**Title:** Coupling Angle Variability in Healthy and Patellofemoral Pain Runners

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**Abstract**

*Background:* Patellofemoral pain is hypothesized to result in less joint coordination variability. The ability to relate coordination variability to patellofemoral pain pathology could have many clinical uses; however, evidence to support its clinical application is lacking. The aim was to determine if vector coding's coupling angle variability, as a measure of joint coordination variability, was less for runners with patellofemoral pain than healthy controls as is commonly postulated.

*Methods:* Nineteen female recreational runners with patellofemoral pain and eleven healthy controls performed a treadmill acclimation protocol then ran at a self-selected pace for fifteen minutes. 3-D kinematics, force plate kinetics, knee pain and rating of perceived exertion were recorded each minute. Data were selected for the: pain group at the highest pain reached (pain≥3/10) in a non-exerted state (exertion<14/20), and; non-exerted healthy group from the eleventh minute. Coupling angle variability was calculated over several portions of the stride for six knee-ankle combinations during five non-consecutive strides.

*Findings:* 46 of 48 coupling angle variability measures were greater for the pain group, with 7 significantly greater ($p<.05$).

*Interpretation:* These findings oppose the theory that less coupling angle variability is indicative of a pathological coordinate state during running. Greater coupling angle variability may be characteristic of patellofemoral pain in female treadmill running when a larger threshold of pain is reached than previously observed. A predictable and directional response of coupling angle variability measures in relation to knee pathology is not yet clear and requires further investigation prior to considerations for clinical utility.

## Introduction

Variability in joint or limb segment coordination has been suggested to be inherent within a healthy motor control strategy (Newell, et al. 1993, Stergiou, et al. 2006). A commonly held interpretation of a dynamical system's application to lower extremity orthopaedic injuries theorizes that a low amount of variation in joint or limb segment coordinative structure may increase the frequency of loading of soft tissue and eventually lead to an overuse condition and pathological state (Hamill, et al. 1999). Patellofemoral Pain (PFP) is theorized to be a condition resultant of this decrease in variability (Hamill, et al. 1999). When originally testing this theory, coordination variability between limb segments was determined using the analysis technique of continuous relative phase (Kelso 1995); however, this technique has limitations in quantifying non-sinusoidal couplings and is not appropriate for most lower extremity couplings during gait (Peters, 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 CAV relation to pathology, Heiderscheit et al.(Heiderscheit, et al. 2002) compared mean CAV values over the entire stride cycle for several lower extremity 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 the mean CAV 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 variable 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, 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, et al. 2002) and a heterogeneous injured population (Ferber, et al. 2005) were cited as possible factors for the limited results.

Previous literature studying joint kinematics of runners with PFP has consistently used a minimum pain level of 3/10 on a numeric pain rating scale 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 change of at least 2 has been recognized as a clinically meaningful change in pain (Crossley, et al. 2004). 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). These limitations should be addressed to further assess the validity of CAV as a clinically useful measure for coordination variability in gait.

It has been suggested that PFP is a condition resulting from a pathological coordinate state which is characterized by a lower amount of coordination variability

than in a healthy population (Hamill, et al. 1999). CAV has been used to test this theory but there is little evidence to suggest that CAV is less in a pathological state regardless of construct. This study aims to address identified limitations of previous literature and determine if CAV measures are less for a population with PFP than a healthy population during running at a self-selected pace; an activity related to development of PFP (Davis and Powers 2010). It was hypothesized that CAV values would be less in individuals with PFP.

## Methods

Twenty-one healthy (Age 25.3(4.0) yrs., Ht. 1.68(0.08) m, Wt. 60.3(7.12) kg)) and twenty injured (Age 25.8(6.0) yrs., Ht. 1.63(0.07) m, Wt. 57.0(6.35) kg) 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 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 using a numeric pain rating scale (Farrar, et al. 2001) 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, Illiotibial band syndrome or meniscal pathology. Potential subjects were excluded if they had a stated neurological disorder or tape allergy. Written informed consent was obtained prior to participation in the study, which was approved by the institute's 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, USA) 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, USA). Subjects wore standardized shoes (ZoomAir; Nike, Beaverton, OR, USA) 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 were captured using a combination of 15 Eagle and Eagle4 cameras at 300 Hz (Motion Analysis Corporation, Santa Rosa CA, USA). A dual belted treadmill instrumented with a force plate under each belt (TM-09-PBertec, Columbus, OH, USA) 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 for reporting level of perceived fatigue during the run. Perceived pain during the run was collected using a verbally administered numeric pain rating scale 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 position (Figure 1 Top). 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 10s of each minute interval. RPE and pain measures were recorded by investigators immediately following each 10s data acquisition.

