66 m/s) and a run (3.33 m/s)) were coded and the

peak value of the cross-correlation coefficient, termed the recognition coefficient (R), was calculated. The resulting R values ranged from -1 to +1 with a -1 indicating two movement patterns 180° out of phase, 0 indicating no correlation between movements and +1 indicating two curvilinear paths have the same size, shape and orientation. (Sparrow, et al. 1987) A substantial finding was that R was dependent on the number of encoding points used. To account for this, R was calculated using 8 different amounts of encoding points (n=35...+5...70) and the mean R value of the 8 amounts was considered a stable representation of the changes in motor control patterns.

There were several limitations to the technique used by Whiting and Zernicke addressed by Sparrow (Sparrow, et al. 1987). Resolution of the encoded sequence was dependent on the resolution of the superimposed grid and unequally spaced data points made comparison between two movements difficult as this method did not account for the length between data points. Additionally, this encoding technique was developed to aid in computational efficiency which at the time would suffer unless an estimate of the

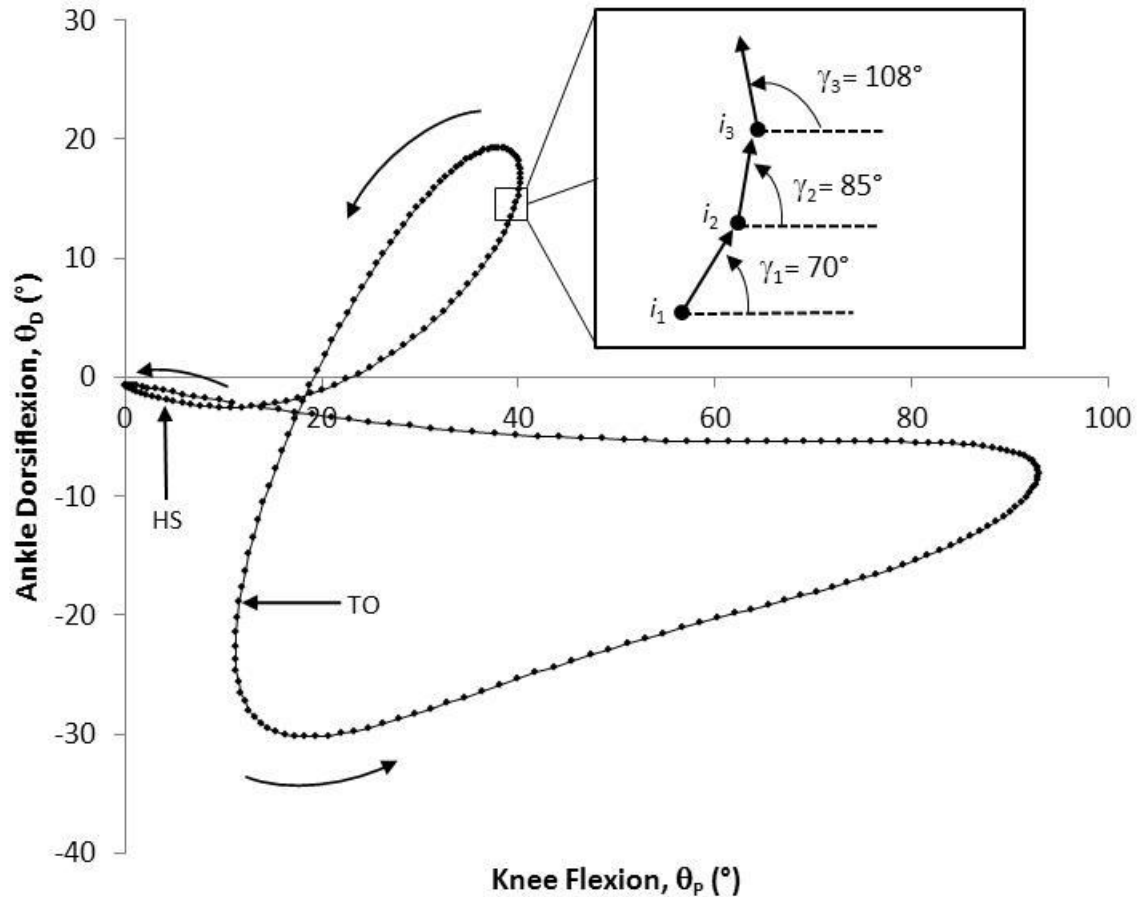


Figure 2.2 Graphical representation of the quantification of an Ankle Dorsiflexion- Knee Flexion (angle-angle) diagram during running gait using methods described by Sparrow.(Sparrow, Donovan et al. 1987) Sparrow calculates the angle γ of a line segment ($i-i+1$) relative to the right horizontal (----) and the length of the corresponding segment and uses the information for trigonometric shape analysis and cross-correlation statistical comparison. Typical convention has the proximal joint/segment (θ_P) on the horizontal axis and the distal joint/segment (θ_D) on the vertical axis.

curve was constrained to the grid intersection points. Considering these limitations and the progression of computational efficiency, Sparrow introduced a modified encoding technique where the angle (γ) was calculated between successive data points relative to the right horizontal (Figure 2.2). The mathematical steps to calculate the angle and length of a segment were not explicitly described by Sparrow (Sparrow, et al. 1987) possibly because they are considered common trigonometric knowledge. Sparrow did however graphically represent angles as ranging from 0-360° inferring this is the range of expected

values but did not explicitly state this. Also introduced by Sparrow (Sparrow, et al. 1987) was a modification to the original cross-correlation function to account for the possible differences in segment lengths between each data point (a_i , b_{i+j}) and incorporate angular data (Eq. 2.2)

$$Ra, b j = \frac{1}{N} \sum_{i=1}^N \cos(\gamma_{a,i} - \gamma_{b,i+j}) * \frac{a_i}{b_{i+j}}, j = 0, 1 \dots N - 1 \quad \text{Eq. 2.2}$$

where $a * b_{i+j}$ is equal to the cosine of the angle between the i th segment of shape **a** and the $i+j$ th segment of shape **b** shown earlier (Eq. 2.1). While this cross-correlation measure was used to assess intersegment variation between two cycles, other shape analysis measures using angular data and segment lengths were also demonstrated. These shape parameters included centroids, areas, heights, widths and perimeters. (Barry 1980, Hershler and Milner 1980, Sparrow, et al. 1987)

Although Sparrow's technique assessed many shortfalls of the chain encoding technique, the use of cross correlation as a variation measure limits the use to linear relationships, (Sidaway, et al. 1995) two curves and only gives a general measure of the similarity between the entire curves. Sidaway et al. (Sidaway, et al. 1995) introduced a measure, normalized root mean squared error (NoRMS), that could be used to measure variation among several curves and be used on both linear and non-linear data. For this calculation, the mean angle–angle plot is calculated and the root mean square is then calculated over the series of trials and normalized with respect to the number of cycles. This calculation as summarized by Mullineaux (Mullineaux, et al. 2001) and descriptions consolidated by Wheat (Wheat and Glazier 2006) is shown (Eq. 2.3);

$$100 * \frac{\sum_{j=1}^k \frac{\sum_{i=1}^n x_A - x_{Ai}^2 + x_B - x_{Bi}^2}{n_j}}{R} \quad \text{Eq. 2.3}$$

where A and B denote the two segment or joint curves of interest, k is the number of cycles, n is the number of data points, R is the resultant excursion of the mean angle-angle curve over the entire cycle, x is the mean position of the segment or joint at the i th data point and x is the position of the segment or joint

at the i th point on the j th cycle. Multiplication by 100 was used for easier management of data. (Sidaway, et al. 1995) Limitations to this technique included use of linear statistics on non-linear data and limiting joint angles between 0° and 360° making the technique invalid if a joint were to rotate through 360° which is rare. (Sidaway, et al. 1995) Most notably, the variability of a movement was summarized into one output measure to summarize the entire cycle, similar to the cross correlation coefficient. If variability characteristics change throughout a movement, as it is theorized to do between phases within a stride in running gait, (Heiderscheit, et al. 2002) this technique would be inappropriate in many common scenarios. Moreover, if means between datasets differ greatly, normalization by the mean can yield misleading results. (Mullineaux, et al. 2001) The NoRMS technique has not been readily adopted as a tool in the biomechanics literature and is used in only a few instances (Crowther, et al. 2008, Crowther, et al. 2008, Crowther, et al. 2009, Robins, et al. 2006, Wheat and Glazier 2006).

Coupling Angle

An alternative approach to encoding was later introduced by Hamill (Hamill, et al. 2000) that adopted the encoding technique utilized by Sparrow terming the output variable γ a “coupling angle” (CA) and relating it to the field of biomechanics as a method to compare oscillating segments and incorporating circular statistics (Batschelet 1981) as a measure of coordination dispersion. Hamill, like Sparrow, did not show or reference the mathematics to calculate the CA but did state that the values should fall within 0° and 360° relative to the right horizontal. Further described by Hamill are interpretations of the meaning of the CA values between 0° and 360° at 45° increments when the proximal joint/segment oscillator is plotted on the horizontal axis and the distal joint/segment oscillator is plotted on the vertical axis of the angle-angle diagram. These descriptions were also represented schematically by Heiderscheit (Heiderscheit 2000)(Figure 2.3).

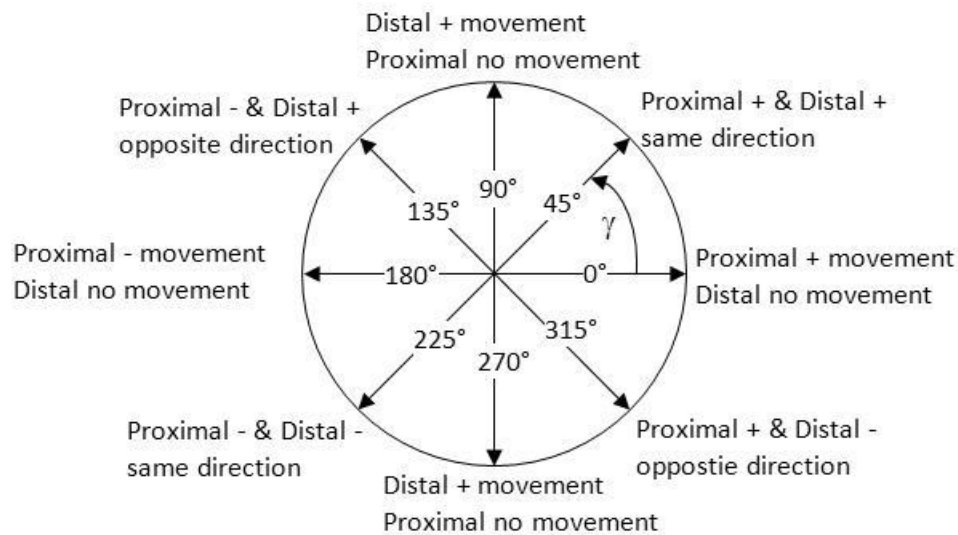


Figure 2.3 Illustration of the meaning of coupling angle values (γ) when a proximal oscillator is plotted on the horizontal axis and the distal oscillator is plotted on the vertical axis of an angle-angle diagram. Movement in the positive direction (+) and movement in the negative direction (–) are indicated. This chart was first presented by Heiderscheit (Heidercheit 2000)

The descriptions given by Hamill are as follows:

1. Values of 0°, 90°, 180° and 270° indicate movement of one oscillator.
 - a. Values of 0° and 180° indicate the distal oscillator is stationary and the proximal oscillator is moving.
 - b. Values of 90° and 270° indicate the proximal oscillator is stationary and the distal oscillator is stationary.
2. Values of 45°, 135°, 225° and 315° indicate equal relative movement between the two oscillators.
 - a. Values of 45° and 225° indicate equal amount of movement in the same direction.
 - b. Values of 135° and 315° indicate equal amount of movement in the opposite direction.

A similar interpretation of this was later presented by Chang et al. (Chang, et al. 2008) where terminology was expanded for intervals encompassing the 45° increments to describe rearfoot-forefoot coordination patterns but could be used for any moving oscillators. These categories of CA were: in-phase (22.5°-67.5° &

202.5°-247.5°), anti-phase (112.5°-157.5° & 292.5°-337.5°), rearfoot-phase (proximal, 337.5°-22.5° & 157.5°-202.5°) and forefoot-phase (distal, 67.5°-112.5° & 247.5°-292.5°). This system recognized that it is rare for only one joint to be moving at a time or equal movements of a joint to occur and wanted to categorize these movements. A more clinical interpretation of CA values has also been applied when interpreting SR-RI CA values where CA have been likened to a continuous excursion ratios between two segments (DeLeo, et al. 2004, Dierks and Davis 2007, Ferber, et al. 2005).

Coupling Angle Calculation & Inconsistencies

There are some areas of inconsistency in the methods presented in the literature that can produce different CA values and possible clinical misinterpretation. As mentioned previously, Sparrow et al. (Sparrow, et al. 1987) implied and Hamill et al. (Hamill, et al. 2000) stated that CA values should be between 0° and 360° but both did not show the mathematics required to calculate the CA. Hamill et al. (Hamill, et al. 2000) provides a figure in which the CA is calculated for a knee-ankle Flexion diagram (Figure 2.4) similar to that shown previously (Figure 2.2). Of note are the dashed lines at 0° and 360°. CA values should only be between 0° and 360° which would leave discontinuities at approximately 10%, 45% and possibly 55% and 58%, respectively. It is unclear as to whether the CA featured was vertically shifted manually or mathematically and the original curve left remaining only for reading purposes. It should be noted that CA should be contained within the 0° and 360° region if to be consistent with statements made by these authors. Mathematics for the calculation of CA were first reported by Heiderscheit et al. (Heidercheit 2000, Heiderscheit, et al. 2002) where it was also stated that values should be between 0° and 360° and calculated as shown (Eq. 2.4);

$$\gamma_i = \tan^{-1} \frac{\theta_{Di+1} - \theta_{Di}}{\theta_{Pi+1} - \theta_{Pi}} * \frac{180}{\pi} \quad \text{Eq. 2.4}$$

where θ_P is the proximal oscillator and θ_D is the distal oscillator and i is a point in the gait cycle. Unfortunately Eq. 2.4 does not help resolve the mentioned issues

as the output does not produce values ranging from 0° to 360° but values from 90° to 90°.

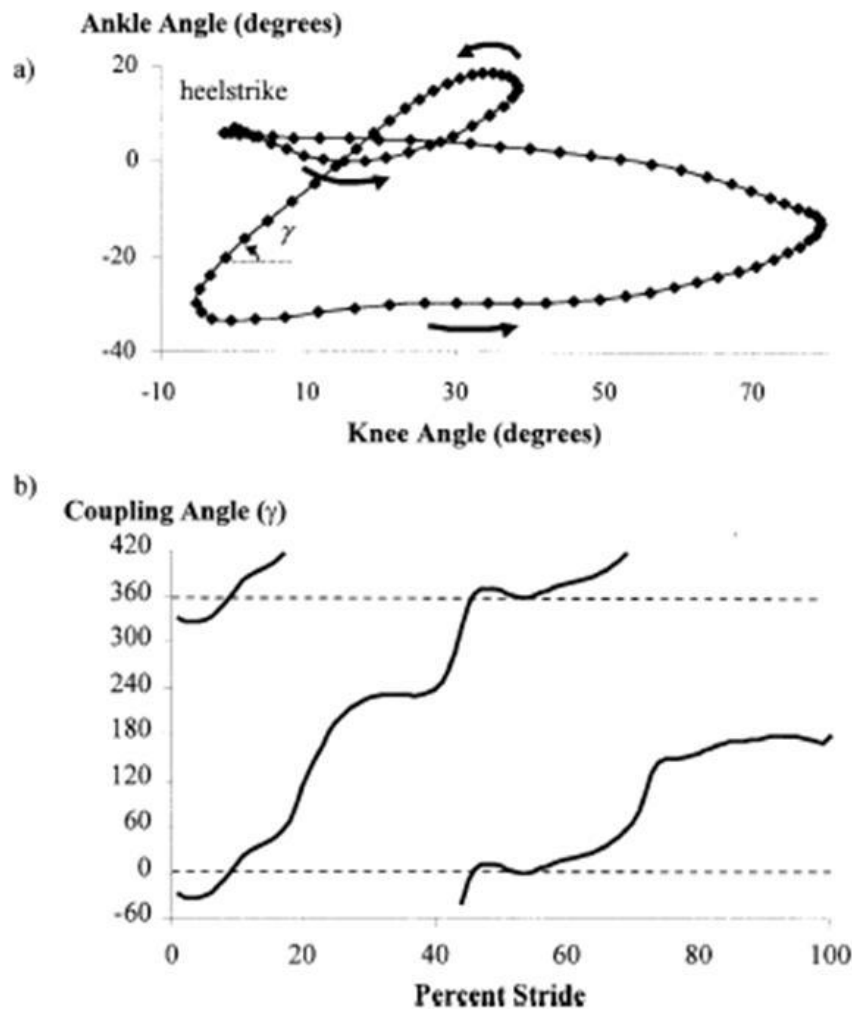


Figure 2.4 Figure describing an angle-angle diagram a) and the corresponding coupling angle b) first depicted by Hamill and reprinted with permission (Hamill, et al. 2000). Of note are the dashed lines at 0° and 360°. Calculated coupling angle values should only be between 0° and 360° which would leave discontinuities at approximately 10%, 45% and possibly 55% and 58%, respectively. It is unclear as to whether the coupling angle has been vertically shifted manually or mathematically and the original curve left remaining for reading purposes.

To achieve values outside a range of -90° to 90°, an additional procedure is required that is not mentioned in these references despite results that are within 0° to 360°. Sample data published (Figure 2.4) (Hamill, et al. 2000) and other studies referencing this procedure have similar conflicting results that show CA

within the stated range but cite methods that will not provide these results.
(Chang, et al. 2008, Gruber, et al. 2009, Wilson, et al. 2009, Wilson, et al. 2008)

Several studies have modified Eq. 2.4 slightly by finding the absolute value of the output (Eq. 2.5) which yields values ranging from 0° to 90° (Dierks and Davis 2007, Ferber, et al. 2005, Pohl and Buckley 2008, Pohl, et al. 2007) and choosing to use opposite angle-angle diagram axis convention. (Dierks and Davis 2007, Ferber, et al. 2005) It is unclear as to the reason for modification of Eq. 2.4 as reasoning is not cited; however, Dierks et al. (Dierks and Davis 2007) interprets meanings of coupling angle values below and above 45° ($\tan(45^\circ) < \text{or} > 1$) possibly implying this is the only information required if CA are to be examined in an excursion ratio context. Reasoning for using opposite angle-angle diagram axis convention was also not mentioned but does effect CA values.

$$\gamma_i = \tan^{-1} \frac{\theta_{Di+1} - \theta_{Di}}{\theta_{Pi+1} - \theta_{Pi}} * \frac{180}{\pi} \quad \text{Eq. 2.5}$$

A similar VC technique introduced by Tepavac and Field-Fote (Tepavac and Field-Fote 2001) similarly expanding on Sparrow's methods (Sparrow, et al. 1987) includes preliminary mathematical steps that can provide correct results. Tepavac and Field-Fote were not concerned with the coupling angle as an output measure; however, they do calculate the components of the CA between each point in a movement cycle which are the numerator and denominator in Eq. 2.4 ($\theta_{Di+1} - \theta_{Di}$ & $\theta_{Pi+1} - \theta_{Pi}$). Tepavac and Field-Fote use these values to calculate the magnitude of the CA (Eq. 2.6).

$$l_{i,i+1} = \sqrt{(\theta_{Di+1} - \theta_{Di})^2 + (\theta_{Pi+1} - \theta_{Pi})^2} \quad \text{Eq. 2.6}$$

The magnitude of the CA, like any length vector, can be used to calculate the sine or cosine of CA relative to the horizontal as presented by Tepavac and Field-Fote (Eq. 2.7 and 2.8).

$$\cos \gamma_i = \frac{\theta_{Pi+1} - \theta_{Pi}}{l_{i-i+1}} \quad \text{Eq. 2.7}$$

$$\sin \gamma_i = \frac{\theta_{Di+1} - \theta_{Di}}{l_{i-i+1}} \quad \text{Eq. 2.8}$$

Although not presented by Tepavac and Field-Fote, CA could then be recomposed using a variety of elementary trigonometric functions that would

yield results in the range of 0° and 360° as Sparrow(Sparrow, et al. 1987) intended and many methods cite as the theoretical CA range.

Corrections

Methods to achieve values between 0° and 360° consistent with Sparrow (Sparrow, et al. 1987) using tangent functions are further discussed in Appendix A in steps 1-3 using equations A.1 with A.2 or just A.3. Results using these equations are demonstrated empirically with sample data in figures A.1thru A.3. Wilson et al. (Wilson, et al. 2009, Wilson, et al. 2008) states that CA values will range between 0° and 180° but math provided does not yield values in that range. Equations A.4 and A.5 in Appendix A are two examples of equations that will yield intended results. The consequences of using inconsistent methods in calculation of CA as presented in this section are detailed in Appendix B. Table B.1 lists scenarios of possible methodical misinterpretations described in the literature and Figure B.1 shows the consequences of the different methods on calculation of CA using theoretical data. As shown in Appendix B, failure to use correct mathematical procedures will affect CA values and subsequent variability values now further described.

Coupling Angle Mean

CA are directional in nature, therefore; Hamill et al. (Hamill, et al. 2000) suggested use of circular statistics (Batschelet 1981) to calculate mean CA and its standard deviation for a number of trials (n). Calculations of the mean CA ($\bar{\gamma}$) as first reported by Hamill et al. (Hamill, et al. 2000) are described (Eqns.2.9-2.11). The CA for each trial collected(γ_i) is first componentized and the mean of each component (x, y) collected at each point in the gait cycle (i) for a required minimum of 3 cycles. (Mahan 1991)

$$x = \frac{1}{n} \sum_{i=1}^n \cos \gamma_i, \quad \text{Eq. 2.9}$$

$$y = \frac{1}{n} \sum_{i=1}^n \sin \gamma_i \quad \text{Eq. 2.10}$$

The mean CA according to Hamill is then calculated (Eq. 2.11).

$$\gamma = \begin{cases} \tan^{-1} \frac{y}{x} , y > 0 \\ 180 + \tan^{-1} \frac{y}{x} , y < 0 \end{cases} \quad \text{Eq. 2.11}$$

The only deviation from this equation as cited in the literature is presented by Chang. (Chang, et al. 2008) where $x_i > 0$ & $x_i < 0$ for $y_i > 0$ & $y_i < 0$ is substituted in Eq. 2.11. These reported equations may have been misinterpreted and employed in the literature. These equations are repeatedly cited in the literature when calculating mean CA but are not correct if using circular statistics as intended by Batschelet. (Batschelet 1981) Eq. 2.11 is modified in Appendix A (Eq. A.7 or A.8) to show equations that will yield intended results.

Coupling Angle Variability

Hamill further suggests that the length of the mean vector (a) can provide an estimate of the variability of the data. a is calculated (Eq. 2.12) and has values that range from 0 to 1 with 0 representative of high variability and 1, uniform data.

$$a = \frac{1}{\sqrt{x^2 + y^2}} \quad \text{Eq. 2.12}$$

Investigating joint coordination variability in healthy runners and runners with patellofemoral pain, Heiderscheit (Heidercheit 2000, Heiderscheit, et al. 2002) suggested the circular standard deviation (s) (Batschelet 1981) of CA as indicative of joint coordination variability. Values for s are bound between 0° and 81.03° ($\frac{\pi}{2}$) after conversion to degrees (Eq. 2.13).

$$s = \frac{1}{2} \sqrt{1 - a} * \frac{180}{\pi} \quad \text{Eq. 2.13}$$

Further described by Heiderscheit (Heidercheit 2000, Heiderscheit, et al. 2002) is calculation of mean CA variability (CAV_{Mean}) values over intervals of gait including swing, stance and stride. In other words, this output variable can be described as the mean circular standard deviation of the mean CA over a desired interval of stride. This has been interpreted as a representative measure of joint coupling variability over selected intervals of a series of movement cycles. Details concerning calculation of these measures are described in Appendix A (Eq. A.11). (Tepavac and Field-Fote 2001)

An alternative, but very similar method to describe variability of a series of movement cycles expanding on Sparrow was presented by Tepavac and Field-Fote (Tepavac and Field-Fote 2001) In their paper, the mean vector (a) is equivalent to a calculated in Eq. 2.12; however, the circular standard deviation of the mean CA is not used as a measure of variability. Instead, a is further manipulated to account for the magnitude of the vector. These calculations continuing from Eq. 2.12 are further described here using nomenclature consistent with what has been presented thus far. The arithmetic average of all the mean vector angles (a) are found (Eq. 2.14);

$$a = \frac{1}{N} (a_{1,2} + a_{2,3} + \dots + a_{N-1,N}) \quad \text{Eq. 2.14}$$

where N is the total number of frames per movement cycle and a signifies the overall variability for all the cycles. The vector lengths for each frame ($l_{i,i+1}$, Eq. 2.6) are then normalized to the maximum vector length observed at each respective interval over the amount of cycles (T) represented as l' (Eq. 2.15);

$$l'_{i,i+1}[t] = \frac{l_{i,i+1}[t]}{\max(l_{1,2}[t])} \quad \text{Eq. 2.15}$$

where $t = 1, 2 \dots T$. This procedure will keep the variance below 1 at each interval. The maximum possible standard deviation for a set of gait cycles is then calculated based on the amount of cycles (Eq. 2.16).

$$\sigma_{max} = \frac{1}{2} \frac{\sqrt{T + \text{mod}(\frac{T}{2})}}{T + \text{mod}(\frac{T}{2}) + 1} \quad \text{Eq. 2.16}$$

The scaled vector magnitude deviation for each frame (m) is then calculated (Eq. 2.17).

$$m_{i,i+1} = 1 - \frac{\sigma(l'_{i,i+1})}{\sigma_{max}} \quad \text{Eq. 2.17}$$

The arithmetic mean of m over the entire cycle (m) is calculated (Eq. 2.18) and indicates the similarity of distances between cycles on a coupling diagram.

$$m = \frac{1}{T} m_{1,2} + m_{2,3} + \dots + m_{N-1,N} \quad \text{Eq. 2.18}$$

The parameters a (Eq. 2.12) and m (Eq. 2.17) are combined to form a parameter termed the “coefficient of correspondence” which is the opposite of deviation (Eq. 2.19).

$$r = m_{i,i+1} * a_{i,i+1} \quad \text{Eq. 2.19}$$

The arithmetic mean over an entire cycle can then be calculated (Eq. 2.20).

$$r = \frac{1}{T} r_{1,2} + r_{2,3} + \dots + r_{N-1,N} \quad \text{Eq. 2.20}$$

Although not mentioned in their paper, r can be subtracted from 1 (Eq. 2.21) to give a measure of variance rather than correspondence (rv).

$$rv = 1 - r \quad \text{Eq. 2.21}$$

a , m and r are singular value output measures comparing similarity of shape, magnitude and a variance of the magnitude of joint coordination over the entire movement cycle, respectively. These measures were compared to Sparrow's correlation coefficient (Eq. 2.2) when analyzed using the same

hypothetical curves first introduced by Sparrow. (Tepavac and Field-Fote 2001) Agreement between these measures was shown to be strong indicating that these measures are valid mathematically; however, the limitation of using these singular values is similar to that of the correlation coefficient detailed by Sparrow (Eq. 2.2).

If variability characteristics are to change substantially within a given movement, which they are theorized to do during running gait, (Hamill, et al. 1999, Heiderscheit, et al. 2002) these summary measures over the entire movement cycle will not be sensitive to changes in variability throughout a movement. r and rv are measures that are continuous over the entire cycle and average values of r and rv within custom intervals of a movement have been reported in the literature (Field-Fote 2003, Field-Fote and Tepavac 2002, Hoch 2011, Mullineaux, et al. 2008, Mullineaux and Uhl 2010, Ness and Field-Fote 2009, Nooijen, et al. 2009) but clinical application of these measures are scarce in comparison to CAV measures.

Summary & Conclusion

Several methods have previously been proposed to measure the variability in joint coordination patterns from a dynamical systems perspective. CRPV initially showed promise as a clinical tool to assess control patterns during simple movements but its validity during complicated movements is questionable. Unfortunately, many limitations exist to CRPV that make application to most movements involving overuse injuries inappropriate. This includes lower extremity limb movement during running gait. VC methods have developed that are mathematically valid in assessing coordination between two moving oscillators and have been applied to lower extremity limb movement if mathematical steps are followed correctly. There are several inconsistencies in the literature concerning calculation of measures of joint coordination (CA) and the variability of joint coordination (CAV & rv), verbiage used to describe these processes and appropriate citations for actual steps used. Detailed mathematical steps and suggested nomenclature for calculation and presentation

of mean CA, CAV and rv are detailed in Appendix A. Previous methods have been summarized with inconsistencies identified. Details concerning these can be seen in Appendix B.

The coefficient of correspondence (r) and CAV are both derived from manipulations of a and are continuous measures of joint coordination. It can be implied that steps leading up to calculation of mean CA and the subsequent CAV provide a mathematically valid model to assess differences in CA among movement cycles. This does not imply validity in a clinical construct or specifically, validity of CAV or rv as a valid measure of coordination variability. Use of CAV as a clinical measure of coordination variability is more prevalent than the output measures of rv and further investigation of its clinical usefulness would reach a wider audience. Clinical validity requires testing sensitivity of an output measure (CAV) in response to an altered change in control parameter (clinical parameter) while controlling extraneous variables. Current evidence of CAV measures ability to detect physiological changes are further described in the next section.

Coupling Angle Variability As A Clinically Useful Measure

Variability of MCS of the lower extremity assessed using CAV as the outcome measure has been examined under several scenarios. In each case, a dynamical systems approach to movement variability (Hamill, et al. 1999) has been cited as the underlying theory warranting investigation. This theory states that a lower amount of variability may be indicative of a constrained or otherwise pathological coordinative state. In the original presentation of this theory, no evidence to statistically support this theory was reported. (Hamill, et al. 1999) For a clinical measure to be useful from a dynamical systems perspective it must behave according to the theoretical construct and predictably respond to changes in a control parameter (Turvey 1990). Investigations citing a dynamical systems approach as a theoretical construct have investigated the response of various CAV measures to changes between healthy and pathological populations (Ferber, et al. 2005, Heidercheit 2000, Heiderscheit, et al. 2002, Maulder 2011), sex (Maulder 2011, Pollard, et al. 2005), skill level (Maulder 2011, Wilson, et al. 2008), locomotion speed (Miller, et al. 2010) and a theoretical control parameter, the Lorenz Attractor (Miller, et al. 2010). Evidence of previous investigations to support use of CAV as clinically useful are further examined in this section emphasizing limitations and considerations that may improve future studies. An overview of these studies including details concerning population, tasks, couplings, comparisons and relevant findings are detailed below for reference (Table 2.2). Each study listed in Table 2.2 is further summarized in the text following Table 2.2. Relevant findings and CAV values found in Table 2.2 correspond to comparisons listed for each respective study for the indicated couplings. Table 2.4 lists all intervals in which CAV_{Mean} were calculated and emphasizes the inconsistency among studies for intervals analyzed.

Table 2.2 Previous literature examining mean Coupling Angle Variability (CAV_{Mean}) for differing populations, tasks, joint or segment couplings over various intervals of stride. Information is grouped by study and relevant findings of each comparison and coupling interval are detailed.

Study	Population	Tasks	Comparisons	Couplings	Relevant Findings
<i>Heiderscheit. 2002</i> And; <i>Heiderscheit 2000</i> <i>Chapter 3</i>	8 PFP Females Pain=1.9(.9) 8 Healthy Females	Treadmill running 2 speeds: Preferred † Fixed (3.3 m/s)	1. Injured leg vs. Non-Injured Leg 2. PFP vs. Healthy	Intra-limb: TR-SR, TF-SF, KR-AI, KF-AI, KF-AF Inter-limb: KF-KF, KR-KR AF-AF, AI-AI	1. Injured<Non-Injured leg TR-SR, Q1 at Preferred $\sim 19^\circ < \sim 27^\circ$, $p=.02$ 2. PFP<Healthy TR-SR, Q1 at Preferred $\sim 19^\circ < \sim 23^\circ$ No p reported
<i>Heiderscheit. 2000</i> <i>Chapter 4</i>	8 PFP Females 8 Healthy Females	Treadmill running 2 speeds: Preferred † Fixed (3.3 m/s)	1. Painful vs. reduced pain 2. PFP vs. Healthy 3. Fixed vs. Preferred speed	Same As above	1. No relation to pain; $p's > .18$ 2. PFP>Healthy KF-AI during Stride, All conditions 9° to $10.8^\circ > 7.6^\circ$ to 9.7° $p=.02$ 3. No difference between speeds $p's > .23$
<i>Ferber et al. 2005</i>	Runners Various injuries 5 Males 6 Females 11 Healthy Controls (CON) NO: No orthotic STD: Orthotic w/no symptom change INV: Inverted orthotic w/symptom relief	Overground running	1. NO vs. CON 2. NO vs. STD 3. INV vs. NO vs. STD	SR-AI	1. None 2. None 3. None
<i>Pollard et al. 2005</i>	College soccer players 12 Males 12 Females	Unanticipated 45° cutting maneuver	1. Males vs. Females	HR-KF HA-KR HR-KV KF-KR TR-SR TA-SA	1. Female<Male TR-SR $16.5^\circ < 24.3^\circ$, $p=.04$ TA-SA $9.7^\circ < 16.2^\circ$, $p=.01$ KF-KR $6.6^\circ < 12.4^\circ$, $p=.05$ HR-KF $7.7^\circ < 13.9^\circ$, $p=.05$

Table 2.2 (continued)

Study	Population	Tasks	Comparisons	Couplings	Relevant Findings
<i>Dierks & Davis 2007</i>	Runners 20 Males 20 Females	Overground Running (3.65 m/s)	Descriptive	KF-AI, KR-AF TR-AF, KF-TR KR-TR	Descriptive: Within subject & group CAV_{Mean}
<i>Maulder 2010 Chapter 3</i>	Netballers 10 elite Females	1. Overground straight run (3.5 to 5 m/s) 2. Unanticipated 180° turn	Between day (7 days) reliability for: Dominate (D) & Non-Dominate(ND)	TR-SR KF-AI KR-AI KF-SR KR-SR SR-AI	Suitable: KR-AI Run D, 20.7° to 19.7°, ICC=.75 KF-AI Run D, 29.4° to 24.8°, ICC=.92 KF-SR Run D, 32.4° to 27.2°, ICC=.82 SR-AI Turn ND, 31.3° to 30.8°, ICC=.89 SR-AI Run ND, 26.2° to 19.8°, ICC=.85 KR-AI Run ND, 23.8° to 18.3°, ICC=.72 KR-AI Turn ND, 37.9° to 34.4°, ICC=.70 KR-SR Turn ND, 37.1° to 33.5°, ICC=.81 Not suitable of note: TR-SR Run D, 21.8° to 20.3°, ICC=.49 TR-SR Run ND, 20.4° to 18.5°, ICC=.40
<i>Maulder 2010 Chapter 4</i>	Netballers. 12 elite Females 12 non-elite Females 12 non-elite Males	1. Overground straight run (3.5 to 5 m/s) 2. Unanticipated 180° turn	1. Elite vs. Non-elite 2. Non-elite vs. males 3. Elite vs. males 4. D vs. ND for all 3 populations & pooled females	Suitable couplings from above.	1. Elite<Non-Elite SR-AI Turn 33.0°<43.2°, p=.05 2. None 3. Males<Elite KR-AI Run D, 10.4°<20.9°, p=.005 4. None

Table 2.2 (continued)

Study	Population	Tasks	Comparisons	Couplings	Relevant Findings
<i>Maulder 2010 Chapter 5</i>	Netballers. 12 elite Females 12 non-elite Females 12 non-elite Males Followed during a season of netball for injury.	1. Overground straight run (3.5 to 5 m/s) 2. Unanticipated 180° turn	Association with injury: 1. All 2. Elite 3. Non-Elite 4. Male	Dominant KR-AI KF-AI KF-SR	1. All may not be associated (-.18 to .43) 2. Association: KR-AI $r=.29$ (.25 to .69), may not be KR-AI $r=.66$ (.24 to .87), very likely KF-SR $r=.12$ (-.4 to .58), may not be 3. Association: KF-AI $r=.47$ (-.04 to .79), likely probable KR-AI $r=.39$ (-.14 to .74), likely probable KF-SR $r=.33$ (-.2 to .71), likely probable 4. Association: KF-AI $r=-.46$ (-.78 to .05), likely probable KR-AI $r=-.35$ (-.72 to .18), likely probable KF-SR $r=-.54$ (-.82 to .06). likely probable
<i>Wilson et al. 2008</i>	Expert triple jumpers 3 Males 2 Females Elite= Personal best of world record $\geq 70\%$ Ranged 70 to 86%	Overground Triple Jump	1. Personal best plotted against CAV	Stance Leg: KF-AF, HF-KF Swing Leg: HF-KF	No statistics. Quadratic fit “U shaped”: Stance Leg: KF-AF $r=.366$, $\sim 12^\circ$ to 24° HF-KF $r=.693$, $\sim 12^\circ$ to 22° Swing Leg: HF-KF $r=.987$, $\sim 10^\circ$ to 21°

Table 2.2 (continued)

Study	Population	Tasks	Comparisons	Couplings	Relevant Findings
Miller et al. 2010	1. Theoretical 2. 18 Healthy Females 4 Healthy Males 3. 5 Healthy Males	1. Change of Lorenz attractor 2. self-selected walking speed 3. 3 walking; 5 running speeds	All comparisons were made between CAV and Continuous Relative Phase (CRP) for each task.	1. N/A 2. AF-FF AI-FI AA-FA 3. TF-SF	1. Acted according to theory: Vector Coding changed with manipulation of Lorenz attractor 2. CAV>CRP Q1 thru 5 for AF-FF, AI-FI Q5, AA-FA CAV_{Max}>CRP_{Max} 3. Descriptive Running>Walking Walking increased with speed. Peak 10% before heel-strike

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Note: All couplings were intra-limb unless denoted. Preferred speed was significantly less in subjects diagnosed with Patellofemoral Pain than healthy runners and both speeds were less than the fixed speed (2.4 to 2.8 m/s) (†). Quintiles of stride as measured from heel-strike when applicable.(Q) Coupling segment abbreviations (Proximal Segment –Distal Segment):TR= Thigh Internal/External Rotation, TF= Thigh Flexion/Extension, TA= Thigh Ab/Adduction, SR=Shank Internal/External Rotation, SF=Shank Flexion/ Extension, SA=Shank Ab/Adduction, FF= Forefoot Flexion/Extension, FI=Forefoot Inversion/Everison, FA=Forefoot Ab/Adduction. Coupling Joint abbreviations (Proximal Joint –Distal Joint): HF= Hip Flexion/Extension, KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion, AF=Ankle Ab/Adduction.

