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Review

Lower extremity joint coupling during running: a current update

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Abstract

Background. The relationship between lower extremity mechanics and injury is not well understood. However, joint coupling studies are beginning to emerge, which may lend further insight into running biomechanics.

Purpose. To provide a current review of the research examining lower extremity joint coupling in running.

Summary. There are various techniques utilized to measure joint coupling, including joint timing, rearfoot eversion/tibial internal rotation ratios, continuous relative phase calculations, and vector coding. The study of joint coupling is of particular interest as it may pertain to running injuries. There is some evidence that joint coupling may be altered with orthotics and/or with footwear. Most studies have included a relatively small sample size and larger scale studies are needed to quantify normal ranges for many of the coupling measures. In addition, prospective studies are needed to clarify the relationship to injury.

Relevance

It is hoped that this update will serve as a review of the current state of thought regarding lower extremity joint coupling during running. As greater insight into the role of joint coupling in injuries is gained, more optimal intervention and prevention strategies can be developed to minimize injury risk.

Keywords: Joint coupling; Running; Review

1. Introduction

Despite the wealth of literature regarding running mechanics, the relationship between running mechanics and injuries is not well understood. Early studies of running generally focused on the movement of individual joints or segments (Bates et al., 1978; Hamill et al., 1992; Levens et al., 1948; Lundberg, 1989; Manter, 1941). While these studies provided a basic understanding of running mechanics, little insight has been gained in the area of injury mechanics. The persistence of unanswered questions has later led researchers to explore the coordination of motion between joints and segments of the lower extremity. Early studies of joint coupling focused on simple relationships involving joint timing of peak frontal plane motion of the rearfoot and sagittal plane knee motion. These studies were followed by investigations of relative joint excursions. More recently, studies have utilized more complex approaches involving Dynamical Systems Theory to examine coupling. All of this research has led to an emerging body of literature relating to lower extremity joint coupling. The purpose of this paper is to provide a current update on the research examining this area. The first section of the paper will provide a review of coupling between lower extremity segments during running. This will be followed by a description of the various methods used to

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measure joint coupling. Studies of normal joint coupling as well as abnormal joint coupling will be included. The final section will review the current literature which explores ways of altering joint coupling.

2. Lower extremity coupling mechanics

The stance phase of running gait may be divided into two functional phases. The first half of stance is commonly referred to as the cushioning, or eccentric phase of gait. The last half of stance is referred to as the propulsion, or concentric phase. Immediately after initial contact with the ground, the foot begins to pronate. Pronation of the subtalar joint is a tri-planar motion consisting of eversion, abduction and dorsiflexion of the calcaneus with respect to the talus (Donatelli, 1993). It has been suggested that pronation occurs, in part, so that the foot is able to accommodate to uneven surfaces as well as to better attenuate shock (Isman and Inman, 1969; Lundberg, 1989; Root et al., 1966). During closed chain pronation, when the calcaneus is fixed to the ground, it cannot abduct relative to the talus. Therefore, in order to obtain the transverse plane component of subtalar joint pronation, the talus adducts or medially rotates. Due to the tight articulation of the ankle mortise, the tibia internally rotates as the talus adducts. During this cushioning phase of stance, the knee joint flexes which is also associated with tibial internal rotation. Thus, pronation, tibial internal rotation, and knee flexion occur relatively synchronously, as demonstrated in Fig. 1 (Buchbinder et al., 1979; Levens et al., 1948; Tiberio, 1987). During the later, propulsive part of stance, these motions reverse, and the calcaneus inverts and the talus and tibia externally rotate as the knee extends. This normal sequence is seen in Fig. 2a.

The earliest and most abundant studies of joint coupling examined the timing of peak rearfoot eversion and peak knee flexion. Authors have assessed these timings under a variety of speeds, footwear conditions and surfaces (Table 1). While there is some variability in timing, most authors reported relative synchrony between these motions. For example, it has been reported that peak eversion occurs between 39.3% and 53.9% of stance, while peak knee flexion occurs between 36.0% and 45.3% of stance (Bates et al., 1978; DeWit and DeClercq, 2000; Hamill et al., 1992; McClay and Manal, 1997, 1998; Stergiou and Bates, 1997; Stergiou et al., 1999; Van Woensel and Cavanagh, 1992). Stergiou and Bates (1997) studied the effect of surface hardness on timing. These authors found that peak eversion and peak knee flexion occurred at 52.4% and 44.3% respectively on the extra hard surface (8.1% difference) and 51.7% and 45.7% respectively on the soft surface (6.0% difference). From these results, it does not appear that surface hardness strongly influences timing between peak rearfoot eversion and knee flexion.