*Data Processing*

Kinematic markers were identified using Cortex 2.0 software (Motion Analysis Corporation, Santa Rosa CA, USA). Three-dimensional marker coordinates and force plate data were exported to Matlab v2009a (Mathworks, Natick MA, USA) 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. 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, and mid-

stance as 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 fourth-order cubic spline function making a 250 point time normalized gait cycle (1 point=0.4% of stride). 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*

One 10s 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 to limit potential effects of exertion within the same RPE level. The average time period of analysis for the PFP group was the eleventh minute of running at a self-selected pace; therefore, healthy data were also analyzed from the eleventh minute for those with a RPE value of less than 14. Two subjects were excluded for missing foot markers and nine did not meet pain or fatigue inclusion criteria.

CAV values were determined using a revised vector coding technique (Heidercheit 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 1) at each point in the gait cycle. The injured limb was analyzed for the PFP group and a limb 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 similar to previous methods (Heiderscheit, et al. 2002) (Table 1). Mean CAV values ($CAV_{Mean}$) were calculated for quintiles (Q), stance, swing and the entirety of stride for each subject and then the mean and SD (mean(SD)) calculated for each group.

*Statistical Analysis*

Independent t-tests were performed to note any differences between population demographics (height, mass, age and average distance run per week), pain, RPE, running speed and all CAV measures. Statistical significance was set *a priori* ($\alpha$<.05) with no correction for multiple comparisons made (Rothman 1990). Effect sizes of the difference in means divided by the pooled SD for each measure were calculated (Cohen, 1998). A Shapiro-Wilk test was used to confirm that all variables were normally distributed ($\alpha$>.05).

**Results**

There were 19 PFP (Age 25.8(6.1) yrs., Ht. 1.63(0.07) m, Wt. 57.1(6.48) kg) and 11 healthy (Age 26.5(13.4) yrs., Ht. 1.66(0.09) m, Wt. 58.0(5.33) kg) female subjects who qualified for analysis. Reported distance run per week was greater for healthy (37.7(13.4) km) than PFP (21.2(9.4) km) (p=0.0008). 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(0.13) m/s) than the PFP population (2.54(0.24) m/s) (p<.0002). Pain values were 4.3(1.3) for the PFP group. RPE levels for the healthy group (12.2(0.9)) and the PFP group (12.4(0.8)) were not significantly different (p=0.41).

$CAV_{Mean}$ were found to be greater in the PFP group compared to the healthy group for 46 of the 48 discrete measures (Figure 2) with only 7 being larger and significantly different (Table 2). Effect sizes (Cohen's d) were reported for each measure. Continuous ensemble averages of the CAV over the entire stride as measured from heel strike for each population are shown with quintiles highlighted (Figure 3). PFP CAV (solid line) were generally greater or the same throughout most portions of stride with few exceptions. There was a brief period in Q1 of KF-AF where the ensemble CAV was larger for the healthy population despite the corresponding $CAV_{Mean}$ measure for the entire period being significantly less.

**Discussion**

The hypothesis that CAV values would be less in individuals in PFP was not supported. Surprisingly, the only statistically significant differences observed showed greater CAV values in PFP than healthy individuals. These findings are contrary to the dynamical systems perspective to lower extremity overuse injuries that suggest lower CAV is indicative of a pathological coordinate state (Hamill, et al. 1999, Heiderscheit, et al. 2002). Previous literature using similar analysis procedures for all $CAV_{Mean}$ intervals in the KR-AI, KF-AI and KF-AF couplings have shown no differences in any $CAV_{Mean}$ values in a PFP population that had less pain (Heiderscheit, et al. 2002). Increases in $CAV_{Mean}$ 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 have preceded the development of PFP. 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 previously observed (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. The nature of this investigation unfortunately cannot determine if the increase in CAV is the result of pathology or precedes development which limits this interpretation (Bartlett, et al. 2007).