Control Parameters

The first use of CAV was introduced by Heiderscheit (Heidercheit 2000) and later peer reviewed(Heiderscheit, et al. 2002) comparing CAV measures between 8 female runners with unilateral PFP and 8 female runners with no symptoms. Intra-limb and inter-limb couplings were investigated at a self-selected and preferred pace during treadmill running. CAV_{Mean} values over the entirety of stride and over quintiles of stride with each quintile containing a functional aspect of running stride were compared between populations and between the symptomatic and asymptomatic leg. Of all comparisons, only CAV_{Mean} for the TR-SR coupling during the preferred running speed was found to be significantly less for the quintile encompassing heel strike in the symptomatic knee when compared to the symptomatic leg and healthy controls. A low pain level observed by the symptomatic group (1.9 using a visual analog scale VAS) led Heiderscheit to suggest that a larger observed pain in a PFP population may have produced more supportive results.

Using the same population and data analysis methods, Heiderscheit investigated the effect of reducing pain through knee taping on CAV_{Mean} . (Heidercheit 2000) No significant relation ($p < .05$) was found between pain and CAV_{Mean} ; however, both the symptomatic and asymptomatic coupling of KF-AI was significantly higher in the PFP group than the healthy controls. This finding is seemingly contrary to the dynamical systems approach to lower extremity injuries which predicts a lower variability and an explanation for these results was not given. A low initial pain value may have contributed to limited results. Additionally, if pain is indicative of a coordinate state, CAV may have a hysteretic effect (Turvey 1990) where a reduction in pain may not result in a change in CAV similar to when there is an increase in pain. The relation between an increase in pain and CAV response has never been examined.

The homogeneity of an injured population may have a large effect on CAV values. A heterogeneous injured and healthy population was used to investigate the effects of orthotics on SR-AI CAV_{Mean} in running injuries. Despite relief of

symptoms, no effects were observed and the conclusion was reached that orthotics had no effect on CAV. This may be a misinterpretation of the results. In this study, 5 males and 6 females composed the injured group and were matched with healthy controls. Additionally, 5 different overuse injuries (posterior tibial tendonitis (1), plantar fasciitis (4), anterior compartment syndrome (4) and PFP (2)) composed the 11 injured subject group. The heterogeneity of sex and injury may have affected CAV masking the effects of the orthotics on CAV.

This is supported by results presented by Pollard et al. (Pollard, et al. 2005) Controlling for sex, Pollard et al. investigated the differences between males and females during an unanticipated cutting maneuver theorizing that the prevalence of ACL injuries seen in females may relate to a coordinate state with lower movement variability. Four of six couplings analyzed were shown to be significantly less in females than males (Table 2.2) suggesting there is a sex effect with CAV. Contrary to this finding, when comparing non-elite male and elite female net ballers, Maulder observed higher CAV values in females than males (Table 2.2). Moreover, when comparing the same males to non-elite females, no differences in CAV values were observed. Further, when sex was held constant, elite females were shown to have less CAV than non-elite females. These conflicting results coincide with the theory that movement variability is context specific (van Emmerik, et al. 2004) and differences in injury, sex and skill might have a large enough influence on CAV to mask possible changes such as those caused by a control parameter change.

Other evidence to support the theory that there is an observable relation of any control parameter to CAV measures is weak. Wilson (Wilson, et al. 2008) assessed the relationship of CAV to skill during stance of the triple jump in 5 elite subjects composed of males and females. CAV and skill were fit with a quadratic curve with correlations reported. (Table 2.4) The swing leg of HF-KF had a “U” shaped curve with a strong correlation to skill ($r=.987$) where authors suggested that as skill increases, variability temporarily decreases to aid in learning a task than increases with further increases in skill. Several limitations were apparent. Upon observation of the data, it is clear that these results would have been

greatly influenced by a single outlier considering the small range of CAV values in the HF-KR coupling (~10 to 21) and low number of subjects. Skill was also determined by the subject's personal best of the current world record and all subjects were considered elite. A more definitive skill as a control parameter or larger change in control parameter might be necessary to extrapolate theories introduced with this study to having any clinical context.

CAV response to changes in speed has also been investigated. (Miller, et al. 2010) In this study, speed was manipulated during treadmill walking and running. Unfortunately, statistical analysis was only performed between CAV measures and CRP values but qualitative observations of CAV response to speed can be made with these data. Generally, CAV measures decreased with an increase in speed within each task and running values were greater than walking indicating that the task of walking and running will differ and speeds within each task can also effect CAV values. Miller et al. also observed CAV measure response to a theoretical control factor, the Lorenz Attractor (Lorenz 1963). Theory dictates that an attractor state will switch at a critical value of the Lorenz attractor (24.28). State space equations were constructed to represent CAV measures and as the critical value was reached, CAV values correspondingly fluctuated. These findings support CAV as a valid construct to represent dynamical systems but this evidence does little to emphasize clinical usefulness.

Table 2.3 Intra-limb couplings used within the literature. Abbreviations used are described in Table 2.1 and the number of authors is indicated (n) when more than one.

Only Segments	Only Joints	Segments & Joints
TR-SR (3)	HR-KF	TR-AF
TF-SF (2)	HR-KV	KF-TR
TA-SA	HF-KF	KR-TR
	HA-KR	KF-SR (2)
	KR-AI (3)	KR-SR
	KR-AF	SR-AI (2)
	KF-AI (4)	AA-FA
	KF-KR	AF-FF
	KF-AF (2)	AI-FI

Couplings

CAV is limited to analyzing the variability between only 2 joint or segment couplings at once. This makes comprehensive analysis of the variability of the lower extremity difficult. For simplicity commonly adopted throughout the literature, the lower extremity contains 3 segments (thigh, shank and foot) and 3 joints (hip, knee and ankle), each with three angular articulations with the exception of the ankle. This makes 28 joint and 36 segment coupling possibilities if combinations remain separate. When allowed to mix joint and segment couplings, possible couplings increase to 136. There are only 21 intra-limb lower extremity combinations used in the literature (Table 2.3) with only 7 being used by more than 1 author. With different authors using various couplings for differing tasks, in addition to conflicting findings in some cases, previously reported values are disparate.

Couplings used must have a theoretical basis for being studied. (Wheat and Glazier 2006) Joint couplings opposed to segment couplings may offer a more thorough representation of the variability of lower extremity because it requires both a proximal and distal segment which can simultaneously effect the joint articulation, consolidating information to interpret. In lower extremity injuries the most commonly studied couplings involve articulations of segments involved in the knee and ankle described by Tiberio (Tiberio 1987). Knee-ankle couplings

of KF-AI, KF-AF, KR-AI and KR-AF have been used previously and involve all three segments of the lower extremity. Surprisingly, no couplings involving KV have been utilized in the literature, a common joint motion with possible clinical implications to knee injuries. (Powers 2003) If these coupling combinations with the inclusion of KV-AI and KV-AF are analyzed at the same time, a more comprehensive understanding of CAV of the distal lower extremity might be understood.

Coupling Angle Variability Mean Sensitivity

Variability fluctuations are spontaneous and relative increases or decreases in value need to be observed to evaluate a dynamical system's perspective to injuries. These fluctuations are thought to occur during functional aspects of stride, particularly near heel-strike(Heiderscheit, et al. 2002) and during stance, the period of joint loading. (Hamill, et al. 1999) Heiderscheit observed no differences in any CAV_{Mean} comparisons over the entire running stride. (Heiderscheit, et al. 2002) Heiderscheit suggested that a more sensitive region for CAV_{Mean} may be required to isolate relative increase in the CAV curves. Quintiles of stride each containing a functional aspect of stride were suggested for use. Upon further analysis, only the quintile surrounding heel-strike showed any difference in CAV_{Mean} values. Authors have used a variety of intervals during stance when finding CAV_{Mean} values (Table 2.4), each smaller and thus more sensitive than the quintiles used by Heiderscheit. These have varied from intervals encompassing 16% to as large as 40% of stance with only one including swing, (Wilson, et al. 2008) a region thought to play an important role in preparation to load the lower limb during stance.(Powers 2003) Assuming stance phase is approximately 40% of running stride(Novacheck 1998) these intervals range from approximately 6% to 16% of stride. This shows a progression in the literature to use more sensitive CAV_{Mean} measurements.

Changes in movement strategy is theorized to be spontaneous(Turvey 1990), occur during specific regions of stride (Heiderscheit, et al. 2002) and characteristics of CAV curves for particular couplings are not fully understood.

Considering this, large intervals for finding CAV_{Mean} constructed *a priori* based on functional events may not be appropriate. These large regions may dampen changes in CAV where more sensitive intervals would yield more promising results. Also, increases in CAV may not occur within these set functional regions. A singular value in the CAV curve is the most sensitive and accurate measurement possible and may give insight to the spontaneous fluctuations and increases in CAV. A maximum value within a region of interest in CAV curves would exclude dampened regions of lesser importance. Regions of interest may best be constructed after observation of the particular CAV curves and intervals chosen on the characteristics of the curve rather than functional regions which haven't yielded promising results thus far. Further, increases in CAV are not instantaneous and like any signal have a time constant. Intervals of a set size specifically encompassing local maximum values may give a more accurate description of CAV increases during a movement for a particular subject.

Table 2.4 Intervals used to calculate mean Coupling Angle Variability(CAV_{Mean}) in the literature.

Study	CAV_{Mean} Interval
Heiderscheit. 2002 And; Heiderscheit 2000	1. Stride 2. Quintiles of stride (Q): Q1: 91 to 10% (heel-strike) Q2: 11 to 30% (mid-stance) Q3: 31 to 50% (toe-off) Q4: 51 to 70% (swing acceleration) Q5: 71 to 90% (swing deceleration) Notes: Heel-strike=0% Each Q contains a (functional aspect of stride) Intervals predetermined and defined using kinematics
Ferber et al. 2005	Intervals of stance (I) I1: ~0–20% (heel-strike to initial loading) I2: ~20–50% (acceptance of body weight) I3: ~50–75% (half distance to toe-off) I4 ~75–100% (to toe-off) Notes: ~predetermined interval % Determined using force plate
Pollard et al.2005	0 to 40% of stance (initial loading/deceleration) Determined using force plate
Dierks & Davis 2007	Intervals of stance (I): I1:~0–16% (heel-strike to impact peak) I2:~16–45% (...to max vertical force) I3:~45–73% (...to half distance to toe-off) I4:~73–100% (...to toe-off). Notes: ~empirically found averages using force plate.
Wilson et al. 2008	Stance: Touchdown to toe-off
Maulder 2010	Stance: Foot strike to maximum vertical force.
Miller et al. 2010	Quintiles of stance (Q1-5) Also CAV max value during stance

Reliability of Coupling Angle Variability measures

The reliability of CAV_{Mean} measures have only been assessed in one instance. (Maulder 2011) Findings from this study were not promising. They revealed volatile CAV_{Mean} values between data collection sessions 7 days apart. Of 24 values analyzed, only 8 were deemed suitable for further analysis (Table 2.2). Of note was the TR-SR coupling during a straight run was deemed unsuitable. This is the same coupling where Heiderscheit (Heiderscheit, et al. 2002) observed significant differences in measures. The lack of reliability values for all CAV measurements found in the literature should raise concerns about the clinical interpretation of any results found until reliability of these values can be established.

There are several methods to assess the test-retest reliability of clinical measurements. (Hopkins 2000) The methods used by Maulder were two-fold. Intra-class correlation coefficients (ICCs) were used to assess the differences between capture sessions reported as a relative percentage of the population mean. Coefficients of variance (CV) were also calculated and used to assess the variability within the measurements for the population and expressed as a percentage of the mean. Results were graded on a qualitative scale and determined whether they were suitable for future use or not. These methods are commonly used throughout the literature; however, give little insight when assessing the clinical validity of these findings. Inherent variability within a MCS and variability introduced by measurement error and noise cannot be separated (Schwartz, et al. 2004) and should be considered when interpreting findings. Also, for an un-established analysis measure such as CAV, little is known as to the actual source of variability. ICC's are limited in that results aren't provided in the original units, cannot assess systematic error and influenced by the range of values. (Mullaney, et al. 2010) Higher measured values are associated with higher ICC values, independent of actual measurement error. (Atkinson and Nevill 1998) A levels of agreement analysis (Bland and Altman 1986) may be an appropriate tool to aid clinical interpretation of findings. In this analysis, a 95%

confidence interval is established among CAV measurements for a population and data remain in their original form. If the amount of change in observed CAV values for a population exceeds the confidence interval established during reliability testing, the change can be considered clinically meaningful and not measurement error. (Mullaney, et al. 2010) These established confidence intervals can be considered analogous to the precision capabilities of CAV measures within a laboratory if considering CAV measures as a type of clinical analysis tool.

Methodical Considerations

There were several parameters of previous studies that may have introduced small amounts of variability caused by analysis techniques which may have contributed to the limited findings. The number of trials used in calculation of CAV_{Mean} has varied anywhere from 5 which is most common (Dierks and Davis 2007, Ferber, et al. 2005, Maulder 2011) upwards as to 10 (Miller, et al. 2010) and 15 (Heiderscheit, et al. 2002) making comparison of values between studies difficult. Discrepancy in the amount of trials is routinely cited as relating to statistical power, which suggests the amount of trial sizes required are 10, 5 and 3 for sample sizes of 5, 10 and 20 subjects to achieve statistical effect sizes greater than 90%. (Bates, et al. 1992) A large amount of trials may also dampen real spontaneous variability. A smaller amount of trials might give a more accurate description of the capabilities of CAV measures. Additionally, data normalized over the entire stride cycle or stance have only used 101 points and been collected anywhere from 120 to 240 Hz. Normalization parameters and collection frequencies can introduce error between subjects and studies, inviting misinterpretation of CAV. (Mullineaux, et al. 2006) High data collection frequencies and minimizing exclusion of real data points in the normalization procedure will increase the validity of the calculated CAV values. Force component data used to determine heel-strike, mid-stance and toe-off has been reported as a sufficiently reliable within subjects (Cavanagh and LaFortune 1980) and constraining normalization periods to these points will decrease deviation

between subjects making comparisons within a population more valid. Controlling for these factors will decrease data reduction error giving a better indication of the CAV characteristics.

Summary & Conclusion

There is little evidence providing credence to a dynamical systems perspective to overuse injuries. These limited findings are thought to be a result of not controlling for factors that can affect CAV such as sex and joint couplings analyzed. Intervals used in calculation of CAV_{Mean} are disparate amongst studies and have trended towards a more sensitive or otherwise smaller interval to calculate CAV_{Mean} . More sensitive measures of CAV may yield more promising results. Methodical consideration in data collection and reduction can also have an effect of CAV and can address several limitations of previous studies. Evidence of CAV measures as being acceptably reliable is also scarce. The reliability of any CAV measure used in clinical interpretation needs to be established. A level of agreement analysis should allow clinically meaningful interpretation of CAV measures. These considerations should be given in clinical analyses involving CAV.

Patellofemoral Pain

A person's inability to adjust to mechanical loads encountered during repeated activity has been suggested to increase susceptibility to lower extremity overuse injuries and manifest in lower amounts of joint coordination variability. (Hamill, et al. 1999) A population of runners with the condition of Patellofemoral Pain (PFP) has been theorized as an appropriate construct to test aspects of injury from this dynamical systems perspective to overuse injuries. (Heiderscheit, et al. 2002) PFP is a condition that encompasses many possible etiologies that present with similar symptoms of the knee. This section will review factors associated with PFP that should be considered to construct a homogenous PFP cohort and test CAV response to physiological changes.

Activity and Sex Prevalence

Although PFP patients usually present with pain in activities such as prolonged sitting, ascent or descent of stairs or squatting, the most common cause of ailment is during or after physical activity. (Thomee, et al. 1999) PFP is commonly termed "runner's knee" due to the abundance of reported cases in runners. Running is one of the most common forms of exercise practiced in the US with approximately 50 million participants. (Novacheck 1998) Of the multitude of injuries that can occur while running it is estimated that Patellofemoral Pain (PFP) accounts for approximately 25% (Devereaux and Lachman 1984, McConnell 1986) of all lower extremity injuries. This prevalence is similar among runners ranging from approximately 20% (Taunton, et al. 2002) to 25% (Clement, et al. 1981) of cases. Repetitive loads endured while running can be demanding on the body, especially the knee. Running may introduce specific kinematic tendencies of the lower extremity that when performed repeatedly, introduce symptoms associated with PFP. Running kinematics and its relation to PFP should be investigated as a task to study lower extremity joint coordination in a PFP population.

The prevalence of PFP does not seem to be restricted to activity but might also relate to sex. In the general population, PFP has been reported to account for 19.6% of female and 7.4% of male injuries. (DeHaven and Lintner 1986) Of reported cases it is estimated that nearly 60% of reported PFP cases are female. (Taunton, et al. 2002) Studying a military population, it was reported that females are more than twice as likely as males to develop PFP. (Boling, et al. 2010) The large difference in prevalence and incidence between sexes may be the result of underlying increases in risk factors for PFP. (Boling, et al. 2010) Females have been shown to have distinct differences in their running joint kinematics when compared to males. (Ferber, et al. 2003) These differences include increased femoral adduction, femoral internal rotation, knee valgus, and tibial external rotation. (Csintalan, et al. 2002) These characteristics can all lead to excessive lateral patellofemoral joint forces, (Lee, et al. 1994, Lee, et al. 2001, Mizuno, et al. 2001) the most common reported location of pain in PFP. (Fulkerson 1983) Mentioned previously in this chapter, smaller values in CAV measures have also been observed in females than males indicating fundamental differences between joint coordination variability between sexes. (Maulder 2011, Pollard, et al. 2005) Separation of sexes for analysis is recommended.

Etiology

Function

PFP can develop from multiple factors and tissue sources. (Fulkerson 2002) The functional role of the patellofemoral joint is to act as a fulcrum to help increase the moment arm of the quadriceps tendon on the tibia during leg extension using the patella. Misplaced forces or malalignment of the patella during this role may introduce a pathological state. (Thomee, et al. 1999) Patella position can be effected by the quadriceps tendon, patellar ligament, medial and lateral retinaculum, and the medial and lateral patellar ligaments. (Thomee, et al. 1999) These structures can affect the kinematics of the patella possibly leading to irregular joint and soft tissue loading and pain (Powers 2003).

Diagnosis

PFP is pain arising from the anterior aspect of the knee and is a diagnosis of exclusion. (Thomee, et al. 1999) Several structures can exhibit pathology and are not associated with PFP and need to be excluded. Subject history is a main focus of diagnosis to identify other sources of pain. (Fulkerson 1994) Sources of pain can come from traumatic injury or dislocation causing damage to the surrounding structures and pain not associated with overuse injury. (Fulkerson 2002) Osteoarthritis can develop in an individual causing peri or retropatellar pain and is uncommon in adults under the age of 40. (Iwano, et al. 1990) Screening for age can reduce the risk of inclusion of an osteoarthritic individual. Other overuse injuries of the Illiotibial band and patellar tendon are associated with running can make subject history difficult to distinguish the source. Pain associated with the Illiotibial band will present over the lateral aspect of the femoral epicondyle or lateral tibial tubercle. (Khaund and Flynn 2005) Similarly, patellar tendonitis will exhibit symptoms of localized tenderness at the tibial tuberosity or just inferior to the patella. (Fulkerson 2002) (Khaund and Flynn 2005) Joint line pain is also a common source of pain that can arise from meniscal tears and also can be assessed with palpation. (Fredericson and Powers 2002) The combination of history and diagnostic procedures are consistently used throughout the literature to determine the presence of PFP and rule out other causes of knee pain. (Bolgia 2005, Crossley, et al. 2004, Powers 2003)

Pain

The location of PFP pain may infer the possible source of pain. Pain is located near the lateral retinaculum of the patella in upwards of 90% of reported cases. (Fulkerson 1983) Only 10% of reported cases exhibit only medial pain, and of cases where medial pain is present it is accompanied with lateral pain at a rate of approximately 50%. (Fulkerson 1983) The location of medial pain is not limited to the medial retinaculum and can be located directly on the patellar facet. (Fulkerson 1983) Due to the high prevalence of pain on the lateral aspect

of the patella, theories of overuse causation have concentrated on possible causes of lateral pain. (Lee, et al. 1994, Lee, et al. 2003) It is possible a cohort of PFP population may display a CAV pattern dependent upon severity of pain. The largest amount of free nerve endings about the knee is located in the lateral retinaculum. (Biedert, et al. 1992) In a preliminary study of 12 females and 1 male with PFP, the severity of pain demonstrated was significantly related to the amount of innervated area in the lateral retinaculum in severe and moderate pain individuals when compared to light and no pain. (Sanchis-Alfonso, et al. 1998) These authors further speculated that lateral retinaculum nerve damage may result in instability of the patella due to proprioceptive deficits. A PFP population capable of reaching a higher pain level may exhibit a coordinate state consistent with a proprioceptive deficit seen only in those with a severe or moderate level of pain. Supporting this theory, as discussed in the previous section, no changes in CAV values were observed when pain decreased from approximately 1.9cm to .7cm on a 10cm VAS pain scale (Heidercheit 2000) and minimal changes were observed between healthy controls and the same PFP population in the painful state. (Heiderscheit, et al. 2002) The limited findings were cited as a result of a possible delay in proprioceptive response which has been observed with a minimal change in environment (Scholz 1990); however, it may also indicate that the PFP population studied may not have reached a severity of pain great enough to elicit a change in coordinate state. A clinically meaningful change of pain is 2cm on a 10cm VAS. (Crossley, et al. 2004) A clinically meaningful change in pain may be required to elicit a clinically meaningful change in CAV.

Assessing pain during running is difficult using the typical paper and pencil VAS. A verbally administered numeric pain rating scale (NPRS) is a valid alternative method to gauge perceived pain. (Williamson and Hoggart 2005) This scale allows assessment of perceived pain by the investigator during activities where marking a paper VAS is not preferred during continuous running. Immediate evaluation of pain is capable with this method rather than at completion of the study. This 11 point scale is described to subjects with the anchor of 0 representing “no-pain” and 10 representing “worst imaginable pain”.

A clinically meaningful change in pain in a PFP population is considered 1.2.
(Piva, et al. 2009)

Kinematics and Kinetics

There are several plausible etiologies of PFP, all of which involve irregular kinematics of the lower extremity involving segments composing the hip, knee, ankle. (Barton, et al. 2009, Powers 2003) Imbalances in muscular control can result in irregular positioning of limb segments or anatomic structures causing large forces to the patellofemoral joint. (Davis and Powers 2010) This may increase the risk of PFP for those who demonstrate these high risk characteristics. (Elias, et al. 2004, Mizuno, et al. 2001) Irregularly large and misplaced kinetics of the patellofemoral joint is commonly believed to be the underlying mechanism of injury for PFP patients, however; there is not currently an understanding of the specific lower extremity joint movements that eventually produces pain about the knee. (Davis and Powers 2010, Powers 2003) This is due to conflicting results consistently reported in the literature even within the same movement tasks. (Barton, et al. 2009)

Reported results of hip motions during running are conflicting. Larger hip adduction (Dierks, et al. 2008, Noehren, et al. 2011, Willson and Davis 2008) and internal rotation (Noehren, et al. 2011, Souza and Powers 2008) angles have been reported in runners with PFP. Contrarily, less hip adduction (Dierks, et al. 2011) and no differences in internal rotation (Dierks, et al. 2008) have also been reported. Discrepancy in hip kinematics may be due to differences in population. Studies by Dierks et al. (Dierks, et al. 2011, Dierks, et al. 2008) included females and males in their population while all others were only females. This indicates further possible characteristic differences between males and females in PFP populations. Regardless of sex differences, increases in hip internal rotation or adduction may be a result of decreased hip strength and may increase knee valgus and lateral forces on the patellofemoral joint. (Bolgla, et al. 2008, Ireland, et al. 2003, Robinson and Nee 2007) Couplings involving knee valgus may add

more insight to CAV response to proximal motions of the lower limb in PFP populations.

A localized view of the knee has also shown conflicting results using traditional kinematic measures. Compression forces observed at the patellofemoral joint vary widely; ranging from half of body weight for walking up to 7 times body weight during squatting. These load magnitudes increase as knee flexion increases. (Mason, et al. 2008) Reduction in knee flexion angle during loading phase of walking gait may be a compensatory mechanism to reduce knee pain. (Powers, et al. 1997) Only one study has found a reduction in knee flexion angle between PFP and healthy individuals while running (Dierks, et al. 2011) with another finding no differences. (Willson and Davis 2008) Conflicting results have also been observed involving knee external rotation that were shown to be larger in a PFP population (Willson and Davis 2008) and in another case no differences between populations. (Dierks, et al. 2011)

The most popular mechanism for injury proposed by Tiberio considers analyzing segments of the knee and ankle in a coupled manner. (Tiberio 1987) Patella position has been suggested to be affected by internal rotation of the tibia coupled with subtalar pronation during loading phases of gait. (Nawaoczenski, et al. 1998, Powers 2003, Tiberio 1987) As described by Tiberio, a chain of simultaneous events occurs with this coupling. When excessive subtalar pronation is present during mid-stance, the tibia is not able to externally rotate as far. External rotation of the tibia is required to extend the knee at this point in the gait. To compensate, the femur internally rotates allowing extension of the knee causing lateral tracking of the patella and excessive lateral forces to the patellofemoral joint. (Tiberio 1987) An externally rotated tibia or internally rotated femur (knee internal rotation) can increase the Q-angle by rotation of the tibial tuberosity and subsequent rotation of the patella. (Mizuno, et al. 2001) It has also been demonstrated that internal rotation of the femur can increase forces to the lateral aspects of patella. (Lee, et al. 1994) CAV involving the knee and ankle might be able to observe this relationship between a healthy and a PFP

population better than past traditional kinematic measures and should be considered for analysis.

Conflicting results in kinematics during running studies may also be related to fatigue, not just sex, pain and population. Observations made by Dierks et al. (Dierks, et al. 2011) during a prolonged run on a treadmill showed many significant differences in multiple joint angle, excursion and velocity measures of the knee between the beginning and completion of an exhaustive run. PFP pain has an insidious onset (Fredericson and Powers 2002) that can worsen during the course of a run and usually not present in a fresh state. (Dierks 2005) Similarly, rating of perceived exertion scale (RPE) (Borg 1982) (Appendix D.2) has shown to steadily increase during the course of an exhaustive run. (Dierks 2005) Perceived exertion, or otherwise fatigue, may have an effect on CAV during the course of a treadmill run and may mask possible CAV changes caused by differences in population or pain level. The task of running on a treadmill should also be described by the fatigue level of the runners.

The gold standard to measure fatigue is the percent of the maximum possible volume of oxygen consumption ($\%VO_{2max}$) (ACSM 2000) but requires specific equipment to implement. An alternative and clinically more feasible approach is the RPE scale which has been shown to have a Pearson's correlation coefficient of $r=.87$ with $\%VO_{2max}$. (Herman, et al. 2006) A state of fatigue should be used that represents a fresher state and still can produce a clinically meaningful state of pain. Also, if pain slowly increases during the course of a run, an amount of time will surpass which might include a change in fatigue. Fatigue change should be kept to a minimum to minimize its effects on CAV as changing fatigue and pain at the same time may confound results. Little is known when kinematic effects occur in relation to the RPE scale for both absolute values or relative change; however, 17 on the RPE has been considered to be exhaustive. (Brown 2011, Dierks 2005) A lower value is recommended to represent a fresher state. 61% of a healthy population has reported values between 11 and 14 for a training range of 60% of maximal heart rate, (Whaley, et al. 1997) typical of a common run which might induce pain.

Also, relative changes on an RPE scale are thought to be linear inferring that the amount of relative change in RPE score is equivalent regardless of the absolute values for an individual (i.e. $12-11=13-12$ =a RPE change of 1). (Borg 1982) It has not been established what is a minimal meaningful change on the RPE score. The lowest unit is a score of 1 therefore this may represent an acceptable relative change of fatigue to delimit during a period of treadmill running where pain increases a meaningful amount.

Summary & Conclusion

Certain cohorts of PFP populations might exist that share similar characteristics and exhibit one or many of the aforementioned gait abnormalities.(Powers 2003) Running is a physical activity that is a common activity in many cases of PFP. Runner's may exhibit motions that cause PFP symptoms at an increased frequency and loading rate making runners a population that may present with similar characteristics and symptoms. Running for a prolonged period may also induce higher pain values further defining a PFP cohort. A prolonged run can elicit symptoms of fatigue that may affect kinematics in PFP individuals and should be accounted for when evaluating gait variability. Additionally, focusing on one sex for evaluation will help to minimize confusion on interpretation of variability data. Females develop PFP at a higher rate so they are a good population to study to generate adequate number of subjects for a study.

Investigation of CAV to describe coordination variability maintains consistency with current etiology theories. The etiologies presented involve joint motion of the hip, knee and ankle, and individual segments and rotations that compose each. These motions are regularly simultaneous which emphasizes the consensus that PFP is multifactorial in nature and more than one issue can be present in an individual with PFP. (Davis and Powers 2010) This suggests that kinematic analysis should include coordination of several joints and segments to fully understand the nature of the irregular movement patterns. Abnormal kinematics of the lower extremity effect the patellofemoral joint during

activity but there are no conclusive measures to discriminate between PFP and healthy individuals during running. The relatively high incidence of PFP makes it a logical problem to study as it affects many people and is relatively common.

Chapter 3: Test-retest reliability of Coupling Angle Variability measures

This chapter aims to present a study which establishes the intra-subject reliability of CAV measures when control parameters remain constant for a PFP and healthy population of runners. This chapter can be read independently from the rest of the text in manuscript format suitable for submission for publication.

Introduction

The variability of lower extremity limb coordination patterns during gait and the theory that it is inherent within a healthy movement control strategy (Hamill, et al. 1999, Newell, et al. 1993, Stergiou, et al. 2006) has become commonly studied. From a dynamical systems perspective of motor control, a movement control strategy is a function of the complex interactions of three multidimensional control parameters; environment, organism and task. (Bernstein 1967, Newell, et al. 1993, Turvey 1990) A large change to a control parameter such as task (walking vs. running), organism (healthy vs. pathological) or environment (treadmill vs. over-ground) may result in a substantial change in strategy and shift in coordinative state. Conversely, if no control parameters have been altered, resulting variation in strategy should be minimal and remain within the same coordinative state. (Kelso 1995) Variability in joint coordination has been suggested as an indirect representation of variability in movement control strategy. (Turvey 1990) A variety of analysis tools to quantify coordination patterns have been used with the most common being continuous relative phase (Kelso 1995). This technique has limitations in quantifying non-sinusoidal joint couplings and may not be appropriate for lower extremity couplings during gait. (Peters, et al. 2003) Vector coding has been introduced as an appropriate method to quantify joint coupling relationships continuously throughout the gait cycle using its output measure of coupling angle. (Heidercheit 2000) Further, coupling angle variability (CAV) has been suggested as a potentially useful measure to distinguish among coordinative states based on certain physiological control parameters. (Dierks and Davis 2007, Ferber, et al. 2005, Heidercheit

2000, Heiderscheit, et al. 2002, Miller, et al. 2010, Pollard, et al. 2005, Wilson, et al. 2008)

Use of CAV as a clinically useful measure is not yet evident. Focusing on lower extremity gait and specifically knee injury mechanics, Patellofemoral Pain (PFP) has been suggested to be the result of decreased joint coupling variability where an over constrained control system leads to an overuse injury and pathological state.(Hamill, et al. 1999) Investigating this theory, Heiderscheit et al. (Heiderscheit, et al. 2002) compared mean CAV values over the entire stride cycle for several joint and segment couplings between injured and healthy individuals while running at a self-selected pace. No differences were found. Further analysis increased sensitivity of measurements by using mean CAV over smaller quintiles of stride. The coupling of thigh-shank long axis rotation near heel strike was found to statistically support the theory. Employing similar analysis methods when assessing the effects of orthotics on an array of injured runners, introduction of an orthotic improved symptoms but no changes in CAV were observed. Minimal pain values (Heiderscheit, et al. 2002) and a heterogeneous injured population (Ferber, et al. 2005) were cited as possible factors for the inconclusive results. Analysis using CAV measures that are more sensitive than the quintiles used might have also yielded more promising results. Regardless of the measure used, if CAV measures are representative of physiological differences occurring during gait, clinically meaningful interpretation of results requires the reliability of these measures to be thoroughly examined prior to investigation between groups. (Schwartz, et al. 2004)

Measurement methods used in gait analysis must be reliable to be clinically useful. (McGinley, et al. 2008) Decreased variability in measurement values is a quality of a reliable method and adds to the extent to which the method is useful. Dynamical Systems theory undermines this standard concept of repeatability of measures focusing on spontaneous increases in joint variability within an individual as the measure of interest. (Stergiou, et al. 2006) The concepts of repeatability and variability are not believed to be mutually exclusive. Sources of variability in all biomechanical measurements are a combination of

intrinsic natural physiological variation and extrinsic error which produces a measurement with a mean and deviation that can be used for comparison. (Schwartz, et al. 2004) This should be no different for CAV measures. CAV measures need to be examined while extrinsic errors are reduced through methods and factors affecting a physiological system remain constant. Test-retest measurement analyses are recommended to quantify the intrinsic repeatability of gait measures within a laboratory (Schwartz, et al. 2004) and limits of precision for clinical measures need to be established to understand the clinical usefulness of any measurement method. (Mullaney, et al. 2010)

Applying dynamical systems theory to running gait analysis affords several possible constructs for test-retest reliability assessment, each with factors that will influence CAV measures. Joint coupling variability of the knee and ankle is thought to be affected by knee pain level (Heiderscheit, et al. 2002), speed (Miller, et al. 2010), population (Ferber, et al. 2005, Heiderscheit, et al. 2002, Miller, et al. 2010, Pollard, et al. 2005, Wheat, et al. 2002) and fatigue. (Dierks 2005) The reliability of CAV measures between the knee and ankle have not been established for a healthy or PFP population when controlling for these factors. Therefore; the purpose of this study is to evaluate the test-retest reliability of multiple vector coding CAV measures when the physiological factors of fatigue and pain have not been altered at a self-selected running pace. Resulting confidence intervals from this study will describe the precision limits of each CAV measure and define thresholds to overcome for a change in CAV measure to be considered clinically meaningful. (Mullaney, et al. 2010)

Methods

Twenty-one healthy and twenty injured female recreational runners participated in the study. To participate, all females had to be between 18 to 45 years of age and run a minimum of 16 km (10 miles) per week. Subjects were included in the healthy group if they had no history of PFP and reported no lower extremity pain while running. Subjects were included in the PFP group if they self-reported a knee pain of a 3 or greater out of 10 during normal running activity and were currently diagnosed with PFP by a certified athletic trainer or licensed

physical therapist after exclusion of knee pain resulting from acute injury, patellar tendonitis, Iliotibial band syndrome or meniscal pathology. Potential subjects were excluded if they had a neurological disorder, tape allergy or felt they could not maintain a minimum pace of 3.3 m/s (8 minute 20 s mile) for 2 minutes. Written informed consent was obtained prior to participation in the study, which was approved by the institute's institutional review board.

Retro-reflective markers were attached to the subjects to model bilateral, hip, knee and ankle articulations. (Figure 3.1) The distal aspects of each thigh and shank were wrapped with elastic straps (ProWrap, Fabrifoam, Exton, PA) and rigid body clusters were then attached to the straps with hook and loop connectors and secured using additional elastic straps (MediPro, Fabrifoam, Exton, PA). Subjects wore standardized shoes (ZoomAir; Nike, Beaverton, OR) modified with windows cut out allowing adhesion of the markers directly to the skin by means of both adhesion spray and toupee tape.

Kinematic data was captured using a combination of 15 Eagle and Eagle4 cameras at 300 Hz (Motion Analysis Corporation, Santa Rosa CA). A dual belted treadmill instrumented with a force plate under each belt (TM-09-PBertec, Columbus, OH) was used to collect ground reaction force data at 1200 Hz. The treadmill belt speed was operated remotely by the investigators with a velocity resolution of 0.01 m/s with each belt being 48 cm wide and 164 cm long. A 15 point Rating of Perceived Exertion scale (RPE)(Borg 1982) was placed on a stand directly in front of the treadmill for subjects to reference to report level of perceived fatigue during the run. Perceived pain during the run was collected using a verbally administered numeric pain rating scale (NPRS) described to subjects as 0 being "no pain" and 10 considered "worst imaginable pain".(Farrar, et al. 2001)

Treadmill Protocol

A one second standing static calibration file was captured while the subjects stood in the anatomical standing position. Subjects then walked on a single belt of the treadmill for 3 minutes at 1.3 m/s to acclimate themselves to the

treadmill. Speed was then increased for 3 minutes to a warm-up pace (2.2-2.3 m/s) followed by 2 minutes at a standard pace of 3.3 m/s. Speed was then set at a self-selected pace where subjects felt they would not become severely fatigued over the course of the next 15 minutes with speed being adjusted upon request (2.2 to 3.3 m/s). To be included in the PFP group, subjects had to reach a minimum knee pain of 3 during the treadmill protocol. Kinematic and kinetic data were acquired for the first 10 seconds of each minute interval. RPE and NPRS measures were recorded by investigators immediately following each 10 s data acquisition.

Data Processing

Kinematic markers were identified using Cortex 2.0 software (Motion Analysis Corporation, Santa Rosa CA). Three-dimensional marker coordinates and force plate data were exported to Matlab v2009a (Mathworks, Natick MA) for gait analysis. A fourth-order lowpass butterworth filter with a cutoff frequency of 8 Hz was applied to kinematic data. Force component data were filtered with a cutoff frequency of 30 Hz for the lateral forces and at 40 Hz for the vertical component. Cut-off frequencies were selected by investigators after visual inspection of a fast fourier transformation performed on the data. Joint coordinate systems were determined using the International Society of Biomechanics recommendations (Grood and Suntay 1983, Wu, et al. 2002). Segment orientations were determined using a singular value decomposition algorithm (Söderkvist and Wedin 1993) and joint angles using an Euler rotation sequence of long axis rotation-abduction-flexion for the knee and ankle.

Consistent gait points of heel-strike, mid-stance and toe-off were determined for each gait cycle for normalization. Heel-strike and toe-off were determined using the vertical component of the ground reaction force with a threshold of 50 N, mid-stance was the transition from braking to propulsion (0 N). (Cavanagh and LaFortune 1980) Both of the two periods of stance were time normalized to 50 points and swing phase to 150 points using a 5th order cubic spline function making a 250 point time normalized gait cycle(1 point=0.4%).

Multiple normalization constraint points were chosen as it may reduce within and between subject variability of CAV measures. (Mullineaux, et al. 2006) The first and last gait cycle from each 10 s trial was discarded to reduce interpolation effects and the first 10 gait cycles were kept for analysis.



Figure 3.1 Marker set used during a static calibration. Only bilateral markers on the lateral aspects of the 5th metacarpal head, base, navicular and both the lateral and medial aspects of the calcaneus were used to model foot movement. Windows are cut out of the shoes allowing markers to be adhered directly to the foot. Rigid clusters were secured to the distal posterior-lateral aspects of each segment to model thigh and shank movement.