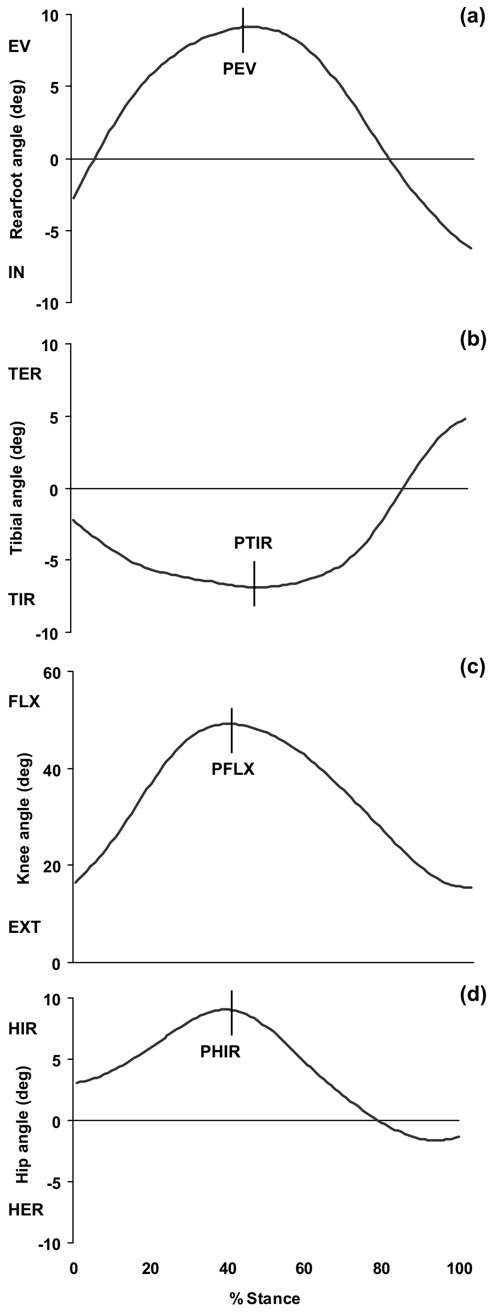

Fig. 1. Lower extremity joint angles during one complete stance phase during running gait, including: (a) rearfoot eversion/inversion, (b) tibial rotation, (c) knee flexion/extension, and (d) hip rotation.

The rationale for studying the timing of joint movements is based on the notion that asynchrony in these motions may result in injury. For example, Tiberio (1987) proposed a mechanism for anterior knee pain that is related to abnormal joint coupling. He theorized that, if pronation of the subtalar joint is prolonged and continues beyond midstance, tibial internal rotation will also be prolonged. This results in a mechanical dilemma at the knee, for knee extension begins around midstance and must be accompanied by tibial external rotation to