This is the first study to document significantly increased CAV for a pathological population to the best of the authors' knowledge, as such; clinical interpretation should be viewed with caution. It is plausible that increases in CAV may just be a result of mathematical artifact as a result of the simple statistical methods employed or a clustering of data capture points in regions where little joint motion occurs; typical near heel-strike (Heiderscheit, et al. 2002). Chosen locations of quintiles used to create

discrete measures from clearly continuous and somewhat volatile CAV curves may have affected the results. For example, if quintiles were chosen to begin at heel strike rather than encompass this event, the brief increase observed within Q1 of KF-AF (Figure 3) for the healthy population may have been found to be significantly greater if located in a separate quintile than the PFP local maxima. On the other hand, this would likely have still resulted in significantly greater variability for the PFP group after heel strike. It is, however; difficult to ignore that not only were there no $CAV_{Mean}$ values that were significantly less in PFP; 85% of the comparisons were observed to be larger in PFP although most of these observations (77%) were only slightly greater and statistically negligible. It may be more appropriate to shift focus to the preponderance of evidence observed in this study that shows no statistical differences between populations. The small differences observed seem to agree with, rather than contrast previous findings. In the first proposal of the dynamical systems perspective to overuse injuries, there is no statistical evidence to support that a PFP population should have less variability than healthy counterparts. Of the discrete variables analyzed in that study, 41% were greater in the PFP population with the largest reported difference actually being four times greater in the PFP population than healthy population (Hamill, et al. 1999). Interpretations or explanations of these larger observed values were not discussed. Of the four CAV measures Ferber et al. used to compare symptomatic runners to controls, none were statistically different (p ranges 0.96 to 0.67) and mean differences between populations only ranged between 0.08° and 2.22°, respectively (Ferber, et al. 2005). Observation of significant differences in CAV measures between PFP and healthy populations seem to be a rarity rather than the norm. The small

amount of evidence to support any particular direction for CAV measures in this construct even seem to conflict. These seemingly conflicting results may, however; coincide with the perspective that there is an optimal amount of variability in running gait; where extreme amounts, too much or too little, are detrimental to a biological system(Stergiou, et al. 2006) and can lead to an overuse condition in the lower extremity.

Slight methodical differences and dependent measures among investigations may have led to different results or a lack of significant findings in support of previous theory. For example, only intralimb knee-ankle joint couplings were analyzed in this study from multitudes of possible segment combinations making direct comparison to previous literature difficult. Further, regions chosen for this analysis are thought to be critically important in the study of movement variability, consistent with previous literature (Clark and Phillips 1993) and accompanied with relative variability increases (Sainburg, et al. 1995), particularly near heel-strike (Heiderscheit, et al. 2002), however; there are other possible CAV measures that may serve as alternative measures than those presented here. Participants wearing their own shoes may have influenced the results, although the experimental control of using standardized shoes that allowed for the application of markers directly to the foot was preferred to provide a better measure of distal segment movement and improve comparisons across our groups (Noehren, et al. 2011). This study surprisingly observed results contrary to a commonly held theory but also, unfortunately, did not observe as many differences as anticipated. To the knowledge of these authors, the clinical precision limits of CAV measures have not yet been defined to evaluate the clinical utility of CAV for linear analyses (Mullaney, et al.

2010). The use of nonlinear statistical methods to analyze these nonlinear dynamical systems may be more appropriate (Stergiou, et al. 2006) and has not been explored within the gait literature using CAV as a measure of interest. Further exploration of CAV measures utilizing nonlinear analysis methods may aid in our clinical interpretation and understanding of the relationship between variability and pathology in the construct of gait and these reported findings.

## Conclusion

Recent debate has arisen to clarify differences in findings for similar analysis methods when interpreting PFP development during prolonged running in the context of dynamical systems (Dierks 2011, Li 2011). The proposed etiology that PFP symptoms are a manifest of less joint coordination variability and observable by CAV measures requires more scrutiny. Although theoretically sound, there is little supporting evidence to suggest less movement variability is indicative of overuse pathology as it relates to running. This can also be said for more movement variability. The clinical utility and applicability of CAV in running analysis is not yet understood or necessarily supported. Future research should concentrate on thoroughly exploring the capability of CAV as a clinically useful measure prior to further interpretation.