Data Reduction

One 10 s trial was chosen for analysis from the 15 minute period of self-selected running pace for each individual. For the PFP group, the trial with the highest pain value with a RPE value less than 14 was chosen. If there was more than one trial that qualified, the trial with the lower RPE value was chosen. If

there was more than one trial with the same RPE and pain value, preference was given to the earlier time point in the run. For the healthy group, trials from the 11th minute of running at the self-selected pace with a RPE value of less than 14 were used. This corresponded to the average trial selected for the PFP group. 19 PFP and 13 healthy participants qualified for analysis with 2 more healthy participants being excluded for missing markers on the foot.

Table 3.1 Common abbreviations and definitions used within the text and tables grouped by Knee-Ankle coupling relationship and coupling angle variability (CAV) measures. Measures are for each quintile and intervals of stride.

Joint Coupling	Definition
KV-AI	Knee Valgus/Varus coupled with Ankle Inversion/Eversion
KV-AF	Knee Valgus/Varus coupled with Ankle Plantar/Dorsi Flexion
KF-AI	Knee Flexion/Extension coupled with Ankle Inversion/Eversion
KF-AF	Knee Flexion/Extension coupled with Ankle Plantar/Dorsi Flexion
KR-AI	Knee Internal/External Rotation coupled with Ankle Inversion/Eversion
KR-AF	Knee Internal/External Rotation coupled with Ankle Plantar/Dorsi Flexion
Measure	Definition
CAV	Coupling Angle Variability. Variation within a set of 5 vector coded, non-consecutive gait cycles for a Knee-Ankle coupling relationship. CAV is a continuous measure for every point in the gait cycle. Units are in degrees.
CAV _{Mean}	Mean CAV value over discrete intervals (Q, I, stance, swing) of stride. Each quintile contains a functional period of stride shown in parentheses.
Quintiles (Q)	Q1: -10 to 10% (heel-strike), Q2: 10-30% (mid-stance), Q3: 30 to 50% (toe-off), Q4: 50 to 70% (swing acceleration), Q5: 70 to 90% (swing deceleration)
Intervals (I)	I1: -10 to 0%, I2: 0 to 10%, I3: 10 to 30%, I4: 30 to 64%, I5: 64 to 90%
Stance	0 to 40%
Swing	40 to 100%
Stride	0 to 100%
CAV _{Max}	Maximum CAV value within a selected interval of stride.
CAV _{Max±2%}	Mean CAV for an interval ±2% of stride about a CAV _{Max}

CAV values were determined using a revised vector coding technique.(Heiderscheit, et al. 2002, Sparrow, et al. 1987) Each 10 s trial contained 2 sets of 5 non-consecutive gait cycles (set 1=cycles 1,3,5,7 and 9; set 2 cycles 2,4,6,8 and 10). CAV values were derived for each set for all knee and ankle coupling combinations (Table 3.1) at each point in the gait cycle. The injured limb was analyzed for the PFP group and a leg was chosen by a random number generator for each of the healthy individuals to reduce systematic error. The normalized gait cycles were divided into quintiles each containing a functional period of stride (Heiderscheit, et al. 2002)(Table 3.1) with the border of each quintile overlapping. Mean CAV values (CAV_{Mean}) were calculated for quintiles (Q), stance, swing and the entirety of stride for each set, respectively.

Observation of the CAV curves revealed there were several locations where CAV increases were not encompassed using the standard 20% quintiles. Therefore, five intervals (I) were created to capture consistent increases in the CAV values among all coupling relationships (Table 3.1). These intervals as well as the quintiles appeared too large to be sensitive enough to distinguish between distinct variability characteristics using traditional CAV_{Mean} values; therefore, the maximum CAV value (CAV_{Max}) and the mean CAV from an interval $\pm 2\%$ of stride about each CAV_{Max} were found within each interval ($CAV_{Max} \pm 2\%$) for each set of gait cycles. Steps involved in construction of the intervals for CAV_{Max} and $CAV_{Max} \pm 2\%$ locations are further detailed (Appendix E and F).

Statistical Analysis

Independent t-tests were performed to note any differences between population demographics (height, mass, age and average distance run per week). Additionally, pain, RPE and running speed were also compared. Reliability of the CAV measures were assessed between the two sets of gait cycles from each trial using a levels of agreement analysis (LOA)(Bland and Altman 1986, Mullineaux, et al. 1999) for the PFP and healthy populations

separately. LOA between set1 and set2 were calculated by finding the difference between the two sets (set1-set2) then calculating the mean difference (\bar{d}) and the 95% confidence interval around the mean difference (standard deviation of \bar{d} (σ) *1.96)(Bland and Altman 1986) for each population. The grand mean (\bar{X}) for each measure was also calculated within each population (\bar{X} = (mean of set1+ mean of set2)/2).

Results

Population demographics are presented in Table 3.2 with only reported distance run found significantly different between populations ($p=.0008$). A wider range of speeds were observed for PFP (2.2-3.1 m/s) than healthy (2.6-3 m/s) with the mean speed for the healthy population being faster (2.89 m/s (.13), (mean (SD)) than the PFP population (2.54 m/s (.24)) ($p<.0002$). Pain values were 4.3 (1.3) for the PFP group. RPE levels for the Healthy (H) group (12.2 (.87)) and the PFP group (12.4 (.77)) were not significantly different ($p=.4091$).

Table 3.2 Subject demographics for Healthy and Patellofemoral Pain (PFP) groups.

Population	Sample Size (n)	Height (m)	Mass (kg)	Age (yrs)	Distance (km/week)
Healthy	11	1.66 (0.09)	58.0 (5.33)	26.5 (3.6)	37.7 (13.4) *
PFP	19	1.63 (0.07)	57.1 (6.48)	25.8 (6.1)	21.2 (9.4)

Note: Means for each measure are displayed with the standard deviation in parentheses. Significant difference between populations denoted at $p<.05$. ()*

Ensemble averages of CAV for each joint coupling over stride are presented for both populations (Figure 2.2). Quintiles used to calculate CAV_{Mean} values (Figure 3.2, A) and chosen intervals used to determine CAV_{Max} and $CAV_{Max\pm2}$ (Figure 3.2, B) are highlighted. Differences in measurement sensitivity capability are elucidated by observing the general increase in CAV within Q1. Within this region, I1 partitions the increase in CAV from the relatively quiescent I2 for several couplings (KV-AI, KF-AI, KF-AF and KR-AI). In the remaining couplings (KV-AF and KF-AF), there is a peak just prior to heel-strike and one immediately following. The I1 and I2 measures separate these events contrasted with Q1 CAV_{Mean} measures which does not identify these characteristics.

LOA results for CAV_{Mean} values for each quintile, stance, swing and stride are presented in Table 3.3. The smallest measurement ranges (1.96σ) excluding systematic error ($|\delta|$) for each population were 1.3° (H, KF-AF,Stride;(Population, Coupling, Interval)) & (I, KF-AF,Q3) while the largest were 14.9° (H,KR-AI,Q5) and 16.1° (PFP,KV-AI,Q5). X ranged from 2.6° (H, KF-AF,Q3) and 2.9° (PFP, KF-AF,Q3) to 15.4° (H, KV-AI,Q5) and 22.3° (PFP,KV-AI, Q5), respectively.

LOA for CAV_{Max} values within each custom interval of gait are presented in Table 3.4 and were generally larger than CAV_{Mean} values. CAV_{Max} ranges for each population varied from 3.7° (H,KF-AF,I2) and 5.2° (PFP,KF-AF,I4) for regions with little CAV activity to 48.8° (H,KV-AI,I1) and 33.2° (PFP,KV-AF,I2) in intervals with noticeably more CAV activity. Similarly, X were larger for CAV_{Max} measures than CAV_{Mean} ranging from 3.3° (H, KF-AF, I2) and 5.4° (PFP, KF-AF,I1) to 52.9° (H, KV-AI,I1) and 54.4° (PFP, KV-AI,I5), respectively.

LOA for $CAV_{Max\pm2}$ values within each custom interval of gait are presented in Table 3.5. $CAV_{Max\pm2}$ ranges for each population varied from 4.6° (H,KF-AF,I2) and 5.9° (PFP,KF-AF,I2) to 33.6° (H,KR-AI,I1) and 27.9° (PFP,KV-AI,I5). Similarly, X ranged from 4.0° (H, KF-AF, I2) and 6.5° (PFP, KF-AF,I4) to 35.3° (H, KR-AI,I4) and 41.6° (PFP, KV-AI,I5), respectively. Ranges and X observed for $CAV_{Max\pm2}$ values were generally larger than CAV_{Mean} values and smaller than CAV_{Max} values. The amount of difference was highly dependent upon the interval and larger differences coincided with general increases in CAV activity.

Table 3.3 Test-retest levels of agreement between CAV_{Mean} values within each quintile (Q1-5) of stride, the entirety of stride, stance and swing phase at a self-selected running pace for six Knee-Ankle joint coupling combinations. Data are displayed separately for Healthy and Patellofemoral Pain populations.

Interval (% of Stride)	KV-AI		KV-AF		KF-AI		KF-AF		KR-AI		KR-AF	
	$\delta \pm 1.96\sigma$	$\bar{X}(^{\circ})$	$\delta \pm 1.96\sigma$	$\bar{X}(^{\circ})$	$\delta \pm 1.96\sigma$	$\bar{X}(^{\circ})$	$\delta \pm 1.96\sigma$	$\bar{X}(^{\circ})$	$\delta \pm 1.96\sigma$	$\bar{X}(^{\circ})$	$\delta \pm 1.96\sigma$	$\bar{X}(^{\circ})$
<i>Healthy</i>												
Q1 (-10 to 10)	-0.8 \pm 8.8	10.0	-1.5 \pm 7.2	11.3	-0.1 \pm 6.5	7.3	0.0 \pm 5.3	6.1	-0.7 \pm 11.3	11.3	-0.4 \pm 11.0	14.6
Q2 (10 to 30)	0.1 \pm 6.0	10.4	-0.5 \pm 5.4	5.7	-0.8 \pm 4.8	5.4	-0.7 \pm 5.0	4.1	-3.7 \pm 10.5	11.9	-1.9 \pm 4.5	7.9
Q3 (30 to 50)	0.2 \pm 8.4	10.1	0.3 \pm 4.0	4.2	0.0 \pm 4.0	5.4	-0.3 \pm 2.4	2.6	-1.3 \pm 10.4	16.1	-0.4 \pm 4.0	6.8
Q4 (50 to 70)	-4.4 \pm 10.6	15.2	-1.4 \pm 5.5	7.0	-0.1 \pm 5.1	5.1	0.2 \pm 3.6	3.7	-3.6 \pm 10.5	18.6	-1.0 \pm 8.4	8.5
Q5 (70 to 90)	-1.6 \pm 12.1	15.4	0.1 \pm 11.6	12.1	-0.5 \pm 3.7	4.1	-0.2 \pm 2.2	2.9	-0.7 \pm 14.9	15.3	0.3 \pm 10.5	12.9
Stride (0 to 100)	-1.0 \pm 4.8	12.1	-0.2 \pm 4.7	8.0	-0.2 \pm 2.2	5.3	-0.1 \pm 1.3	3.8	-1.9 \pm 4.4	14.6	-0.7 \pm 3.6	10.2
Stance (0 to 40)	0.0 \pm 5.2	8.2	-0.8 \pm 3.4	4.9	-0.5 \pm 3.1	4.5	-0.5 \pm 2.7	2.9	-2.2 \pm 9.1	11.5	-1.2 \pm 3.2	7.8
Swing (40 to 100)	-1.7 \pm 7.3	14.6	0.2 \pm 7.6	10.0	0.1 \pm 3.0	5.9	0.2 \pm 2.1	4.3	-1.6 \pm 8.7	16.8	-0.3 \pm 5.7	11.8
<i>Patellofemoral Pain</i>												
Q1 (-10 to 10)	1.5 \pm 8.9	10.6	0.8 \pm 11.1	13.0	1.6 \pm 4.9	7.3	1.5 \pm 4.4	7.2	2.6 \pm 7.2	13.2	2.1 \pm 11.4	14.7
Q2 (10 to 30)	1.1 \pm 8.3	12.0	-0.4 \pm 5.2	7.7	0.1 \pm 3.3	6.3	-0.6 \pm 4.1	5.1	1.2 \pm 12.4	15.4	0.5 \pm 7.6	10.1
Q3 (30 to 50)	1.0 \pm 6.6	9.9	0.1 \pm 4.4	3.5	0.5 \pm 3.5	6.0	0.3 \pm 1.6	2.9	1.1 \pm 8.7	16.0	0.5 \pm 5.9	7.0
Q4 (50 to 70)	1.1 \pm 12.0	17.3	-0.7 \pm 10.7	11.0	0.0 \pm 5.4	5.8	0.1 \pm 5.0	4.2	0.0 \pm 10.9	17.0	-1.1 \pm 8.0	10.4
Q5 (70 to 90)	2.4 \pm 16.1	22.3	0.5 \pm 13.5	14.3	-0.6 \pm 2.5	5.0	-0.6 \pm 3.5	3.8	-1.0 \pm 7.7	17.1	-1.4 \pm 10.5	12.6
Stride (0 to 100)	1.1 \pm 6.4	14.3	-0.2 \pm 4.6	10.0	0.0 \pm 2.0	5.9	-0.2 \pm 1.8	4.5	0.2 \pm 5.7	15.5	-0.3 \pm 4.0	10.9
Stance (0 to 40)	0.7 \pm 6.1	9.3	0.0 \pm 3.7	6.9	0.1 \pm 1.9	5.4	-0.3 \pm 2.2	3.6	1.0 \pm 7.5	12.6	0.5 \pm 5.8	8.6
Swing (40 to 100)	1.4 \pm 10.0	17.6	-0.3 \pm 8.6	12.0	-0.1 \pm 2.8	6.3	-0.2 \pm 3.0	5.1	-0.4 \pm 6.4	17.5	-0.9 \pm 5.9	12.4

Note. CAV_{Mean} = mean coupling angle variability over a selected interval of stride for a set of 5 non-consecutive gait cycles taken from a 10 s running trial. δ = mean difference between two CAV_{Mean} values taken from the same 10 s running trial for each subject within a population; 1.96σ =95% confidence interval of δ ; \bar{X} =grand mean for CAV_{Mean} values within a population. All units are in degrees ($^{\circ}$). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion

Table 3.4 Test-retest levels of agreement for CAV_{Max} values within five intervals of stride (I1-5) at a self-selected running pace for six Knee-Ankle joint coupling combinations. Data are displayed separately for Healthy and Patellofemoral Pain populations.

Interval (% of stride)	KV-AI		KV-AF		KF-AI		KF-AF		KR-AI		KR-AF	
	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$
<i>Healthy</i>												
I1 (-10 to 0)	-2.7 \pm 48.0	52.9	-0.7 \pm 36.0	46.9	-2.6 \pm 32.3	45.4	-1.3 \pm 26.4	46.5	-3.8 \pm 34.8	36.3	-1.7 \pm 43.5	35.0
I2 (0 to 10)	-3.9 \pm 12.7	7.4	-5.6 \pm 14.3	18.4	-1.3 \pm 5.8	5.4	-0.5 \pm 3.7	3.3	-4.4 \pm 15.2	12.3	2.4 \pm 21.6	36.8
I3 (10 to 30)	-1.4 \pm 21.4	28.1	0.8 \pm 27.9	39.5	-1.1 \pm 18.8	18.6	-0.9 \pm 10.6	9.6	-6.0 \pm 24.1	31.3	-4.5 \pm 15.1	33.0
I4 (30 to 64)	-12.6 \pm 34.8	31.8	6.3 \pm 28.8	27.2	-1.1 \pm 18.2	24.8	-0.5 \pm 7.1	6.5	-3.8 \pm 18.2	49.0	-2.1 \pm 30.2	44.3
I5 (64 to 90)	-3.1 \pm 18.9	45.0	-1.2 \pm 25.6	24.7	-7.1 \pm 25.6	40.8	2.4 \pm 18.3	15.2	-2.3 \pm 28.7	46.2	2.9 \pm 22.7	33.3
<i>Patellofemoral Pain</i>												
I1 (-10 to 0)	1.1 \pm 23.2	43.3	1.7 \pm 28.8	41.2	6.8 \pm 26.7	40.0	4.7 \pm 29.9	53.0	10.6 \pm 23.8	35.7	5.0 \pm 28.7	36.1
I2 (0 to 10)	1.1 \pm 16.1	14.0	3.8 \pm 33.2	36.6	-0.4 \pm 6.0	5.8	0.0 \pm 5.8	5.4	-2.2 \pm 17.3	15.8	0.9 \pm 29.4	25.1
I3 (10 to 30)	0.5 \pm 23.8	29.2	0.5 \pm 22.7	49.5	-0.2 \pm 16.7	31.6	-1.6 \pm 10.7	11.6	0.9 \pm 18.7	44.4	0.2 \pm 27.0	34.9
I4 (30 to 64)	1.8 \pm 24.5	40.6	-3.2 \pm 30.7	29.9	0.4 \pm 17.6	22.5	1.0 \pm 5.2	7.5	1.2 \pm 33.1	50.8	-1.0 \pm 29.1	38.3
I5 (64 to 90)	4.4 \pm 29.0	54.4	-1.2 \pm 24.8	31.1	-1.6 \pm 29.0	49.0	-1.5 \pm 19.3	17.6	-1.8 \pm 21.1	46.6	-3.0 \pm 22.3	29.9

Note. CAV_{Max} = maximum coupling angle variability value over a selected interval of stride for a set of 5 non-consecutive gait cycles taken from a 10 s running trial. δ = mean difference between two sets of CAV_{Max} values for each subject within a population; 1.96σ =95% confidence interval of δ ; \bar{X} =grand mean for both sets of CAV_{Max} values within a population. All units are in degrees ($^\circ$). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion

Table 3.5 Test-retest levels of agreement for $CAV_{Max\pm 2\%}$ values within five intervals of stride (I1-5) at a self-selected running pace for six Knee-Ankle joint coupling combinations. Data are displayed separately for Healthy and Patellofemoral Pain populations.

Interval (% of stride)	KV-AI		KV-AF		KF-AI		KF-AF		KR-AI		KR-AF	
	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$	$\delta \pm 1.96\sigma$	$\bar{X}(^\circ)$
<i>Healthy</i>												
I1 (-10 to 0)	-1.1 \pm 26.0	28.4	-1.1 \pm 25.0	26.0	-1.4 \pm 22.4	20.6	-0.3 \pm 18.4	19.4	0.7 \pm 33.6	25.8	-2.2 \pm 24.3	26.7
I2 (0 to 10)	-4.2 \pm 13.8	8.6	-6.4 \pm 17.1	13.0	-1.3 \pm 7.1	6.1	-0.7 \pm 4.6	4.0	-4.7 \pm 17.5	12.1	-3.3 \pm 13.3	23.8
I3 (10 to 30)	-0.1 \pm 13.4	18.2	-0.7 \pm 15.5	17.0	-2.5 \pm 14.6	12.0	-0.9 \pm 9.5	8.1	-4.5 \pm 18.2	22.9	-4.0 \pm 9.6	18.5
I4 (30 to 64)	-7.8 \pm 18.3	24.8	1.9 \pm 19.7	15.2	-0.9 \pm 10.1	12.7	-0.4 \pm 6.5	5.7	-3.4 \pm 14.7	35.3	-0.7 \pm 20.2	22.5
I5 (64 to 90)	-5.7 \pm 18.7	35.1	-2.2 \pm 19.3	21.6	-1.1 \pm 23.4	19.5	1.5 \pm 13.9	10.9	-2.6 \pm 21.4	34.6	2.3 \pm 22.6	27.4
<i>Patellofemoral Pain</i>												
I1 (-10 to 0)	2.8 \pm 24.4	25.4	1.4 \pm 24.9	26.2	5.5 \pm 15.3	19.7	5.2 \pm 17.2	22.9	6.0 \pm 22.8	27.3	3.8 \pm 19.2	27.4
I2 (0 to 10)	1.3 \pm 11.3	10.6	0.2 \pm 21.5	20.3	0.1 \pm 8.8	6.9	0.4 \pm 6.2	8.4	-1.7 \pm 12.9	14.4	0.7 \pm 22.9	20.6
I3 (10 to 30)	0.2 \pm 18.9	20.5	-0.5 \pm 20.6	24.1	-0.2 \pm 12.6	17.4	-1.5 \pm 9.6	10.2	1.7 \pm 14.5	29.9	0.5 \pm 18.9	23.3
I4 (30 to 64)	2.5 \pm 20.5	32.1	-3.6 \pm 26.4	22.0	1.0 \pm 9.3	13.7	1.2 \pm 5.9	6.5	1.1 \pm 24.6	36.4	-1.0 \pm 21.3	25.0
I5 (64 to 90)	3.8 \pm 27.9	41.6	-0.3 \pm 21.4	25.9	-1.6 \pm 17.6	24.3	-0.8 \pm 15.4	13.6	-2.1 \pm 15.0	33.7	-2.6 \pm 19.2	23.5

Note. $CAV_{Max\pm 2\%}$ =mean coupling angle variability value within selected an interval $\pm 2\%$ of stride about a given CAV_{Max} stride location. Data were collected for 2 sets of 5 non-consecutive gait cycles taken from a 10 s running trial. δ = mean difference between two sets of $CAV_{Max\pm 2\%}$ values for each subject within a population; 1.96σ =95% confidence interval of δ ; \bar{X} =grand mean for both sets of $CAV_{Max\pm 2\%}$ values within a population. All units are in degrees ($^\circ$). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion

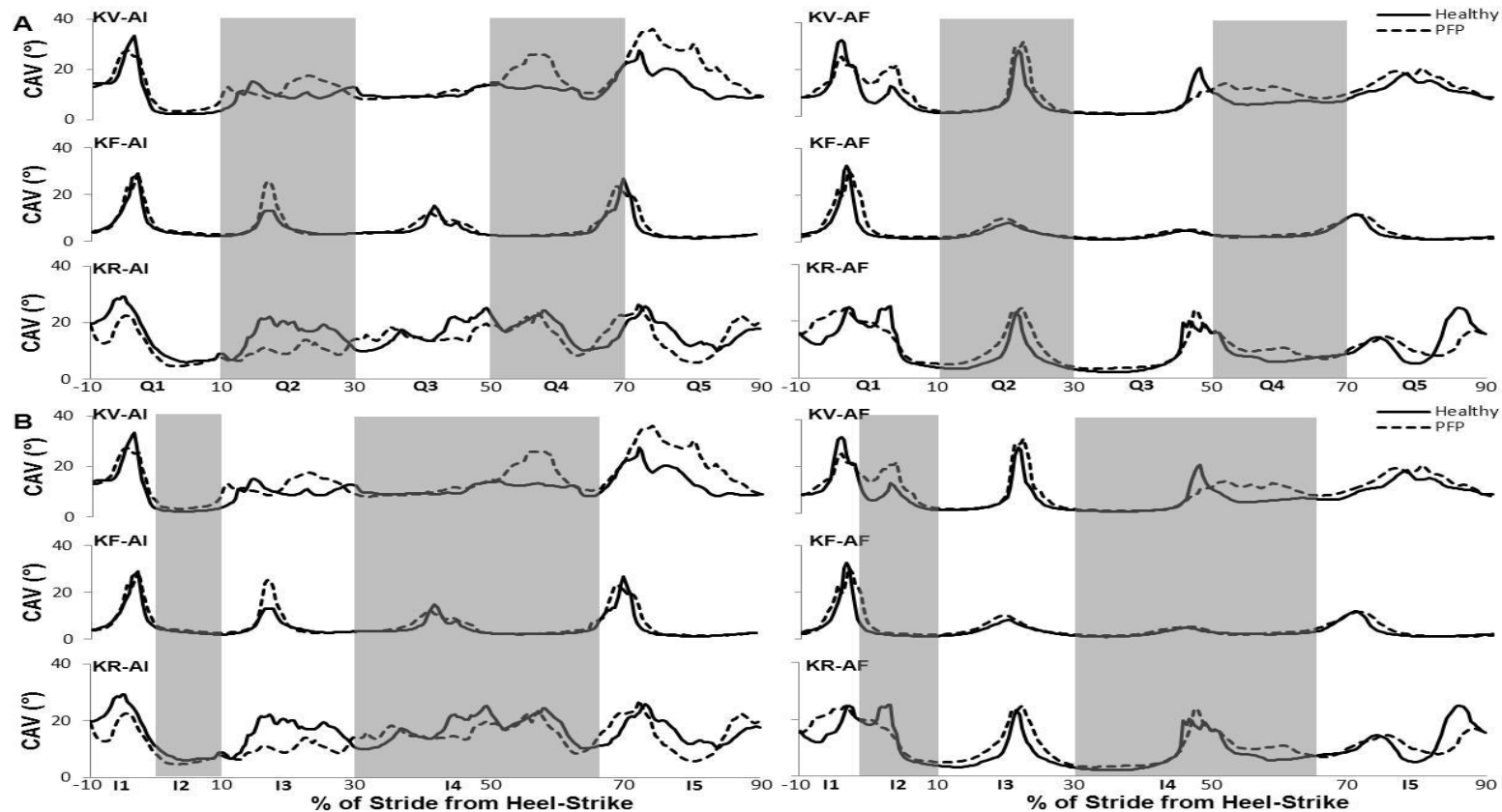


Figure 3.2 Ensemble averaged Coupling Angle Variability (CAV) curves for Healthy and Patellofemoral Pain (PFP) populations taken from 1 set of 5 non-consecutive gait cycles for six Knee-Ankle coupling combinations. (A) highlights stride quintiles (Q1-5) labeled on the horizontal axis with Q2 and Q4 shaded in the plotting area. Q1 begins at -10% stride as measured from heel-strike (0%). (B) shows identical curves highlighting 5 custom intervals of stride (I1-5) with I2(0 to 10%) and I4(30 to 64%) being shaded. Not labeled is toe-off (40%) and the transition from braking to propulsion (i.e. mid-stance, 20%). All vertical axis units are in degrees (°). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion

Discussion

This is the first study to assess the reliability of lower extremity CAV measures in healthy and PFP recreational runners during a treadmill running event, an identified construct to test the relation between joint coordination variability and physiological control factors. LOA values varied greatly among CAV measures, joint couplings and intervals of stride being analyzed. CAV_{Mean} values were generally more reliable than the more sensitive CAV_{Max} and $CAV_{Max\pm 2\%}$ measures but also exhibited substantially lower mean magnitudes in most instances. This may indicate that there is a tradeoff between reliability and sensitivity when observing changes in CAV. The wide range of CAV_{Max} and $CAV_{Max\pm 2\%}$ values throughout the different intervals of stride agreed with previous literature that increases in CAV occur at distinct locations of stride (Heiderscheit, et al. 2002) and the more sensitive CAV_{Max} and $CAV_{Max\pm 2\%}$ seem to assess the amount of variability at these locations quite well, although the wide range of values also suggests CAV magnitude increases are highly volatile and observing clinically meaningful differences in later work may prove difficult. Some portions of the CAV curve might best be described using less sensitive measures giving a general indication of CAV while others might benefit from sensitive measurements for distinct spontaneous increases. An assessment of reliability and sensitivity for each CAV measure and coupling relationship should be considered to choose the appropriate measures for the specific clinical question being studied.

Most surprising was the large amount of inherent variability in all CAV measures between sets of data when no delimited physiological changes were observed. The precision of these measures were poor in the context of the full scale range of CAV values (81°) (Batschelet 1981) with only a few exceptions reporting values less than 10% of capable values. This would be considered irregularly large for clinical instrument standards and may be considered unacceptable. (Mullineaux, et al. 1999) Regardless of the source of variation, physiological or methodical, or the CAV measure used; large changes in CAV

values will need to be observed to consider changes in CAV outside the range of error and clinically meaningful.

The large amount of variability observed coincides with the lack of previous findings using CAV_{Mean} to observe changes in a movement control strategy. (Ferber, et al. 2005, Heidercheit 2000, Heiderscheit, et al. 2002) Reliability of KV-AI CAV_{Mean} during stride ($2.0 - 2.2^\circ$) was larger than the 1.4° significant change reported by Heiderscheit (Heidercheit 2000) using the same coupling and CAV_{Mean} measures when comparing healthy and PFP female runners. This brings into question the clinical validity of Heiderscheit's only observed difference in CAV_{Mean} value for that particular study. Further, long axis thigh-shank coupling in Q1 has also shown differences between PFP and healthy individuals (change of 4°) and between injured and non-injured limbs (change of $\sim 8^\circ$). (Heiderscheit, et al. 2002) Although these couplings were not analyzed in this study, KR-AI and KR-AF which involve the long axis rotation of the thigh and shank, demonstrated precision larger than these differences during Q1 (7.2° to 11.3°). Speculation in these comparison should be viewed with caution as direct comparisons to previous findings are difficult due to the context specific nature of CAV (Maulder 2011) and differences between control factors between studies were likely. Factors thought to affect CAV such as sex (Pollard, et al. 2005), fatigue (Dierks and Davis 2007) and speed (Miller, et al. 2010) were similar; however, pain level in the PFP group in the current study was reported as 4.3 compared to 1.9 in the previous Heiderscheit study. (Heidercheit 2000) Investigation on the effect of this large discrepancy in pain level might reveal a relationship to CAV and pain contrary to previous findings. (Heidercheit 2000) Observed grand means were also smaller in this study which may be attributed to pain level. Rather, this may suggest that joint CAV measures are generally less than individual segment couplings or treadmill running may reduce CAV when compared to overground running. (Wheat, et al. 2005) This further emphasizes the context specificity of couplings and CAV values for this task. Further investigation is warranted to establish reliability of any coupling or CAV measures prior to clinical interpretation.

There were several limitations to this study. Only six joint couplings were analyzed of the 136 possible joint and segment couplings of the lower extremity. Couplings used are consistent with previous literature with the exception of KV-AI, KV-AF and the exclusion of segment couplings.(Heiderscheit, et al. 2002) Long axis rotation of the thigh, shank and eversion of the ankle are thought to lead to overuse injury of the knee.(Hamill, et al. 1999, Tiberio 1987) These specific segments weren't studied; however, segment variability may manifest in joint CAV and patterns may become apparent when viewed comprehensively as it is in this study. CAV is thought to be context specific with slight variations in physiological control parameters possibly having a large effect on CAV.(Maulder 2011) Test-retest methods used in this study minimized the changes that can occur within a subject but all extraneous variables within each population may not have been controlled which could have increased the confidence intervals observed. Additionally, the reliability values reported for this study are delimited to the population and tasks performed. Extrapolation of these values should be viewed with caution.

In summary, the precision of several CAV measures has now been established for a PFP and healthy population under the reported constructs. Two new CAV measures were presented as measurements more sensitive to CAV increases during the stride cycle than previously reported CAV_{Mean} measures. Confidence interval limits reported in this study represent the amount of change a respective CAV measure must present to be considered clinically meaningful. CAV measure changes in response to a control parameter change can now be interpreted clinically with less conjecture in future analyses.

Chapter 4: Coupling Angle Variability measure response to an increase in pain

This chapter aims to present a study to determine the clinical validity of CAV measures to observe possible changes in coordinate state when the physiological control parameter of knee pain is increased for runners with PFP. This chapter can be read independently from the rest of the text in manuscript format suitable for submission for publication.

Introduction

A movement control strategy is a function of the complex interactions of three multidimensional control parameters; environment, organism and task. (Bernstein 1967, Newell, et al. 1993, Turvey 1990) A large change to a control parameter such as knee pain (organism) may result in a substantial change in strategy and shift in coordinative state. Conversely, if no control parameters have been altered, resulting variation in strategy should be minimal and remain within the same coordinative state. (Kelso 1995) Patellofemoral Pain (PFP) has been suggested to be the result of decreased joint coupling variability where an over constrained control system does not allow the shifting of coordinate states and can lead to an overuse injury. (Hamill, et al. 1999) The prevailing symptom amongst individuals with PFP is an increase in pain experienced as a result of a physical activity, particularly running. (Clement, et al. 1981, Taunton, et al. 2002) Pain commonly develops about the knee and worsens during the course of a run possibly indicating a pathological coordinate state responsive to the exhibited amount of pain. Studies examining possible relationships between PFP symptoms and traditional joint kinematic gait measures during running have yielded inconclusive or conflicting results. (Barton, et al. 2009, Davis and Powers 2010, Dierks, et al. 2011, Noehren, et al. 2011) Coupling angle variability (CAV) has been suggested as a potentially useful measure to distinguish among coordinative states dependent on physiological control parameters, (Dierks and Davis 2007, Ferber, et al. 2005, Heidercheit 2000, Heiderscheit, et al. 2002, Miller, et al. 2010, Pollard, et al. 2005, Wilson, et al. 2008) including runners diagnosed with PFP. (Heidercheit 2000)

Pain as a control parameter has previously been investigated using CAV measures with no correlation found between a reduction in pain and CAV. (Heidercheit 2000) Additionally, mean CAV values were significantly greater for Knee Flexion-Ankle Inversion joint couplings over the entire stride cycle in a PFP population when compared to healthy controls. This is contradictory to the dynamical systems approach to overuse injuries which predicts lower variability is indicative of a pathological state. (Hamill, et al. 1999) The only mean CAV measures to support the theory of lower variability in a PFP population was found in the long axis rotation of the thigh-shank segment coupling at heel-strike for both pain conditions. This evidence; however, should be viewed with caution as angular measures in the transverse plane are the least reliable during running gait. (Ferber, et al. 2002)

Several limitations to this study may have contributed to the limited findings. Average pain levels only decreased from 1.9 to approximately .7 as measured using a visual analogue scale (VAS). This small amount of change in pain may not have been large enough to elicit observable changes in CAV. (Crossley, et al. 2004) Order parameter response to an identified control parameter is theorized to exhibit hysteretic characteristics. (Turvey 1990) Applied to a dynamical systems perspective to PFP, CAV measures may respond differently with an increase in pain than with a reduction in pain if a coordinate state change occurs. Methodical issues such as foot marker set, gait normalization procedures, amount of stride cycles analyzed and motion capture parameters could also have effected CAV measure reliability (Mullineaux, et al. 2006) decreasing chances of identifying possible differences.

Responsiveness of CAV measures must predictably correspond to a physiological change to be considered clinically useful. (Crossley, et al. 2004) Use of CAV as a clinically useful measure is not yet evident. Perceived knee pain is a valid measure when used to diagnose PFP and evaluate effectiveness of treatment programs. (Crossley, et al. 2004) The concurrent validity of CAV measures to physiological changes as described by perceived knee pain level is not established. Validation of CAV measures as clinically useful requires identification of a control parameter and alteration of that control parameter with a measureable change in CAV values. It is theorized that knee pain in a PFP population is a control parameter or

directly related to a control parameter effecting joint coordination patterns and can be measured using CAV. Few changes were observed in previous literature when pain was reduced; however, CAV measure sensitivity to an increase in pain has never been investigated. This study aims to investigate the relationship between an increase in knee pain and a variety of CAV measures during running at a self-selected pace; an activity related to development of PFP. (Davis and Powers 2010) In accordance with a dynamical systems theory of overuse injury, it is hypothesized that CAV values will decrease with a clinically meaningful increase in knee pain. Findings contrary to this hypothesis will suggest those CAV measures are not adequate to observe possible changes in response to a change in knee pain and support for dynamical systems theory in this construct is not appropriate.

Table 4.1 Common abbreviations and definitions used within the text and tables grouped by Knee-Ankle coupling relationship and coupling angle variability (CAV) measures. Measures are for each quintile and intervals of stride.

Joint Coupling	Definition
KV-AI	Knee Valgus/Varus coupled with Ankle Inversion/Eversion
KV-AF	Knee Valgus/Varus coupled with Ankle Plantar/Dorsi Flexion
KF-AI	Knee Flexion/Extension coupled with Ankle Inversion/Eversion
KF-AF	Knee Flexion/Extension coupled with Ankle Plantar/Dorsi Flexion
KR-AI	Knee Internal/External Rotation coupled with Ankle Inversion/Eversion
KR-AF	Knee Internal/External Rotation coupled with Ankle Plantar/Dorsi Flexion
CAV Measure	
CAV	Coupling Angle Variability. Variation within a set of 5 vector coded, non-consecutive gait cycles for a Knee-Ankle coupling relationship. CAV is a continuous measure for every point in the gait cycle. Units are in degrees.
CAV _{Mean}	Mean CAV value over discrete intervals (Q, I, stance, swing) of stride. Each quintile contains a functional period of stride shown in parentheses.
Quintiles (Q)	Q1: -10 to 10% (heel-strike), Q2: 10-30% (mid-stance), Q3: 30 to 50% (toe-off), Q4: 50 to 70% (swing acceleration), Q5: 70 to 90% (swing deceleration)
Intervals (I)	I1: -10 to 0%, I2: 0 to 10%, I3: 10 to 30%, I4: 30 to 64%, I5: 64 to 90%
Stance	0 to 40%
Swing	40 to 100%
Stride	0 to 100%
CAV _{Max}	Maximum CAV value within a selected interval of stride.
CAV _{Max±2%}	Mean CAV for an interval $\pm 2\%$ of stride about a CAV _{Max}

Methods

Twenty injured female recreational runners originally participated in the treadmill protocol of this study. To participate, all females had to be between 18 to 45 years of age and run a minimum of 16 km (10 miles) per week, if they self-reported a knee pain of a 3 or greater out of 10 during normal running activity and were currently diagnosed with PFP by a certified athletic trainer or licensed physical therapist after exclusion of knee pain resulting from acute injury, patellar tendonitis, Iliotibial band syndrome or meniscal pathology. Potential subjects were excluded if they had a neurological disorder, tape allergy or felt they could not maintain a minimum pace of 3.3 m/s (8 minute 20 s mile) for 2 minutes. Written informed consent was obtained prior to participation in the study, which was approved by the institute's institutional review board.

Retro-reflective markers were attached to the subjects to model bilateral, hip, knee and ankle articulations. (Figure 1) The distal aspects of each thigh and shank were wrapped with elastic straps (ProWrap, Fabrifoam, Exton, PA) and rigid body clusters were then attached to the straps with hook and loop connectors and secured using additional elastic straps (MediPro, Fabrifoam, Exton, PA). Subjects wore standardized shoes (ZoomAir; Nike, Beaverton, OR) modified with windows cut out allowing adhesion of the markers directly to the skin by means of both adhesion spray and toupee tape.