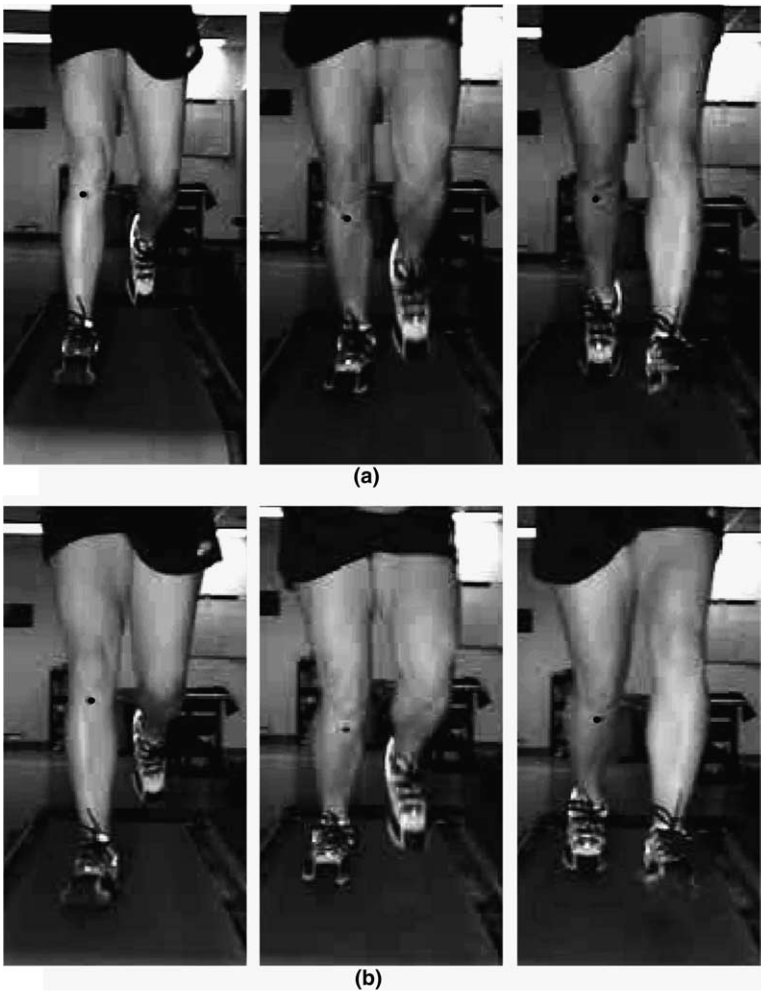

Fig. 2. (a) Lower extremity alignment at footstrike, midstance and later stance in normal running. (b) Lower extremity alignment at footstrike, midstance, and late stance in a patient exhibiting prolonged pronation. Note the internal rotation of the tibia in late stance and the associated compensatory femoral internal rotation as described by Tiberio (1987).

maintain joint congruity. However, since the tibia is continuing to internally rotate with the talus, the femur must excessively internally rotate to obtain the relative knee external rotation needed. This abnormal sequence of events is illustrated in Fig. 1b. Tiberio (1987) suggested that the compensatory femoral internal rotation may alter normal patellofemoral alignment and cause excessive contact pressures at the lateral facet of the patella. This excessive pressure is thought to eventually lead to cartilage breakdown and anterior knee pain (Buchbinder et al., 1979). Though often cited, Tiberio's (1987) theory has not been tested experimentally. Hutchison and Ireland (1995) suggest that females exhibit a femoral anteversion, as well as tibial external rotation. They propose that this rotatory limb malalignment contributes to the increased incidence of patellofemoral disorders compared to their male counterparts.

The degree of coupling between the rearfoot and knee is believed to be influenced by the orientation of the subtalar joint axis in the sagittal plane. Using telescoping pins placed into the calcaneus and talus of cadaveric feet, Manter (1941) measured the orientation of the subtalar joint axis in the sagittal plane at 42°, with a range from 29° to 47°. Employing a similar technique, Root et al. (1966) reported a subtalar joint (STJ) axis orientation of 41° but with a larger range (22–55°). With a subtalar joint axis oriented at 45° in the sagittal plane, one would expect approximately equal amounts of frontal motion (eversion) and transverse motion (tibial internal rotation). These early studies were performed on feet measured in a non-weight bearing position, which likely resulted in a higher orientation of the STJ axis than is present during stance. Nearly 40 years later, Lundberg et al. (1989a,b,c) quantified the orientation of the subtalar joint axis during stance in eight healthy subjects. Three tantalum balls were injected into numerous bones of the foot, and relative joint positions were measured using bi-planar radiographs. Subjects were positioned

