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New Concepts

A dynamical systems approach to lower extremity running injuries

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Abstract

In this paper, we are presenting an alternative approach to the investigation of lower extremity coupling referred to as a dynamical systems approach. In this approach, we calculate the phase angle of each segment and joint angle. Pairing the key segment/joint motions, we use phase angles to determine the continuous relative phase and the variability of the continuous relative phase. Data from two studies illustrate the efficacy of the dynamical systems approach. Individuals who were asymptomatic, even though they may have anatomical aberrant structural problems (i.e. high Q-angle vs low Q-angle) showed no differences in the pattern of the continuous relative phase or in the variability of the continuous phase. However, differences in the variability of the continuous relative phase were apparent in comparing individuals who were symptomatic with patellofemoral pain with non-injured individuals. Patellofemoral pain individuals showed less variability in the continuous relative phase of the lower extremity couplings than did the healthy subjects. We hypothesize that the lower variability of the couplings in the symptomatic individuals indicates repeatable joint actions within a very narrow range.

Relevance

We claim that the traditional view of the variability of disordered movement is not tenable and suggest that there is a functional role for variability in lower extremity segment coupling during locomotion. While the methods described in this paper cannot determine a cause of the injury, they may be useful in the detection and treatment of running injuries.

Keywords: Injury; Lower extremity; Coupling; Dynamical systems; Continuous relative phase; Variability

1. Introduction

Clinical studies [1,2] reported that the most prevalent site of running injury was the knee. In these studies, knee injuries accounted for over 25% of all running injuries. The etiology for knee injuries in running has not been reported but excessive rearfoot pronation is quite often associated with knee injuries [3]. However, the coupling of excessive rearfoot motion and knee mechanics remains unknown. In fact, there is no clear clinical definition of excessive rearfoot pronation. A mechanism for knee injuries has been alluded to in the biomechanics literature but, to date, there has been little evidence linking the mechanics of the lower extremity to knee injuries.

Bates et al. [3] suggested that injury to the knee may be the result of a disruption of the timing of the normal events in closed chain pronation. The mechanism that was suggested related to the timing of the subtalar and knee joint actions. In the normal situation, they suggested that the maxima of rearfoot eversion, internal tibial rotation and knee flexion should all occur at the same instant in time. It was suggested, therefore, that if the subtalar joint continued to pronate while the knee began to extend, or if the subtalar joint began to resupinate while the knee continued to flex, timing discrepancies between the joint actions would occur. It would appear, therefore, that the tibia would undergo antagonistic counter rotations at the proximal and distal ends. This would lead to excessive stress at the knee joint and, over many running cycles, may lead to knee injury.

The majority of studies in the biomechanics literature that investigated lower extremity actions have reported on the kinematics of individual lower extremity joints rather than addressing the interaction between the joints [3,4]. Far fewer studies, however, have investigated the coupling of the subtalar joint and the knee joint during running. Those studies that have investigated lower

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extremity joint coupling did so by investigating the relative timing of the joint actions [5–8] or by reporting the calcaneal eversion/tibial internal rotation ratio [9,10].

Bates et al. [5,6] reported no significant difference between the time of peak knee flexion and peak calcaneal eversion in runners with normal mechanics and those who over-pronated. However, the mean rearfoot angle for the group that over-pronated was 11°, a value well within the 8–15° range considered to be normal [11].

The subtalar joint is thought to act as a mitre joint with the axis of the joint inclined in the sagittal plane to approximately 45° [12]. In theory, therefore, equal amounts of calcaneal eversion should result in equal amounts of internal tibial rotation during pronation. This would result in a ratio of 1.0 between the excursions of calcaneal eversion and internal tibial rotation [10]. An increase or decrease in the subtalar joint axis, respectively, would decrease or increase this ratio. Nigg et al. [9] investigated the relationship between arch structure and the calcaneal eversion/internal tibial rotation ratio. They reported that calcaneal eversion was no different in the high and low arched groups but internal tibial rotation was less in the low arched group thus the low arched group had greater ratios.

McClay and Manal [10] used the same calcaneal eversion/internal tibial rotation ratio to investigate lower extremity coupling parameters in runners with normal rearfoot mechanics and in runners with excessive pronation. They reported no statistically significant differences between the groups in the timing between the maximum angles of the knee and the rearfoot. While not statistically significant, the values, however, were more closely timed in the normal group. The calcaneal eversion/internal tibial rotation ratio was significantly less in the excessive pronation group due to the greater internal tibial rotation. Most importantly, linear regression analyses revealed significant relationships among calcaneal eversion, internal tibial rotation and internal knee rotation.

It is clear that the actions of the lower extremity are coupled and it is most likely that perturbations to the system can result in injury, particularly to the knee. It is our thesis that the coupling relationships have not been clarified using traditional spatial constructs. A different approach to spatial angles and timing relationships is one that involves dynamical systems. This approach is not new and has been used before in biomechanical studies [13,14] but it has not been previously used to investigate orthopaedic injuries to runners. In this paper we will demonstrate the use of a dynamical systems approach to investigate the coupling relationships in the lower extremity.

2. The dynamical systems approach

In a biomechanical system, the high number of available degrees of freedom is reduced through the formation of coordinative structures. Coordinative structures can be defined as muscle synergies, often spanning several joints, that are functionally linked to satisfy the task demands [15]. Coordinative structures enable the organism to achieve (a) the same goal by using different degrees of freedom (e.g. muscles, joints) and (b) use the same degrees of freedom to reach different movement goals. The concept of coordinative structures has been derived from the important work of Bernstein [16] in the area of movement coordination and control.

Kelso et al. [17–19] developed a dynamical systems approach that has been particularly fruitful to study movement coordination and stability features of coordinative structures. The approach developed by Kelso and his colleagues offers tools to identify the nature of transition processes and stability in human movement. From this perspective, an essential element in transition processes is the degree of variability in the coordination dynamics. In more traditional (engineering) viewpoints variability in patterns are regarded as noise and to be eliminated. Within dynamical systems approaches, variability can be essential in inducing a coordination change and to establish a combination of stability and flexibility of movement.

Within the dynamical systems approach, instabilities can be used to demarcate different movement patterns [18]. Different qualitative patterns are specified by collective variables or order parameters. These order parameters identify low-dimensional qualitative states of the system dynamics. The full identification of the order parameter dynamics can be obtained through manipulation of a control parameter.