Kinematic data was captured using a combination of 15 Eagle and Eagle4 cameras at 300 Hz (Motion Analysis Corporation, Santa Rosa CA). A dual belted treadmill instrumented with a force plate under each belt (TM-09-PBertec, Columbus, OH) was used to collect ground reaction force data at 1200 Hz. The treadmill belt speed was operated remotely by the investigators with a velocity resolution of 0.01 m/s with each belt being 48 cm wide and 164 cm long. A 15 point Rating of Perceived Exertion (RPE)(Borg 1982) was placed on a stand directly in front of the treadmill for subjects to reference to report level of perceived fatigue during the run. Perceived pain during the run was collected using a verbally administered numeric pain rating scale (NPRS) described to

subjects as 0 being “no pain” and 10 considered “worst imaginable pain”.
(Farrar, et al. 2001)



Figure 4.1 Markerset used during a static calibration. Only bilateral markers on the lateral aspects of the 5th metacarpal head, base, navicular and both the lateral and medial aspects of the calcaneus were used to model foot movement. Windows are cut out of the shoes allowing markers to be adhered directly to the foot at locations consistent with Pohl et al.(Pohl, et al. 2007). Rigid clusters were secured to the distal posterior-lateral aspects of each segment to model thigh and shank movement.

Treadmill Protocol

A one second standing static calibration file was captured while the subjects stood in the anatomical standing position. Subjects then walked on a single belt of the treadmill for 3 minutes at 1.3 m/s to acclimate themselves to the treadmill. Speed was then increased for 3 minutes to a warm-up pace (2.2-2.3 m/s) followed by 2 minutes at a standard pace of 3.3 m/s. Speed was then set at a self-selected pace where subjects felt they would not become severely fatigued

over the course of the next 15 minutes with speed being adjusted upon request (2.2 to 3.3 m/s). To be included in the PFP group, subjects had to reach a minimum knee pain of 3 during the treadmill protocol. Kinematic and kinetic data were acquired for the first 10 seconds of each minute interval. RPE and NPRS measures were recorded by investigators immediately following each 10 second data acquisition.

Data Processing

Kinematic markers were identified using Cortex 2.0 software (Motion Analysis Corporation, Santa Rosa CA). Three-dimensional marker coordinates and force plate data were exported to Matlab v2009a (Mathworks, Natick MA) for gait analysis. A fourth-order lowpass butterworth filter with a cutoff frequency of 8 Hz was applied to kinematic data. Force component data were filtered with a cutoff frequency of 30 Hz for the lateral forces and at 40 Hz for the vertical component. Cut-off frequencies were selected by investigators after visual inspection of a fast fourier transformation performed on the data. Joint coordinate systems were determined using the International Society of Biomechanics recommendations (Grood and Suntay 1983, Wu, et al. 2002). Segment orientations were determined using a singular value decomposition algorithm (Söderkvist and Wedin 1993) and joint angles using an Euler rotation sequence of long axis rotation-abduction-flexion for the knee and ankle.

Consistent gait points of heel-strike, mid-stance and toe-off were determined for each gait cycle for normalization. Heel-strike and toe-off were determined using the vertical component of the ground reaction force with a threshold of 50 N, mid-stance was the transition from braking to propulsion using the lateral component (0 N)(Cavanagh and LaFortune 1980). Both of the two periods of stance were time normalized to 50 points and swing phase to 150 points using a 5th order cubic spline function making a 250 point time normalized gait cycle (1 point=0.4%). Multiple normalization constraint points were chosen as it may reduce within and between subject variability of CAV measures. (Mullineaux, et al. 2006) The first and last gait cycle from each 10 s trial was

discarded to reduce interpolation effects and the first 10 gait cycles were kept for analysis.

Data Reduction

Two 10 s trials were chosen from the 15 minute period of self-selected running pace to represent a high pain (HP) and a low pain (LP) condition for each individual. First, the trial with the highest pain value with a RPE value of less than 14 was selected as the HP. If there was more than one trial that qualified, the trial with the lowest RPE value was chosen. If there was more than one trial with the same RPE and pain value, preference was given to the earlier time point during the run to reduce time between measurements. Secondly, the trial with the lowest pain value at the same running speed as the HP with no more than a change of 1 on the RPE scale was selected. If more than one 10 s trial qualified, the later time point was selected. A minimum change in pain of 2 between LP and HP was required as it signifies a clinically meaningful change in pain for a PFP.(Piva, et al. 2009) 13 of the 20 subjects qualified for analysis based on the pain and RPE criterion set.

CAV values were determined using a revised vector coding technique. (Heiderscheit 2000, Sparrow, et al. 1987) Five non-consecutive stride cycles from each 10 s trial of the injured leg were used for analysis. CAV values were derived for LP and HP for all knee and ankle coupling combinations (Table 4.1) at each point in the gait cycle. The normalized stride cycles were divided into quintiles each containing a functional period of stride (Heiderscheit, et al. 2002)(Table 4.1) with the border of each quintile overlapping. Mean CAV values (CAV_{Mean}) were calculated for quintiles (Q), stance, swing and the entirety of stride for each set, respectively.

Further observation of the CAV curves revealed there were several locations where CAV increases were not encompassed using the standard 20% quintiles. Therefore, five intervals (I) were created to capture consistent increases in the CAV values among all coupling relationships (Table 4.1). These intervals as well as the quintiles appeared too large to be sensitive enough to

distinguish between distinct variability characteristics using traditional CAV_{Mean} values; therefore, the maximum CAV value (CAV_{Max}) and the mean CAV from an interval $\pm 2\%$ of stride about each CAV_{Max} were found within each interval ($CAV_{Max\pm 2\%}$) for each set of gait cycles. Steps involved in construction of the intervals for CAV_{Max} and $CAV_{Max\pm 2\%}$ locations are further detailed (Appendix E and F).

Statistical Analysis

Paired t-tests were performed between all LP and HP CAV measures. Significance levels were set a priori ($\alpha \leq .05$).

Results

Demographics for the qualifying subjects are presented in Table 4.2. Ensemble averages of LP and HP CAV curves for each coupling relationship are shown in Figure 4.2. Qualitatively, little differences were observed between populations with the exception of the local peaks in I5 KF-AI and I1 KV-AF which saw slightly greater LP values.

Paired t-tests between CAV measures revealed few significant differences from LP to HP. Of the few observed changes, all were increases in CAV measures with an increase in pain. CAV_{Mean} values increased in Q1 KF-AI from 6.9° (2.3) (mean (SD) to 8.3° (2.8) ($p=.036$) and in Q1 KF-AF 6.6° (2.0) to 8.2° (2.1) ($p=.037$). $CAV_{Max\pm 2\%}$ values were shown to increase in I1 KF-AI from 18.2° (8.3) to 23.5° (9.6) ($p=.037$). No significant differences were observed with any CAV_{Max} measures. All CAV_{Mean} (Figure 4.3), CAV_{Max} (Figure 4.4) and $CAV_{Mean\pm 2\%}$ (Figure 4.5) values are reported.

Table 4.2 Subject demographics, Rated Perceived Exertion (RPE) and pain values for PFP subjects at a low pain (LP) and high pain (HP) condition during a treadmill run at a preferred running speed.

Height (m)	Mass (kg)	Age (yrs)	Distance (km/wk)	Speed (m/s)	LP RPE (6-20)	LP Pain (0-10)	HP RPE (6-20)	HP Pain (0-10)
1.63 (0.07)	58.4 (6.7)	27.2 (6.8)	23.1 (10.6)	2.49 (0.2)	11.8 (1.2)	1.5 (1.5)	12.2 (0.9)	4.2 (1.4)

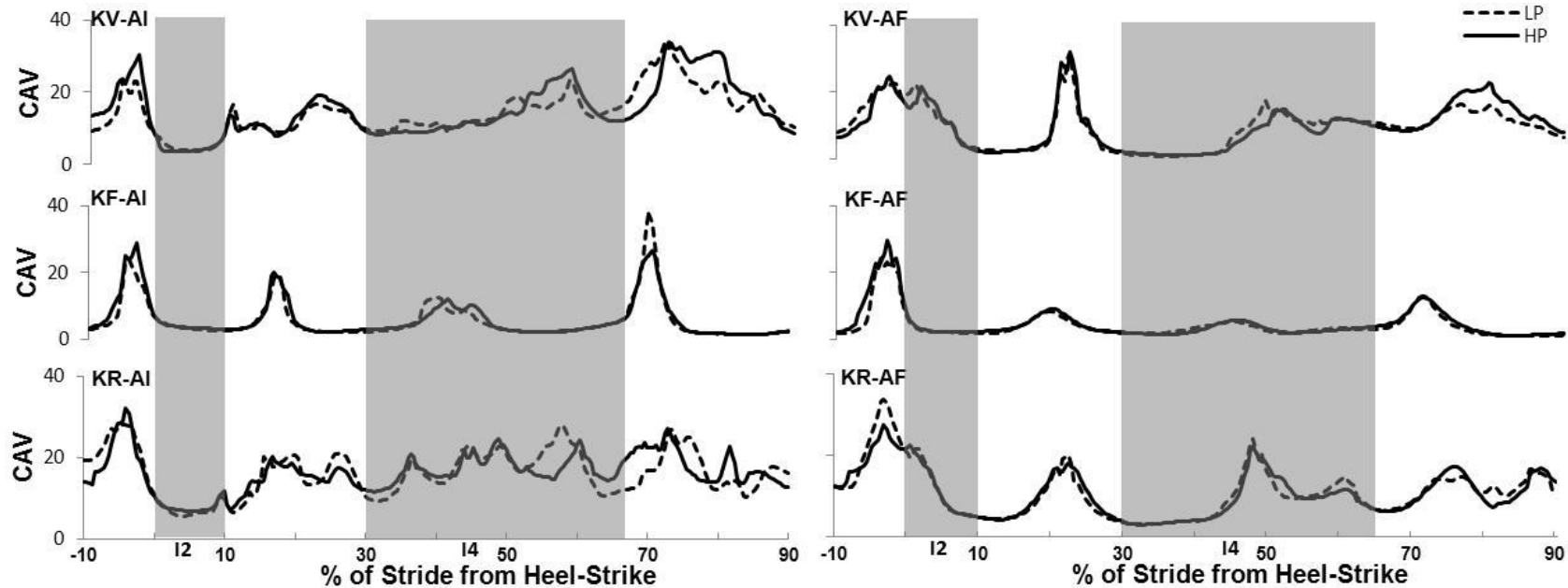


Figure 4.2 Ensemble averaged Coupling Angle Variability (CAV) curves for a low pain (LP) and high pain (HP) condition for female runners with patellofemoral pain taken from 5 non-consecutive gait cycles for six Knee-Ankle coupling combinations. There are five custom intervals of stride (I1-5) with I2(0 to 10%) and I4(30 to 64%) shaded. Quintiles (Q1-5) are every 20% of stride starting at -10%. Not labeled are heel-strike (0%), mid-stance (20%) and toe-off (40%). All vertical axis units are in degrees ($^{\circ}$). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

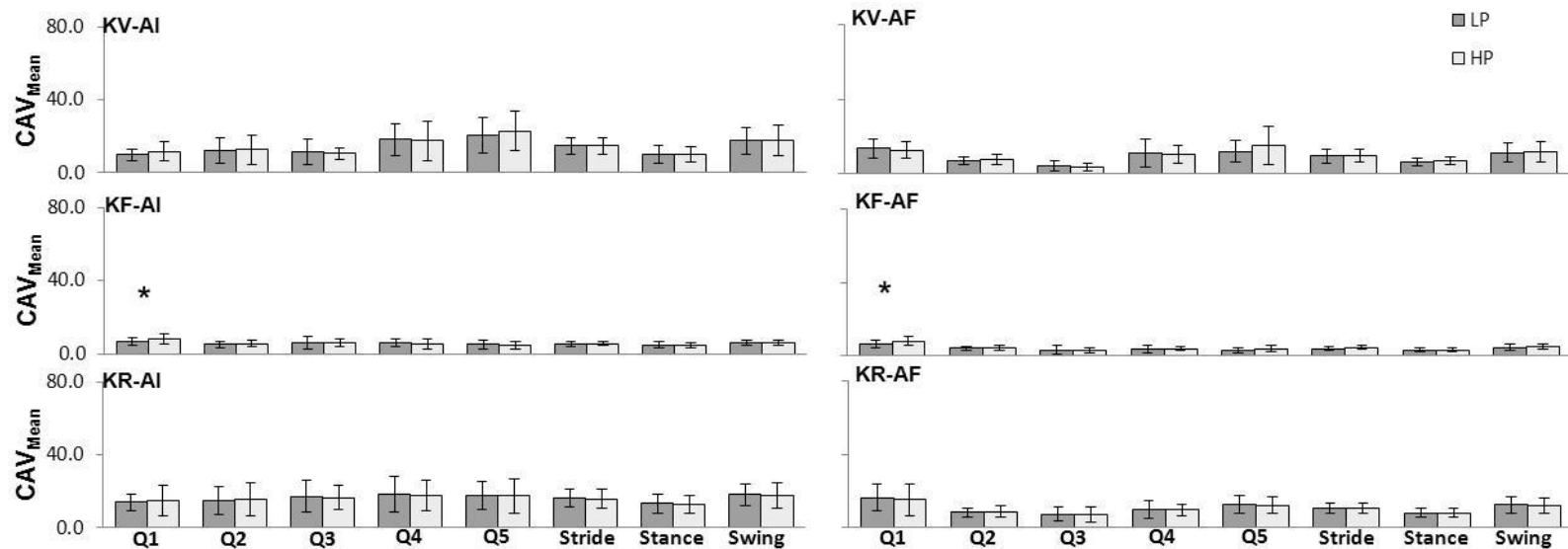


Figure 4.3 Mean Coupling Angle Variability (CAV_{Mean}) values within each quintile (Q1-5, Table 1) of stride, the entirety of stride, stance and swing phase at a self-selected running pace for six Knee-Ankle joint coupling combinations for female runners with patellofemoral pain. Data are taken from 5 non-consecutive gait cycles from a 10 s running trial for a low pain (LP) and high pain (HP) condition. Units of CAV_{Mean} values are in degrees. Significant differences are denoted $p < .05$ (*). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

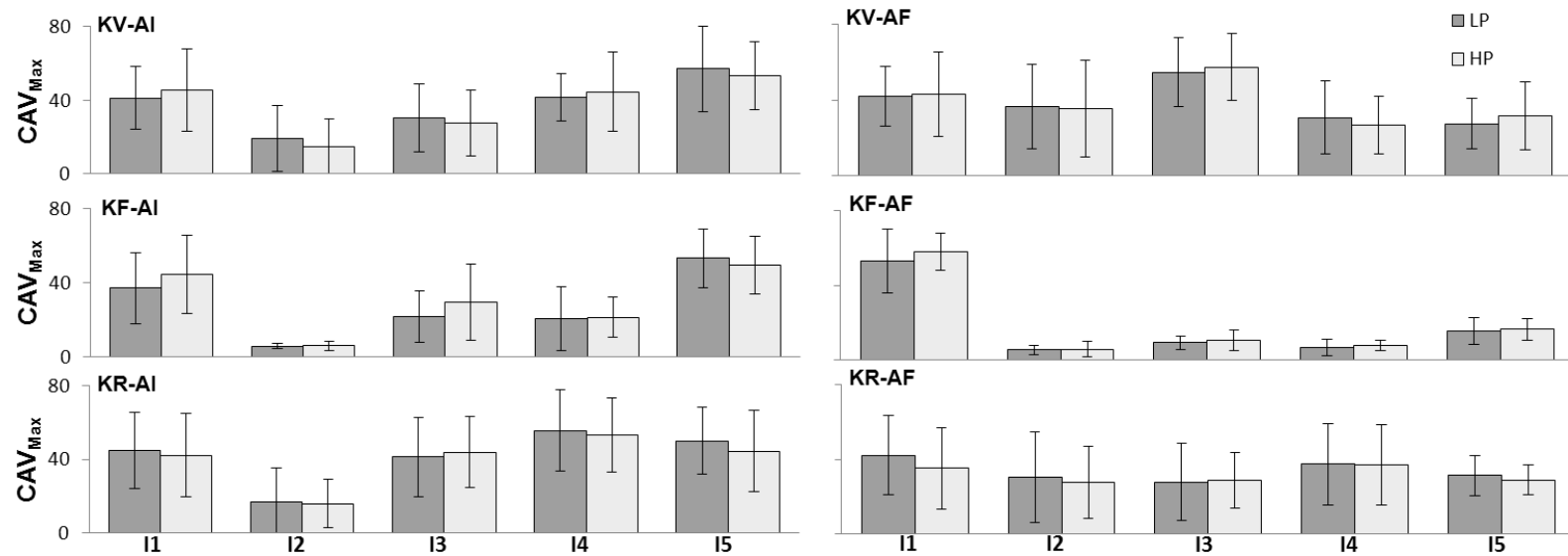


Figure 4.4 Maximum Coupling Angle Variability ($CAV_{Max} \pm SD$) values for five intervals (I1-5, Table 1) of stride at a self-selected running pace for six Knee-Ankle joint coupling combinations for female runners with patellofemoral pain. Data are taken from 5 non-consecutive gait cycles from a 10 s running trial for a low pain (LP) and high pain (HP) condition. Units of CAV_{Max} values are in degrees. No significant differences were observed. Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

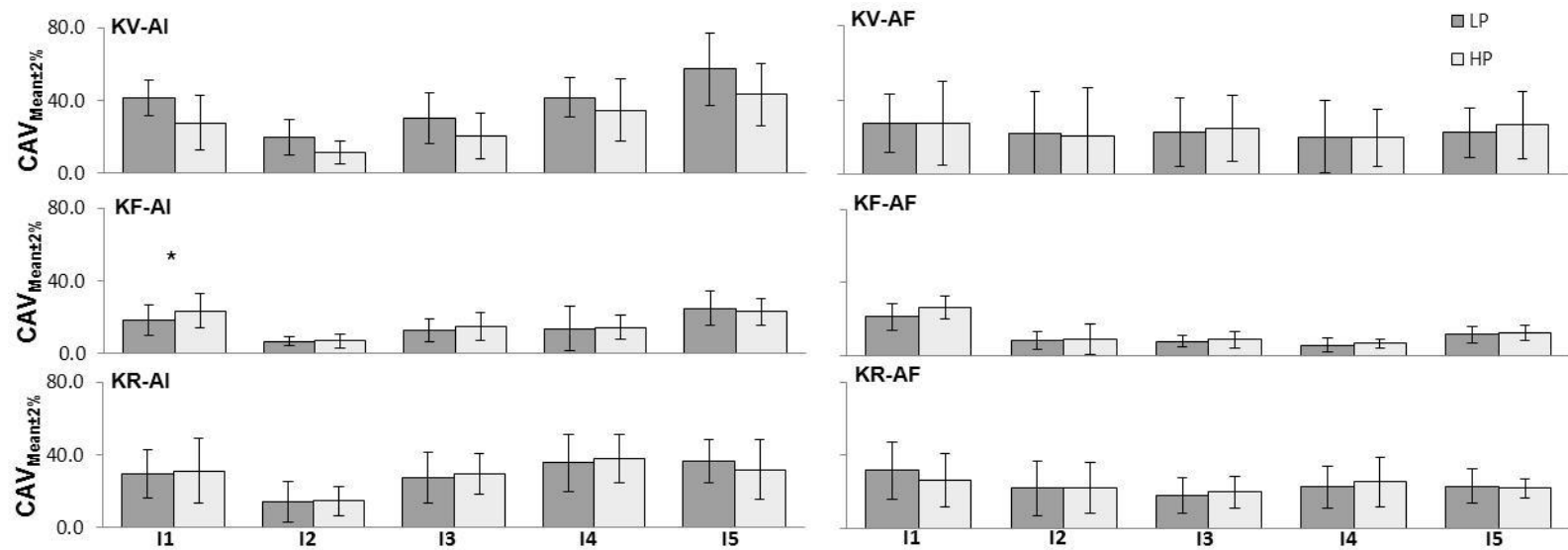


Figure 4.5 Mean Coupling Angle Variability over an interval of $\pm 2\%$ of stride about a CAV_{Max} stride location ($CAV_{Mean \pm 2\% \pm SD}$) for five intervals (I1-5, Table 4.1) of stride at a self-selected running pace for six Knee-Ankle joint coupling combinations for female runners with patellofemoral pain. Data are taken from 5 non-consecutive gait cycles from a 10 s running trial for a low pain (LP) and high pain (HP) condition. Units of CAV_{Mean} values are in degrees. Significant differences are denoted at $p = .037$ (*). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

Discussion

The hypothesis that CAV values would decrease with increasing knee pain was not supported. There appeared to be no relationship between pain level and CAV for all knee-ankle joint couplings for the entire PFP population with the exception of KF-AF and KF-AI near heel-strike. Most surprising were the increases in KF-AF and KF-AI CAV measures with increasing pain. This is seemingly contradictory to the dynamical systems approach to overuse knee injuries which suggests that lower variability is a product of a pathological state. (Hamill, et al. 1999) Q1 and I1 are regions that encompass and precede heel-strike, an identified region where reduced CAV could be detrimental when preparing for high impact of the stance phase loading. (Heiderscheit 2000, Heiderscheit, et al. 2002) Reduction of knee flexion angle has been observed in walking gait (Nadeau, et al. 1997, Powers, et al. 1999) and running gait (Dierks, et al. 2011) in PFP populations which may be a compensatory mechanism to reduce forces to the knee. (Dillon, et al. 1983) Likewise, the observed increase in KF-AF and KF-AI CAV at this point might be a compensatory mechanism as a response to reduce knee pain. Heiderscheit et al. (Heiderscheit 2000) observed larger Stride KF-AI CAV_{Mean} values in a PFP population than healthy controls in a painful and reduced pain state while running at a preferred running speed. Average values observed in their study (9° to 10.9°) were slightly higher than observed in this study (4.0° to 5.7°) with differences possibly attributed to treadmill vs. overground running which may decrease coupling variability. (Wheat, et al. 2002) This adds evidence to suggest that a joint coordinate state exists in a PFP population that differs than that from a healthy population and may even exhibit larger values. These values may be context specific to the task being performed.

Increases in pain with no observed changes in CAV values could be the result of remaining in the same pathological coordination pattern regardless of pain level. Pain values increased between the two conditions on average of 2.7 on an NPRS which is considered a clinically meaningful change in pain. (Piva, et al. 2009) This large change in pain did not show a decrease in CAV despite

assessing the limitations of previous work that also showed no changes in CAV_{Mean} values when pain changed by 1.2 on a similar VAS scale. (Heidercheit 2000) Assuming the CAV measures studied are capable of measuring changes in a coordinative state, it is possible the coordinate state never switched after an initial onset of pain was reached or there was no coordinative switch at all. The initial CAV measurement for each respective study was after an initial onset of pain on average (1.5 and 1.9). Following a dynamical system's perspective, this indicates that once a pain threshold is reached a pathological coordinative structure remains intact. This would suggest that CAV values are not related to the current state of pain but rather representative of the PFP population.

The clinical interpretation of the significant changes in KF-AF and KF-AI CAV measures should be interpreted with caution. Knee and ankle flexion are the most reliable angle measurements during running gait (Ferber, et al. 2002) and although analysis of angles was not performed in this study, values were visually observed and determined to be consistent with previous literature and highly repeatable within and between subjects. The high reliability of these measurements infers that these measurements were not a result of measurement error. Observed changes in CAV_{Mean} and $CAV_{Mean \pm 2}$ were within the limits of precision reported during reliability testing of these same measures when physiological parameters were held constant. (Chapter 3) This suggests that these results, although statistically significant, may not be clinically useful when extrapolated to represent an entire population for clinical use. Further, CAV measures investigated may not be a valid representation of changes in joint coordination variability as a result of physiological changes that may occur with an increase in knee pain.

Results of this study yielded little evidence to support the theory that there is a coordinative structure change in relation to knee pain that can be measured using CAV. This study addressed several methodical limitations identified in the literature but had its own. Only knee-ankle couplings and a few discrete CAV measures were analyzed, making comparison to previous literature and extrapolation of results difficult. Couplings involving rotation of the thigh and tibia

in the transverse plane and rearfoot eversion are thought to be involved in the development of knee pain (Tiberio 1987) and is the basis of a dynamical systems approach to overuse injuries. (Hamill, et al. 1999) This study was delimited to knee-ankle joint couplings but other couplings might yield more promising results. The warm-up running protocol used, limitations of minimal fatigue change and higher pain values helped homogenize the population analyzed reduced the amount of qualified subjects and resulted in LP pain values of 0 for only 4 of the 13 subjects. Future analysis of subjects from a fresh state to a painful state would provide more information regarding the presented theory that a shift coordinate state might occur after an onset of pain. There are many possible PFP pain scenarios and the currently studied construct is only one of them. Other scenarios and methods may lead to different results and should be investigated before disregarding a knee pain CAV relationship.

The proposed etiology that PFP symptoms are a manifest of less joint coordination variability and CAV measures are able to observe changes to the physiological control parameter of knee pain is not promising. If joint coordination variability is related to knee injuries as theorized, (Hamill, et al. 1999) less variability might be involved in developing PFP symptoms rather than a result of the an immediate increase in observed symptoms. Knee pain may not induce a large enough physiological change that can be accurately measured using CAV. Investigations of control parameters that might have a larger effect on CAV are warranted. Evidence relating any joint variability measures to overuse injuries is extremely weak. (Ferber, et al. 2005, Hamill, et al. 1999, Heiderscheit, et al. 2002) The only clinically meaningful CAV differences ever reported identified differences in gender, (Maulder 2011, Pollard, et al. 2005) which involves cohorts of exceptional differences. Analysis methods to identify differences between individuals with PFP from healthy are still needed. (Davis and Powers 2010) Future investigations of CAV measures should concentrate on control parameters that may induce large changes to CAV values, such as population, as a possible construct to validate vector coding as a viable clinical tool.

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Chapter 5: Coupling Angle Variability measure differences between a PFP and healthy population

The aim of this chapter is to present a study to determine the clinical validity of CAV measures to distinguish between possible different coordinate states between healthy runners and runners with PFP. This chapter can be read independently from the rest of the text in manuscript format suitable for submission for publication.

Introduction

For a measure to be clinically useful from a dynamical systems perspective, it must behave according to a theoretical construct, predictably respond to changes in a control parameter (Turvey 1990) and be sufficiently reliable to observe real change. A movement control strategy is a function of the complex interactions of three multidimensional control parameters; environment, organism (population) and task. (Bernstein 1967, Newell, et al. 1993, Turvey 1990) Variability in joint coordination has been suggested as an indirect representation of variability in movement control strategy (Turvey 1990) and inherent within a healthy control strategy. (Newell, et al. 1993, Stergiou, et al. 2006) A dynamical system's perspective to lower extremity orthopaedic injuries suggests that a low amount of variation in joint coordinative structure may increase the frequency of loading of soft tissue and eventually lead to an overuse condition and pathological state. (Hamill, van Emmerik et al. 1999) Patellofemoral Pain (PFP) is thought to be a condition resultant of this decrease in variability. (Hamill, et al. 1999) In the original investigation of this theory continuous relative phase was used to assess movement variability (Kelso 1995); however, this technique has limitations in quantifying non-sinusoidal joint couplings and may not be appropriate for lower extremity couplings during gait. (Peters, Haddad et al. 2003) Coupling angle variability (CAV) has been suggested as an alternative measurement method to observe changes in coordinative state between PFP and healthy populations. (Heiderscheit, et al. 2002)

Previous literature using CAV has found little evidence to support its use as a clinically useful measure in relation to overuse injury. (Ferber, et al. 2005, Heiderscheit, et al. 2002, Maulder 2011) Investigating this theory, Heiderscheit et al. (Heiderscheit, et al. 2002) compared mean CAV (CAV_{Mean}) values over the entire stride cycle for several joint and segment couplings between PFP and healthy individuals while running at a self-selected pace. No differences between populations were found. Further analysis using CAV_{Mean} over smaller quintiles of stride only revealed less variability in the PFP population for the coupling of thigh-shank long axis rotation near heel strike. The clinical relevance of this evidence is unclear and should be interpreted with caution (DeLeo, et al. 2004) as angular measures in the transverse plane are the least reliable during running gait. (Ferber, Davis et al. 2002) Employing similar analysis methods when assessing the effects of orthotics on injured runners with an array of overuse injuries, introduction of an orthotic improved symptoms but no changes in CAV were observed. Minimal pain values reached (Heiderscheit, Hamill et al. 2002) and a heterogeneous injured population (Ferber, Davis et al. 2005) were cited as possible factors for the limited results. CAV is thought to be context specific (Maulder 2011) and variability through methodical error needs to be minimized to understand the physiological variability within a joint coordinate system.

Several limitations to these studies may have contributed to the limited findings. Previous literature studying joint kinematics of runners with PFP has consistently used a minimum pain level of 3 as an inclusion criterion. (Dierks, et al. 2011, Dierks, et al. 2008, Noehren, et al. 2011, Willson and Davis 2008) An average pain level of only 1.9 was reached in the population analyzed by Heiderscheit et al. (Heiderscheit, et al. 2002) A population capable of achieving a larger amount of pain or a critical threshold of pain may be required to observe a pathological coordinative state. Methodical issues such as foot marker set, gait normalization procedures, amount of stride cycles analyzed, small sample sizes and motion capture parameters effect the precision and accuracy of CAV measures (Mullineaux, et al. 2006) decreasing the likelihood of identifying real differences. (Maulder 2011) Analysis using CAV measures that are more

sensitive to changes in spontaneous increases in CAV than the previously used CAV_{Mean} intervals might also lead to more promising results. These limitations should be addressed to further assess the validity of CAV as a clinically useful measure for joint coordination variability.

Validation of CAV measures as clinically useful from a dynamical systems approach requires identification of a control parameter, alteration of that control parameter and observing measureable changes in CAV within an identified theoretical construct. It has been suggested that PFP is a condition resulting from a pathological coordinate state which has a lower amount of joint coordination variability than a healthy population. This is a plausible theoretical construct which alters the population to test the theory that a lower amount of joint coordination variability is indicative of overuse injury. There is little evidence to suggest that CAV is a clinically useful measure when population has been treated as a control parameter. This study aims to address identified limitations of previous literature and determine if a variety of CAV measures can observe a meaningful change in value between a healthy population and a population with PFP during running at a self-selected pace; an activity related to development of PFP. (Davis and Powers 2010) It is hypothesized that CAV values will be less for PFP individuals. Accepting this hypothesis would support the theoretical construct put forth in the dynamical systems model that lower variability places excessive load on structures about the lower extremity. Rejection of this hypothesis would suggest CAV measures are not adequate to observe changes in population consistent with a dynamical systems approach.

Methods

Twenty-one healthy and twenty injured female recreational runners originally participated in the study. To participate, all females had to be between 18 to 45 years of age and run a minimum of 16 km (10 miles) per week. Subjects were included in the healthy group if they had no history of PFP and reported no lower extremity pain while running. Subjects were included in the PFP group if they self-reported a knee pain of a 3 or greater out of 10 during normal running activity and were currently diagnosed with PFP by a certified

athletic trainer or licensed physical therapist after exclusion of knee pain resulting from acute injury, patellar tendonitis, Iliotibial band syndrome or meniscal pathology. Potential subjects were excluded if they had a neurological disorder, tape allergy or felt they could not maintain a minimum pace of 3.3 m/s (8 minute 20 s mile) for 2 minutes. Written informed consent was obtained prior to participation in the study, which was approved by the institute's institutional review board.

Table 5.1 Common abbreviations and definitions used within the text and tables grouped by Knee-Ankle coupling relationship and coupling angle variability (CAV) measures. Measures are for each quintile and intervals of stride.

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KF-AF	Knee Flexion/Extension coupled with Ankle Plantar/Dorsi Flexion
KR-AI	Knee Internal/External Rotation coupled with Ankle Inversion/Eversion
KR-AF	Knee Internal/External Rotation coupled with Ankle Plantar/Dorsi Flexion
CAV Measure	
CAV	Coupling Angle Variability. Variation within a set of 5 vector coded, non-consecutive gait cycles for a Knee-Ankle coupling relationship. CAV is a continuous measure for every point in the gait cycle. Units are in degrees.
CAV _{Mean}	Mean CAV value over discrete intervals (Q, I, stance, swing) of stride. Each quintile contains a functional period of stride shown in parentheses.
Quintiles (Q)	Q1: -10 to 10% (heel-strike), Q2: 10-30% (mid-stance), Q3: 30 to 50% (toe-off), Q4: 50 to 70% (swing acceleration), Q5: 70 to 90% (swing deceleration)
Intervals (I)	I1: -10 to 0%, I2: 0 to 10%, I3: 10 to 30%, I4: 30 to 64%, I5: 64 to 90%
Stance	0 to 40%
Swing	40 to 100%
Stride	0 to 100%
CAV _{Max}	Maximum CAV value within a selected interval of stride.
CAV _{Max±2%}	Mean CAV for an interval ±2% of stride about a CAV _{Max}

Retro-reflective markers were attached to the subjects to model bilateral, hip, knee and ankle articulations. (Figure 5.1) The distal aspects of each thigh and shank were wrapped with elastic straps (ProWrap, Fabrifoam, Exton, PA) and rigid body clusters were then attached to the straps with hook and loop connectors and secured using additional elastic straps (MediPro, Fabrifoam, Exton, PA). Subjects wore standardized shoes (ZoomAir; Nike, Beaverton, OR) modified with windows cut out allowing adhesion of the markers directly to the skin by means of both adhesion spray and toupee tape.

Kinematic data was captured using a combination of 15 Eagle and Eagle4 cameras at 300 Hz (Motion Analysis Corporation, Santa Rosa CA). A dual belted treadmill instrumented with a force plate under each belt (TM-09-PBertec, Columbus, OH) was used to collect ground reaction force data at 1200 Hz. The treadmill belt speed was operated remotely by the investigators with a velocity resolution of 0.01 m/s with each belt being 48 cm wide and 164 cm long. A 15 point Rating of Perceived Exertion scale (RPE)(Borg 1982) was placed on a stand directly in front of the treadmill for subjects to reference to report level of perceived fatigue during the run. Perceived pain during the run was collected using a verbally administered numeric pain rating scale (NPRS) described to subjects as 0 being “no pain” and 10 considered “worst imaginable pain”. (Farrar, et al. 2001)

Treadmill Protocol

A one second standing static calibration file was captured while the subjects stood in the anatomical standing position. Subjects then walked on a single belt of the treadmill for 3 minutes at 1.3 m/s to acclimate themselves to the treadmill. Speed was then increased for 3 minutes to a warm-up pace (2.2-2.3 m/s) followed by 2 minutes at a standard pace of 3.3 m/s. Speed was then set at a self-selected pace where subjects felt they would not become severely fatigued over the course of the next 15 minutes with speed being adjusted upon request (2.2 to 3.3 m/s). To be included in the PFP group, subjects had to reach a minimum knee pain of 3 during the treadmill protocol. Kinematic and kinetic data

were acquired for the first 10 seconds of each minute interval. RPE and NPRS measures were recorded by investigators immediately following each 10 s data acquisition.

Data Processing

Kinematic markers were identified using Cortex 2.0 software (Motion Analysis Corporation, Santa Rosa CA). Three-dimensional marker coordinates and force plate data were exported to Matlab v2009a (Mathworks, Natick MA) for gait analysis. A fourth-order lowpass butterworth filter with a cutoff frequency of 8 Hz was applied to kinematic data. Force component data were filtered with a cutoff frequency of 30 Hz for the lateral forces and at 40 Hz for the vertical component. Cut-off frequencies were selected by investigators after visual inspection of a fast fourier transformation performed on the data. Joint coordinate systems were determined using the International Society of Biomechanics recommendations (Grood and Suntay 1983, Wu, et al. 2002). Segment orientations were determined using a singular value decomposition algorithm (Söderkvist and Wedin 1993) and joint angles using an Euler rotation sequence of long axis rotation-abduction-flexion for the knee and ankle.

Consistent gait points of heel-strike, mid-stance and toe-off were determined for each gait cycle for normalization. Heel-strike and toe-off were determined using the vertical component of the ground reaction force with a threshold of 50 N, mid-stance was the transition from braking to propulsion (0 N). (Cavanagh and LaFortune 1980) Both of the two periods of stance were time normalized to 50 points and swing phase to 150 points using a 5th order cubic spline function making a 250 point time normalized gait cycle(1 point=0.4%). Multiple normalization constraint points were chosen as it may reduce within and between subject variability of CAV measures. (Mullineaux, et al. 2006) The first and last gait cycle from each 10 s trial was discarded to reduce interpolation effects and the first 10 gait cycles were kept for analysis.



Figure 5.1 Markerset used during a static calibration. Only bilateral markers on the lateral aspects of the 5th metacarpal head, base, navicular and both the lateral and medial aspects of the calcaneus were used to model foot movement. Windows are cut out of the shoes allowing markers to be adhered directly to the foot. Rigid clusters were secured to the distal posterior-lateral aspects of each segment to model thigh and shank movement.

Data Reduction

One 10 s trial was chosen for analysis from the 15 minute period of self-selected running pace for each individual. For the PFP group, the trial with the highest pain value with a RPE value less than 14 was chosen. If there was more than one trial that qualified, the trial with the lowest RPE was chosen. If there was more than one trial with the same RPE and pain value, preference was given to the earlier time point in the run. For the healthy group, trials from the 11th minute of running at the self-selected pace with a RPE value of less than 14 were used. This corresponded to the average trial selected for the PFP group. Based on the above exclusion criteria, 19 PFP and 13 healthy participants qualified for analysis with 2 more healthy participants being excluded for missing markers on the foot.

CAV values were determined using a revised vector coding technique. (Heiderscheit 2000, Sparrow, et al. 1987) Five non-consecutive stride cycles from each 10 s trial were used for analysis. CAV values were derived for all knee and ankle coupling combinations (Table 5.1) at each point in the gait cycle. The injured limb was analyzed for the PFP group and a leg was chosen randomly for each of the healthy individuals to reduce systematic error. The normalized gait cycles were divided into quintiles each containing a functional period of stride (Heiderscheit, et al. 2002)(Table 2) with the border of each quintile overlapping. CAV_{Mean} values were calculated for quintiles (Q), stance, swing and the entirety of stride for each 10 s set, respectively.

Further observation of the CAV curves revealed there were several locations where CAV increases were not encompassed using the standard 20% quintiles. Therefore, five intervals (I) were created to capture consistent increases in the CAV values among all coupling relationships (Table 5.1). These intervals as well as the quintiles appeared too large to be sensitive enough to distinguish between distinct variability characteristics using traditional CAV_{Mean} values; therefore, the maximum CAV value (CAV_{Max}) and the mean CAV from an interval $\pm 2\%$ of stride about each CAV_{Max} were found within each interval ($CAV_{Max\pm 2\%}$) for each set of gait cycles. Steps involved in construction of the intervals for CAV_{Max} and $CAV_{Max\pm 2\%}$ locations are further detailed (Appendix E and F).