Table 1
Rearfoot and Knee Kinematic Parameters from Various Studies

Authors	*N*	Speed (m/s)	Condition	Time to peak EV (s)	% Stance to PEV	Time to peak knee flexion (s)	% Stance to PKF	EV/TIR
Hamill et al. (1992)	12	3.35	Hard shoe		43.8		44.2	
			Medium shoe		42.8		44.1	
			Soft shoe		38.7		45.9	
Stergiou et al. (1997)	5	Self selected	Extra hard surface		52.4		44.3	
			Hard surface		61.2		45.3	
			Medium surface		53.9		45.3	
			Soft surface		51.7		45.7	
Stergiou et al. (1999)	8	Self selected			44.65		40.02	
		10% slower			43..31		41.74	
		10% faster	Shod		43.87		38.80	
		20% faster			42.49		40.02	
Bates et al. (1978)	10	3.35	Shod	0.0986	39.56	0.1021	40.94	
		3.83–4.47	Shod	0.0821	38.58	0.0855	40.40	
		3.83–4.47	Barefoot	0.0953	46.84	0.0700	37.05	
DeWit and DeClercq (2000)	9	3.5	Shod	0.112	45	0.104	41	
			Barefoot	0.098	41	0.087	36	
		4.5	Shod	0.100	46	0.097	43	
			Barefoot	0.082	41	0.076	38	
		5.5	Shod	0.090	46	0.085	45	
			Barefoot	0.075	43	0.070	40	
Van Woensel and Cavanagh (1992)	9	3.8	Neutral shoe	0.0983		0.090		
			10° varus shoe	0.1183		0.0878		
			10° valgus shoe	0.1089		0.0933		
McClay and Manal (1997)	9	3.35	Normal	0.11		0.112		1.53
	9		Pronator	0.098		0.108		1.23
McClay and Manal (1998)	9	3.35	Normal	0.091		0.102		1.33
	9		Pronator	0.092		0.103		1.43
Williams et al. (2001)	20	3.35	High arch					
	20		Low arch					1.29
			High arch, no orth					1.71
Nawoczenski et al. (1995)	10	Self selected	High arch, orth					1.0
			High arch, no orth					1.4
	10		Low arch, no orth					1.8
			Low arch, orth					2.2
Nigg et al. (1993)	30	4.0	Shod					1.32
Stacoff et al. (2000a,b)	5	2.5–3.0	Shod					1.72

on a platform that was moved through various degrees of frontal and sagittal plane motions. These authors reported an average orientation of subtalar joint axis of only 32° in the sagittal plane, ranging from 14° to 39.8°. Contrary to the earlier studies, Lundberg et al. (1989a,b,c) data suggest relatively greater frontal plane (eversion) versus transverse plane (tibial internal rotation) motion. These results may be more relevant as they were obtained during functional positions.

It is difficult to measure the orientation of the subtalar joint axis directly without invasive techniques. However, a number of authors have examined the relative amounts of both rearfoot eversion (EV) and tibial internal rotation (TIR) motion, which is suggestive of the orientation of the subtalar joint. The EV/TIR ratio provides a measure of the relative motion between eversion excursion and tibial internal rotation excursion from heel strike to the respective peaks which occur around midstance. For example, a ratio of 2.0 suggests that for every two degrees of eversion, there is one degree of tibial internal rotation. Using an intracortical bone pin protocol, Stacoff et al. (2000a,b,c) measured EV/TIR in five uninjured runners. These authors reported a mean EV/TIR ratio of 1.72 for the loading phase of gait. Similar to Stacoff's results, McClay and Manal (1997) reported an EV/TIR ratio of 1.42 for nine uninjured runners. Both studies suggest there is a greater amount of eversion as compared to tibial internal rotation during running. It is important to note that TIR was measured as transverse plane motion of the foot with respect to the tibia in a joint coordinate system as defined by Grood and Suntay (1983). However, since the foot is fixed throughout most of stance, this transverse plane motion is a function of tibial rotation.