Kelso [20] and Haken et al. [17] consider phase relations or the relative phase between body segments as a potential order parameter. Relative phase between component oscillators (arms, legs, etc.) can identify different qualitative states of the system dynamics (for example, galloping and trotting as different gaits in quadrapedal locomotion) on which basis changes in coordination patterns can be evaluated. In this example, locomotor speed can act as a control parameter. Transitions or changes in movement patterns can both be evaluated through continuous and discrete measures of relative phase [18,21].

In the *continuous relative phase* measure the relative phase throughout the entire movement cycle is obtained. This is done by plotting the position signal of one segment or joint angle versus the angular velocity of that segment or joint in the so-called 'phase-plane' (see Fig. 1). After different normalization procedures the phase angle in this phase-plane is obtained and the relative phase is the difference between the phase angles of the two segments or joints. In the discrete relative phase measure, the difference in timing of two segments or joints is calculated (for example when both reach

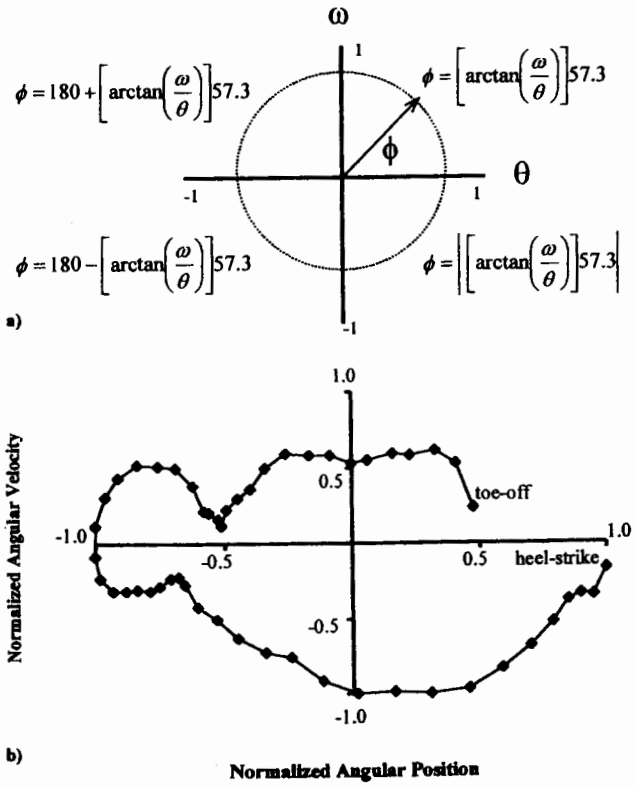

Fig. 1. (a) Phase angle (ϕ) definition based on a phase plot of normalized spatial angle (θ) and normalized angular velocity (ω). The resulting phase angle range is 0–180°. (b) A typical normalized phase angle of a lower extremity coupling.

maximum values) and divided by the cycle time of one of the segments or joints. In sinusoidal type movements, as in the above example of finger motion, these two relative phase measures should and do provide similar information regarding coordination changes. However, in more complex coordinative patterns as in locomotion these measures could provide different sources of information.

Firstly, the continuous relative phase measure includes both continuous spatial and temporal information. As such it can give higher dimensional information through a more detailed analysis of the behavior in state space. The *state space* is made up from the relative system variables, and these are often times unknown (in the example of Fig. 1, the state space is the phase-plane in which the angular position is plotted against angular velocity).

Secondly, previous research that has examined the coupling between joints or segments (such as the relation between calcaneal eversion and tibial rotation as described in the studies above) has only addressed coupling at single occurrences during the gait cycle, such as maximal internal or external tibial rotation. The absence of significant differences in coupling at these discrete moments between pathological and healthy subjects does not mean that during the stride cycle no differences exist. The continuous relative phase measure as described below will provide such a continuous measure of the coupling throughout the entire stride cycle. In addition, this measure can also provide information regarding the stability/flexibility of the coordination patterns.

Phase transitions can be characterized by abrupt jumps between different coordinative modes; these abrupt jumps can occur for very small changes in the control parameter. An important feature of these phase transitions is the emergence of variability in the order parameter dynamics: this instability occurs before the transition point. The instability can be measured by means of critical fluctuations (increase in standard deviation) in relative phase or the relaxation time after a transient perturbation. In the original phase transition model developed by Haken et al. [17], stability properties of different coordination patterns are directly related to the emergence of transitions between coordination patterns.

The paradigmatic example of these concepts can be seen in phase transitions in human finger movements [18]. Subjects were asked to move both index fingers anti-phase and then increase movement frequency. The result is a sudden transition in which the coordination pattern spontaneously shifted to an in-phase pattern. A critical feature of the observed transition was a loss of stability (increased variability) of the relative phase when the control parameter frequency was increased. This increase in variability around the transition point is termed 'critical fluctuations'. The central message from Kelso's research is that variability is a necessary ingredient for coordination change.

The notion that pattern variability might be functional and not considered to be simply noise is slowly gaining ground in a wide variety of disciplines. In the studies of Goldberger and colleagues [22,23] a new perspective on abnormal cardiac dynamics has emerged, namely that healthy cardiac functioning might be more variable and irregular than previously thought, and that cardiac pathologies are characterized by more regular, stable oscillations. In other words, they observed that patients at high risk of sudden death show periodic trajectories in the heartbeat with a loss in heart rate availability. If these periodic dynamics are highly regular, than this could lead to a loss of adaptability. Recently, Tsuji et al. [24] in a prospective study reanalyzed data from the Framingham heart study, and demonstrated a significant link between heart rate variability and survival rate. This study shows that low heart rate variability can predict mortality in a population-based sample of elderly subjects. Similar reductions in variability in EEG signals have been associated with neurological disease [25].

The important role of movement variability has not only been demonstrated in rhythmical processes as

described above but also in the control of posture and orientation. In a clinical setting, Newell et al. [26] observed that in patients with tardive dyskinesia, a movement disorder that emerges after prolonged use of neuroleptics, the movement trajectory of center of foot pressure was less variable as compared to healthy control subjects. The authors concluded that the postural control dynamics in these patients were less variable and this reduced variability could be related to observed postural problems. In more expert systems, what distinguishes expert from less expert marksmen is the degree of variability in the arm to stabilize the end-effector [27].