Statistical Analysis

Independent t-tests were performed to note any differences between population demographics (height, mass, age and average distance run per week). Additionally, pain, RPE and running speed were also compared. Independent t-tests were performed between populations for all CAV measures with significance set *a priori* ($\alpha \leq .05$) with no correction for multiple comparisons made. (Rothman 1990)

Results

19 PFP and 11 healthy subjects qualified for analysis. Population demographics are presented in Table 4.2 with only reported distance run found significantly different between populations ($p=.0008$). A wider range of speeds were observed for PFP (2.2-3.1 m/s) than healthy (2.6-3 m/s) with the mean speed for the healthy population being faster (2.89 m/s (.13), (mean (SD)) than the PFP population (2.54 m/s (.24)) ($p<.0002$). Pain values were 4.3 (1.3) for the PFP group. RPE levels for the healthy group (12.2 (.87)) and the PFP group (12.4 (.77)) were not significantly different ($p=.41$).

Table 5.2 Subject demographics for Healthy and Patellofemoral Pain (PFP) groups.

Population	Sample Size (n)	Height (m)	Mass (kg)	Age (yrs)	Distance (km/week)
Healthy	11	1.66 (0.09)	58.0 (5.33)	26.5 (3.6)	37.7 (13.4) *
PFP	19	1.63 (0.07)	57.1 (6.48)	25.8 (6.1)	21.2 (9.4)

Note. Means for each measure are displayed with the standard deviation in parentheses. Significant differences are denoted between populations at $p<.05$ ().*

The majority of the variability was found to be greater in the PFP group compared to the healthy group for the measures; CAV_{Mean} (Figure 5.2), CAV_{Max} (Figure 5.3) and $CAV_{Max\pm 2\%}$ (Figure 5.4). All significant differences for these measures demonstrated higher CAV in the PFP group compared to the Healthy group. Ensemble averages of the CAV over the entire stride for each population are shown in Figure 5.5 with quintiles and intervals highlighted.

CAV_{Mean} values were greater for PFP in KF-AF at Q1 $6.1^\circ(1.8)<7.9^\circ(2.0)$ ($p=.020$) (healthy mean(SD)<PFP mean(SD)). CAV_{Mean} values in Q2 were also larger in PFP than H for KR-AI $10.1^\circ(4.0)<16.0^\circ(8.9)$ ($p=.050$) and KR-AF $7.0^\circ(2.5)<10.3^\circ(4.6)$ ($p=.038$). Increases were also observed in Q4 for KV-AF $6.2^\circ(1.9)<10.6^\circ(5.0)$ ($p=.010$) and Q5 for KV-AI $14.6^\circ(5.0)<23.5^\circ(9.6)$ ($p=.008$). Larger values were also observed during the regions of stance for KV-AF $4.5^\circ(1.5)<6.9^\circ(2.4)$ ($p=.008$) and stride for KV-AI $11.6^\circ(2.2)<14.8^\circ(4.5)$ ($p=.031$).

The proposed more sensitive measure of CAV_{Max} showed several differences in population throughout the stride cycle. I2 showed greater values in

KV-AI $5.5^{\circ}(3.0) < 14.5^{\circ}(14.4)$ ($p=.050$), KV-AF $15.6^{\circ}(11.3) < 38.5^{\circ}(26.0)$ ($p=.010$) and KF-AF $3.1^{\circ}(1.3) < 5.4^{\circ}(3.6)$ ($p=.048$). I3 showed greater values only for KR-AI $28.2^{\circ}(15.6) < 44.8^{\circ}(18.4)$ ($p=.018$) and I4 only showed differences in KV-AI $25.5^{\circ}(10.2) < 41.5^{\circ}(20.8)$ ($p=.024$).

All $CAV_{Max\pm 2\%}$ measures that showed significant differences between population were in the same intervals as the CAV_{Max} measures with the exception of I1 KF-AF which had greater $CAV_{Max\pm 2\%}$ values in the PFP population $19.2^{\circ}(5.7) < 25.5^{\circ}(6.0)$ ($p=.009$). Similar to corresponding CAV_{Max} values, I2 showed greater values in KV-AI $6.5^{\circ}(3.5) < 11.3^{\circ}(6.2)$ ($p=.027$), KV-AF $9.8^{\circ}(3.7) < 20.5^{\circ}(11.8)$ ($p=.007$) and KF-AF $3.7^{\circ}(1.7) < 8.6^{\circ}(6.8)$ ($p=.027$). I3 showed greater values only for KR-AI $20.7^{\circ}(10.4) < 30.7^{\circ}(12.7)$ ($p=.034$) and I4 only showed differences in KV-AI $20.9^{\circ}(6.7) < 33.4^{\circ}(16.9)$ ($p=.027$).

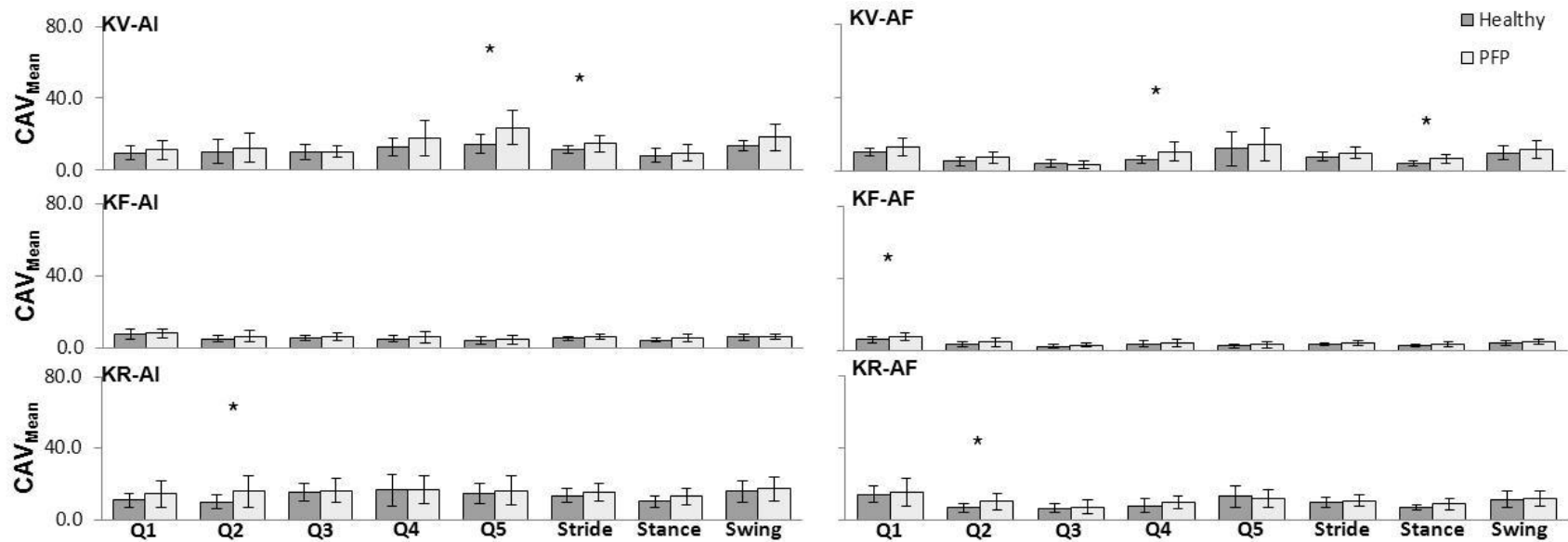


Figure 5.2 Mean Coupling Angle Variability ($CAV_{Mean} \pm SD^\circ$) values within each quintile (Q1-5, Table 5.1) of stride, the entirety of stride, stance and swing phase at a self-selected running pace for six Knee-Ankle joint coupling combinations for female runners with patellofemoral pain and healthy controls. Data are taken from 5 non-consecutive gait cycles from a 10 s running trial in a high pain condition for PFP and the 11th minute of a self-selected pace for healthy. Significant difference between populations denoted at $p < 0.05$ (*). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

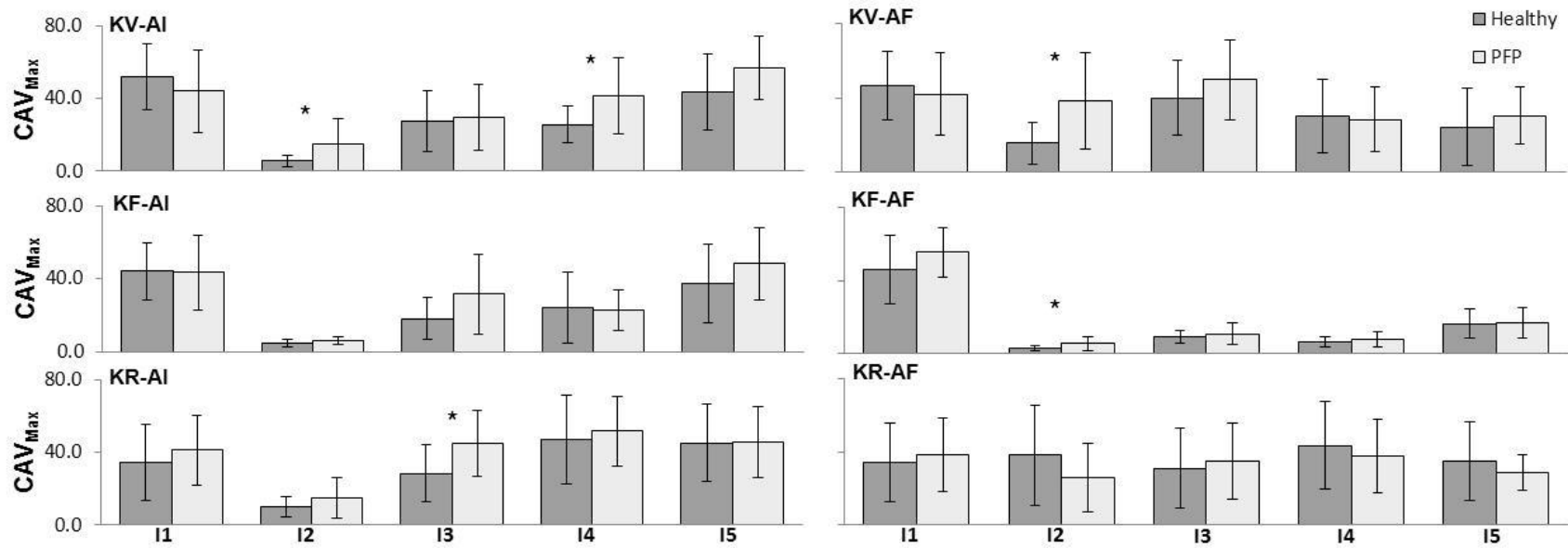


Figure 5.3 Maximum Coupling Angle Variability ($CAV_{Max} \pm SD^\circ$) values for five intervals (I1-5, Table 5.1) of stride at a self-selected running pace for six Knee-Ankle joint coupling combinations for female runners with patellofemoral pain and healthy controls. Data are taken from 5 non-consecutive gait cycles from a 10 s running trial in a high pain condition for PFP and the 11th minute of a self-selected pace for healthy. Significant difference between populations denoted at $p < 0.05$ (*). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

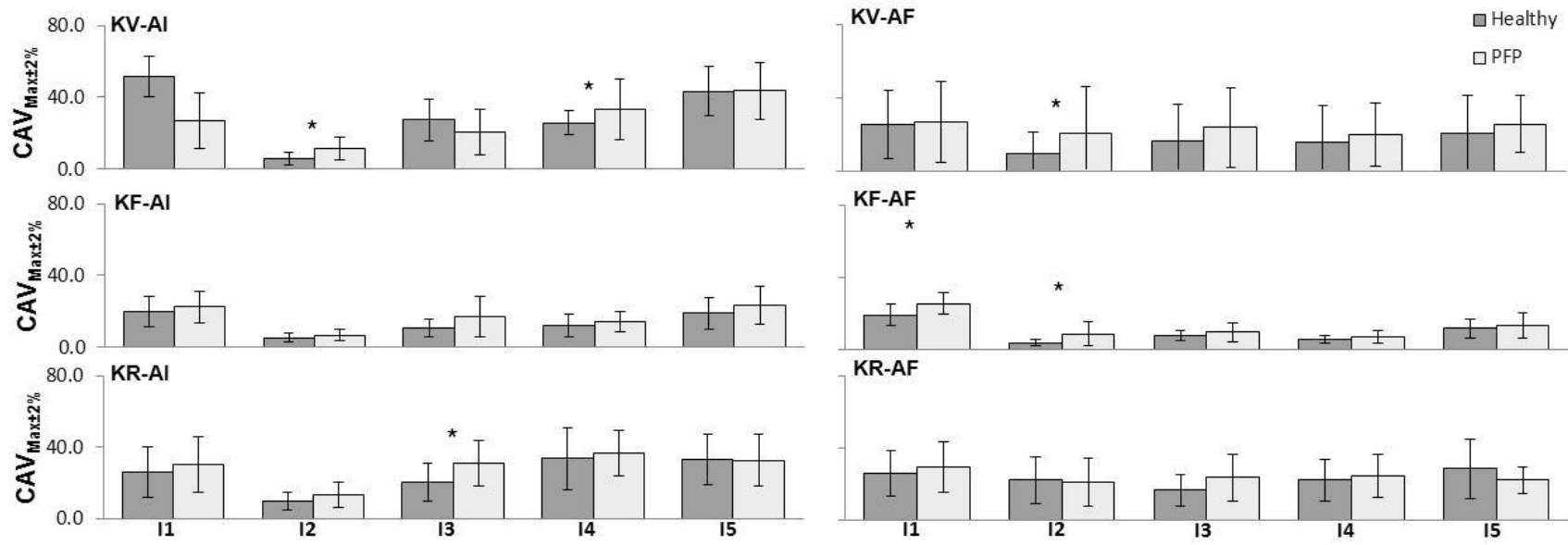


Figure 5.4 Mean Coupling Angle Variability over an interval of $\pm 2\%$ of stride about a CAV_{Max} stride location ($CAV_{Max\pm 2\% \pm SD^\circ}$) for five intervals (I1-5, Table 5.1) of stride at a self-selected running pace for six Knee-Ankle joint coupling combinations for females with patellofemoral pain and healthy controls. Data are taken from 5 non-consecutive gait cycles from a 10 s running trial in a high pain condition for PFP and the 11th minute for of the self-selected pace for healthy. Significant difference between populations denoted at $p < 0.05$ (*). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

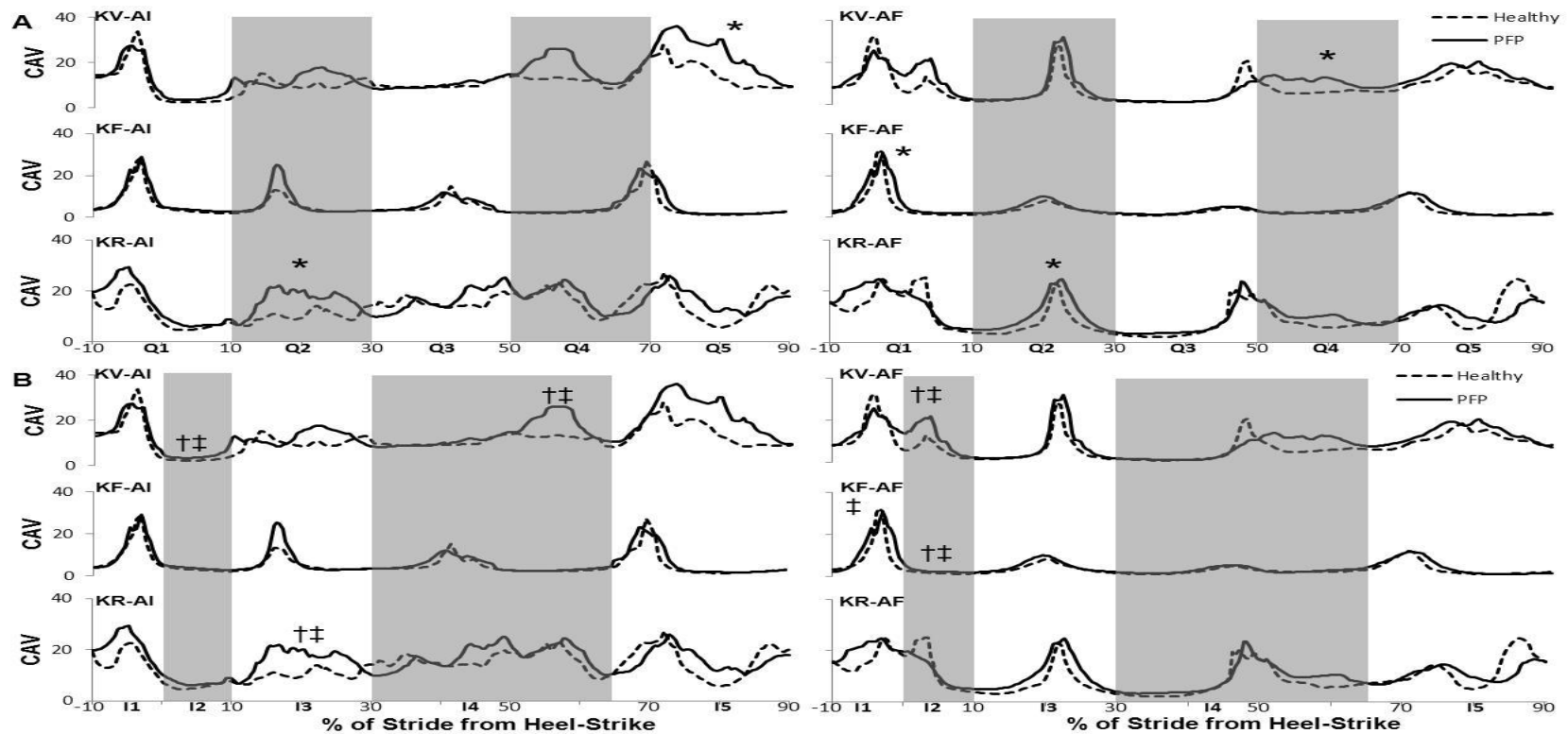


Figure 5.5 Ensemble averaged Coupling Angle Variability (CAV) curves for Healthy and Patellofemoral Pain (PFP) populations taken from 5 non-consecutive gait cycles for six Knee-Ankle coupling combinations. (A) highlights stride quintiles (Q1-5) labeled on the horizontal axis with Q2 and Q4 shaded in the plotting area. Q1 begins at -10% stride as measured from heel-strike (0%). (B) shows identical curves highlighting 5 custom intervals of stride (I1-5) with I2(0 to 10%) and I4(30 to 64%) being shaded. All vertical axis units are in degrees ($^{\circ}$). Significant differences between populations at $p < .05$ for quintiles: CAV_{Mean} (*); within intervals: CAV_{Max} (†), $CAV_{Max \pm 2\%}$ (‡). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

Discussion

The hypothesis that CAV values would be less in individuals in PFP was not supported. Surprisingly, the only differences that were observed showed greater CAV values in PFP than healthy individuals. These findings are contrary to the dynamical systems perspective to lower extremity overuse injuries taken by Hamill et al. which suggested lower CAV is indicative of a pathological coordinate state. (Hamill, et al. 1999, Heiderscheit, et al. 2002) Previous literature using identical analysis procedures for all CAV_{Mean} intervals in the KR-AI, KF-AI and KF-AF couplings showed no differences in any CAV_{Mean} values in a PFP population that had less pain. (Heiderscheit, et al. 2002) Increases in CAV values observed in the current study suggest that a PFP population that reports with a higher level of pain may exhibit a coordinative structure different than that observed previously. (Heiderscheit, et al. 2002) The increase in CAV observed after development of PFP may describe an adaptive coordinative structure that is compensating to a painful state to reduce stress among inflamed structures. Reduction of knee flexion has been observed in walking gait (Nadeau, et al. 1997, Powers, et al. 1999) and running gait (Dierks, et al. 2011) in PFP populations which may be a compensatory mechanism to reduce forces to the knee. (Dillon, et al. 1983) Similarly, increases in CAV involving knee flexion may help reduce loads to the knee.

The observed increases in variability may also have preceded the development of PFP. This would coincide with the perspective that there is an optimal amount of variability where extreme amounts, too much or too little, are detrimental to a biological system (Stergiou, et al. 2006) and lead to an overuse condition in the lower extremity. Dierks et. al. (Dierks, et al. 2011) theorized that increased variability in the lower extremity might be a result of decreased muscular control due to running in an exerted state coinciding with an observed increase in knee valgus. Increased femur internal rotation and adduction can effect peak knee valgus and internal rotation during running (Dierks, et al. 2011,

Noehren, et al. 2011, Powers 2003) Similarly, the couplings of KV-AF, KV-AI, KR-AF and KR-AI, each saw an increase in CAV during early stance but at a lower exertion state than observed by Dierks et al. (Dierks, et al. 2011) This suggests that increased variability resulting from femoral adduction and internal rotation may be a result of decreased muscular control inherent in a PFP population leading to a painful state.

Clinical interpretation of these results should be viewed with caution. The reliability of CAV measures in this specific context have been shown to be poor within a population (Chapter 3) and these results, like any biomechanical measure, should be viewed in the context of the precision limits of each CAV measure. None of the differences observed were outside the precision limits at a 95% confidence established for both populations. (Chapter 3) This indicates that these results are statistically significant but may not be clinically meaningful when distinguishing between populations. (Mullaney, et al. 2010) It is plausible that increases in CAV may also just be a result of mathematical artifact resulting from a clustering of data capture points in regions where little joint motion occurs such as heel-strike. (Heiderscheit, et al. 2002) Conflicting results of this study and limited findings of previous research brings into question the validity of CAV measures in distinguishing between inherent variability in joint coordination and measurement error when applied to lower extremity motion. Validity testing of CAV measures in human application has never been thoroughly investigated despite prevalently being applied and interpreted as such throughout the literature. (Heiderscheit, et al. 2002, Maulder 2011, Miller, et al. 2010, Wilson, et al. 2008) . This is similar to the promising initial use and eventual disregard of CRP measures which were deemed not appropriate for most lower extremity motions. (Hamill, et al. 1999, Peters, et al. 2003) CAV has shown promise in responding to the theoretical constructs of dynamical systems (Miller, et al. 2010) and may be a valid method in assessing joint coordination variability; however, its clinical usefulness is not yet understood.

This study addressed several limitations identified in the literature but also had several of its own. The nature of this investigation cannot determine if the increase in CAV is the result of pathology or precedes development limiting interpretation. (Bartlett, et al. 2007) Only knee-ankle couplings and a few discrete CAV measures were analyzed making comparison to previous literature difficult. The intervals chosen for the CAV_{Max} and $CAV_{Max \pm 2\%}$ were thought to encompass locations of stride that involve the reversal of joint movement. These regions are thought to be critically important in the study of movement variability (Clark and Phillips 1993) and accompanied with relative variability increases, (Sainburg, et al. 1995) particularly near heel-strike. (Heiderscheit, et al. 2002) These measures were able to more accurately describe the increases in variability prior to heel-strike in KF-AF (I1,I2) than CAV_{Mean} over the entire Q1 region which included the relatively less relevant and quiescent I2 region of the CAV curve. There are other possible measures that may serve as alternative measures than those presented here.

The proposed etiology that PFP symptoms are a manifest of less joint coordination variability and observable by CAV measures is not promising. Evidence statistically supporting the theory that joint variability is related to overuse injuries is scarce. (Ferber, et al. 2005, Hamill, et al. 1999, Heiderscheit, et al. 2002) Surprisingly, this study showed increases in CAV for several couplings at several different locations providing evidence that less joint variability is not indicative of pathology. The reliability of CAV seems to be extremely poor and unpredictable and it is unclear whether the sources of variation are physiological or methodological. Regardless, linear statistical comparison and discrete CAV measures used in previous literature and this study do not seem to allow meaningful clinical application. If further analysis is pursued, analysis of CAV measures may benefit from comparison methods that take into account the poor reliability of these discrete CAV measures, individual variability of subjects and comprehensive investigation of coupling patterns. It is

recommended that future studies using CAV do not focus on clinical applications that involve violent maneuvers such as running but rather focus on the validity of CAV during simpler motions that will reduce methodological error and clinical conjecture.

Chapter 6: Summary, Conclusion and Recommendations

Summary

One perspective on musculoskeletal injuries of the lower extremity from a dynamical systems perspective has been suggested by Hamill et al. (Hamill, et al. 1999) They suggest that a person's inability to exhibit variations in their joint coordination patterns can increase the frequency of loading of soft tissue and eventually lead to an overuse condition and pathological state. (Hamill, et al. 1999) There is little evidence to support this theory that lower joint coordination variability is indicative of overuse injury and pathological coordinate state.

Coupling angle variability (CAV), derived from a vector coding technique, (Sparrow, et al. 1987) has been proposed as a suitable measure to quantify lower extremity joint coordination variability to test this theory. (Heiderscheit, et al. 2002) The clinical usefulness of CAV measures is not yet understood. For a CAV measure to be clinically useful from a dynamical systems perspective it must be reliable and predictably respond to a change in an identified control parameter under a valid construct.

The goals of this study were to identify a valid construct to test the clinical usefulness of CAV measures from a dynamical systems perspective by altering a single control parameter and observe a predictable response. Patellofemoral Pain (PFP) in female runners has been suggested as an overuse injury that may result from less variability in lower extremity joint coordination patterns. (Hamill, et al. 1999) CAV measures used previously in the literature (CAV_{Mean}) have been insensitive to many changes in control parameters in most previous lower-extremity gait analyses. Therefore, two more CAV measures (CAV_{Max} and $CAV_{Max\pm 2}$) thought to be more sensitive to change were investigated to establish their clinical utility when interpreted from a dynamical systems perspective. Runners with PFP and healthy controls ran on a treadmill at a preferred speed. Knee pain increased a clinically meaningful amount during the run for the PFP

population. CAV were calculated at a low pain level and the highest pain level in the PFP population and at a similar portion of the run as the high pain condition for the healthy population. CAV for all six joint couplings of the Knee-Ankle were investigated allowing comprehensive assessment of the clinical usefulness of several CAV measures for joint couplings thought to effect the development of knee pain. (Tiberio 1987)

The intra-subject reliability of CAV measures had not been established for healthy runners and runners with PFP. CAV were calculated for two sets of 5 non-consecutive stride cycles from the same 10 second capture period. A level of agreement analysis was performed for each population between the two sets of data. Data reported established the precision limits for all CAV measures analyzed for each population. A change in CAV larger than the established limits would indicate that the change was beyond measurement error and indicates a true sensorimotor change to the system. Changes larger than the precision limits would be considered clinically meaningful. (Mullaney, et al. 2010)

It was unknown if CAV measures can delineate a possible change in coordinate state when a clinically significant increase in knee pain occurs in a PFP population. Paired t-tests were performed between CAV measures calculated from 5 stride cycles in a low state of pain and 5 stride cycles from a high state of pain. Only 3 of 108 CAV measures investigated were found to have a significant change. These three CAV values demonstrated significant increase but did not exceed the precision limits of the CAV measures. Further, these increased variability disagree with previous theoretical concepts put forward that lower variability would indicate a pathology.

It was also unknown if CAV measures can detect a different coordinate state between a PFP population in a painful state from healthy controls. Independent t-tests were performed between CAV measures calculated from 5 stride cycles in a high state of pain and a similar portion of the run for healthy controls. Several CAV measures were shown to be significantly larger in PFP

while none were shown to be significantly less. None of the CAV measures were different by a clinically meaningful amount.

Conclusions

The purpose of the first analysis was to determine the intra-subject reliability of CAV measures when control parameters remain constant for a PFP and healthy population of runners. It was hypothesized that changes to CAV measures in both healthy and PFP populations with physiological variables held constant will be less than 10% of the 81° full scale range capable of CAV measures (8.1°). This hypothesis was supported for 70% of CAV_{Mean} measures and only 10% of CAV_{Max} and 8% of CAV_{Max±2%} measures.

The purpose of the second analysis was to determine the clinical validity of CAV measures when a physiological state control parameter of knee pain was increased for runners with PFP. It was hypothesized that there would be a statistically significant decrease and clinically meaningful decrease in CAV. These hypotheses were not supported for any measure.

The purpose of the third analysis was to determine the clinical validity of CAV measures to distinguish between runners with PFP and runners who were healthy. It was hypothesized that CAV values would be significantly less for runners with PFP and magnitude differences would be clinically meaningful. These hypotheses were not supported for any measure.

Recommendations for Future Research

Discrete measures of CAV did not support the theory that less coordination variability was indicative of overuse injury. Future analysis of CAV measures may benefit from statistical methods that take into account the poor reliability and the volatile nature of CAV measures if to be clinical useful. CAV

was shown to increase for several measures when hypothesized to decrease. This should be further investigated and results may provide important insight regarding the debate between optimal amounts of variability as being indicative of a healthy coordinate state with extremes, regardless of direction, being indicative of pathology. Most increases in CAV measures were in knee couplings of the transverse and frontal plane which may be due to less control of hip musculature. Future analysis investigating hip-knee couplings might yield more promising clinical applications of CAV measures. CAV measures have only been used during violent lower extremity motions which can increase measurement error and further inhibit the ability to distinguish sources of variability. Future research may benefit from similar research designs to those used by Kelso to first determine the clinical validity of CAV measures from a dynamical systems perspective prior to application. A study is still needed that can effectively identify and manipulate a control parameter and observe a CAV response consistent with dynamical systems theory of motor control. This must be accomplished before CAV can be considered a valid representation of a coordinate state from a dynamical systems perspective.

Appendices

Appendix A: Calculation of Coupling Angle and Coupling Angle Variability measures

Detailed in this appendix are VC CA, CAV and rv calculations described in Chapter 2. Knee flexion and ankle dorsiflexion data for five gait cycles during treadmill running are used at each step for demonstration (Figure A.1).

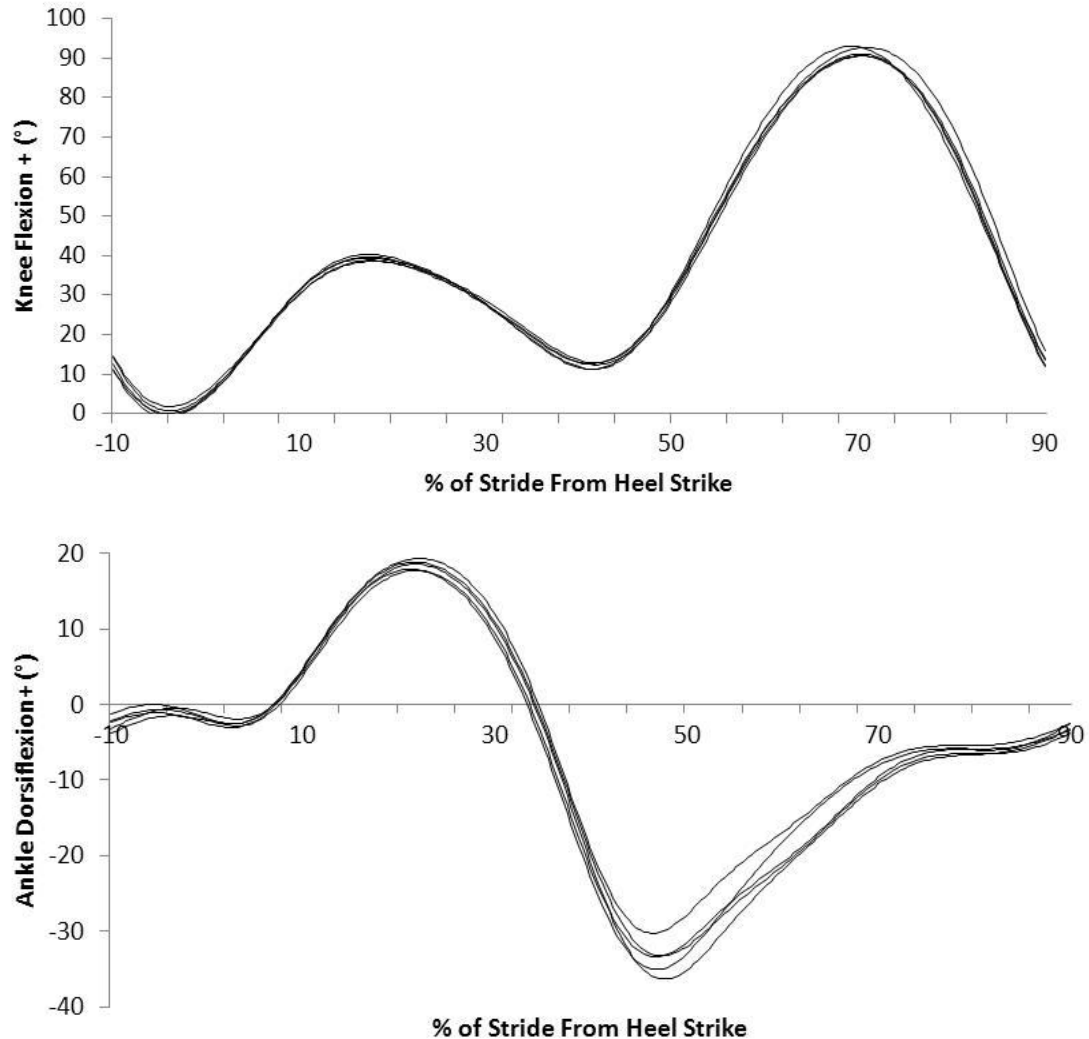


Figure A.1 Knee flexion (top) and Ankle dorsiflexion (bottom) during stride. Data shown are from five treadmill running gait cycles at a self-selected pace.

Step 1: Create angle-angle diagram with the proximal oscillator on the horizontal axis and the distal oscillator on the vertical axis.

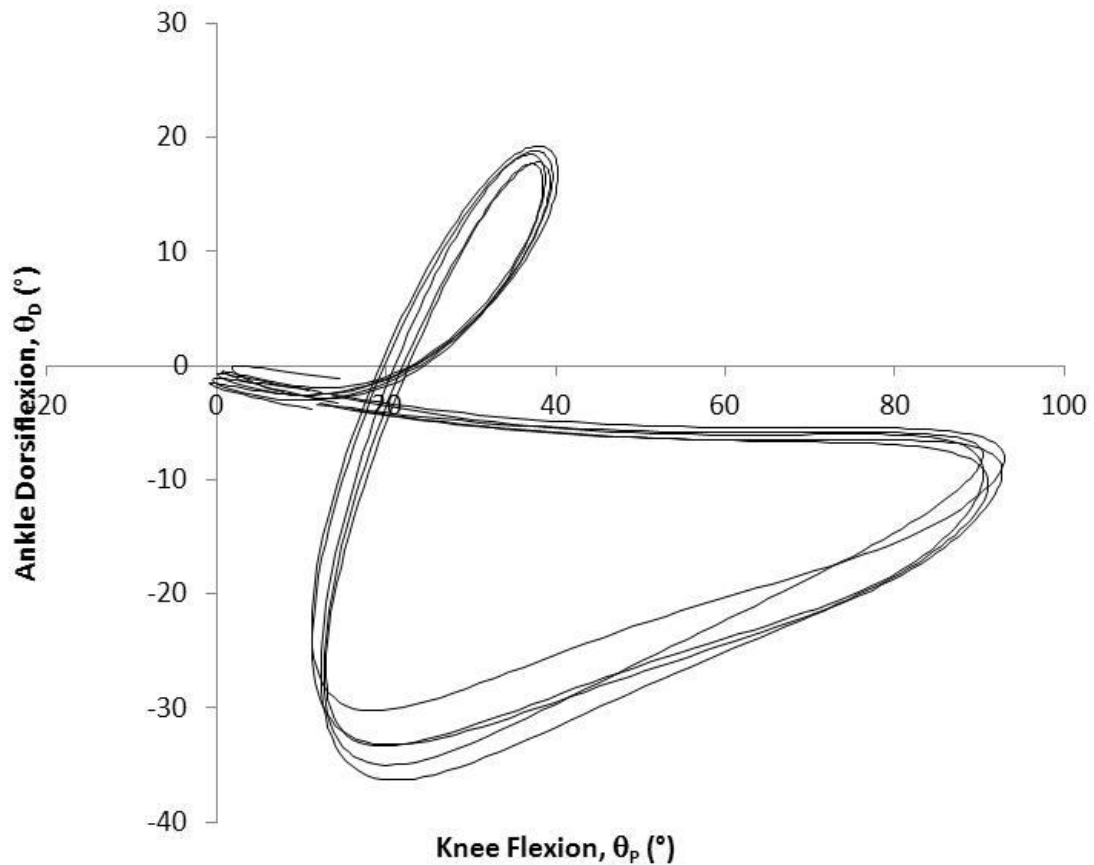


Figure A.2 Angle-angle diagram for five stride cycles during running gait. The distal joint is plotted on the vertical axis (Ankle Dorsiflexion, θ_D) and the proximal joint is plotted on the horizontal axis (Ankle Dorsiflexion, θ_P). These results are consistent with the example reported by Hamill 2000 (Figure 2.4).

Step 2: Calculate CA for each point in the normalized gait cycle (γ_i). This can be done using equation A.1 followed by A.2 or using a modulus 2π atan2 function in MATLAB (2009a, The Mathworks, Natick, MA) shown in A.3. These equations output a coupling angle between 0° and 360° as intended by Sparrow. (Sparrow, et al. 1987)

$$\begin{aligned}
& \tan^{-1} \frac{\theta_{Di+1} - \theta_{Di}}{\theta_{Pi+1} - \theta_{Pi}} \frac{180}{\pi}, \theta_{Pi+1} - \theta_{Pi} > 0 \\
& \tan^{-1} \frac{\theta_{Di+1} - \theta_{Di}}{\theta_{Pi+1} - \theta_{Pi}} \frac{180}{\pi} + 180, \theta_{Pi+1} - \theta_{Pi} < 0, \theta_{Di+1} - \theta_{Di} > 0 \\
\gamma_i = & \tan^{-1} \frac{\theta_{Di+1} - \theta_{Di}}{\theta_{Pi+1} - \theta_{Pi}} \frac{180}{\pi} - 180, \theta_{Pi+1} - \theta_{Pi} = 0, \theta_{Di+1} - \theta_{Di} > 0 \\
& 90, \theta_{Pi+1} - \theta_{Pi} > 0, \theta_{Di+1} - \theta_{Di} > 0 \\
& -90, \theta_{Pi+1} - \theta_{Pi} = 0, \theta_{Di+1} - \theta_{Di} < 0 \\
& \text{undefined}, \theta_{Pi+1} - \theta_{Pi} = 0, \theta_{Di+1} - \theta_{Di} = 0
\end{aligned} \tag{A.1}$$

$$\gamma_i = \begin{cases} \gamma_i + 360, & \gamma_i < 0 \\ \gamma_i, & \gamma_i \geq 0 \end{cases} \tag{A.2}$$

$$\gamma_i = \text{mod } \text{atan2 } \theta_{Di+1} - \theta_{Di}, \theta_{Pi+1} - \theta_{Pi}, 2\pi * \frac{180}{\pi} \tag{A.3}$$

In two studies, Wilson et al. (Wilson, et al. 2009, Wilson, et al. 2008) states that coupling angles obtained ranged from 0° to 180°. A.4 demonstrates an equation that would provide that result. This equation will yield a result with a discontinuity at 180°.

$$\gamma_i = \text{mod } \text{atan2 } y_i, x_i, \pi * \frac{180}{\pi} \tag{A.4}$$

A.5 is another example that will yield results between 0° and 180° that will not have a discontinuity at 180°.

$$\gamma_i = \text{atan2 } y_i, x_i * \frac{180}{\pi} \tag{A.5}$$

Step 2: Resulting coupling angles from step 1 can then be graphed (Figure A.3).