There is mounting evidence that arch structure influences EV/TIR ratios. As arch height increases, the orientation of the subtalar increases, thus altering the EV/TIR ratio. A high arch would be associated with relatively greater tibial internal rotation and a lower EV/TIR ratio then would be associated with a low arch. A number of studies have examined the EV/TIR ratio in runners with differing arch structure. Nigg et al. (1993) reported that runners with high and low arches exhibited similar rearfoot eversion excursion. However, the high arch runners exhibited greater tibial internal rotation excursions, resulting in lower EV/TIR ratios compared to the low arch runners. Nawoczenski et al. (1998) used radiographic measurements to classify arch structure and investigated differences in joint coupling in 20 runners with either high or low arches. Again, these authors found no difference in rearfoot eversion between the groups, however the high arch group exhibited increased tibial internal rotation, resulting in a lower EV/TIR ratio compared to the low arch group. Williams et al. (2001) also reported that individuals with high arches exhibit lower EV/TIR ratios as compared to their low arch counterparts. Unlike the previous studies, Williams et al. (2001) reported that the difference in EV/TIR ratio was due to a decreased eversion excursion in the high arch group, rather than a greater tibial internal rotation excursion. While they did not assess arch structure, McClay and Manal (1997) compared EV/TIR ratios in runners who were excessive pronators and those with normal rearfoot mechanics. These authors also reported similar rearfoot eversion excursions between groups. However, tibial internal rotation was greater in the pronator group, resulting in a significantly lower EV/TIR ratio in the excessive pronator group (1.33) as compared to the normal group (1.42). Based on the majority of these studies, it appears that variations in EV/TIR ratio can be attributed to tibial internal rotation excursion to a greater extent than rearfoot eversion excursion.

Researchers have focused on the EV/TIR ratio because it was thought to provide insight into where an injury is most likely to occur. It has been suggested that runners with high EV/TIR ratios (relatively greater rearfoot eversion motion) would be at greater risk for foot injuries. Conversely, those with lower EV/TIR ratios (relatively more tibial motion) would be at greater risk for knee related injuries (McClay and Manal, 1997; Nawoczenski et al., 1998; Williams et al., 2001). However, only two studies have actually related EV/TIR ratios to injury site. Contrary to previous thought, Williams et al. (2001) reported that low arched individuals with higher EV/TIR ratios had a higher incidence of knee related injuries. Those with high arches and low EV/TIR ratios experienced more foot related injuries. Nawoczenski et al. (1998) also found that high arch runners with an associated lower EV/TIR ratio, reported a greater incidence of foot injuries. The greater number of foot injuries in these high arched individuals may be due to other factors, such as foot stiffness, rather than relative excursions. Therefore, while EV/TIR ratio does appear to be related to arch structure, it may not be helpful in understanding the anatomic distribution of injuries in runners. Bellchamber and van den Bogert (2000) suggested that proximal joint moments have a greater influence on tibial rotation compared to distal joint moments. This coupling from proximal to distal may be more important than the coupling from distal to proximal which is captured in the EV/TIR ratio.

Joint timing values, as well as joint excursion ratios are measures of discrete points or ranges in the gait cycle. They do not provide a description of continuous joint coupling throughout the gait cycle. This has led biomechanists to borrow a method from Dynamical Systems Theory that is often used in motor control science. In this method, a continuous relative phase (CRP) measure is computed. In brief, the CRP is calculated by first generating a phase plane portrait of normalized angular velocity plotted against normalized angular position for two segments or joints of interest (Fig. 3). Phase angles are then calculated for all points in the phase plane portrait. Finally, the CRP angle is plotted by subtracting the phase angle of the distal segment from the phase angle from the proximal segment. CRP values can range between −180° and 180° with CRP values of 0° indicating complete in-phase coupling, and 180° or −180° indicating complete out-of-phase coupling (Hamill et al., 1999; Li et al., 1999; Stergiou et al., 2001).

Hamill et al. (1999) were among the first to introduce the use of CRP into the biomechanics literature. In a group of healthy runners, these authors reported a CRP angle of approximately 45° for EV/TIR at foot strike. This transitioned quickly into a more in-phase relationship (CRP approximately 10°) that was maintained throughout the remainder of stance. Transitions, such as foot strike and toe off, have been characterized as periods associated with out-of-phase coupling (Kelso, 1984). Hamill et al. (1999) also reported out-of-phase relationships at foot strike for knee flexion–TIR (CRP approximately 60°), knee adduction–TIR (CRP approximately 100°) and femoral internal rotation–TIR (CRP approximately 100°). Unlike EV/TIR, both of these relationships remained uncoupled throughout stance. CRP values were approximately 90° for knee flexion–TIR, 150° for knee adduction–TIR and 160° for femoral internal rotation–TIR. These results are surprising, as the motions of knee flexion and tibial internal rotation as well as knee adduction and tibial internal rotation