Van Emmerik and Wagenaar [21] have used the dynamical systems approach to examine the stability in the trunk during locomotion at a variety of walking velocities. At low walking velocities the relative phase between pelvis and thorax is close to in-phase (about 20°). With increasing walking velocity this phase relation changes to more out-of-phase (about 120–140°). This counter rotation in the trunk is essential for maintaining stability at higher velocities and to avoid excessive left or right rotations in the entire upper body. In patients with Parkinson's disease, these transitions in the trunk are affected due to the presence of axial rigidity. Wagenaar and Van Emmerik [28] show that this axial rigidity and inability to make transitions in coordination patterns in Parkinson's disease is associated with a reduced variability (hyperstability) of the coordination patterns.

Diedrich and Warren [29] recently used a dynamical systems approach to study the dynamics of the walk to run transition in human bipedal locomotion. Their analysis focused on transitions in the lower extremity. They used discrete relative phase as a measure of coordination and observed abrupt switches in ankle–hip and ankle–knee relative phase around the transition from walking to running. These transitions were also accompanied by increases in fluctuations in the variability of the relative phase between joints of the lower extremity.

In the following sections, we will describe the methodology of dynamical systems and present examples in which this approach is used to investigate orthopaedic injuries in runners. Of particular importance in this analysis is the role of movement variability in coordination among lower extremity segments in healthy and pathological cases.

3. Methods

3.1. Subjects

Representative data from individual subjects in two different studies will be presented in this paper. In each case, subjects were required to complete a Physical Activity Questionnaire to verify their health status and to give their consent by signing an informed-consent form. Both procedures were in accordance with University policy. In the first study, subjects with Q-angles greater than 15° and those with Q-angles less than 15° were compared. Individuals with Q-angles greater than 15° have been thought to be at a greater risk of lower extremity injury than those with Q-angles less than 15° [30]. In the second study, a comparison of symptomatic individuals with patellofemoral pain and individuals with no pain was undertaken.

3.2. Experimental set-up

Kinematic data were collected using two high-speed (200 Hz) video cameras (NAC) and recorders. Both cameras recorded a right sagittal view while allowing for minimal view convergence. A 1.0 m cube containing 25 markers of known coordinates was used to calibrate the room coordinate system. A force platform (Advanced Mechanical Technologies, Inc., Newton, MA, USA) was located flush with the running surface in a 35.0 m runway. Foot–ground contact was indicated by an illuminated light-emitting diode (LED) interfaced with the vertical ground reaction force component from the force platform. The LED was placed in view of both cameras. Subject velocity was monitored via photoelectric sensors interfaced with a digital counter and placed on either side of the force platform at a known distance from each other.

3.3. Protocol

A comprehensive lower extremity evaluation was performed on each subject by a physical therapist. All goniometric measures were recorded from the right lower extremity and included such parameters as genu valgum, tibial varum, static rearfoot eversion, femoral anteversion and Q-angle. Using a standard anthropometric clinical measure, the existence of a lower extremity length discrepancy was also determined [31]. A bilateral difference of greater than 1.27 cm (0.5 m) prevented subject participation. Using a three-segment model, a triad of three non-collinear markers was securely placed on the lateral surface of the right thigh, leg and foot.

The first study comparing high and low Q-angles was conducted with subjects running overground. Subjects ran at a locomotor velocity ranging from 3.60 m s^{-1} (7:30 min $mile^{-1}$) to 3.83 m s^{-1} (7:00 min $mile^{-1}$) across the force platform. They were instructed not to target the force platform with their right foot as forced kinematic changes may result. A total of ten trials were

recorded for each subject with a trial deemed acceptable if proper velocity was maintained and the entire right foot contacted the force platform. The second study was performed on a treadmill using a locomotor speed ranging from 2.5 to 3.5 m s^{-1}. Finally, as a reference for the running trials in both situations, a static calibration trial was recorded of each subject standing.

3.4. Data reduction

The Motion Analysis VP 110 Expert Vision System (Motion Analysis Corp., Santa Rosa, CA, USA) was used to digitize the kinematic data. In the first study, the stance phase only of the running stride was digitized while the complete stride cycle was digitized in the second study. In the overground study, the illuminated LED indicated the stance phase defined as right foot heel-strike to right toe-off. In the treadmill running study, algorithms to predict right heel contact were used to define the stride cycle. The digitized data produced the 2-D Cartesian coordinates (x,y) of each marker at each frame during the stance phase or the stride cycle. Three additional frames before and after the event of concern were included in the analysis to minimize filtering effects. Higher frequencies associated with noise were eliminated using a 16 Hz low-pass 4th order zero lag Butterworth filter. The frequency cutoff was determined based on power spectral densities of the x and y marker paths of each segment and selecting a common frequency that contained 95% of the power. A direct linear transformation (DLT) was employed to reconstruct the three-dimensional image from the x and y coordinates collected from the right and left cameras [32]. Average calibration errors ranged between 2 and 3 mm for the calibration volume.

The local segment coordinate systems were constructed from the triads throughout the stance phase and stride cycles respectively and oriented to the calibration position of each subject [33]. Areblad et al. [33] described a method of calculating local coordinate systems of the foot and leg segments with triads placed in the frontal plane. In the present study, this protocol was modified by placing the triads in the sagittal plane. In addition to the foot and leg segments, the thigh segment was included by extending the protocol for the leg. Three-dimensional segment angles were calculated as described in Areblad et al . [33] and three-dimensional joint coordinate angles were calculated according to a method outlined by Grood and Suntay [34].