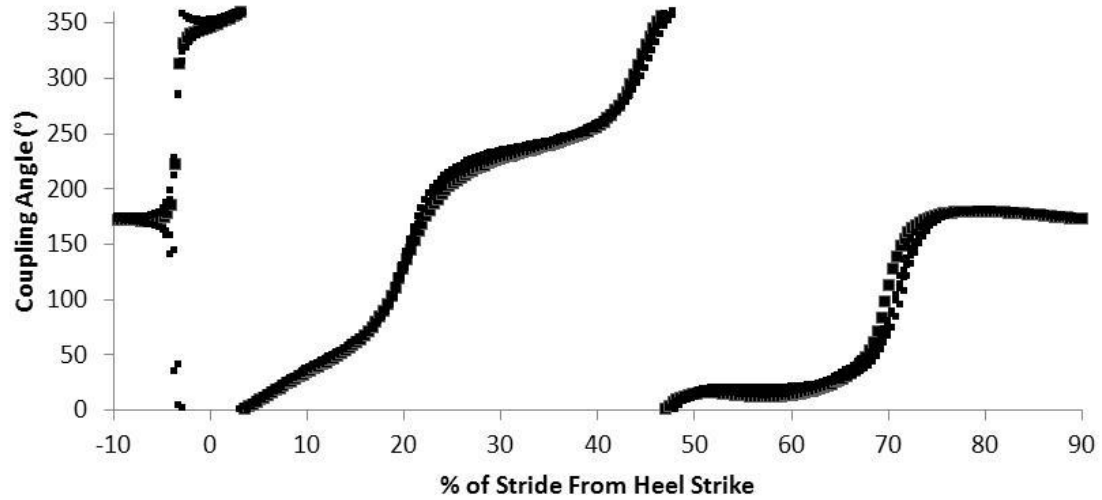


Figure A.3 Coupling angles for five stride cycles during running gait for a Knee-Ankle flexion coupling. Coupling angles were consistent other than two trials prior to heel strike (0%) which trended towards 0° indicating a different coupling pattern for that portion of gait for those two cycles. Equation A.3 was used in this calculation.

Step 3: Calculate the mean coupling angle using circular statistics for the five trials (n=5) shown in equations A.5 to A.7.

$$x_i = \frac{1}{n} \sum_{i=1}^n \cos \gamma_i, \quad \text{A. 5}$$

$$y_i = \frac{1}{n} \sum_{i=1}^n \sin \gamma_i \quad \text{A. 6}$$

The mean coupling angle (γ_i) is then described using equation A.7.

$$\gamma_i = \begin{cases} \tan^{-1} \frac{y_i}{x_i} \frac{180}{\pi}, & x_i > 0, y_i > 0 \\ \tan^{-1} \frac{y_i}{x_i} \frac{180}{\pi} + 180, & x_i < 0 \\ \tan^{-1} \frac{y_i}{x_i} \frac{180}{\pi} + 360, & x_i > 0, y_i < 0 \\ 90, & x_i = 0, y_i > 0 \\ 270, & x_i = 0, y_i < 0 \\ \text{undefined}, & x_i = 0, y_i = 0 \end{cases} \quad \text{A.7}$$

Or once again using an atan2 modulus 2π in MATLAB shown in A.8,

$$\gamma_i = \text{mod} \left(\text{atan2} \left(y_i, x_i \right), 2\pi \right) * \frac{180}{\pi} \quad \text{A.8}$$

Step 4: Graph the mean coupling angle (Figure A.4).

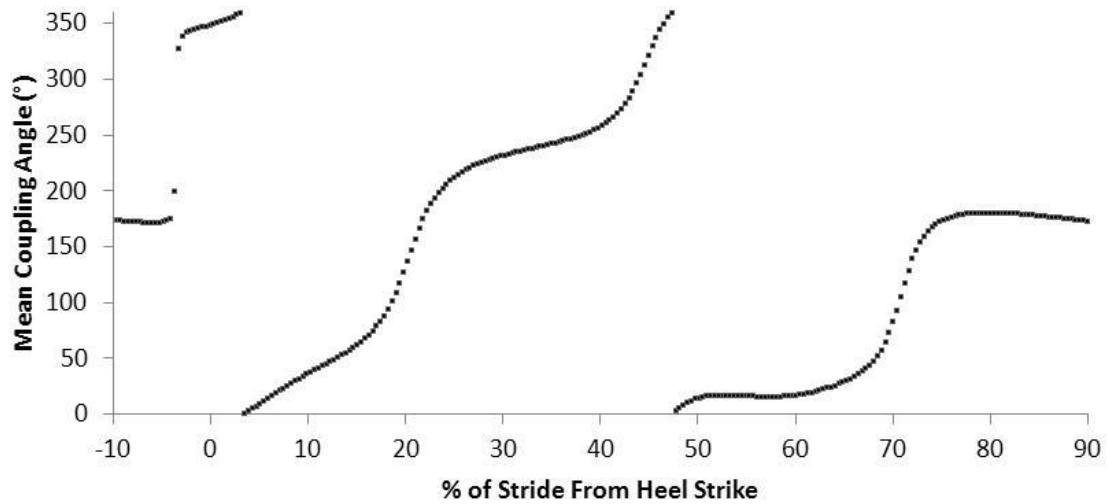


Figure A.4 Mean coupling angle over five stride cycles during running gait for a Knee-Ankle flexion coupling for one subject. Equation A.8 was used. Notice the discontinuities not depicted in Figure 2.4.

Step 5: Calculate the length of CA (a_i) using circular statistics.

$$a_i = \sqrt{x_i^2 + y_i^2} \quad \text{A. 9}$$

Step 6: Calculate the standard deviation (s) of CA from the five trials using circular statistics. This is termed Coupling Angle Variability (CAV).

$$CAV_i = s_i = \sqrt{\frac{1}{2} (1 - r_i)} * \frac{180}{\pi} \quad \text{A. 10}$$

Step 7: Graph the CAV as shown in Figure A.5

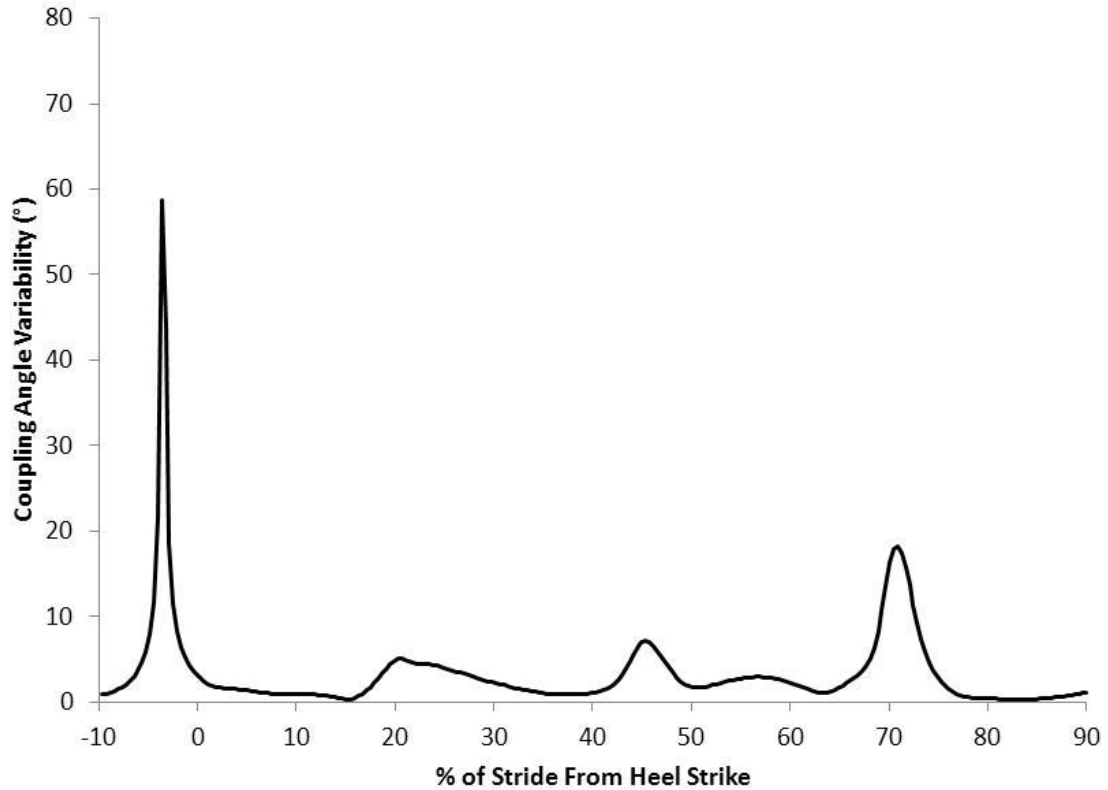


Figure A.5 Coupling angle variability for five stride cycles during running gait for a Knee-Ankle flexion coupling for one subject. Equation A.10 was used.

Step 8: Traditional linear discrete dependent measures can then be created from the continuous CAV_i curve shown in Figure A.5. The mean CAV over a desired interval of the stride cycle from a first location of the stride cycle ($i_1\%$) to a second location in the stride cycle ($i_2\%$) can be calculated using equation A.11 where (n) is the amount of points being averaged over. This gives a measure of within-subject variability over the selected portion of the stride cycle.

$$CAV_{i_1\%-i_2\%} = \frac{1}{n} \sum_{i_1}^{i_2} CAV_i \quad A.11$$

Values for some common dependent measures from the sample data are shown below:

CAV_{Mean} values:

Stride from heel-strike to heel-strike (0% to 100%): Stride $CAV_{Mean}=3.36^\circ$

Stance phase (0% to 40%): Stance $CAV_{Mean}=1.97^{\circ}$

Swing phase (40% to 100%): Swing $CAV_{Mean}=4.26^{\circ}$

Prior to heel-strike (Interval -10% to 0%): $CAV_{Mean}=9.10^{\circ}$

CAV_{Max} values:

Maximum during stride: Stride $CAV_{Max}=58.69^{\circ}$

Local maximum during stance: Stance $CAV_{Max}=5.07^{\circ}$

Locations of common events such as the maximum, minimum, or local maximum or local minimum within a certain period can also be used as a CAV measure.

These dependent measures can be interpreted as a location of events within the CAV curve.

CAV_{MaxLoc} values:

Location of maximum during stride: Stride $CAV_{Max} = -3.6\%$

Location of local maximum during stance: Stance $CAV_{Max}=18.96\%$

Step 9: Traditional dispersion statistics such as mean and standard deviation can then be calculated for the CAV measures comparing population means giving a measure of between subject variability for a sample population.

Step 10: An alternative approach can be taken to quantify variability using Tepavac and Field-Fote's methods continuing from equation A.9 as shown in Chapter 2 Eqns. 2.13-2.21. The coefficient of correspondence subtracted from 1 (rv) is shown in Figure A.6.

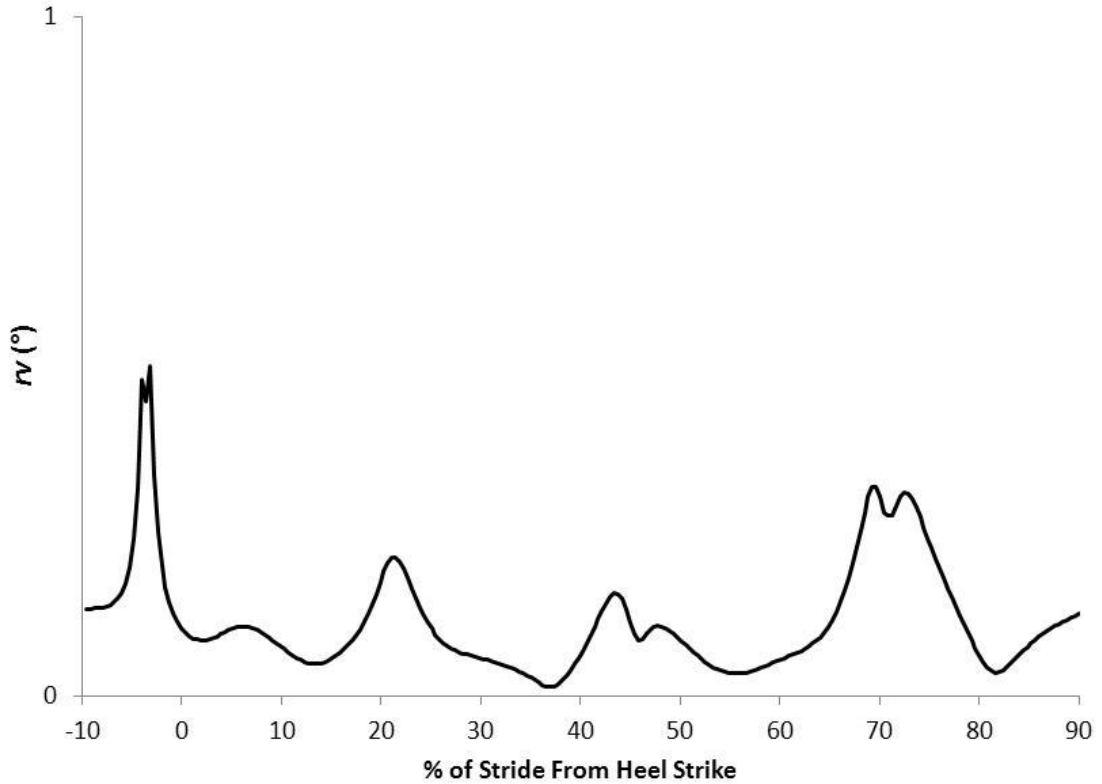


Figure A.6 Variability as described by Tepavac and Field-Fote (rv) for five stride cycles during running gait for a Knee-Ankle flexion coupling for one subject. rv is the coefficient of correspondence subtracted from 1. Equation 2.21 was used.

Step 11: Similar to A.11, average values of variability can be calculated over periods of the gait cycle using rv . These are the equivalent to r (Eq. 2.20) for selected intervals.

$$rv_{i_1\%-i_2\%} = \frac{1}{n} \sum_{i_1}^{i_2} rv_i \quad \text{A.12}$$

Calculations in step 8 but using equation A.12 are shown below:

Stride from heel-strike to heel-strike: Stride $rv_{Mean}=0.1095^\circ$

Stance phase: Stance $rv_{Mean}=0.0793^\circ$

Swing phase: Swing $rv_{Mean}=0.1291^\circ$

Prior to heel-strike: I1 $rv_{Mean}=0.1963^\circ$

Maximum: Stride $rv_{Max}=0.4849^\circ$

Local maximum during stance: Stance $rv_{Max}=0.1830^\circ$

Appendix B: Coupling Angle and Coupling Angle Variability inconsistencies

Inconsistencies in methods and citations of methods in calculating CA and CAV are possible sources of unwarranted error within and between studies. Identified inconsistencies in the literature as described in Chapter 2 are presented using CA and CAV curves calculated with two sample sets of data to identify differences in results. The first set of data are hypothetical and present CA of an angle-angle diagram with relative movement changes of 5° increments ranging from 0° to 360° , a CA input range intended by Sparrow (Sparrow, et al. 1987). The second set of data is from empirical data collected over 5 strides of running gait.

Table B.1 details the mathematical methods used to calculate these values. Scenario 1 is the only method that is mathematically valid for both CA and CAV measurements. Scenarios 1 & 2 are valid for calculation of CAV measures as CAV is shown to only be affected by the input range calculations used.

Table B.1 Ten scenarios presented in the literature that can affect coupling angle and coupling angle variability values depending on interpretation of the methods cited. Equations used for each scenario and studies that have used these are referenced and described in the text.

Scenario	Reference	Input CA Range	Mean CA Math
1		0° to 360° (A.1, A.2)	Corrected Math (A.7)
2		0° to 360° (A.1, A.2)	Cited Math (2.8)
3		-90° to 90° (2.4)	Corrected Math (A.7)
4		-90° to 90° (2.4)	Cited Math (2.8)
5		0° to 90° (2.5)	Corrected Math (A.7)
6		0° to 90° (2.5)	Cited Math (2.8)
7		0° to 180° (A.5)	Corrected Math (A.7)
8		0° to 180° (A.5)	Cited Math (2.8)
9		0° to 180° (A.4)	Corrected Math (A.7)
10		0° to 180° (A.4)	Cited Math (2.8)

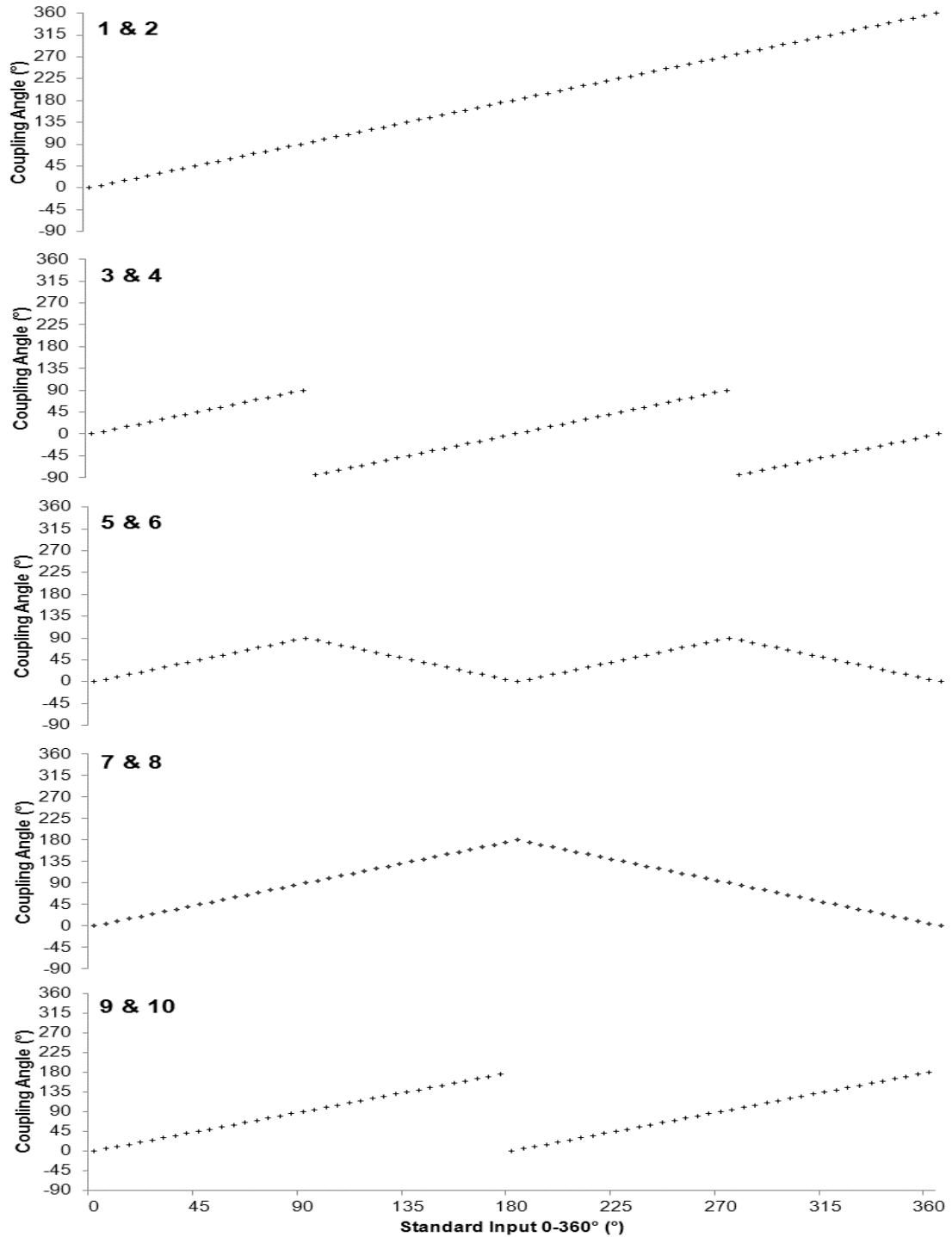


Figure B.1 Coupling angles are shown for possible outcomes of theoretical angle-angle diagrams at 5° intervals ranging from 0° to 360° for 10 scenarios labeled in bold (Table B.1). Coupling angles should equal the standard input for the entire range if valid.

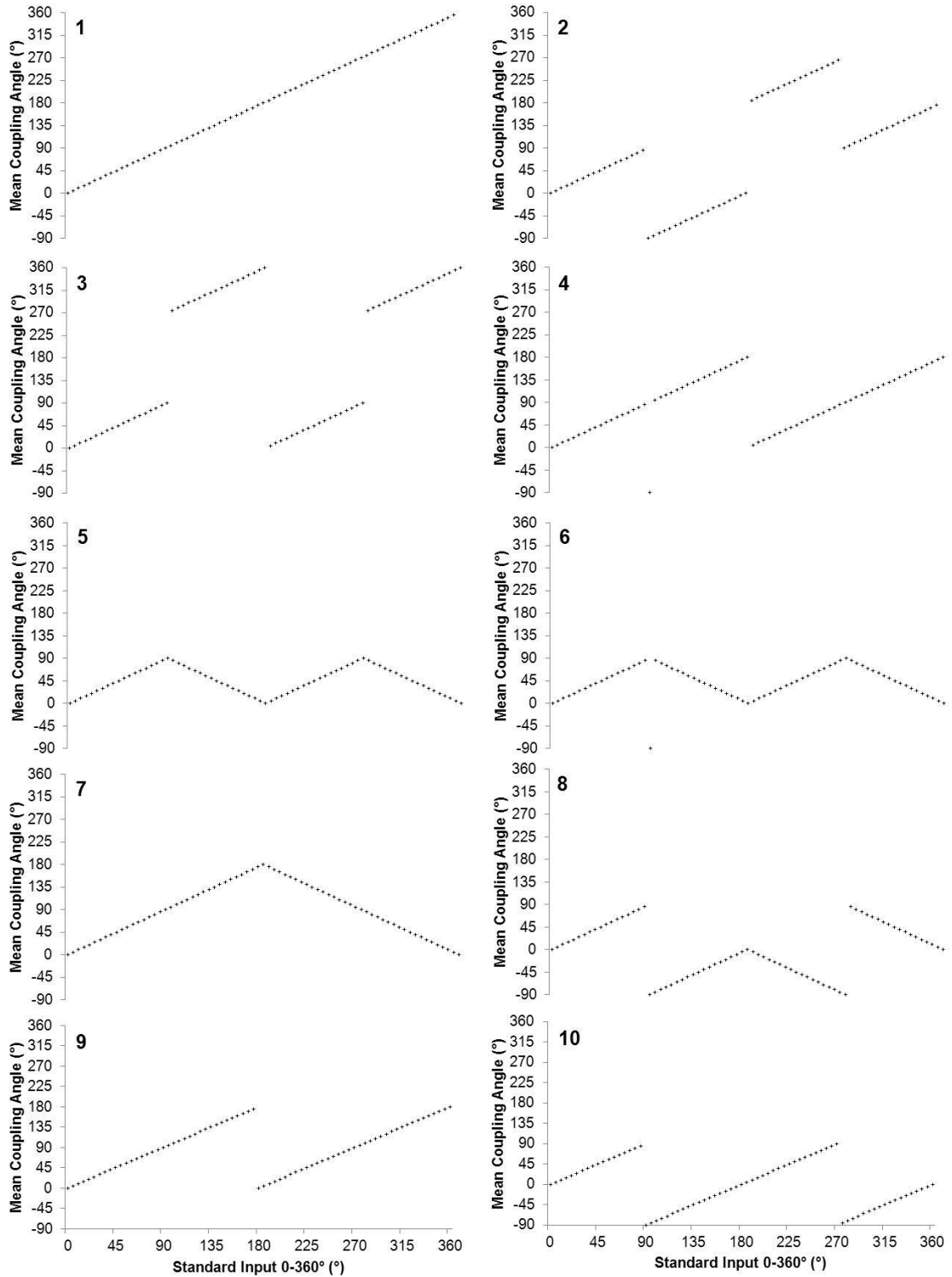


Figure B.2 Mean coupling angles are shown for all possible outcomes of a theoretical angle-angle diagram composed of 5° ratio intervals ranging from 0° to 360° for 10 scenarios labeled in bold (Table B.1). Mean coupling angles should equal the standard input if valid.

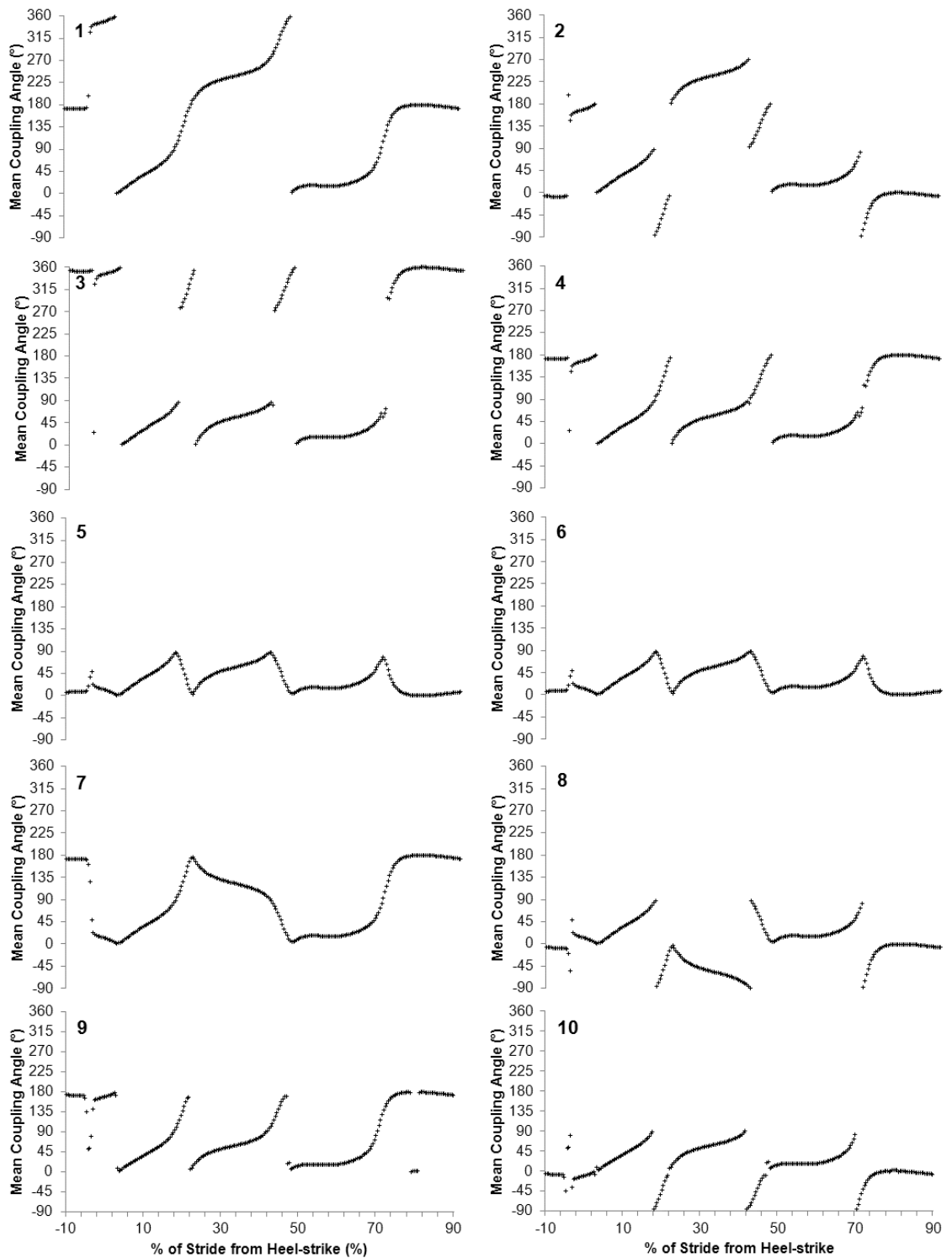


Figure B.3 Mean coupling angles are shown for possible outcomes using Knee and Ankle flexion angles for five gait cycles for 10 methodical scenarios labeled in bold (Table B.1). All graphs on the left use corrected coupling angle equations while the right uses equations cited in the literature.

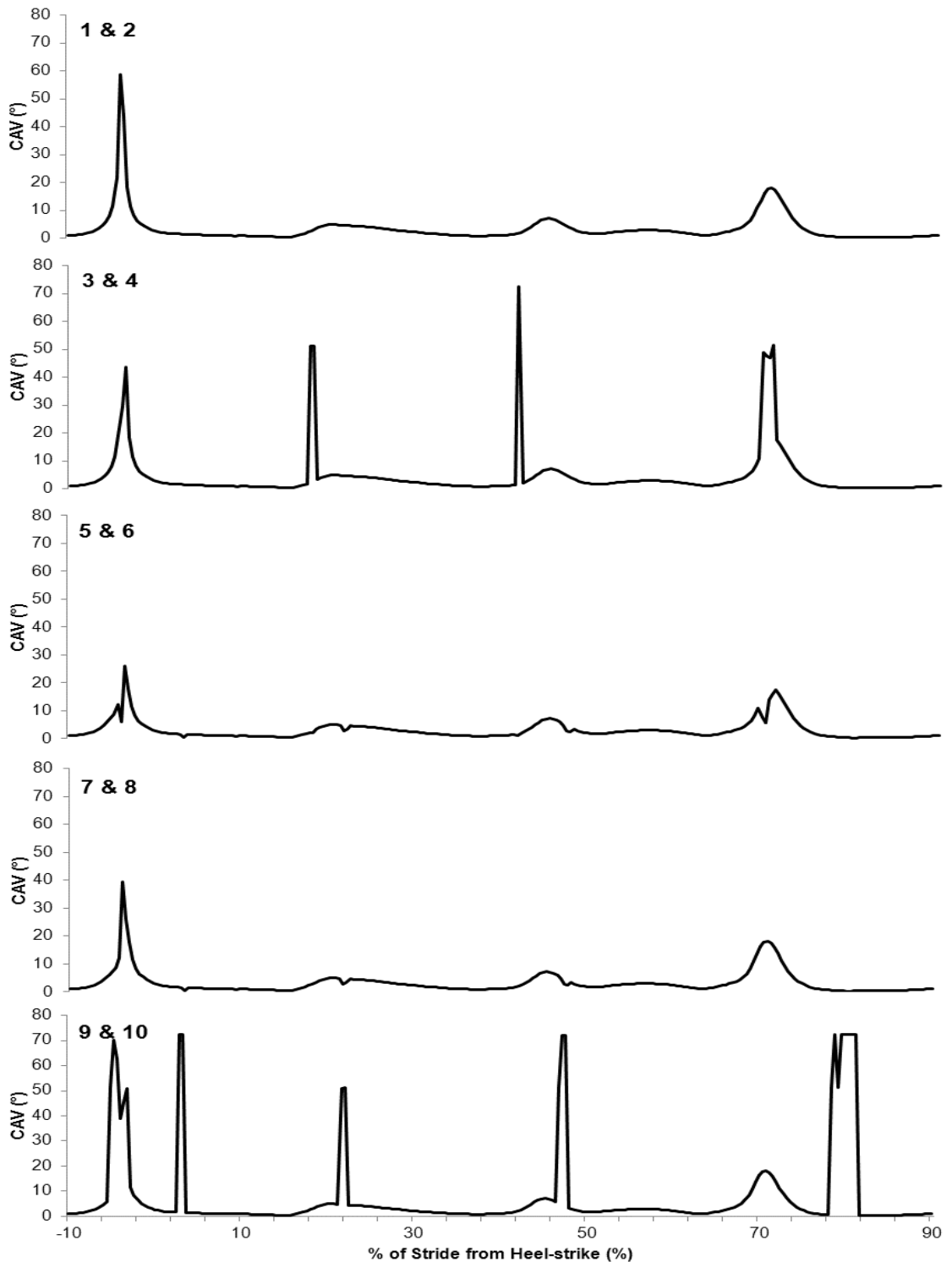


Figure B.4 Coupling angle variability (CAV) curves are shown for possible outcomes using Knee and Ankle flexion angles from five gait cycles for 10 calculation scenarios labeled in bold (Table B.1). CAV is affected by input range of CA values and not mean coupling angle. Only scenarios 1 & 2 produce valid results through the entirety of the stride cycle for this coupling.

Appendix C: Data Collection Sheet

Subject #																									
Condition	Healthy (H)	Injured (I)		Knee	Left	Right	Other Information																		
Height (in)				Mileage (miles/week)																					
Weight (lbs)				Resting HR (bpm)																					
Age (yrs)																									
Category	Trial#	Time (min)	Speed (m/s)	Pain							Fatigue							HR							
W (walking)	1	1	1.3	0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	2	2		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	3	3		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
RW (running warmup)	1	4	2 to 2.6	0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	2	5		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	3	6		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
RS (running)	1	7	3.3	0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	2	8		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
RO (Running Other)	1	9	2.5 to 3.3	0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	2	10	2.2 to 3.5	0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	3	11		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	4	12		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	5	13		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	6	14		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	7	15		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	8	16		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	9	17		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	10	18		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	11	19		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	12	20		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	13	21		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	14	22		0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
RO (Running Other)	15		2.2 to 3.8	0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	16			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	17			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	18			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	19			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	20			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	21			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	22			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	23			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	24			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
WP (Walking Post)	1		1.3	0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	2			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	3			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	4			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	5			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	6			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	7			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	8			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	
	9			0	1	2	3	4	5	6	7	5	6	7	8	9	10	11	12	13	14	15	16	17	

Figure C.1 Data collection sheet. Protocol is in chronological order from top to bottom. Trials were collected at one minute intervals. Speeds listed are the ranges observed during the study.

Appendix D: Laboratory Layout

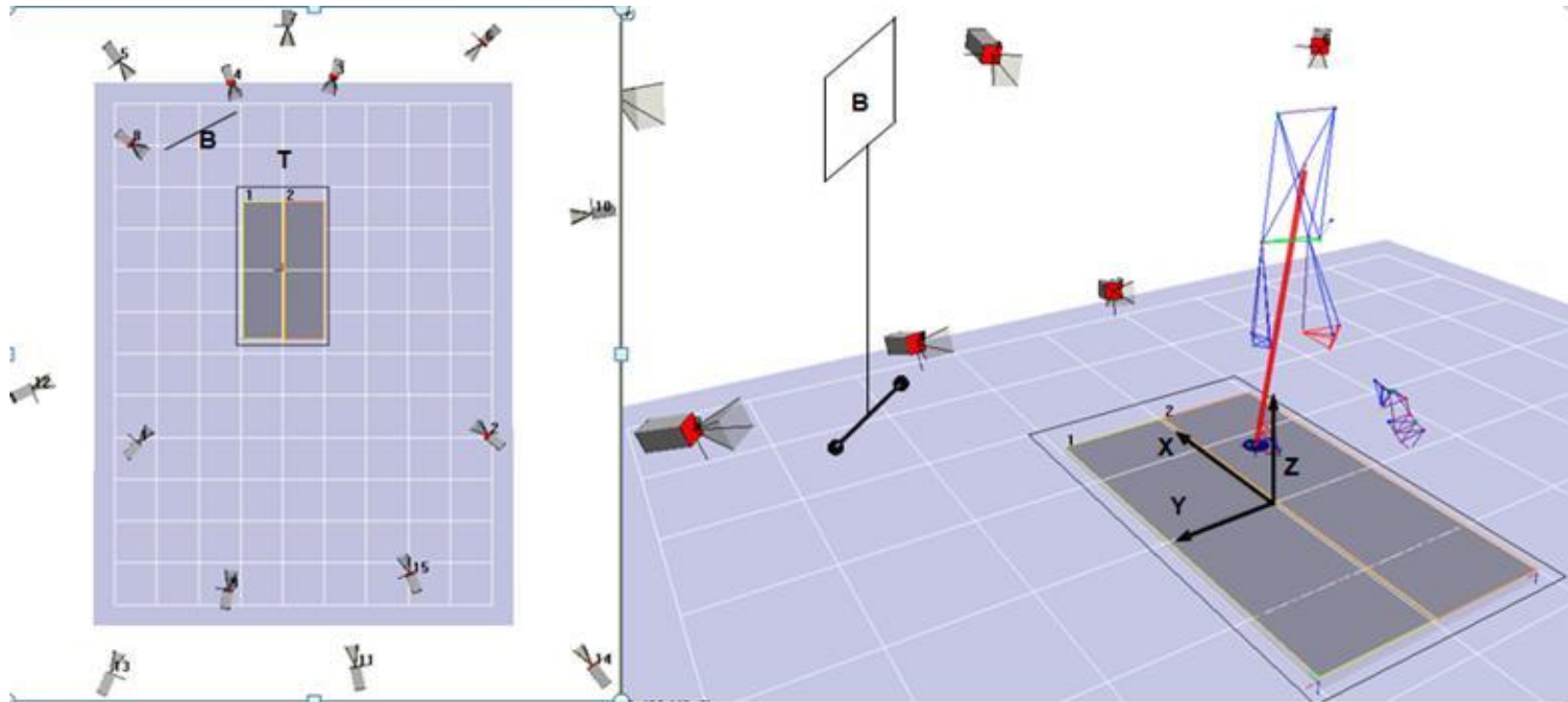


Figure D.1 Aerial view of the laboratory shown on the left with the dual belted treadmill (T) and the corresponding independent force plates (1,2) with the Borg RPE scale (B) positioned in front and to the left of the treadmill. On the right is a perspective view of the laboratory setup showing the treadmill and the corresponding laboratory coordinate system and RPE scale. Also shown is a digitized subject running on force plate 2 and the resultant ground reaction force measured by the treadmill.

6	
7	Very, Very Light
8	
9	Very Light
10	
11	Fairly Light
12	
13	Somewhat Hard
14	
15	Hard
16	
17	Very Hard
18	
19	Very, Very Hard
20	Maximal Exertion

Figure D.2 Scale of rated perceived exertion displayed in the laboratory. For this study, 14 was considered fatigued and data at this level or beyond were not considered for analysis.



Figure D.3 Layout of the laboratory during a participant's data collection session.