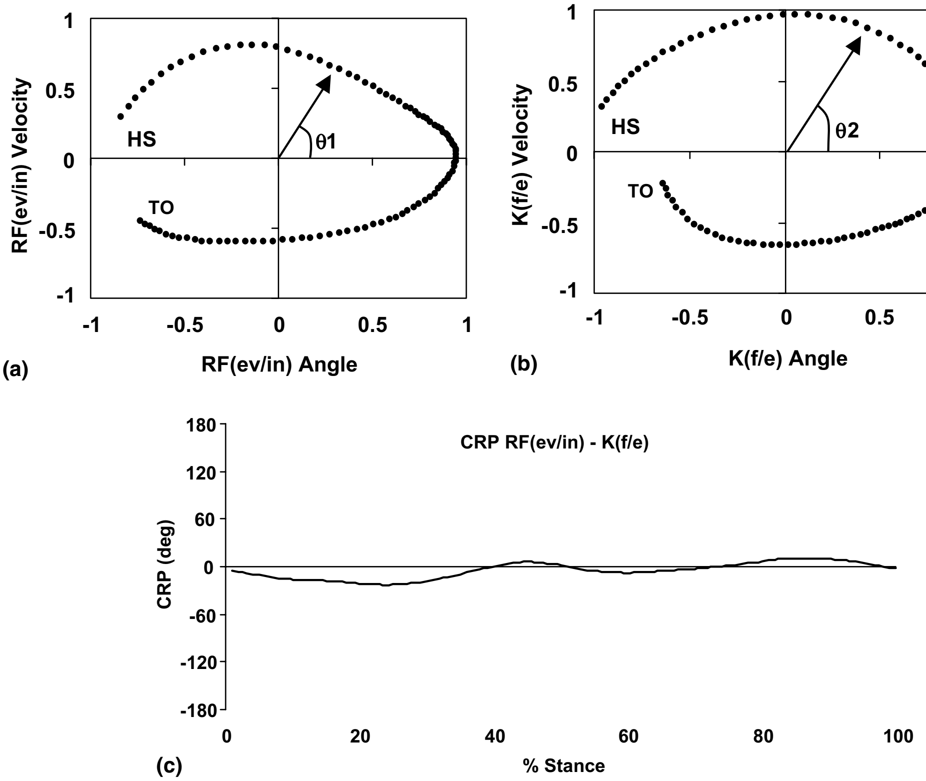

Fig. 3. (a) Phase angle plots of position versus velocity curves for rearfoot eversion/inversion, (b) tibial rotation, and (c) the CRP plot for EV/TIR coupling.

are thought to occur synchronously. The out-of-phase nature of these relationships may be a function of the velocities of these motions, which also factors into the CRP. Ferber et al. (2002) used CRP to examine joint coupling in healthy and injured runners. These authors reported an in-phase relationship for EV/TIR for the healthy group and more out-of-phase relationship for the injured group throughout stance. These data suggest that a more out-of-phase relationship for EV/TIR may be related to injury. However, this was a retrospective analysis and prospective studies are needed to determine whether these mechanics actually led to the injury. Stergiou et al. (2001) studied EV-tibial abduction coupling in the frontal plane and reported an out-of-phase relationship at heel strike which transitioned into an in-phase relationship by midstance. From midstance to toe-off, EV-tibial abduction transitioned back to an out-of-phase relationship. These data suggest that coupling relationships are different for different segment or joint combinations, and may change throughout stance.

Variability in coupling has also been suggested to play a role in injury. Hamill et al. (1999) proposed that injured runners may also exhibit reduced variability of coupling, thereby reducing the flexibility in the system and increasing the risk of overuse injuries. In their study, CRP variability was calculated for EV/TIR, knee flexion–TIR and knee adduction–TIR. These authors reported lower variability in a group of runners with a history of anterior knee pain as compared to their healthy counterparts for all three coupled relationships. The greatest difference in variability occurred at the transition periods of heel strike and toe-off. However, the injured subjects actually demonstrated more variability than the uninjured runners during the majority of stance, when the lower extremity is most loaded and injuries are thought to occur. Therefore, the clinical relevance of these findings is still unclear.