Phase plots were then calculated for the relevant segment and joint angles. Each phase plot consisted of the angle (θ) on the horizontal axis with its first derivative, angular velocity (ω), on the vertical axis. To allow for calculation of the phase angle (ϕ), phase plots were normalized for each trial using the following equations:

$$\text{Horizontal axis (angle)}: \quad \theta_i = \frac{2 * [\theta_i - \min(\theta_i)]}{\max(\theta_i) - \min(\theta_i)}$$

θ = segment angle; i = data point within stance phase or within stride cycle

Vertical axis (angular velocity) :

$$\omega_i = \frac{\omega_i}{\max\{\max(\omega_i), \max(-\omega_i)\}}$$

ω = segment angular velocity; i = data point within stance phase or within stride cycle

Following the normalization process, each phase plot had four quadrants with the origin of the x-axis at mid-range, and the minimum and maximum values being -1.0 and 1.0, respectively. The y-axis was normalized to its greatest absolute value to maintain zero velocity at the origin. The phase angle was defined as the angle between the right horizontal and a line drawn from the origin to a specific data point (θ,ω) and was calculated as follows:

$$\phi = \tan^{-1} \frac{\omega(t)}{\theta(t)}$$

The lower extremity components for which the phase angles were calculated included thigh adduction/abduction, thigh flexion/extension, tibial rotation and foot inversion/eversion. A schematic of the method of calculation of the phase angle and an exemplar phase plot are presented in Fig. 1.

Phase plots of the relevant components of the lower extremity segment and joint angles were used to illustrate the coupling of the lower extremity segments during running via the calculation of the continuous relative phase (CRP). The CRP was defined as the difference between the normalized phase angles of two segment motions throughout the stance phase or the entire stride phase. In each coupling, the distal segment was subtracted from the proximal. A CRP of 0° indicated that the respective segments were in-phase. As the CRP increased the segments would be more out-of-phase until a CRP of 180° would indicate an anti-phase coupling. A positive CRP indicated that the proximal segment had a greater phase angle while a negative CRP indicated that the distal segment had a greater phase angle. CRPs were calculated from the normalized phase plots for thigh flexion/extension and tibial rotation ($Th_{F/E}$–Tib_{Rot}), thigh abduction/adduction and tibial rotation ($Th_{Ab/Ad}$–Tib_{Rot}), tibial rotation and foot eversion/inversion (Tib_{Rot}–$Ft_{Ev/In}$) and femoral rotation and tibial rotation (Fem_{Rot}–Tib_{Rot}). For example:

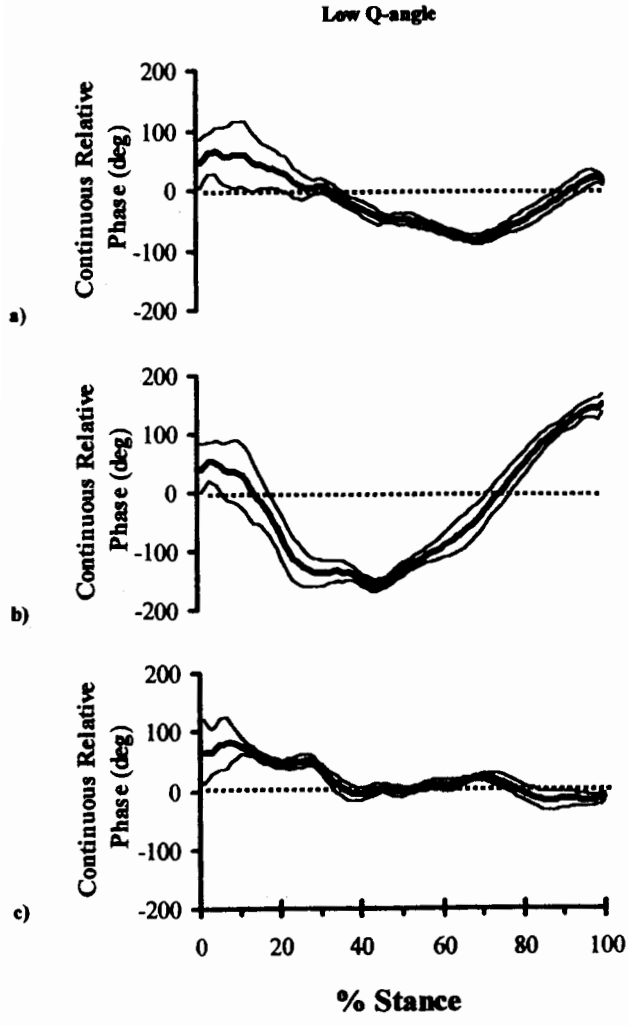

Fig. 2. Continuous relative phase (CRP) patterns of lower extremity couplings present in an exemplar healthy (low Q-angle) individual during running: (a) $Th_{F/E}$–Tib_{Rot}; (b) $Th_{Ab/Ad}$–Tib_{Rot}; and (c) Tib_{Rot}–$Ft_{Ev/In}$. The thick line indicates the CRP with the thin lines representing the variability of the CRP across trials. Note the higher CRP variability displayed during initial stance relative to the remainder of stance. Zero percent stance indicates heel strike with toe-off occurring at 100%.

$$CRP(t) = \phi Th_{F/E}(t) - \phi Tib_{Rot}(t)$$

To measure between trial variability of the relative phase, each CRP profile for each coupling relationship was interpolated to 100 data points using a polynomial procedure. Ensemble curves were calculated from each coupling relationship for each subject as the mean from the 10 trial CRP curves. The variation of CRP was calculated as the standard deviation of each point on the ensemble curve and was quantified by calculating the average standard deviation over the complete profile (either support phase only or the complete stride), over the support and swing phases of the stride (where applicable) and over portions of the support phase. The portions of the support phase were based on key events that occurred in subtalar joint inversion/eversion. The portions of the support phase include: (1) from foot contact to neutral position; (2) from neutral position to maximum eversion; (3) from maximum eversion to neutral position; and (4) from neutral position to toe-off. Figure 2 illustrates ensemble CRP plots of lower extremity couplings for a single, healthy, representative subject.