Appendix E: Selection of intervals for calculation of Max Coupling Angle Variability measures

Previous literature has calculated CAV_{Mean} values over several intervals of stride. CAV_{Mean} over the entire stride cycle can be considered the least sensitive measure to spontaneous change in CAV values. If sensitivity is considered to have an indirect relationship with CAV_{Mean} interval size, a single point measurement would then be the most sensitive to change. The progression from less to more sensitivity to CAV change in a stride cycle would progress from taking the CAV_{Mean} over the entire stride cycle to smaller intervals such as functional periods of swing or stance, followed by smaller intervals of the previously used quintiles of stride; and finally, a single point within the stride. CAV is thought to behave differently dependent upon the location of stride, hence choosing quintiles each containing a functional aspect of stride. Likewise, the most sensitive measurement within a quintile is a singular point. A local maximum within a quintile would represent the most sensitive measure of a spontaneous increase in CAV during a functional period. Upon observation of CAV values over a period of stride for 13 healthy and 19 PFP runners at a self-selected running pace on a treadmill (Figure E.1) it was noticed that;

1. CAV curves were highly volatile (Q2, KV-AI, Stride KR-AI),
2. quintiles might not encompass consistent increases in CAV entirely (Q4 & Q5, KR-AI),
3. many quintiles include regions of stride where CAV has large increases but also encompasses large regions of low values (Q1, KF-AF, KF-AI),
4. contain separate increases in CAV (Q1, KV-AF), and
5. generally, location of increases in CAV differed among couplings.

It was concluded that quintiles may not be the optimal intervals in which to find CAV_{Mean} values and local maximums within these quintiles may not be acceptably repeatable within these intervals.

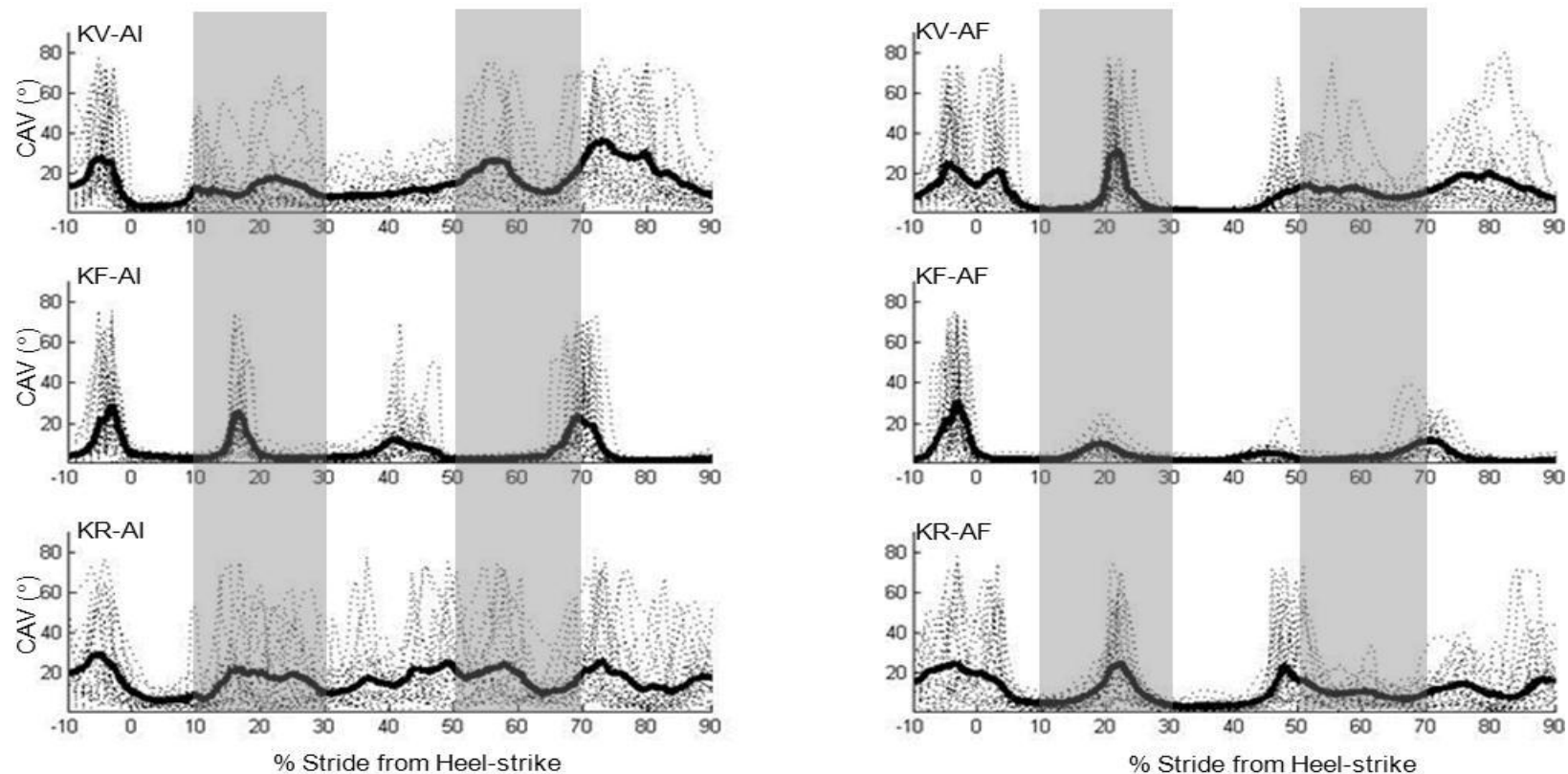


Figure E.1 Coupling Angle Variability (CAV) curves for 13 Healthy and 19 Patellofemoral Pain (PFP) subjects taken from 1 set of 5 non-consecutive gait cycles (---) for six Knee-Ankle coupling combinations. The ensemble averages for all subjects are shown in bold. Stride quintiles (Q1-5) with Q2 and Q4 shaded in the plotting area where CAV_{Mean} is calculated over the entire quintile. All vertical axis units are in degrees ($^{\circ}$). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

Five custom intervals (I1-5) were created to encompass general increases in CAV curves amongst all studied joint couples (Figure E.2). These intervals are located where consistent local maximums occur. CAV curves for all subjects were visually inspected and all consistent increases in CAV were encompassed using these intervals. Most consistent increases are located in I1. I2 mostly contains low CAV values of lesser importance except for KV-AF. I3 contains local maxima in KF-AI, KV-AF, KF-AF and KF-AF also possibly at KV-AI. I4 contained the beginning of swing and encompassed an increase in KV-AI and possibly a slight increase in KF-AF. A more general increase was seen in KV-AF and KR-AF. I5 focused on increases in all couplings. These intervals were chosen to select local maxima, not find the entire mean of the interval.

The local maximum CAV value within each coupling (CAV_{Max}) can be considered the most sensitive measure to CAV change within each interval and may give a more precise description of CAV characteristics. An example of the different sensitivity between these measurements can be observed in the -10 to 30% of a representative subject's data (Figure E.3). In the KV-AF coupling, there is a consistent increase in CAV prior to heel-strike (0%) then a further increase just following heel-strike. Q1 CAV_{Mean} clearly pools these two characteristics together as a singular increase. By separating this quintile into I1 and I2, CAV_{Max} measures are able to separate these characteristics into two CAV measures; I1 and I2 CAV_{Max} .

CAV_{Max} values are clearly more accurate is assessment of CAV magnitude. The CAV_{Max} values located in I3 have values of 53.8° (A) and 71.3° (B), respectively. This is substantially higher than Q2 CAV_{Mean} values of 7.1° (A) and 9.7° (B), respectively. These large discrepancies in values are likely due to the large portion of the CAV curve Q2 encompasses that are of minimal value (*). CAV_{Max} disregards these regions focusing only on the increases in CAV values. These measurements; however, might be too sensitive. The difference between I3 A and B CAV_{Max} values was 17.5° opposed to 1.6° in the Q2 CAV_{Mean} measurements. This was common throughout many intervals; therefore, the mean CAV of an interval $\pm 2\%$ of stride about each CAV_{Max} ($CAV_{Max \pm 2\%}$) was

calculated as an additional measure to quantify CAV. The size of this interval was chosen upon visual inspection of all coupling increases which generally ranged $\pm 2\%$ of CAV_{Max} locations before tapering to a relatively quiescent level. These measures using the same curves and coupling locations are shown (Figure E.4). $CAV_{Max\pm 2\%}$ values for these I3 points were 20.3° (A) and 35.8° (B), respectively.

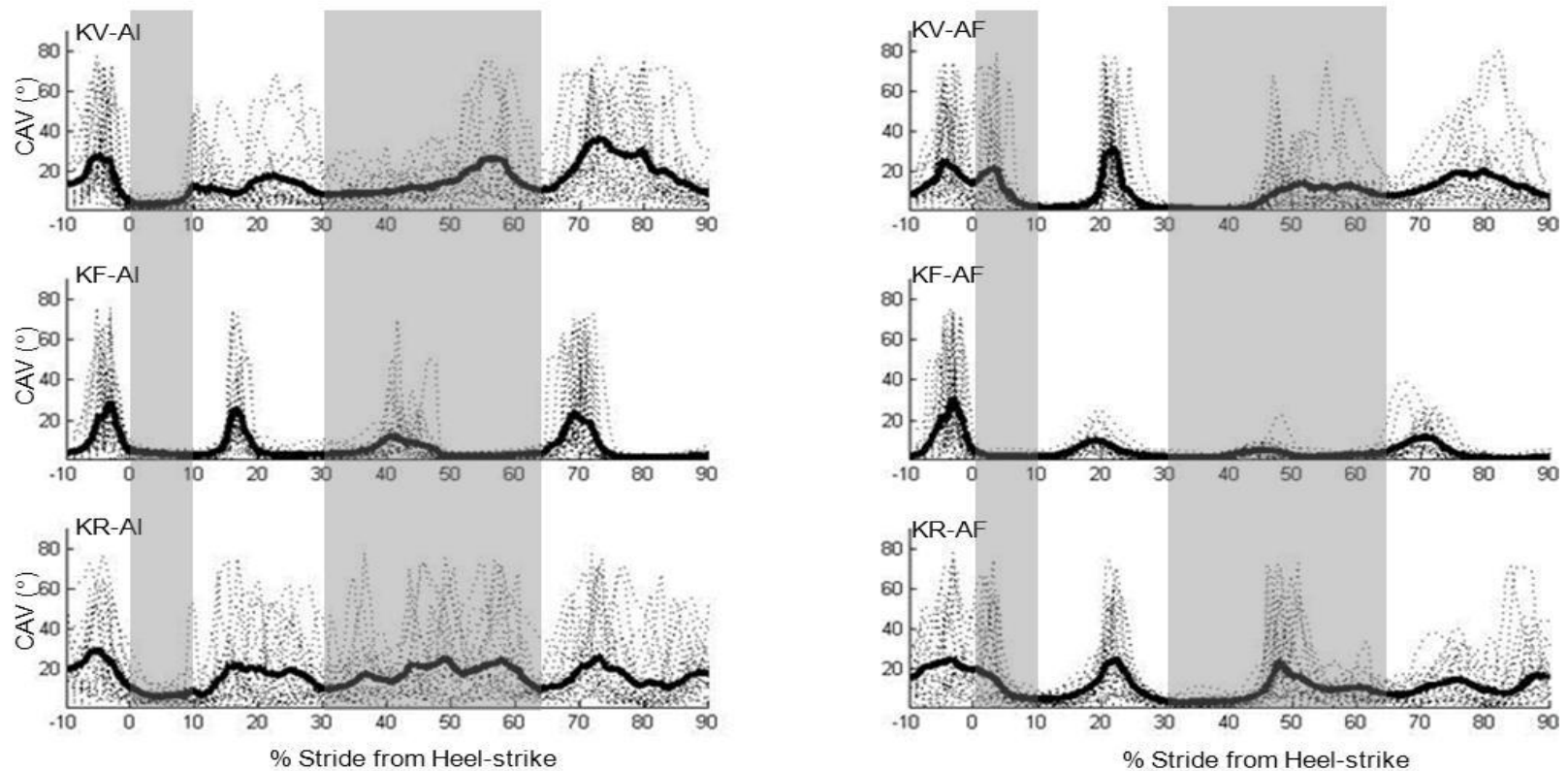


Figure E.2 Coupling Angle Variability (CAV) curves for 13 Healthy and 19 Patellofemoral Pain (PFP) subjects taken from 1 set of 5 non-consecutive gait cycles (---) for six Knee-Ankle coupling combinations. Ensemble averages for all subjects are shown in bold. 5 custom intervals of stride (I1-5) with I2 (0 to 10%) and I4(30 to 64%) shaded where CAVMax and CAVMax \pm 2% are located within each interval. Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

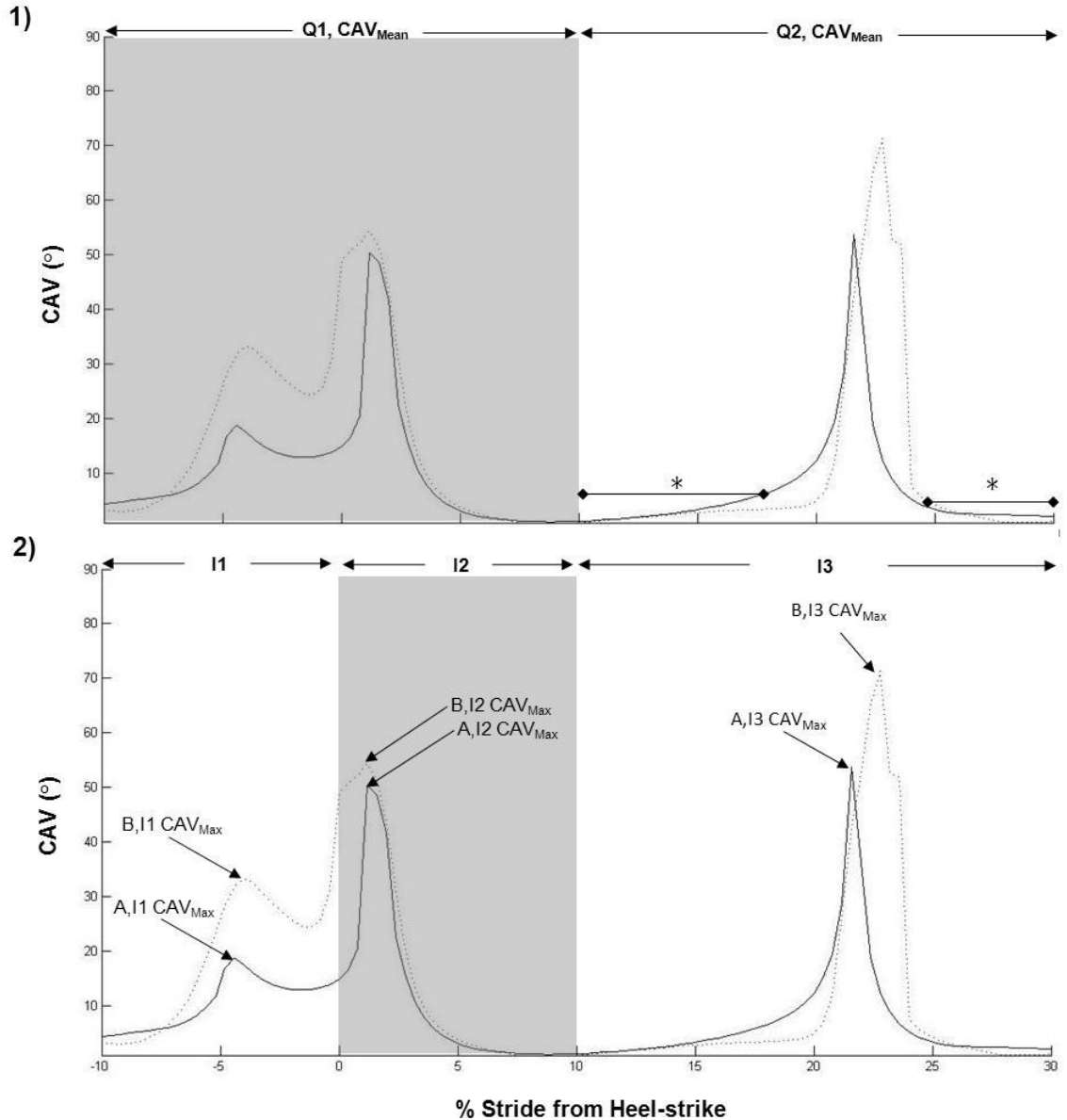


Figure E.3 Coupling Angle Variability (CAV) curves of a representative PFP subject from a first set (A) and second set (B) of 5 non-consecutive gait cycles of the Knee Valgus-Ankle Flexion coupling from -10 to 30% of stride. 1) shows the interval of Q1(shaded) and Q2 where the mean CAV (CAV_{Mean}) will be taken to represent the respective portions of the curve. Q1 will not distinguish the two local maximum values in the curve while Q2 includes CAV characteristics which will decrease the CAV_{Mean} value (*). 2) The same curves are shown except custom intervals (I1,I2, & I3) locate regions to find a local maximum (CAV_{Max}), a more sensitive measure of CAV.

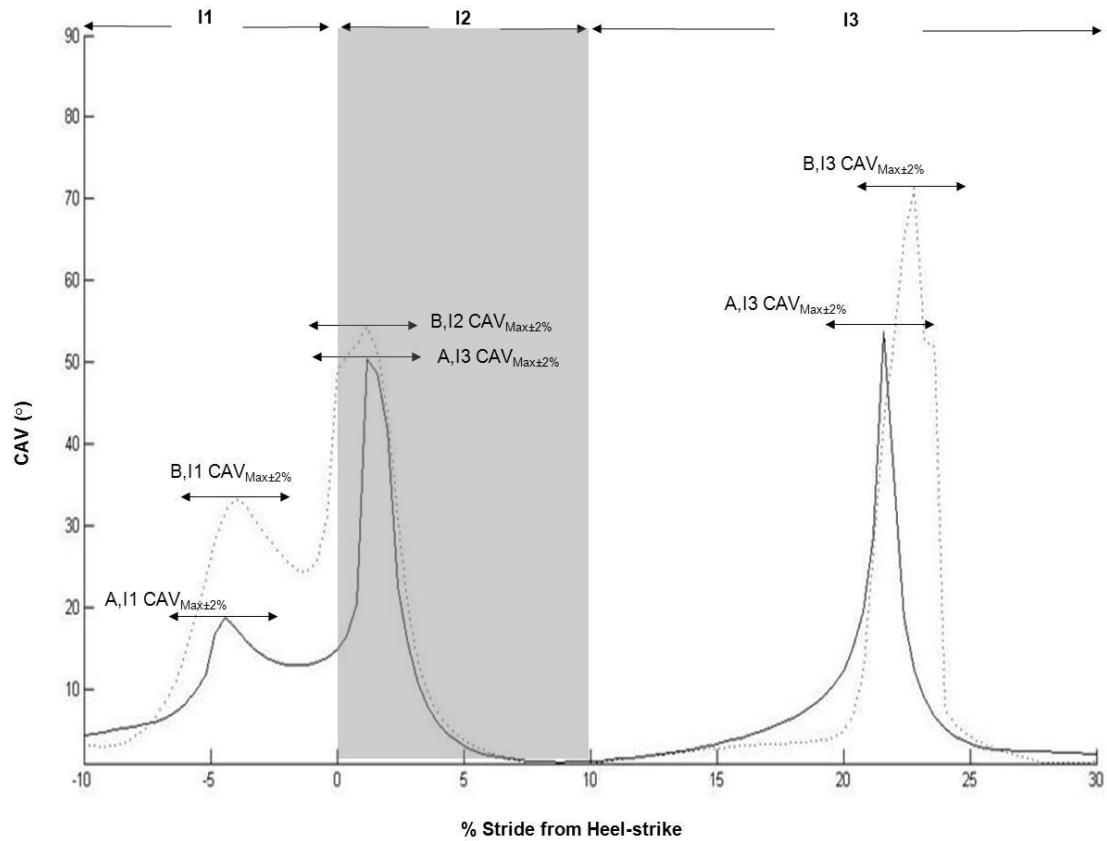


Figure E.4 Locations of $CAV_{Max \pm 2\%}$ measures for a representative PFP subject from a first set (A) and second set (B) of 5 non-consecutive gait cycles of the Knee Valgus-Ankle Flexion coupling from -10 to 30% of stride. Custom intervals (I1, I2(shaded), & I3) locate regions to find CAV_{Max} values and the subsequent CAV_{Mean} intervals $\pm 2\%$ of stride about a CAV_{Max} ($CAV_{Max \pm 2\%}$). Interval widths are indicated by horizontal arrows.

Appendix F: Max Coupling Angle Variability stride location results for each study

Table F.1 Test-retest levels of agreement for CAV_{MaxLoc} values within five intervals of stride (I1-5) at a self-selected running pace for six Knee-Ankle joint coupling combinations. Data are displayed separately for Healthy and Patellofemoral Pain populations.

Interval (% of stride)	KV-AI		KV-AF		KF-AI		KF-AF		KR-AI		KR-AF	
	$\delta \pm 1.96\sigma$	$\bar{X}(\%)$	$\delta \pm 1.96\sigma$	$\bar{X}(\%)$	$\delta \pm 1.96\sigma$	$\bar{X}(\%)$	$\delta \pm 1.96\sigma$	$\bar{X}(\%)$	$\delta \pm 1.96\sigma$	$\bar{X}(\%)$	$\delta \pm 1.96\sigma$	$\bar{X}(\%)$
<i>Healthy</i>												
I1 (-10 to 0)	-0.2 ± 1.3	-4.5	-0.4 ± 1.3	-3.6	0.0 ± 0.8	-3.3	0.1 ± 0.8	-3.2	-1.8 ± 5.8	-5.0	-0.3 ± 6.3	-4.2
I2 (0 to 10)	2.5 ± 8.6	7.1	0.6 ± 2.3	3.1	-0.1 ± 4.8	2.2	-0.4 ± 8.9	2.6	1.3 ± 8.6	5.2	1.3 ± 4.4	2.3
I3 (10 to 30)	2.7 ± 11.9	18.0	-0.1 ± 0.7	21.9	-0.2 ± 1.9	17.1	0.1 ± 0.9	19.8	2.9 ± 12.8	19.8	-0.1 ± 2.0	21.1
I4 (30 to 64)	-1.0 ± 23.9	51.9	-0.6 ± 3.7	50.1	0.2 ± 1.6	41.4	4.7 ± 14.3	48.8	0.1 ± 19.9	48.5	1.3 ± 11.1	50.1
I5 (64 to 90)	1.5 ± 6.5	73.9	1.1 ± 19.8	76.7	-0.1 ± 1.4	70.1	-0.6 ± 2.1	70.4	1.7 ± 13.1	79.2	-1.1 ± 13.3	85.8
<i>Patellofemoral Pain</i>												
I1 (-10 to 0)	-0.3 ± 1.5	-4.1	-0.4 ± 4.4	-3.5	-0.2 ± 1.2	-3.0	-0.2 ± 1.5	-2.9	0.4 ± 5.7	-4.9	-0.7 ± 5.7	-3.3
I2 (0 to 10)	-1.5 ± 10.7	7.9	0.2 ± 1.1	3.3	-0.5 ± 7.4	2.1	-0.1 ± 4.7	1.2	-0.5 ± 9.8	4.3	-0.6 ± 7.6	2.7
I3 (10 to 30)	0.8 ± 9.8	17.7	0.1 ± 1.3	22.1	0.1 ± 1.1	17.0	-0.2 ± 2.0	19.4	0.0 ± 11.8	19.8	0.5 ± 2.2	22.1
I4 (30 to 64)	-2.8 ± 12.0	52.0	-1.1 ± 8.5	52.8	0.3 ± 2.3	43.4	2.9 ± 11.6	47.0	-1.5 ± 16.0	49.1	-1.3 ± 7.3	51.8
I5 (64 to 90)	-0.8 ± 7.5	78.1	-0.3 ± 3.6	77.9	-0.2 ± 1.0	70.5	-0.3 ± 2.9	70.7	0.4 ± 11.2	77.7	2.4 ± 15.9	82.7

Note: CAV_{MaxLoc} = location in the stride of the maximum coupling angle variability value over a selected interval of stride for a set of 5 non-consecutive gait cycles taken from a 10 s running trial. δ = mean difference between two sets of CAV_{MaxLoc} values for each subject within a population ($CAV_{MaxLoc1} - CAV_{MaxLoc2}$); 1.96σ = 95% confidence interval of δ ; \bar{X} = grand mean for both sets of CAV_{MaxLoc} values within a population. All units are in % of stride from heel strike (%). Heel-strike=0%. Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion

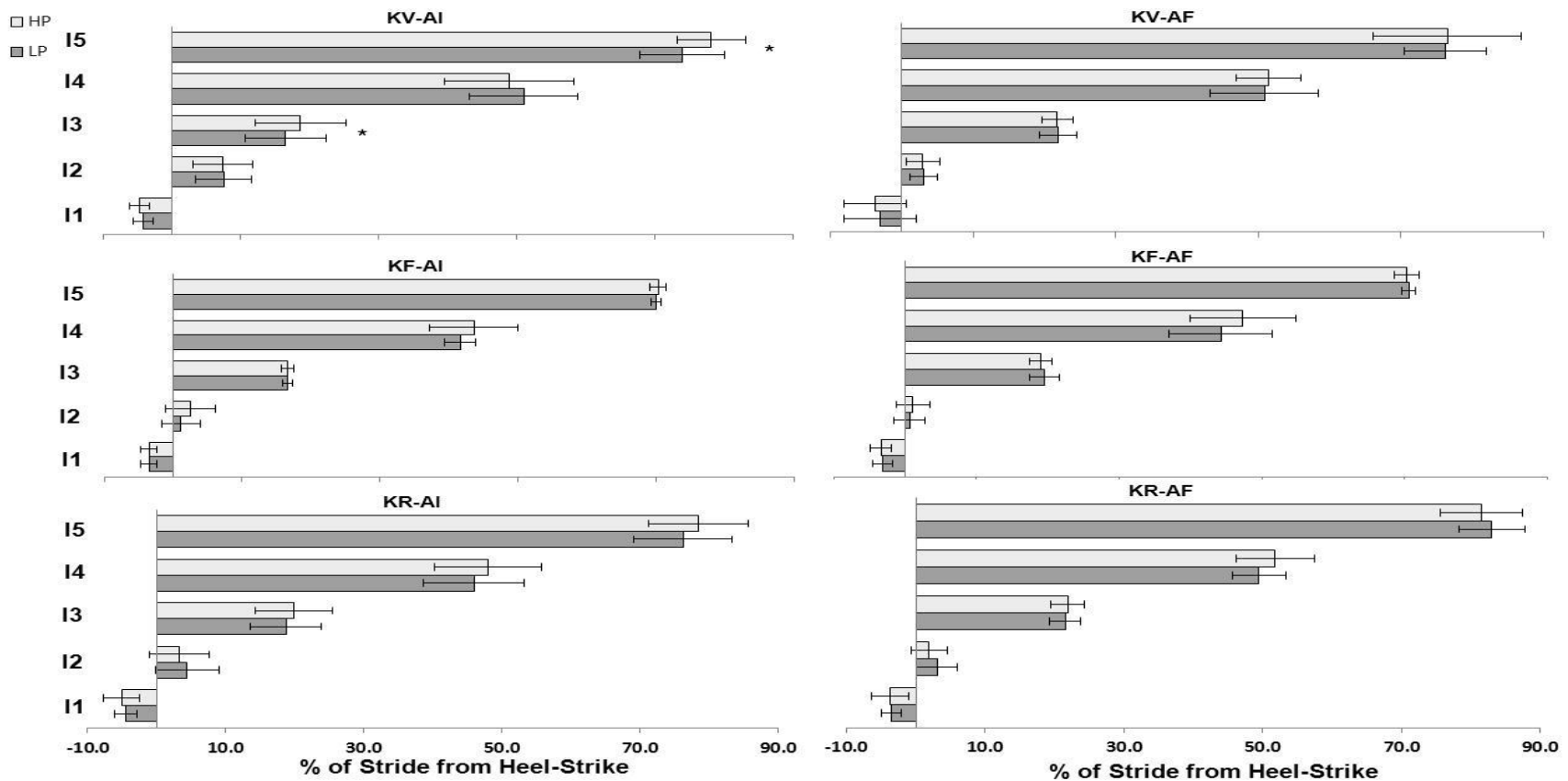


Figure F.1 Max Coupling Angle Variability stride location (CAV_{MaxLoc} ± SD) for five intervals (I1-5, Table 1) of stride at a self-selected running pace for six Knee-Ankle joint coupling combinations for female runners with patellofemoral pain. Data are taken from 5 non-consecutive gait cycles from a 10 s running trial for a low pain (LP) and high pain (HP) condition. Units of CAV_{MaxLoc} values are in percentage of stride as measured from heel-strike. Significant differences are denoted at p < .05(*). Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

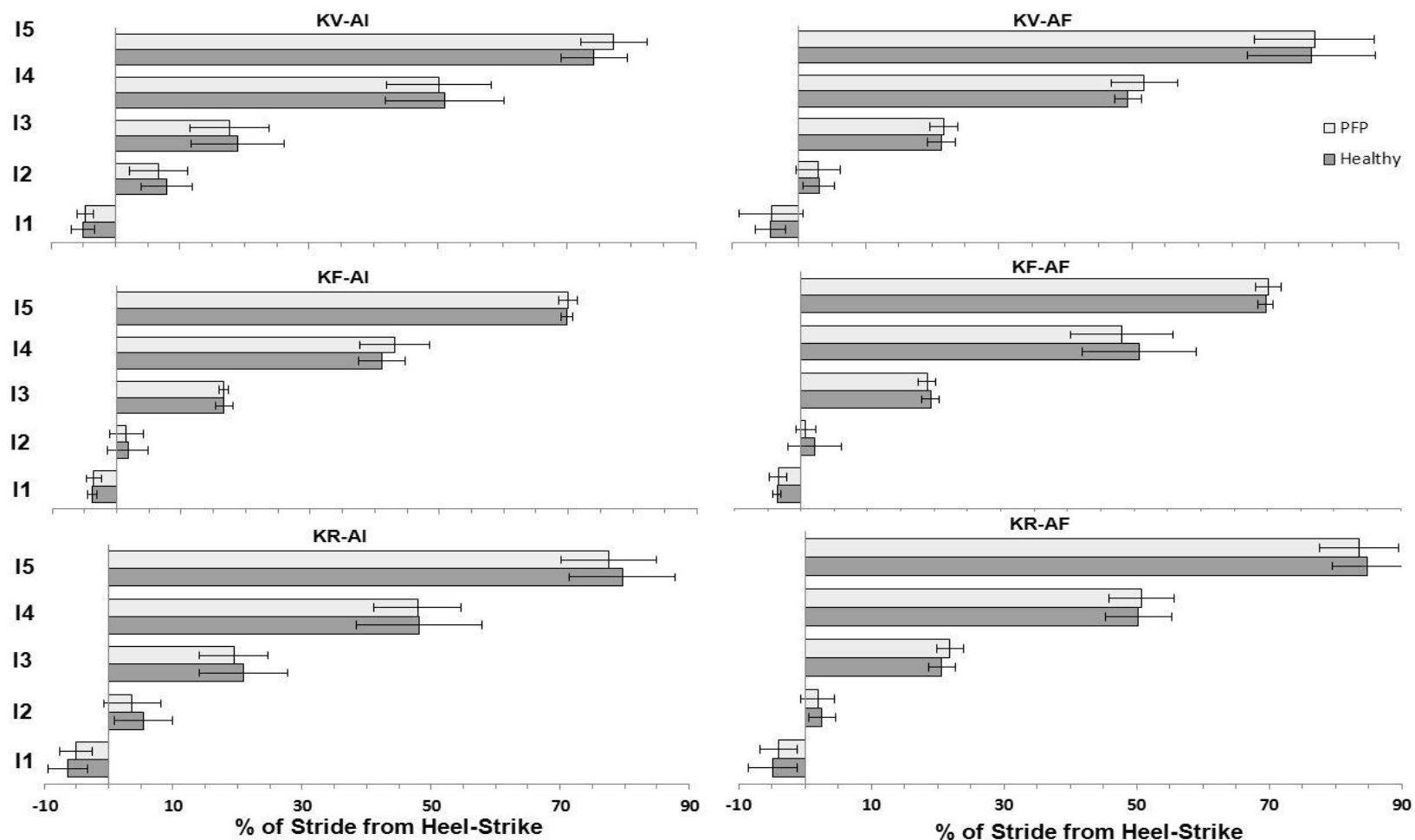



Figure F.2 Max Coupling Angle Variability stride location ($CAV_{MaxLoc} \pm SD^\circ$) for five intervals (I1-5, Table 1) of stride at a self-selected running pace for six Knee-Ankle joint coupling combinations for females with patellofemoral pain (PFP) and Healthy controls. Data are taken from 5 non-consecutive gait cycles from a 10 s running trial in a high pain condition for PFP and the 11th minute of a self-selected pace for healthy. No significant differences between populations were observed. Coupling angle abbreviations: KV=Knee Valgus/Varus, KF=Knee Flexion/Extension, KR= Knee Internal/External Rotation, AI=Ankle Inversion/Eversion, AF=Ankle Plantar/Dorsi Flexion.

Appendix G: Recruitment Flyers

UNIVERSITY OF KENTUCKY RESEARCH

A photograph of four female runners in a race. They are wearing athletic gear and bib numbers. From left to right, the bib numbers are 200, 197, 196, and 102. They are running on a track with trees in the background.

Female Runners Needed for Research Study

Researchers at the University Of Kentucky College Of Health Sciences are conducting a research study to identify differences in hip, knee and ankle movement associated with Patellofemoral Pain Syndrome (knee pain under or around the knee cap), one of the most common injuries associated with active females. Both runners with knee pain and healthy runners are needed for this study.

You may be eligible to participate if you:

- are a female runner;
- are 18 - 45 years of age;
- are running at least six miles per week; and
- with or without knee pain

Research will be performed at the University of Kentucky Biodynamic Laboratory in Lexington, KY.

Principal Investigator is Brian Noehren PT, PhD,
Research coordinator is Zack Sanchez.

For more information, call (813) [REDACTED]
or email [REDACTED]@uky.edu.

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Figure G.1 Recruitment flyer distributed throughout local community.

Appendix H: IRB Approved Informed Consent

Consent to Participate in a Research Study

A COMPARISON OF THE JOINT KINEMATICS CHARACTERISTICS OF THE LOWER EXTREMITY IN FEMALE HEALTHY RUNNERS AND FEMALE RUNNERS PRESENTING WITH PATELLOFEMORAL PAIN SYNDROME

WHY ARE YOU BEING INVITED TO TAKE PART IN THIS RESEARCH?

You are being invited to take part in a research study that involves understanding how the hip, knee and ankle joints act together to possibly cause knee pain. You are being invited to take part in this research study because you are a female runner that has knee pain or are a female runner that does not have a history of knee pain. If you volunteer to take part in this study, you will be one of about 60 people to do so.

WHO IS DOING THE STUDY?

The person in charge of this study is Brian Noehren, PT, PhD (PI) of University of Kentucky, Department of Rehabilitation Sciences. There may be other people on the research team assisting at different times during the study.

WHAT IS THE PURPOSE OF THIS STUDY?

By doing this study, we hope to learn how to identify differences in movement between healthy runners and runners with patellofemoral pain syndrome. By looking at runners' hip, knee and ankle movements during running on a treadmill, we can hopefully use this information in the future to help clinicians develop better methods of treating patellofemoral pain syndrome (pain under the knee cap) in active individuals. Additionally, we are working on developing simple methods of determining leg movements that can be used by clinicians.

ARE THERE REASONS WHY YOU SHOULD NOT TAKE PART IN THIS STUDY?

You should not participate in this study if you have had knee surgery, have lower extremity injury/s other than knee pain, have an allergy to tape, and are under 18 years of age or over 45 years of age. If you are a healthy control subject, then you should not participate if you have had any lower extremity injury that affects your running or are under 18 years of age or over 45 years of age.

WHERE IS THE STUDY GOING TO TAKE PLACE AND HOW LONG WILL IT LAST?

The research procedures will be conducted at the University of Kentucky's Biodynamics Laboratory located in the central part of campus. You will need to come to the Wenner-Gren building where the lab is centrally located 1 time during the study. The total amount of time you will be asked to volunteer for this study is 2 hours.

WHAT WILL YOU BE ASKED TO DO?

A member of the research team will first screen you to determine if you can participate in the study. If you are a control subject, then you may participate as long as you do not meet any exclusion criteria. If you have knee pain you will be screened by a licensed physical therapist or athletic trainer. They will determine if you have the knee condition that we are looking for. Then you will be asked to complete the rest of the study. If you do not meet the criteria then you will be excluded from the study. We will ask that you wear athletic shorts and shirt (which we will provide if necessary). Running shoes will be provided for you.

Initial Pain Assessment:

You will also be asked to rate your knee pain at its worst while running in the past week on a scale between 0-10 (0 means no pain; 10 means the worst pain imaginable).

2D foot, hip analysis:

In order to better understand how the hip and foot contribute to the development of knee pain we will take a picture of them while you perform a squat. The picture will only be of your foot and hip. We will compare these pictures to the data we collect with the motion analysis system.

Motion Analysis:

Motion analysis will provide a means for evaluating motion of your hip, knee and ankle joints during walking and running on a treadmill. You will have approximately 40 reflective markers placed on certain landmarks of your legs and lower back to allow the motion analysis system to record your hip, knee and ankle movements. Markers will be applied with sticky tape to the skin and if necessary, athletic tape to limit marker movement during activities. A stationary trial will be collected to help us identify anatomical landmarks. After this trial, some markers might be removed that won't be necessary for the activities you will later perform. Motion data will be collected for 10 seconds every minute. Additionally, we will take a video from your lower back down of your running form 5 minutes into the run.

Pain Assessment during Activities:

In order to monitor your pain throughout the activities of this study, you will be asked to rate your current pain from 0 to 10. (0 means no pain; 10 means worst pain imaginable) We will ask you to rate your pain once every minute you are walking or running on the treadmill. If your pain ever reaches the value of a 7, the data collection session will be terminated immediately.

Exertion Assessment during Activities

In order to control for fatigue throughout the activities of this study, you will be asked to wear a heart rate monitor and rate your perceived exertion from 0 to 20. This lets us approximate how fatigued you might be. You will not be asked to perform activity over a level of 15. An example of the chart we will show you is shown below:

6 No exertion at all	12
7 Extremely light	13 Somewhat hard
8	14
9 Very light	15 Hard (heavy)
10	16
11 Light	17 Very hard (very strenuous, very fatigued)

We will ask you to rate your perceived exertion once every minute you are walking or running on the treadmill.

Treadmill Activity:

You may walk on the treadmill as long as necessary to feel comfortable with it. Once you are ready you will run at a self selected warm up pace for 3 minutes. We will then gradually increase the speed of the treadmill to 3.35 meters/second or an 8 minute mile pace. You will run at this specific pace for two minutes. After the two minutes, you will be able to self-select the pace of the rest of your run. The run will continue for 30 minutes. We will then reduce the speed and have you walk until you are below a fatigue level of 11/20 on RPE scale for 2 minutes. You may request to stop walking or running at any time. Additionally, if your pain goes above 7 out of 10 or you become too fatigued (15/20 on RPE scale) we will stop the study.