There are a number of limitations to the CRP approach. First, the CRP approach is based on the assumption that the data of interest are sinusoidal in nature. While some variables, such as knee flexion–extension are relatively sinusoidal, many are not. The second limitation relates the controversy of whether the data should be normalized. Normalization is often performed so that each of the variables of interest has a range of 0–1. This is thought to decrease the possibility that one segment will dominate the CRP (Hamill et al., 1999; Kurz and Stergiou, 2002; Li et al., 1999). However, some authors have shown that the normalization process alters both the numeric and graphic representations of the CRP, and therefore do not recommend it be done (Clark and Phillips, 1993; Kurz and Stergiou, 2002). This controversy remains, leading some authors to normalize data while others do not, limiting the ability to make comparisons. One final limitation of the CRP method is the difficulty in interpreting the results as they relate to injury. The resultant CRP angle is a function of the position and velocity of one segment relative to the position and velocity of another segment. Proposing a mechanical cause for musculoskeletal injury from these data is difficult.

Due to these issues related to the CRP approach, Heiderscheit et al. (2002) proposed the use of a vector coding technique modified from Sparrow et al. (1987). In this technique, an angle–angle diagram is constructed from the movement of two segments or joints of interest. The angle of the trajectory between two successive data points is calculated as:

$$\Theta = \tan^{-1}(\text{excursion angle1}/\text{excursion angle2})$$

This procedure is continued for all successive points. An example of the vector coding technique for EV/TIR is illustrated in Fig. 4. Because this is computed from relative excursions, it represents a value somewhat analogous to a continuous excursion ratio.

Variability of these angles can be assessed across multiple trials. Based on the same data set used by Hamill et al. (1999); Heiderscheit et al. (2002) employed the vector coding technique and compared the tibiofemoral joint coupling in a group of runners with a history of patellofemoral pain (PFP) to a group of uninjured controls. These authors reported that PFP subjects exhibited less variability of tibiofemoral coupling in the transverse plane at heel strike when compared to the uninjured group. These authors hypothesized that lower variability at the transition periods of heel strike and toe off was a reflection of a less flexible and adaptive system. However, it remains unclear whether the decreased variability was merely a compensation mechanism for injury, since the study was retrospective in nature.

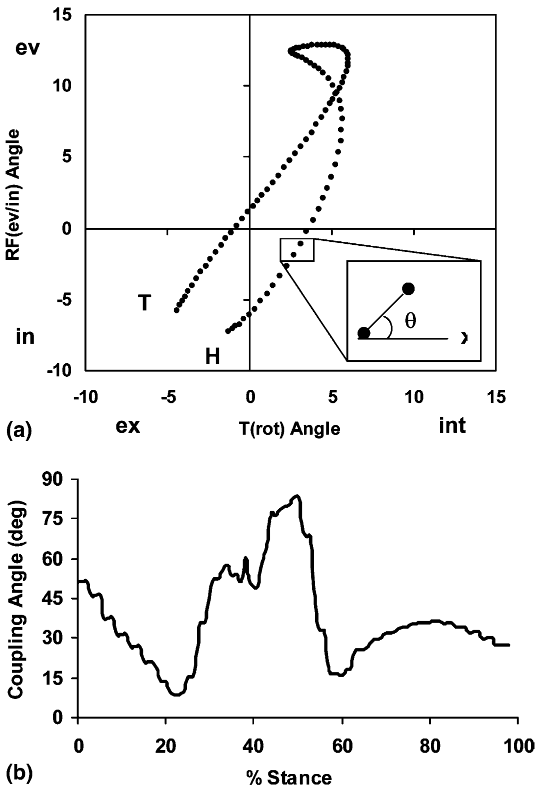

Fig. 4. (a) A sample vector coding plot including an angle–angle plot for rearfoot eversion/inversion versus tibial internal/external rotation. (b) The resultant continuous excursion ratio coupling angle for EV/TIR.