4. Results and discussion

The purpose of this paper was to demonstrate the use of a dynamical systems approach to investigate the coupling relationships in the lower extremity. Particularly, we wish to show that the variability of the CRP is an effective means of discriminating between symptomatic and asymptomatic individuals. To begin this discussion, however, we will describe the coupling patterns of individuals who have no lower extremity injury and a Q-angle less than 15°. Figure 2 illustrates the CRP for the segment interactions for $Th_{F/E}$–Tib_{Rot}, $Th_{Ab/Ad}$–Tib_{Rot} and Tib_{Rot}–$Ft_{Ev/In}$ for an exemplar individual who had no history of lower extremity injury. While these profiles are a single representative of a larger data set, the distinct patterns shown here are similar for all subjects without a lower extremity involvement. Data showing the segment coupling for this subject are presented in Table 1. For all three couplings, the CRP was out-of-phase at heel contact and for $Th_{F/E}$–Tib_{Rot} and $Th_{Ab/Ad}$–Tib_{Rot} remained out-of-phase during the remainder of the support phase. Further in the support phase, the Tib_{Rot}–$Ft_{Ev/In}$ coupling changed to a rela-

Table 1
Mean continuous relative phase (CRP) and CRP variability over the complete support phase for individuals representing the high Q-angle (> 15°) and low Q-angle (< 15°) groups

Segment coupling	Low Q-angle		High Q-angle	
	CRP	CRP variability	CRP	CRP variability
$Th_{F/E}$–Tib_{Rot}	41.72	18.45	50.02	13.88
$Th_{Ab/Ad}$–Tib_{Rot}	89.66	23.90	92.49	20.85
Tib_{Rot}–$Ft_{Ev/In}$	25.22	12.85	14.67	16.87

tively in-phase coupling. In each case, there was a relatively high variability of the CRP from heel contact to foot flat ($CRP_{SD} = 38.66°$ for $Th_{F/E}$–Tib_{Rot}; 40.33° for $Th_{Ab/Ad}$–Tib_{Rot} and 32.55° for Tib_{Rot}–$Ft_{Ev/In}$) indicating that this was a period of greatest fluctuation in CRP. For the remainder of the support period in all couplings, there was generally less variability of the CRP.

Various static goniometric measures, such as the quadriceps angle or Q-angle have been defined to quantify lower extremity segment alignment. Greater Q-angles result in greater genu valgum, hip adduction and foot pronation. While these measures are determined with the patient in a static position, greater Q-angles have been associated with lower extremity injuries such as infrapatellar tendinitis and chrondromalacia patella [35–37]. For this study, we compared individuals with a Q-angle greater than 15° to individuals with a Q-angle less than 15° [30]. It should be noted that all subjects in this study were asymptomatic regardless of the Q-angle measurement. Figure 3 displays the lower extremity couplings of an individual representing the high Q-angle group who was typical of the group data. Similar to the low Q-angle data in Fig. 2, the CRP for all couplings was more out-of-phase at heel contact. The Tib_{Rot}–$Ft_{Ev/In}$ then became strongly in-phase for the remainder of support. Relatively high variability of the CRP from heel contact to foot flat was again revealed for all three couplings. For the remainder of the support period, there was generally less variability of the CRP. The $Th_{Ab/Ad}$–Tib_{Rot} coupling was generally out-of-phase by a mean value of 89.6° throughout the support period with the greatest variability occurring from heel contact to foot flat.

When comparing the subject samples representing low and high Q-angle individuals, it becomes evident that there is little difference in either the pattern or the variability of the CRP of any of the couplings. The mean CRP and mean standard deviations over the total support phase for the high and low Q-angle groups are presented in Table 1. There were no statistically significant differences in the mean CRP and the variability in CRP between the groups for all couplings ($P > 0.05$).

While there were no differences beween groups, from Figs. 2 and 3 it is apparent that there are systematic differences in CRP and CRP variability across specific regions of stance. To further analyze group differences, the stance phase was divided into four intervals based on rearfoot angular events. Fig. 4 displays the means of the CRP and the variability of the CRP for the four stance intervals of a single subject who was representative of the group. All couplings initiated stance about 60° out-of-phase with the Tib_{Rot}–$Ft_{Ev/In}$ coupling becoming more in-phase as stance advances. The opposite was true for the $Th_{Ab/Ad}$–Tib_{Rot} coupling where, following initial stance, the phase difference increased between segments.

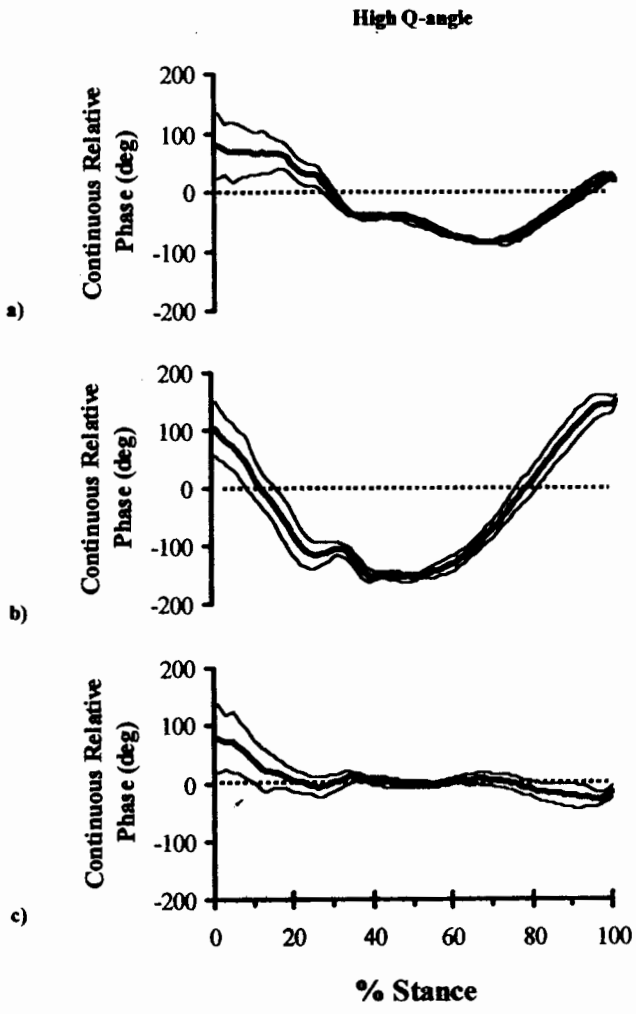

Fig. 3. Continuous relative phase (CRP) patterns of lower extremity couplings during running present in an exemplar individual with high Q-angle: (a) $Th_{F/E}$–Tib_{Rot}; (b) $Th_{Ab/Ad}$–Tib_{Rot}; and (c) Tib_{Rot}–$Ft_{Ev/In}$. The thick line indicates the CRP with the thin lines representing the variability of the CRP across trials. Note the higher CRP variability displayed during initial stance relative to the remainder of stance. Minimal difference can be found in comparison to the healthy individual displayed in Fig. 2.