Future Studies:

I give permission to Brian Noehren PT, Ph.D. and his research team to contact me regarding future research studies involving orthopedic data. If you agree to be contacted for future research, your contact and consent information will be kept in a separate locked filing cabinet in the principal investigators office. This office has limited access and is kept locked when not occupied. Your information will only be available to the principal investigator. Your records will be kept for three years following the conclusion of this study at which point they will be shredded and disposed of with appropriate care.

☐☐

Yes

No

WHAT ARE THE POSSIBLE RISKS AND DISCOMFORTS?

Risks are minimal in this study. You may experience a skin reaction from the adhesive markers or joint or muscle soreness from activities that you will perform. These discomforts should be

minimal. If you have patellofemoral pain syndrome you will experience pain in your knee typical to what you experience when you ran. This pain will be monitored throughout your testing session. In addition to the risks listed above, you may experience a previously unknown risk or side effect.

WILL YOU BENEFIT FROM TAKING PART IN THIS STUDY?

You will not get any personal benefit from taking part in this study.

DO YOU HAVE TO TAKE PART IN THE STUDY?

If you decide to take part in the study, it should be because you really want to volunteer. You will not lose any benefits or rights you would normally have if you choose not to volunteer. You can stop at any time during the study and still keep the benefits and rights you had before volunteering. If you decide not to take part in this study, your decision will have no effect on your grades or standing at the University of Kentucky.

IF YOU DON'T WANT TO TAKE PART IN THE STUDY, ARE THERE OTHER CHOICES?

If you do not want to be in the study, there are no other choices except not to take part in the study.

WHAT WILL IT COST YOU TO PARTICIPATE?

Cost of parking expenses will be paid by funds from the investigator.

WHO WILL SEE THE INFORMATION THAT YOU GIVE?

We will make every effort to keep private all research records that identify you to the extent allowed by law.

Your information will be combined with information from other people taking part in the study. When we write about the study to share it with other researchers, we will write about the combined information we have gathered. You will not be personally identified in these written materials. We may publish the results of this study; however, we will keep your name and other identifying information private. Officials of the University of Kentucky may look at or copy pertinent portions of your records that identify you.

We will make every effort to prevent anyone who is not on the research team from knowing that you gave us information, or what that information is. Electronic data will be stored on password protected computers in the Biodynamics laboratory and on storage devices. The storage devices and hard copies of the data when not in use will be stored in a locked filing cabinet in the principal investigator's office or the laboratory. The laboratory and office have limited access, and are kept locked shut when not occupied. All electronic data will be coded with only the subject's initials, and of the hard copied data only the informed consent forms will contain the name of the subject. These consent forms will be kept in a filing cabinet separate to the one containing the storage devices and hard copies of the data.

CAN YOUR TAKING PART IN THE STUDY END EARLY?

If you decide to take part in the study you still have the right to decide at any time that you no longer want to continue. You will not be treated differently if you decide to stop taking part in the study.

The individuals conducting the study may need to withdraw you from the study. This may occur if you are not able to follow the directions they give you, or if they find that your being in the study is more risk than benefit to you.

ARE YOU PARTICIPATING OR CAN YOU PARTICIPATE IN ANOTHER RESEARCH STUDY AT THE SAME TIME AS PARTICIPATING IN THIS ONE?

You may take part in this study if you are currently involved in another research study. It is important to let the investigator/your doctor know if you are in another research study. You should also discuss with the investigator before you agree to participate in another research study while you are enrolled in this study.

WHAT HAPPENS IF YOU GET HURT OR SICK DURING THE STUDY?

If you believe you are hurt or if you get sick because of something that is due to the study, you should call Brian Noehren, PT, PhD (PI) at 859-218-0581 immediately.

Brian Noehren, PT, PhD (PI) will determine what type of treatment, if any, that is best for you at that time. This may include referral to your primary care physician for treatment.

It is important for you to understand that the University of Kentucky does not have funds set aside to pay for the cost of any care or treatment that might be necessary because you get hurt or sick while taking part in this study. Also, the University of Kentucky will not pay for any wages you may lose if you are harmed by this study.

The medical costs related to your care and treatment because of research related harm will be your responsibility;

Or

May be paid by your insurer if you are insured by a health insurance company (you should ask your insurer if you have any questions regarding your insurer's willingness to pay under these circumstances);

Or

May be paid by Medicare or Medicaid if you are covered by Medicare, or Medicaid (if you have any questions regarding Medicare/Medicaid coverage you should contact Medicare by calling 1-800-Medicare (1-800-633-4227) or Medicaid 1-800-635-2570. A co-payment/deductible from you may be required by your insurer or Medicare/Medicaid even if your insurer or Medicare/Medicaid has agreed to pay the costs). The amount of this co-payment/deductible may be substantial.

You do not give up your legal rights by signing this form.

WILL YOU RECEIVE ANY REWARDS FOR TAKING PART IN THIS STUDY?

You will not receive any rewards or payment for taking part in the study. However, your parking expenses will be paid by funds from the PI.

WHAT IF YOU HAVE QUESTIONS, SUGGESTIONS, CONCERNS, OR COMPLAINTS?

Before you decide whether to accept this invitation to take part in the study, please ask any questions that might come to mind now. Later, if you have questions, suggestions, concerns, or complaints about the study, you can contact the investigator, Brian Noehren PT, PhD at 859-218-0581. If you have any questions about your rights as a volunteer in this research, contact the staff in the Office of Research Integrity at the University of Kentucky at 859-257-9428 or toll free at 1-866-400-9428. We will give you a signed copy of this consent form to take with you.

WHAT IF NEW INFORMATION IS LEARNED DURING THE STUDY THAT MIGHT AFFECT YOUR DECISION TO PARTICIPATE?

If the researcher learns of new information in regards to this study, and it might change your willingness to stay in this study, the information will be provided to you. You may be asked to sign a new informed consent form if the information is provided to you after you have joined the study.

WHAT ELSE DO YOU NEED TO KNOW?

There is no external group providing financial support and/or material for this study.

Signature of person agreeing to take part in the study

Date

Printed name of person agreeing to take part in the study

Name of [authorized] person obtaining informed consent

Date

Signature of Investigator

References

- ACSM, 2000. ACSM's guidelines for exercise testing and prescription. Lippincott Williams & Wilkins, Philadelphia.
- Amazeen, P. G., Amazeen, E. L., Turvey, M. T., 1998. Dynamics of human intersegmental coordination: theory and research. (edited by 238-259. MIT Press, Cambridge.
- Arutyunyan, G. H., Gurfinkel, V. S., Mirskii, M. L., 1968. Investigation of aiming at a target. *Biophysics* 13, 536-538.
- Atkinson, G., Nevill, A. M., 1998. Statistical methods for assessing measurement error (reliability) in variables relevant to sports medicine. . *Sports Medicine* 26, 217-238.
- Barry, E. B., 1980. Characteristics of Gait. MS Thesis, University of New South Wales, Australia.
- Bartlett, R., Wheat, J., Robins, M., 2007. Is movement variability important for sports biomechanists? *Sports Biomechanics* 6, 224-243.
- Barton, C. J., Levinger, P., Menz, H. B., Webster, K. E., 2009. Kinematic gait characteristics associated with patellofemoral pain syndrome: A systematic review. *Gait & Posture* 30, 405-416.
- Bates, B. T., Dufek, J. S., Davis, H. P., 1992. The effect of trial size on statistical power. *Medicine & Science in Sports & Exercise* 24, 1059-1065.
- Batschelet, E., 1981. Circular statistics in biology. Academic Press, New York.
- Bernstein, N. A., 1967. The co-ordination and regulation of movements. Pergamon Press, Oxford, England.
- Biedert, R. M., stauffer, E., Friederich, N. F., 1992. Occurrence of free nerve endings in the soft tissue fo the knee joint: a histologic investigation. *American Journal of Sports Medicine* 20, 430-433.
- Bland, J. M., Altman, D. G., 1986. Statistical methods for assessing agreement between two methods of clinical measurement. *Lancet* i, 307-310.
- Bolglia, L. A., 2005. Etiology of patellofemoral pain syndrome: A proximal link to a distal problem. PhD thesis. University of Kentucky, Lexington KY. 134

Bolgia, L. A., Malone, T., Umberger, B. R., Uhl, T. L., 2008. Hip strength and hip and knee kinematics during stair descent in females with and without patellofemoral pain syndrome. *Journal of Orthopaedic Sports Physical Therapy* 38, 12-18.

Boling, M., Padua, D., Marshall, S., Guskiewicz, K., Pyne, S., Beutler, A., 2010. Gender differences in the incidence and prevalence of patellofemoral pain syndrome. *Scandinavian Journal of Medicine & Science in Sports* 20, 725-730.

Borg, G. A. V., 1982. Psychophysical basis of perceived exertion. *Medicine and Science in Sports and Exercise* 14, 377-381.

Brown, A. M., 2011. The effects of fatigue on pathomechanics and electromyography in female runners with iliotibial band syndrome. PhD, Temple University, Ann Arbor, MI. 149.

Cavanagh, P. R., LaFortune, M. A., 1980. Ground reaction forces in distance running. *Journal of Biomechanics* 13, 397-406.

Chang, R., van Emmerik, R. E. A., Hamill, J., 2008. Quantifying rearfoot-forefoot coordination in human walking. *Journal of Biomechanics* 41, 3101-3105.

Clark, J. E., Phillips, S. J., 1993. A longitudinal study of intralimb coordination in the first year of independent walking: a dynamical systems analysis. *Child Development* 64, 1143-1157.

Clement, D. B., Tauton, J. E., Smart, G. W., McNichol, N. L., 1981. A survey of overuse running injuries. *Physician Sportsmedicine* 9, 47-58.

Crossley, K., Bennell, K., Cowan, S., Green, S., 2004. Analysis of outcome measures for persons with patellofemoral pain: which are reliable and valid? *Archives of Physical Medicine and Rehabilitation* 85, 815-822.

Crowther, R. G., Spinks, W. L., Leicht, A. S., Quigley, F., Golledge, J., 2008. Intralimb coordination variability in peripheral arterial disease. *Clinical Biomechanics* 23, 357-364.

Crowther, R. G., Spinks, W. L., Leicht, A. S., Quigley, F., Golledge, J., 2008. Lower limb movement variability in patients with peripheral arterial disease. *Clinical Biomechanics* 23, 1080-1085.

Crowther, R. G., Spinks, W. L., Leicht, A. S., Sangla, K., Quigley, F., Golledge, J., 2009. The influence of a long term exercise program on lower limb movement variability and walking performance in patients with peripheral arterial disease. *Human Movement Science* 28, 494-503.

- Csintalan, R. P., Schultz, M. M., Woo, J., McMahon, P. J., Lee, T. Q., 2002. Gender differences in patellofemoral joint biomechanics. *Clinical Orthopaedics and Related Research* 402, 260-269.
- Davids, K., Glazier, P., Araujo, D., Barlett, R., 2003. Movement systems as dynamic systems: the functional role of variability and its implications for sports medicine. *Sports Medicine* 33, 245-260.
- Davids, K., Shuttleworth, R., Button, C., Renshaw, I., Glazier, P., 2004. "Essential noise" - enhancing variability of informational constraints benefits movement control: a comment on Waddington and Adams (2003). *British Journal of Sports Medicine* 38, 601-605.
- Davis, I. S., Powers, C. M., 2010. Patellofemoral pain syndrome: proximal, distal, and local factors. *Journal of Orthopaedic & Sports Physical Therapy* 40, A1-A48.
- DeHaven, K. E., Lintner, D., 1986. Athletic injuries: comparison by age, sport, and gender. *American Journal of Sports Medicine* 14, 218-224.
- DeLeo, A. T., Dierks, T. A., Ferber, R., Davis, I. S., 2004. Lower extremity joint coupling during running: a current update. *Clinical Biomechanics* 19, 983-991.
- Derrick, T. R., Dereu, D., McLean, S. P., 2002. Impacts and kinematic adjustments during an exhaustive run. *Medicine & Science in Sports & Exercise* 34, 998-1002.
- Devereaux, M., Lachman, S., 1984. Patellofemoral arthralgia in athletes attending a sports injury clinic. *British Journal of Sports Medicine* 20, 434-440.
- Diedrich, F. J., Warren, W. H., 1995. Why change gaits? Dynamics of the walk-run transition. *Journal of Experimental Psychology* 21, 183-202.
- Dierks, T. A., 2005. Kinematics and joint coupling in runners with patellofemoral pain during a prolonged run. PhD. thesis, University of Delaware, Newark. 177.
- Dierks, T. A., Davis, I., 2004. Lower extremity joint coupling and patellofemoral joint pain during running. *Medicine & Science in Sports & Exercise* 36, S56.
- Dierks, T. A., Davis, I., 2007. Discrete and continuous joint coupling relationships in uninjured recreational runners. *Clinical Biomechanics* 22, 581-591.
- Dierks, T. A., Manal, K., Hamill, J., Davis, I., 2011. Lower extremity kinematics in runners with patellofemoral pain during a prolonged run. *Medicine & Science in Sports & Exercise* 43, 693-700.

- Dierks, T. A., Manal, K., Hamill, J., Davis, I. S., 2008. Proximal and distal influences on hip and knee kinematics in runners with patellofemoral pain during a prolonged run. *Journal of Orthopaedic & Sports Physical Therapy* 38, 448-456.
- Dillon, P., WF, U., Allen, W., 1983. Gait analysis with reference to chondromalacia patellae. *Journal of Orthopaedic & Sports Physical Therapy* 5, 127-131.
- Elias, J. J., Cech, J. A., Weinstein, D. M., Cosgrea, A. J., 2004. Reducing the lateral force acting on the patella does not consistently decrease patellofemoral pressures. *The American Journal of Occupational Therapy* 32, 1202-1208.
- Farrar, J. T., Young Jr., J. P., LaMoreaux, L., Werth, J. L., Poole, R. M., 2001. Clinical importance of changes in chronic pain intensity measured on an 11-point numerical pain rating scale. *Pain* 94, 149-158.
- Ferber, R., Davis, I. M., Williams III, D. S., 2003. Gender differences in lower extremity mechanics during running. *Clinical Biomechanics* 18, 350-357.
- Ferber, R., Davis, I. M., Williams III, D. S., 2005. Effect of foot orthotics on rearfoot and tibia joint coupling patterns and variability. *Journal of Biomechanics* 38, 477-483.
- Ferber, R., Davis, I. M., Williams III, D. S., Lughton, C., 2002. A comparison of within- and between-day reliability of discrete 3D lower extremity variables in runners. *Journal of Orthopaedic Research* 20, 1139-1145.
- Field-Fote, E. C., 2003. Quantification of functional behavior in humans and animals: Time for a paradigm shift. *Journal of Rehabilitation Research and Development* 40, 19-24.
- Field-Fote, E. C., Tepavac, D., 2002. Improved intralimb coordination in people with incomplete spinal cord injury following training with body weight support and electrical stimulation. *Physical Therapy* 82, 707-715.
- Fitch, H. L., Tuller, B., Turvey, M. T., 1982. The Bernstein perspective: III. Tuning of coordinative structures with special reference to perception. (edited by Kelso, J.A.S.), 271-287. Lawrence Erlbaum Associates, Hillsdale, NJ.
- Fredericson, M., Powers, C. M., 2002. Practical management of patellofemoral pain. *Clinical Journal of Sport Medicine* 12, 36-38.
- Freeman, H. A., 1961. A technique for the classification and recognition of geometric patterns. Namur, Belgium.

- Fulkerson, J. P., 1983. The etiology of patellofemoral pain in young, active patients. *Clinical Orthopaedics and Related Research* 179, 128-133.
- Fulkerson, J. P., 1994. Patellofemoral pain disorders: Evaluation and management *Journal of American Academy of Orthopaedic Surgeons* 2, 124-132.
- Fulkerson, J. P., 2002. Diagnosis and treatment of patients with patellofemoral pain. *The American Journal of Sports Medicine* 30, 447-456.
- Gittoes, M. J. R., Wilson, C., 2010. Intralimb joint coordination patterns of the lower extremity in maximal velocity phase sprint running. *Journal of Applied Biomechanics* 26, 188-195.
- Glazier, P. S., Wheat, J. S., Pease, D. L., Bartlett, R. M., 2006. The interface of biomechanics and motor control: Dynamic systems theory and the functional role of movement variability. (edited by Davids, K., Bennett, S., Newell, K.), 49-69. *Human Kinetics, Champaign, Illinois, USA*.
- Grieve, D. W., 1968. Gait pattern and the speed of walking. *Biomedical Engineering* 3, 119-122.
- Good, E. S., Suntay, W. J., 1983. A joint coordinate system for the clinical description of three-dimensional motions: Application to the knee. *Transactions of the ASME* 105, 136-144.
- Gruber, A. H., Russell, E. M., Miller, R. H., Chang, R., Hamill, J., 2009. Segment coordination response to alterations in foot strike pattern. *American Society of Biomechanics Annual Meeting, State College, Pennsylvania: Penn State University*.
- Haken, H., Kelso, J., Bunz, H., 1985. A theoretical model of phase transitions in human hand movements. *Biological Cybernetics* 51, 347-56.
- Hamill, J., Haddad, J. M., Heiderscheit, B. C., Van Emmerik, R. E. A., Li, L., 2006. Clinical relevance of variability in coordination. (edited by Davids, K., Bennett, S., Newell, K.), 153-165. *Human Kinetics, Champaign, Illinois, USA*.
- Hamill, J., Haddad, J. M., McDermott, W. J., 2000. Issues in quantifying variability from a dynamical systems perspective. *Journal of Applied Biomechanics* 16, 407-418.
- Hamill, J., van Emmerik, R. E. A., 2000. Variability and stability: A dynamical systems perspective. *Journal of Applied Biomechanics* 16, 392-393.

Hamill, J., van Emmerik, R. E. A., Heiderscheit, B. C., Li, L., 1999. A dynamical systems approach to lower extremity running injuries. *Clinical Biomechanics* 14, 297-308.

Heiderscheit, B. C., 2000. Locomotion variability and joint pain. PhD Thesis, University of Massachusetts Amherst, Cambridge MA. 177

Heiderscheit, B. C., 2000. Movement variability as a clinical measure for locomotion. *Journal of Applied Biomechanics* 16, 419-427.

Heiderscheit, B. C., Hamill, J., van Emmerik, R. E. A., 1999. Q-angle influences on the variability of lower extremity coordination during running. *Medicine & Science in Sports & Exercise* 31, 1313-1319.

Heiderscheit, B. C., Hamill, J., van Emmerik, R. E. A., 2002. Variability of stride characteristics and joint coordination among individuals with unilateral patellofemoral pain. *Journal of Applied Biomechanics* 18, 110-121.

Herman, L., Foster, C., Maher, M. A., Mikat, R. P., Porcari, J. P., 2006. Validity and reliability of the session RPE method for monitoring exercise training intensity. *American Journal of Sport Medicine* 18, 14-17.

Hershler, C., Milner, M., 1980. Angle-angle diagrams in above-knee amputee cerebral palsy gait. *American Journal of Physical Medicine* 59, 165-183.

Hoch, M. C., 2011. The effect of joint mobilization on functional outcomes associated with chronic ankle instability. PhD thesis, University of Kentucky, 1-158.

Hopkins, W. G., 2000. Measures of reliability in sports medicine and science. *Sports Medicine* 30,1-15

Ireland, M. L., Willson, J. D., Ballantyne, B. T., Davis, I. M., 2003. Hip strength in females with and without patellofemoral pain *Journal of Orthopaedic & Sports Physical Therapy* 33, 671-675.

Iwano, T., Kurosawa, H., Tokuyama, H., Hoshikawa, 1990. Roentgenographic and clinical findings of patellofemoral arthrosis. *Clinical Orthopaedics* 252, 190-7.

Kelso, J. A. S., 1984. Phase transitions and critical behavior in human bimanual coordination. *American Journal of Physiology* 246, R1000-R1004.

Kelso, J. A. S., 1995. *Dynamic Patterns: The Self-Organization of Brain and Behavior*. MIT Press, Cambridge, MA.

Kelso, J. A. S., Buchanan, J. J., Wallace, S. A., 1991. Order parameters for the neural organization of single, multijoint limb movement patterns. *Experimental Brain Research* 85, 432-444.

Kelso, J. A. S., Southard, D. L., Goodman, D., 1979. On the coordination of two-handed movements. *Journal of experimental psychology* 5, 229-238.

Khaund, R., Flynn, S. H., 2005. Iliotibial Band Syndrome: A common source of knee pain. *American Family Physician* 15, 1545-1550.

Kurz, M. J., Stergiou, N., 2002. Effect of normalization and phase angle calculation on continuous relative phase. *Journal of Biomechanics* 35, 369-374.

Lee, T. Q., Anzel, S. H., Bennett, K. A., Pang, D., Kim, W. C., 1994. The influence of fixed rotational deformities of the femur on the patellofemoral contact pressures in human cadver knees. *Clinical Orthopaedics and Related Research* 69-74.

Lee, T. Q., Morris, G., Csintalan, R. P., 2003. The influence of tibial and femoral rotation on patellofemoral contact area and pressure. *Journal of Orthopaedic and Sports Physical Therapy* 33, 686-693.

Lee, T. Q., Yang, B. Y., Sandusky, M. D., McMahon, P. J., 2001. The effects of tibial rotation on the patellofemoral joint: assesment of the changes in *in situ* strain in the peripatellar retinaculum and the patellofemoral contact pressures and areas. *Journal of Rehabilitation Research and Development* 38, 463-469.

Li, L., 2000. Stability landscapes of walking and running near gait transition speed. *Journal of Applied Biomechanics* 16, 428-435.

Li, L., Haddad, J. M., Hamill, J., 2005. Stability and variability may respond differently to changes in walking speed. *Human Movement Science* 24, 257-267.

Li, L., van den Bogert, E. C. H., Caldwell, G. E., van Emmerik, R. E., Hamill, J., 1999. Coordination patterns of walking and running at similar speed and stride frequency. *Human Movement Science* 18, 67-85.

Lorenz, E. N., 1963. Deterministic nonperiodic flow. *Journal of Atmospheric Science* 20, 130-141.

Mahan, R. P., 1991. Circular statistical methods:applicatoin in spatial and temporal performance analysis 1-38.

Mann, R. A., Hagy, J., 1980. Biomechanics of walking, running and sprinting. *The American Journal of Sports Medicine* 8, 345-350.

- Mason, J. J., Leszko, F., Johnson, T., Komistek, R. D., 2008. Patellofemoral joint forces. *Journal of Biomechanics* 41, 2337-2348.
- Maulder, P. S., 2011. What is the role of joint coupling variability and joint stiffness in lower limb injury? PhD thesis, Auckland University of Technology, 203
- McConnell, J., 1986. The management of chondromalacia patell: a long term solution. *Australian Journal of Physiotherapy* 32, 215-223.
- McGinley, J. L., Baker, R., Wolfe, R., Morris, M. E., 2008. The reliability of three-dimensional kinematic gait measurements: A systematic review. *Gait & Posture* 29, 360-369.
- Miller, R. H., Chang, R., Baird, J. L., van Emmerik, R. E. A., Hamill, J., 2010. Variability on kinematic coupling assessed by vector coding and continuous relative phase. *Journal of Biomechanics* 43, 2554-2560.
- Miller, R. H., Meardon, S. A., Derrick, T. R., Gillette, J. C., 2008. Continuous relative phase variability during an exhaustive run in runners with a history of iliotibial band syndrome. *Journal of Applied Biomechanics* 24, 262-270.
- Mizuno, Y., Kumagai, M., Mattessich, S. M., Elias, J. J., Ramrattan, N., Cosgarea, A. J., 2001. Q-angle influences tibiofemoral and patellofemoral kinematics. *Journal of Orthopaedic Research* 19, 834-840.
- Mullaney, M. J., McHugh, M. P., Johnson, C., Tyler, T. F., 2010. Reliability of shoulder range of motion comparing a goniometer to a digital level. *Physiotherapy Theory and Practice* 26, 327-333.
- Mullineaux, D. R., Barnes, C. A., Batterham, A. M., 1999. Assessment of bias in comparing measurements: A reliability example. *Measurement in Physical Education and Exercise Science* 3, 195-205.
- Mullineaux, D. R., Bartlett, R., Bennett, S., 2001. Research design and statistics in biomechanics and motor control. *Journal of Sports Sciences* 19, 739-760.
- Mullineaux, D. R., Dierks, T. A., Davis, I. S., 2006. Variability in coordination of leg kinematics in males and females during treadmill running. 36. *Proceedings of ACL injuries research retreat III: The gender bias*, Lexington, KY: Kentucky Sports Medicine Center.
- Mullineaux, D. R., McKeon, P. O., Kerrigan, D. C., Hartel, J., 2008. Methodical issues in quantifying coordination-variability. 420-424. Seoul National University, Seoul Korea: Korean Society of Sports Biomechanics and Department of Physical Education.

- Mullineaux, D. R., Uhl, T. L., 2010. Coordination-variability and kinematics of misses versus swishes of basketball free throws. *Journal of Sports Sciences* 28, 1017-1024.
- Nadeau, S., Gravel, D., Herbert, L. J., Arsenault, A. B., Lepage, Y., 1997. Gait study of patients with patellofemoral pain syndrome. *Gait and Posture* 5, 21-27.
- Nawaoczenski, D. A., Saltzman, C. L., Cook, T. M., 1998. The effect of foot structure on the three-dimensional kinematic coupling behavior of the leg and rear foot. *Physical Therapy* 78, 404-416.
- Ness, L. L., Field-Fote, E. C., 2009. Whole-body vibration improves walking function in individuals with spinal cord injury: A pilot study. *Gait & Posture* 30, 436-440.
- Newell, K. M., van Emmerik, R. E. A., Lee, D., Sprague, R. L., 1993. On postural variability and stability. *Gait and Posture* 1, 225-230.
- Noehren, B., Pohl, M. B., Sanchez, Z., Cunningham, T., Lattermann, C., 2011. Proximal and distal kinematics in female runners with patellofemoral pain. *Clinical Biomechanics* 27, 366-371.
- Nooijen, C. F. J., Hoeve, N. t., Field-Fote, E. C., 2009. Gait quality is improved by locomotor training in individuals with SCI regardless of training approach. *Journal of NeuroEngineering and Rehabilitation* 6, 1-8.
- Novacheck, T. F., 1998. The biomechanics of running. *Gait and Posture* 7, 77-95.
- Peters, B. T., Haddad, J. M., Heiderscheit, B. C., van Emmerik, R. E. A., Hamill, J., 2003. Limitations in the use and interpretation of continuous relative phase. *Journal of Biomechanics* 36, 271-274.
- Piva, S. R., Fitzgerald, G. K., Irrgang, J. J., Fritz, J. M., Wisniewski, S., McGinty, G. T., Childs, J. D., Domenech, M. A., Jones, S., Delitto, A., 2009. Associates of physical function and pain in patients with patellofemoral pain syndrome. *Archives of Physical Medicine and Rehabilitation* 90, 285-295.
- Piva, S. R., Gil, A., Moore, C., Fitzgerald, K., 2009. Responsiveness of the activities of daily living scale of the knee outcome survey and numeric pain rating scale in patients with patellofemoral pain. *Journal of Rehabilitation Medicine* 41, 129-135.
- Pohl, M. B., Buckley, J. G., 2008. Changes in foot and shank coupling due to alterations in foot strike pattern during running. *Clinical Biomechanics* 23, 334-341.

Pohl, M. B., Messenger, N., Buckley, J. G., 2007. Forefoot, rearfoot and shank coupling: effect of variations in speed and mode of gait. *Gait & Posture* 25, 295-302.

Pollard, C. D., Heiderscheit, B. C., van Emmerik, R. E. A., Hamill, J., 2005. Gender differences in lower extremity coupling variability during an unanticipated cutting maneuver. *Journal of Applied Biomechanics* 21, 143-152.

Post, A. A., Dafferthofer, A., Beek, P. J., 2000. Principal components in three-ball cascade juggling. *Biological Cybernetics* 82, 143-152.

Powers, C. M., 2003. The influence of altered lower-extremity kinematics on patellofemoral joint dysfunction: A theoretical perspective. *Journal of Orthopaedic & Sports Physical Therapy* 33, 639-646.

Powers, C. M., Heino, J. G., Rao, S., Perry, J., 1999. The influence of patellofemoral pain on lower limb loading during gait. *Clinical Biomechanics* 14, 722-728.

Powers, C. M., Landel, R., Sosnik, T., Kirby, J., Mengel, K., Cheney, A., Perry, J., 1997. The effects of patellar taping on stride characteristics and joint motion in subjects with patellofemoral pain. *Journal of Orthopaedic & Sports Physical Therapy* 26, 286-291.

Robins, M., Wheat, J., Irwin, G., Bartlett, R., 2006. The effect of shooting distance on movement variability in basketball. *Journal of Human Movement Studies* 50, 217-238.

Robinson, R. L., Nee, R. J., 2007. Analysis of hip strength in females seeking physical therapy treatment for unilateral patellofemoral pain syndrome. *The Journal of Orthopaedic and Sports Physical Therapy* 37, 232-8.

Rothman, K. J., 1990. No adjustments are needed for multiple comparisons. *Epidemiology* 1, 43-46.

Sainburg, R. L., Ghilardi, F., Poizner, H., Ghez, C., 1995. Control of limb dynamics in normal subjects and patients without proprioception. *Journal of Neurophysiology* 73, 820-835.

Sanchis-Alfonso, V., Rosello-Sastre, E., Monteagudo-Castro, C., Esquardo, J., 1998. Quantitative analysis of nerve changes in lateral retinaculum in patients with isolated symptomatic patellofemoral malalignment: a preliminary study. *American Journal of Sports Medicine* 26, 703-709.

Scholz, J. P., 1990. Dynamic Pattern Theory--Some Implication for therapeutics. *Physical Therapy* 70, 827-843.

Schoner, G., Jiang, W. Y., Kelso, J., 1990. A synergistic theory of quadrupedal gaits and gait transitions. *Journal of Theoretical Biology* 142, 359-391.

Schwartz, M. H., Trost, J. P., Werver, R. A., 2004. Measurement and management of errors in quantitative gait data. *Gait and Posture* 20, 196-203.

Sidaway, B., Heise, G., Schoenfelder-Zohdi, 1995. Quantifying the variability of angle-angle plots. *Journal of Human Movement Studies* 29, 181-197.

Söderkvist, I., Wedin, P. A., 1993. Determining the movements of the skeleton using well-configured markers. *Journal of Biomechanics* 26, 1473-1477.

Souza, R. B., Powers, C. M., 2009. Differences in hip kinematics, muscle strength and muscle activation between subjects with and without patellofemoral pain. *Journal of Orthopaedic & Sports Physical Therapy* 39, 12-19

Sparrow, W. A., Donovan, E., van Emmerik, R. E. A., Barry, E. B., 1987. Using relative motion plots to measure changes in intra-limb and inter-limb coordination. *Journal of Motor Behavior* 19, 115-129.

Stergiou, N., Harbourne, R. T., Cavanaugh, J. T., 2006. Optimal movement variability: A new theoretical perspective for neurologic physical therapy. *Journal of Neurologic Physical Therapy* 30, 120-129.

Taunton, J. E., Ryan, M. B., Clement, D. B., McKenzie, D. C., Lloyd-Smith, D. R., Zumbo, B. D., 2002. A retrospective case-control analysis of 2002 running injuries. *British Journal of Sports Medicine* 36, 95-101.

Tepavac, D., Field-Fote, E. C., 2001. Vector coding: A technique for quantification of intersegmental coupling in multicyclic behaviors. *Journal of Applied Biomechanics* 17, 259-270.

Thomee, R., Augustsson, J., Karlsson, J., 1999. Patellofemoral pain syndrome: A review of current issues. *Sports Medicine* 28, 245-262.

Tiberio, D., 1987. The effect of excessive subtalar joint pronation on patellofemoral joint mechanics: a theoretical model. *Journal of Orthopaedic & Sports Physical Therapy* 9, 160-169.

Tuller, B., Turvey, M. T., Fitch, H. L., 1982. The Bernstein Perspective: II. The concept of muscle linkage of coordinative structure. (edited by Kelso, J.A.S.), 253-270. Lawrence Erlbaum Associates, Hillsdale, NJ.

- Turvey, M. T., 1990. Coordination. *American Psychologist* 45, 938-953.
- Turvey, M. T., Fitch, H. L., Tuller, B., 1982. The Bernstein Perspective: I. The problems of degree of freedom and context-conditioned variability. (edited by Kelso, J.A.S.), 239-251. Lawrence Erlbaum Associates, Hillsdale, NJ.
- van Emmerik, R. E., Wagenaar, R. C., 1996. Effects of walking velocity on relative phase dynamics in the trunk in human walking. *Journal of Biomechanics* 29, 1175-1184.
- Van Emmerik, R. E., Wagenaar, R. C., Winogrodzka, A., Wolters, E. C., 1999. Identification of axial rigidity during locomotion in Parkinson's disease. *Archives of Physical Medicine and Rehabilitation* 80, 186-191.
- van Emmerik, R. E. A., Hamill, J., McDermott, W. J., 2005. Variability and coordinative function in human gait. *Quest* 57, 102-123.
- van Emmerik, R. E. A., Rosenstein, M. T., McDermott, W. J., Hamill, J., 2004. A nonlinear dynamics approach to human movement. *Journal of Applied Biomechanics* 20, 396-420.
- Whaley, M., Brubaker, P., Kaminski, L., Miller, C., 1997. Validity of rating of perceived exertion during graded exercise testing in apparently healthy adults and cardiac patients. *Journal of Cardiopulmonary Rehabilitation* 17, 261-267.
- Wheat, J., Glazier, P., 2006. Measuring coordination and variability in coordination. (edited by Davids, K., Bennett, S., Newell, K.M.), 167-181. Human Kinetics, Champaign, IL.
- Wheat, J., Mullineaux, D. R., Bartlett, R. M., Milner, C. E., 2002. Quantifying variability in coordination during running. 519-522. Caceres, Spain: University of Extremadura.
- Wheat, J. S., Baltzopoulos, V., Milner, C. E., 2005. Coordination variability during overground, treadmill and treadmill-on-demand running. 781-784. ISBS, Beijing, China.
- Whiting, W. C., Zernicke, R. F., 1982. Correlation of movement patterns via pattern recognition. *Journal of Motor Behavior* 14, 135-142.
- Williams, K. R., 1985. Biomechanics of running. *Exercise and Sport Sciences Reviews* 13, 389-441.
- Williamson, A., Hoggart, B., 2005. Pain: a review of three commonly used pain rating scales. *Journal of Clinical Nursing* 14, 798-804.

Willson, J. D., Davis, I. S., 2008. Lower extremity mechanics of females with and without patellofemoral pain across activities with progressively greater task demands. *Clinical Biomechanics* 23, 203-211.

Wilson, C., Simpson, S., Hamill, J., 2009. Movement coordination patterns in triple jump training drills. *Journal of Sports Sciences* 27, 277-282.

Wilson, C., Simpson, S. E., van Emmerik, R. E., Hamill, J., 2008. Coordination variability and skill development in expert triple jumpers. *Sports Biomechanics* 7, 2-9.

Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., Whittle, M., D'Lima, D. D., Cristofolini, L., Witte, H., Schmid, O., Stokes, I., 2002. ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion-part I: ankle, hip and spine. *Journal of Biomechanics* 35, 543-548.

Vita

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General Information

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Education

- | | |
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Exercise Science: Biomechanics Doctoral Program |
| 2005-2007 | <u>The University of Kentucky</u> , College of Education,
Master of Science, Exercise Science: Biomechanics
Thesis: Three-dimensional quantitative analysis of
the trajectory of the foot while running |
| 2001-2005 | <u>The University of Kentucky</u> , College of Engineering
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Professional Positions

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| 2011-Present | Biomechanical Engineer
ERMI Medical Devices
Atlanta, GA |
| 2006-Present | President
TC Motions Inc.
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| 2005-2009 | Research Assistant
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Publications

Peer-Reviewed Manuscripts

Noehren, B., Pohl, M. B., Sanchez, Z., Cunningham, T., Lattermann, C., 2011. Proximal and distal kinematics in female runners with patellofemoral pain. *Clinical Biomechanics* 27, 366-371.

Kibler, W., Sciascia, A., Uhl, T. L., Tambay, N., Cunningham, T., 2008. Electromyographic analysis of specific exercises for scapular control in early phases of shoulder rehabilitation. *American Journal of Sports Medicine* 36, 1789-1798.

Patents

Cunningham, T., 2011. Patent 8043195 B2, Exercise Apparatus. TC Motions Inc.

Cunningham, T., 2011. Patent 7744507, Exercise Apparatus. TC Motions Inc.

Oral Presentations

Cunningham, T., Branch, T., Stinton, S., Mills, T., Albahar, A., Jacobs, C., 2012. A robotic in vivo comparison of 3-dimensional patellar kinematics between healthy volunteers and those with patellofemoral pain syndrome using dynamic computed tomography. Orthopaedic Research Society Annual Meeting, San Francisco, CA.

Cunningham, T., Mullineaux, D. R., Stinton, S., 2009. Intra-limb joint coupling patterns during the use of three lower extremity exercise machines. 368-371. Limerick, Ireland.

Cunningham, T., Butterfield, T., 2009. A device to quantify cyclic compressive loads applied to soft tissue for in-vivo animal models. State College, PA.

Cunningham, T., Uhl, T. L., Shapiro, R., Mattacola, C., 2008. Three-dimensional quantitative analysis of the trajectory of the foot while running. Lexington, KY.

Poster Presentations

Capilouto, G., T. Butterfield, Cunningham, T. 2011. Effects of non-oral feeding on infant tongue muscle mass and force. Joint conference 7th Kentucky innovation and entrepreneurship conference and 16th Kentucky EPSCoR, Louisville, KY.

Abshire, S., Dupont-Versteegden, E., Kelly, D., Cunningham, T. J., Butterfield, T. A., 2009. Lymphangiogenesis is dependent upon the magnitude of cyclic compressive load: Implications for manual lymphatic drainage. Baltimore, MD.

Kibler, W., Uhl, T. L., Cunningham, T., 2009. The effect of the scapular assistance test on scapular kinematics in the clinical exam. Journal of Orthopaedic & Sports Physical Therapy 39, A12.

Spigelman, T., Cunningham, T., Mair, S., Shapiro, R., Uhl, T. L., Mullineaux, D. R., 2009. Motor control patterns in elite swimmers' freestyle stroke during dryland swimming. Limerick, Ireland.

Uhl, T. L., Cunningham, T., Kibler, W., 2009. Kinematic and neuromuscular actions during the scapular retraction test (SRT). Journal of Orthopaedic & Sports Physical Therapy 39, A12.

Cunningham, T., Uhl, T. L., Shapiro, R., Mattacola, C., 2008. Three-dimensional analysis of the trajectory of the ankle while running. North American Congress of Biomechanics, Ann Arbor, MI.

Spigelman, T., Uhl, T. L., Mullineaux, D. R., Cunningham, T., Mair, S., Shapiro, R., 2008. Effect of ability on freestyle swimbench stroke characteristics. Ann Arbor, MI.