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**Table 1** Common abbreviations and definitions used within the text and tables grouped by Knee-Ankle coupling relationship and coupling angle variability (CAV) intervals.

| 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 over discrete intervals (Q, 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) |
| Stance | 0 to 40% |
| Swing | 40 to100% |
| Stride | 0 to 100% |

**Table 2** Significant differences observed for mean Coupling Angle Variability values (mean (SD)) within quintiles (Q1-5) of stride, the entirety of stride or stance phase for runners with Patellofemoral Pain (PFP) and healthy controls. Couplings include: Knee (K) Flexion (F), Rotation (R) and Valgus (V) – Ankle (A): Flexion (F) and Inversion (I).

| Interval | Coupling | PFP (°),n=19 | Healthy (°),n=11 | *P* value | Effect Size (Cohen's d) |
|---|---|---|---|---|---|
| Q1 | KF-AF | 7.9(2.0) | 6.1(1.8) | .020 | .97 |
| Q2 | KR-AI | 16.0(8.9) | 10.1(4.0) | .049 | .80 |
| Q2 | KR-AF | 10.3(4.6) | 7.0(2.5) | .038 | .85 |
| Q4 | KV-AF | 10.6(5.0) | 6.2(1.9) | .010 | 1.09 |
| Q5 | KV-AI | 23.5(9.6) | 14.6(5.0) | .008 | 1.12 |
| Stance | KV-AF | 6.9(2.4) | 4.5(1.5) | .008 | 1.21 |
| Stride | KV-AI | 14.8(4.5) | 11.6(2.2) | .031 | .89 |

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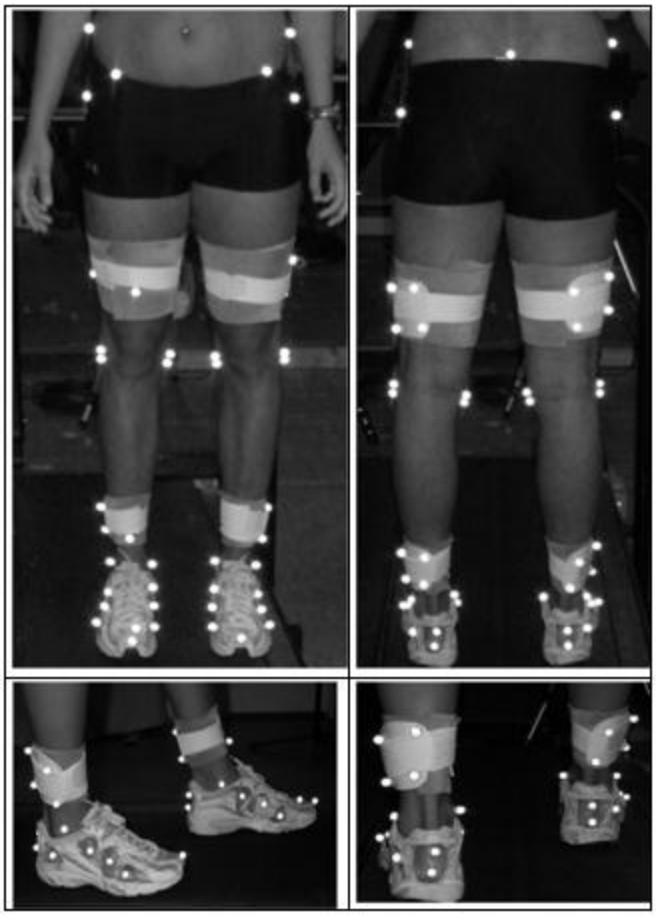
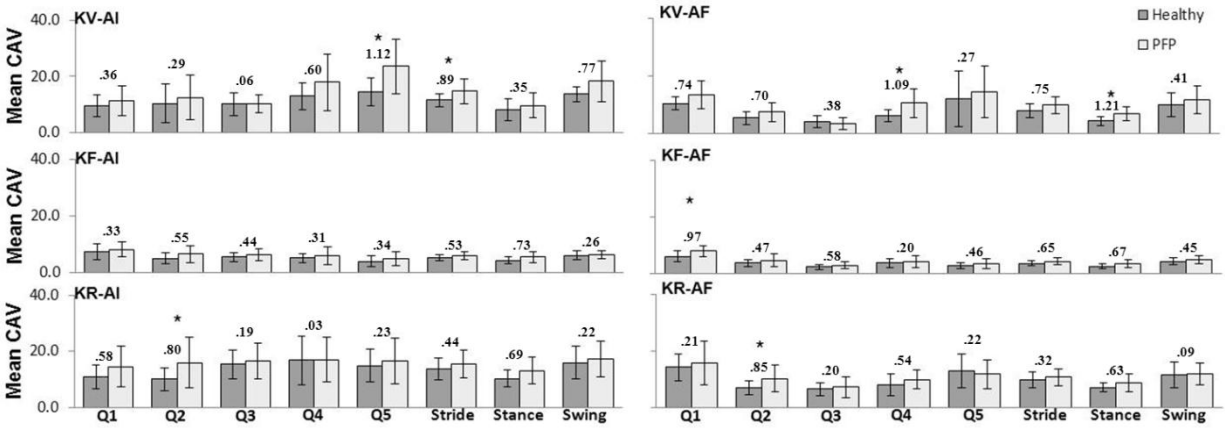
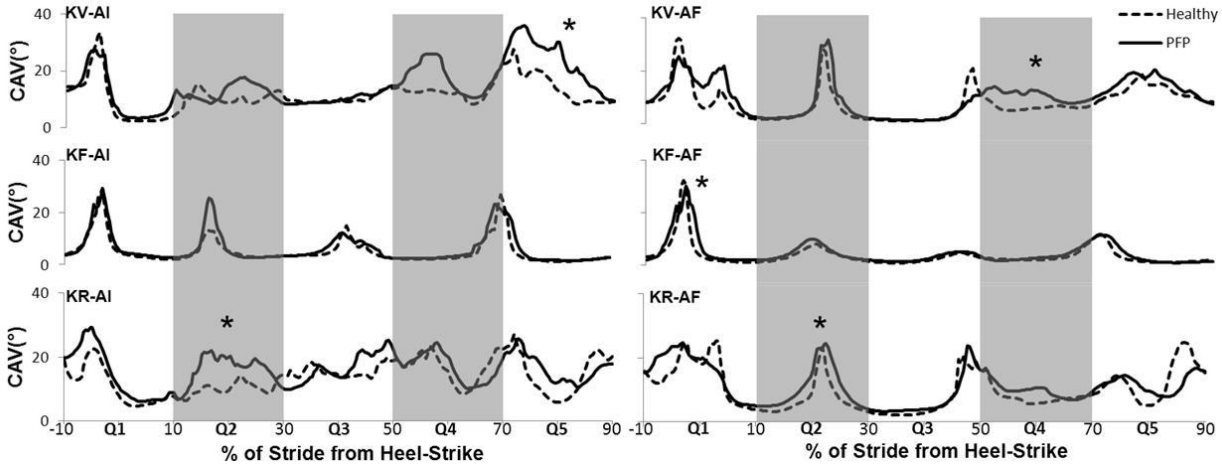
**Figure 1** Markerset used during a static calibration in anatomical position (Top). Only bilateral markers on the lateral aspects of the 5th metacarpal head, base, navicular and both the lateral and medial aspects of the calcaneus were used to model foot movement. Windows are cut out of the shoes allowing markers to be adhered directly to the foot (Bottom). Rigid clusters were secured to the distal posterior-lateral aspects of each segment to model thigh and shank movement.