The variability of the CRP was similar for all couplings with the highest variability found during the initial stance period from heel contact to pronation. Unlike the other couplings, the Tib_{Rot}–$Ft_{Ev/In}$ coupling displayed a slight increase in variability as toe-off neared.

In the second study, symptomatic individuals with patellofemoral pain (PFP) were compared with individuals who had no history of PFP. For this study, subjects performed multiple trials at three velocities (2.5, 3.0 and 3.5 m s^{-1}). Phase angles of the relevant lower extremity angles were calculated over the complete stride cycle. The CRP and the variability of the CRP of these subjects were compared at each running velocity for the same couplings as the previous investigations

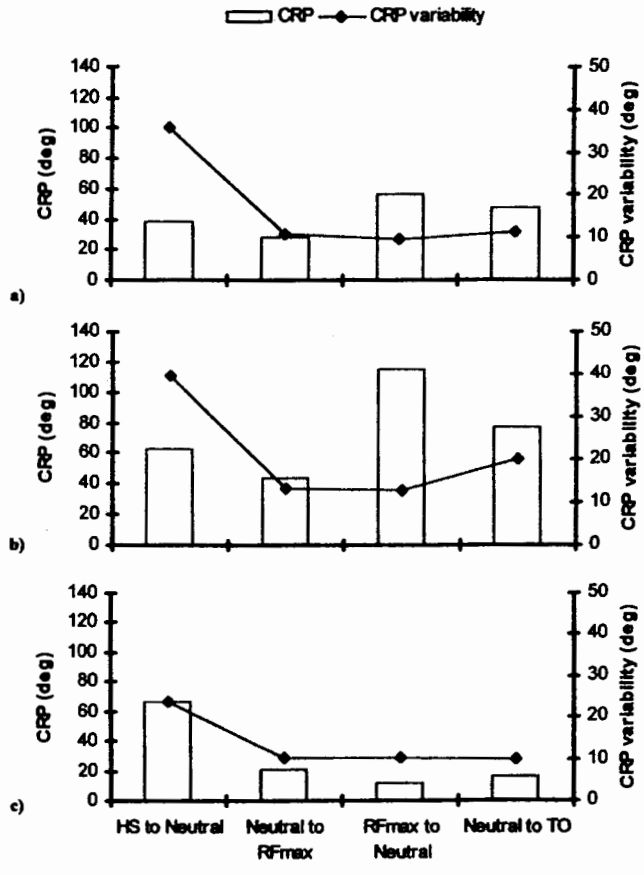

Fig. 4. Continuous relative phase (CRP) and variability of the CRP during the four stance intervals. Data from low Q-angle subject and high Q-angle subject were pooled as no group differences were present. Note the highest variability of CRP present during the initial period of stance. HS = heel-strike, RF_{max} = maximum subtalar joint eversion, TO = toe-off.

with the addition of the thigh internal/external rotation and tibial internal/external rotation coupling (Th_{Rot}–Tib_{Rot}). This additional coupling was added to address possible antagonistic rotations in PFP of the two segments that comprise the knee. Mean values for the CRP and CRP variability were calculated over the entire stride cycle as well as the swing and stance phases. In addition, the stance phase was analyzed over the four previously defined stance phase intervals.

The data for each progression velocity followed similar trends and, as such, only the 2.5 m s^{-1} data will be presented hereafter. Fig. 5 shows the variability of the CRP for each lower extremity coupling of representative individuals from both groups. Unlike the previous investigation that revealed much group similarity, differences can be found in both the pattern and magnitudes of the respective couplings. From the data in Fig. 5 and Table 2, it appears that the variability of the four CRP couplings was less in the symptomatic PFP group as compared to the asymptomatic healthy data. From Fig. 5 it can be seen that these differences were especially strong in the transition from stance to swing (40–50% of the stride) and from swing to stance (80–100% of the stride). Separating the stride cycle into stance and swing portions, the strongest differences in CRP variability appear during the swing portion in which the healthy group has consistently greater variability in all four couplings (see Table 2). The implication of these data is that there is a greater degree of repeatability of action between the respective segment couplings in the PFP data, an observation that could be related to inflexible patterns of coordination and possible emergence of patellofemoral pain.

To allow comparison with the previous study on low vs high Q-angle subjects, a further analysis of the coordination differences during the stance intervals was performed on the healthy and PFP subjects. The progression of the CRP pattern through the stance phase is similar to that present in the two previous investigations (Fig. 6). The PFP data, however, showed a marked difference in the $Th_{Ab/Ad}$–Tib_{Rot} pattern during midstance. Following heel-strike, the asymptomatic data reveal more out-of-phase couplings while the PFP data show a progression toward in-phase. Further differences between groups were present in the comparison of the CRP variability (Fig. 7). Generally, the healthy asymptomatic data in Fig. 7 indicated that there was approximately 15° of CRP variability for all couplings at heel-strike. During midstance, a slight decline in variability was noted, followed by a return to higher values at terminal stance. This global pattern of increase of variability in the transition regions between stance and swing was not displayed in most of the PFP couplings.

5. Conclusions

There are important inferences that can be drawn from the CRP standard deviation profiles. The CRP and the variability of the CRP do not appear to define the differences in anatomical structure between the Q-angle groups during running. For example, whether the individual has a Q-angle greater than or less than 15° does not appear to be contained in these data. It is our position that differences in the CRP or the variability of the CRP indicated that no pathology was present. Functionally, the two groups had similar CRPs and similar variability of the CRPs. The fact that one group had a greater Q-angle, however, did not mean that this group was necessarily symptomatic of any lower extremity pathology. We maintain, however, that the variability of the CRP gives insight into the dynamic function of the lower extremity.

