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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 15 min. 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 $\geq$ 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.

1. 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 orthopedic 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. (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

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observed. Minimal pain values reached (Heiderscheit et al., 2002) and heterogeneous injured populations (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., 2008, 2011; 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. (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 affect 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.

2. 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 and 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, iliotibial 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 (Fig. 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 that are 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 (RPE) scale (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).

2.1. Treadmill protocol

A one second standing static calibration file was captured while the subjects stood in the anatomical position (Fig. 1 Top). Subjects then walked on a single belt of the treadmill for 3 min at 1.3 m/s to acclimate themselves to the treadmill. Speed was then increased for 3 min to a warm-up pace (2.2–2.3 m/s) followed by 2 min 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 min 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 s of each minute interval. RPE and pain measures were recorded by investigators immediately following each 10 s data acquisition.

2.2. 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 the last gait cycle from each 10 s trial was discarded to reduce interpolation effects and the first 10 gait cycles were kept for analysis.

2.3. 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 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