**Figure 2** Discrete mean Coupling Angle Variability (CAV, mean(SD)) 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 for female runners with patellofemoral pain (PFP) and healthy controls. Significant difference between populations denoted at $P<0.05$ (*) and effect size (Cohen's d) is reported for each measure. Couplings include: Knee (K) Flexion (F), Rotation (R) and Valgus (V) – Ankle (A): Flexion (F) and Inversion (I).

**Figure 3** Continuous ensemble averaged Coupling Angle Variability (CAV) curves for Healthy and Patellofemoral Pain (PFP) populations for six Knee-Ankle coupling combinations. Even quintiles of stride are highlighted starting at Heel-Strike (0%). Significant differences between populations at $P<.05$ for quintiles are indicated (*). Couplings include: Knee (K) Flexion (F), Rotation (R) and Valgus (V) – Ankle (A): Flexion (F) and Inversion (I).




KV-AI
KV-AF
KF-AI
KF-AF
KR-AI
KR-AF
Healthy
PFP
Mean CAV
40.0
20.0
0.0
.36
.29
.06
.60
1.12
.89
.35
.77
.74
.70
.38
1.09
.27
.75
1.21
.41
.33
.55
.44
.31
.34
.53
.73
.26
.97
.47
.58
.20
.46
.65
.67
.45
.58
.80
.19
.03
.23
.44
.69
.22
.21
.85
.20
.54
.22
.32
.63
.09
Q1
Q2
Q3
Q4
Q5
Stride
Stance
Swing


KV-AI
KV-AF
KF-AI
KF-AF
KR-AI
KR-AF
CAV(°)
40
20
0
*
Healthy
PFP
-10
Q1
10
Q2
30
Q3
50
Q4
70
Q5
90
% of Stride from Heel-Strike

1