The notion that an individual with a lower extremity pathology would be less variable than an individual without a pathology may appear to be inconsistent with

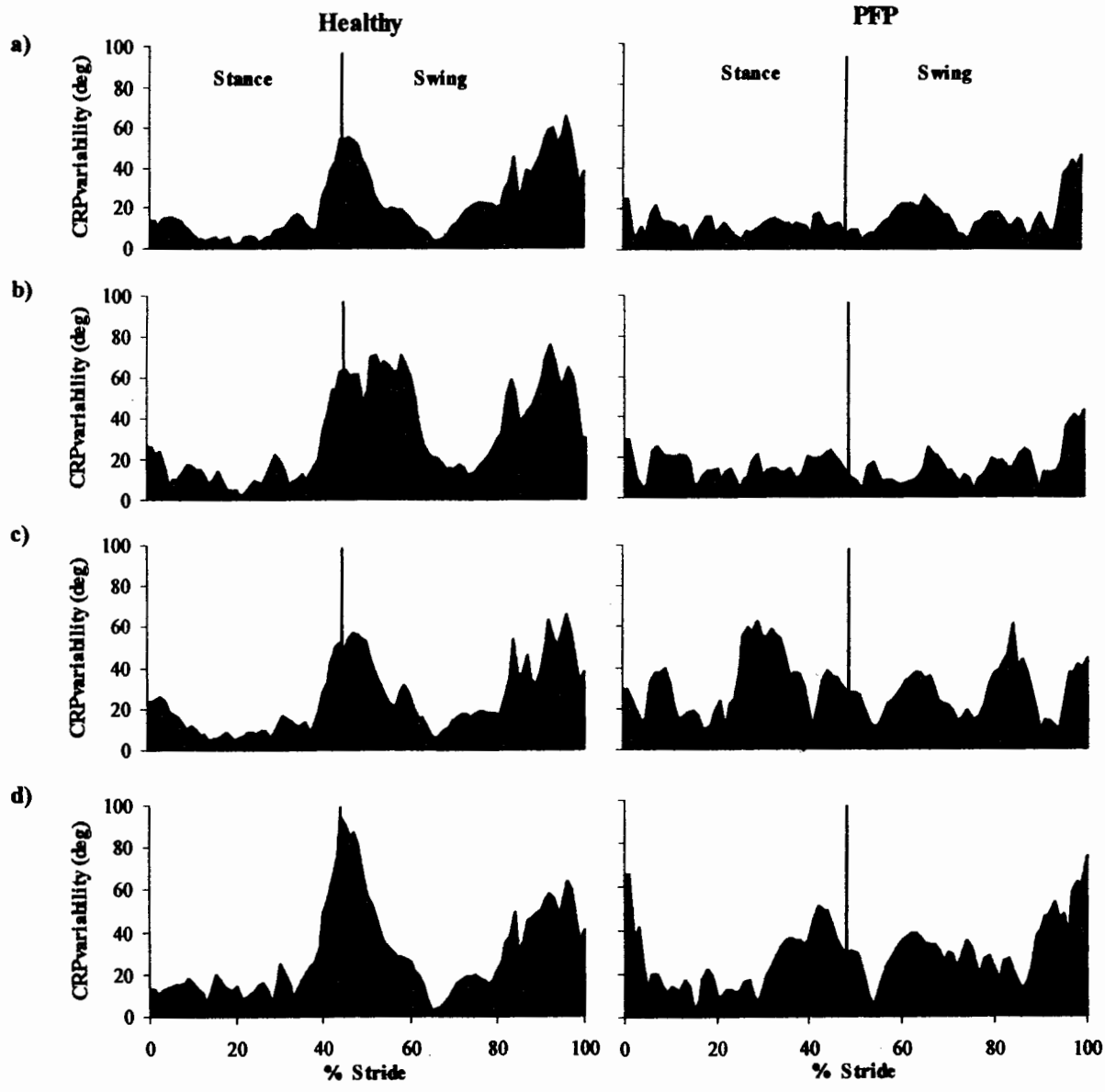

Fig. 5. CRP variability of lower extremity couplings during running present in individuals with and without patellofemoral (PF) pain: (a) $Th_{F/E}$–Tib_{Rot}; (b) Th_{Rot}–Tib_{Rot}; (c) $Th_{Ab/Ad}$–Tib_{Rot}; and (d) Tib_{Rot}–$Ft_{Ev/In}$. Note the higher CRP variability displayed during terminal stance for the healthy individual compared to the individual with PF pain. Further the individual with the PF pain displays less variability throughout the stride cycle.

Table 2
Continuous relative phase (CRP) variability of each segment coupling over the complete stride cycle, as well as the stance and swing phases for the healthy and patellofemoral pain (PFP) individuals

Segment coupling	Healthy			PFP		
	Stride	Stance	Swing	Stride	Stance	Swing
$Th_{F/E}$–Tib_{Rot}	21.96	10.48	30.28	14.26	11.68	16.20
Th_{Rot}–Tib_{Rot}	32.27	14.85	44.88	15.49	14.90	15.93
$Th_{Ab/Ad}$–Tib_{Rot}	24.55	13.43	32.59	29.17	31.22	27.63
Tib_{Rot}–$Ft_{Ev/In}$	29.54	18.08	37.84	28.01	22.73	31.99

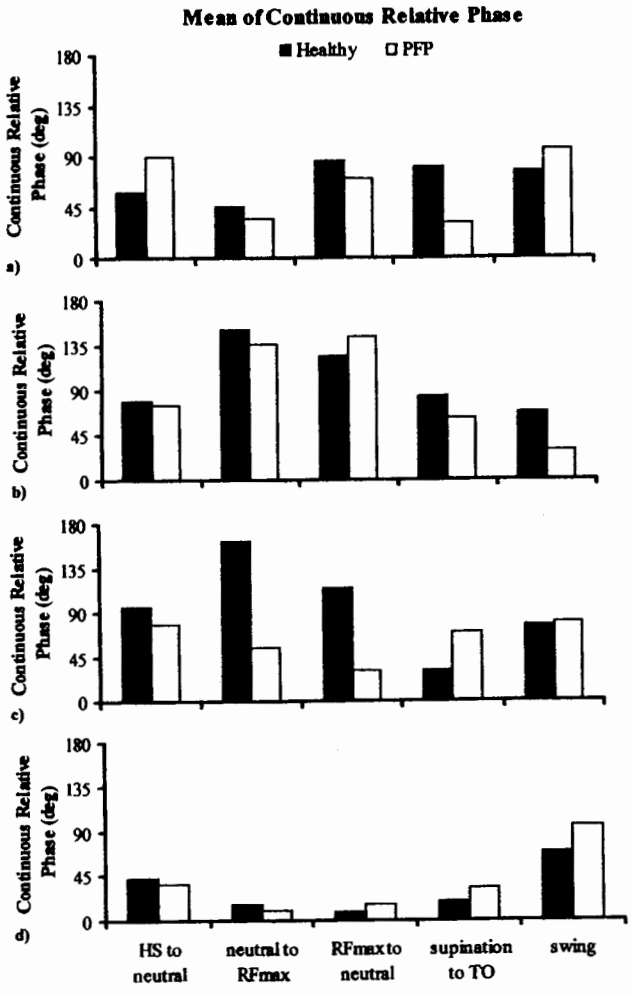

Fig. 6. Continuous relative phase (CRP) at discrete intervals of the stance phase during running: (a) $Th_{F/E}$–Tib_{Rot}; (b) Th_{Rot}–Tib_{Rot}; (c) $Th_{Ab/Ad}$–Tib_{Rot}; and (d) Tib_{Rot}–$Ft_{Ev/In}$. The comparison is between individuals with and without patellofemoral pain (PFP). HS = heel-strike, RF_{max} = maximum subtalar joint eversion, TO = toe-off.

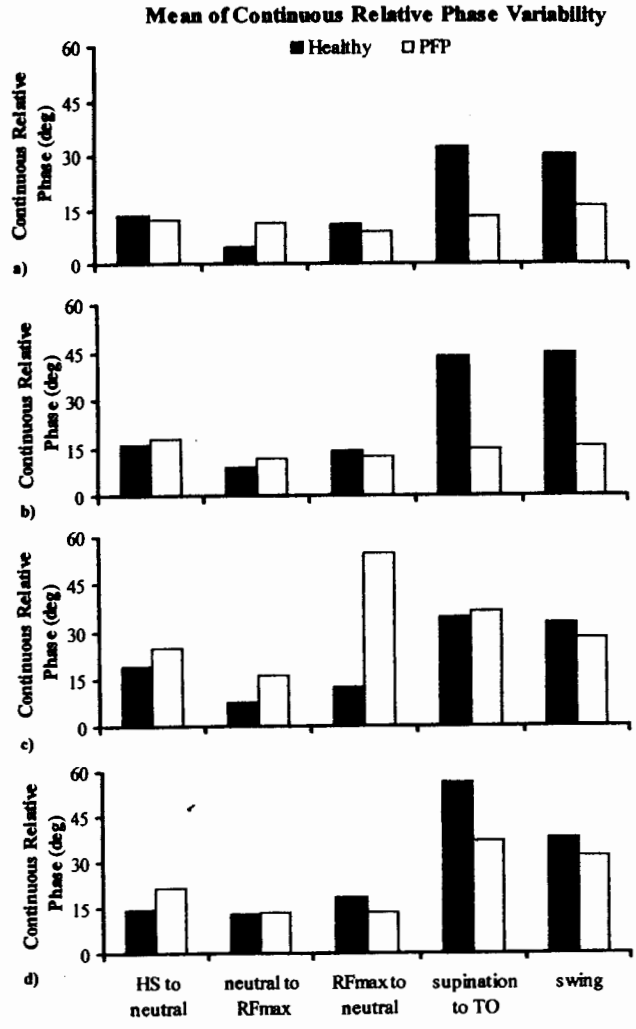

Fig. 7. Continous relative phase (CRP) variability at discrete intervals of the stance phase during running: (a) $Th_{F/E}$–Tib_{Rot}; (b) Th_{Rot}–Tib_{Rot}; (c) $Th_{Ab/Ad}$–Tib_{Rot}; and (d) Tib_{Rot}–$Ft_{Ev/In}$. The comparison is between individuals with and without patellofemoral pain (PFP). Note the discrepancy in the CRP variability present between the two data sets during terminal stance. HS = heel-strike, RF_{max} = maximum subtalar joint eversion, TO = toe-off.

what is generally accepted in the biomechanics literature. We have seen that healthy individuals have greater variability of the CRP in the segment couplings that directly contain the knee joint. The obvious question is why is there greater variability in the healthy norm or, to the contrary, why is there less variability in the pathological state? Lower CRP variability appears to be an indicator of a non-healthy state. In the case of the PFP individuals, while symptomatic, they were able to perform the required task of running multiple trials at three different speeds. The lack of variability of the joint couplings indicated that segment actions were repeatable within a very narrow range and enabled these individuals to accomplish this task with a minimum of pain. The coordination of the lower extremity segments was such that there could be little deviation in the relative actions of the segments to produce a relatively pain-free performance. The narrow CRP variability range indicates the presence of an injury but does not determine the cause of the injury. The determination of the cause of the injury and the resulting change in variability of the CRP as the injury progresses must be found in a prospective study rather than a case-cohort study.

However, while the lack of variation indicates an injury state, it also could result in a further constant stress on soft tissue such as cartilage, tendon and ligaments. Repeated stress may result in pain in these tissues ultimately causing degenerative changes. In effect, individuals with little variability in the lower extremity CRPs may produce an overuse situation that compounds the original problem. For example, fatigue wear

of articular cartilage results from the accumulation of microscopic damage of the material under repetitive stressing [38]. The repetitive stressing occurs with the repeated application of low loads over an extended period even though the loads may be much lower than the material's ultimate strength. With the PFP individuals, the repetitive stressing would appear to occur in the same general area of the cartilage and thus produce localized stress in the tissue.

On the other hand, the healthy individuals do not experience pain and were not constrained in any manner. The variety of each coupling action would indicate there were multiple combinations of coupling patterns that could be utilized. This situation would be an optimal solution because it would mean that no soft tissue would be repeatedly stressed. The healthy state, therefore, is one in which no tissue is repeatedly stressed which results from the relatively greater variability of joint couplings.

In summary, we claim that the traditional view of disordered movement as showing more variability is no longer tenable. As indicated previously, recent development in the motor control literature [18] and in the clinical domain [22] all suggest a functional role for variability in system dynamics. The present findings demonstrate that this also may be the case for lower extremity coupling patterns during locomotion. As such, the methods described in this paper may prove very useful in the detection and treatment of lower extremity running injuries.

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