3. Alteration of joint coupling

There appears to be suggestion in the literature that running injuries may be due, in part, to abnormalities in joint coupling (Hamill et al., 1999; McClay and Manal, 1997, 1998; Nigg et al., 1993; Nawoczenski et al., 1995; Stergiou et al., 1999). Due to the closed chain nature of gait, abnormalities in foot function may influence the joint coupling throughout the lower extremity. Therefore, factors influencing foot mechanics, such as footwear, may lead to a change in joint coupling. Bates et al. (1978) compared barefoot to shod conditions in runners that were visually classified as pronators. These authors reported that the shod condition significantly reduced the period of pronation compared to the barefoot condition. Additionally, rearfoot to knee coupling became more synchronous as EV–knee flexion decreased from 9.79% to −1.82% (Bates et al., 1978). DeWit and DeClercq (2000) also reported that maximum knee flexion and peak pronation occurred more closely to one another in shod running (43 ± 11 ms) when compared to a barefoot condition (62 ± 16 ms). Both of these studies concluded that running shoes may provide for a more optimal coupling relationship between the foot and knee compared to running barefoot.

Studies examining the effects of different types of shoes on joint coupling have also been performed. Hamill et al. (1992) tested three shoes of varying midsole hardness and reported that the softest midsole resulted in larger joint timing differences between maximum knee flexion and maximum subtalar joint pronation. Therefore, changing midsole durometer in footwear may influence the synchrony between the subtalar joint and the knee placing some runners at greater risk for injury. Van Woensel and Cavanagh (1992) assessed the effect of altering shoes with 10° valgus wedged and 10° varus midsoles on joint coupling in nine healthy runners. The timing difference between maximum pronation and maximum knee flexion was greatest in the varus wedged shoe (30.5 ms), followed by the valgus wedged shoe (15.6 ms), and was smallest in the neutral shoe (8.3 ms). Peak knee flexion occurred significantly earlier in the varus wedged shoe (87.8 ms) as compared to the valgus wedged shoe (93.3 ms).

Joint coupling may also be altered with the use of foot orthotic devices. The majority of the literature reports that orthotic devices are effective in reducing rearfoot motion (Baitch et al., 1991; Bates et al., 1979; Johanson et al., 1994; Smith et al., 1986). Bates et al. (1979) investigated the effect of orthotics on joint timing in excessive pronators. They reported that the period of pronation was significantly and progressively reduced from the barefoot to shod to orthotic conditions (Bates et al., 1979). Nawoczenski et al. (1995) assessed the effect of semi-rigid orthotics on EV/TIR ratios in runners with high and low arches. Contrary to other investigations, these authors reported negligible differences in eversion excursion for both high and low arch groups when comparing the orthotic and non-orthotic device conditions. However, a significant decrease in tibial internal rotation excursion in the orthotic condition (5.2°) as compared to the non-orthotic condition (8.1°) was noted in the high arch group. Therefore, it appears that changes in tibial internal rotation are responsible for changes in EV/TIR ratios induced by orthotic devices.

Using a CRP approach, Ferber et al. (2002) compared the effects of foot orthotic devices on joint coupling and coupling variability in 11 injured runners. These runners were initially treated with standard custom orthotic devices without resolution of their symptoms. They were then treated with inverted orthotic devices as described by Blake (1986), resulting in pain reduction. When running without the orthotic devices, the injured group demonstrated out-of-phase coupling and high CRP variability. However, when running in either the standard orthotic devices or the inverted orthotic devices, these same subjects exhibited a more in-phase relationship and lower coupling variability. It was hypothesized that the pattern seen in the inverted orthotic device condition would most closely approximate that of a group of uninjured controls used for comparison. This was based on the fact that the inverted orthotic devices provided pain relief and the standard orthotic devices did not. However, it was the coupling observed in the standard orthotic device condition that more closely approximated that of the control group. Therefore, while both orthotic devices altered joint coupling, the results did not provide an explanation for the resolution of the symptoms seen in the inverted device.

4. Summary

The assessment of joint coupling has provided an alternative way to examine gait mechanics and explore the mechanisms of running injuries. In terms of relative timing, it appears that there is synchrony between peak eversion, peak tibial internal rotation and peak knee flexion, which takes place near mid-stance in healthy runners. Normal EV/TIR excursions during running are greater than 1, suggesting relatively greater rearfoot eversion excursion than tibial internal rotation. However, EV/TIR also appears to be related to the orientation of the STJ axis, but does not appear to lend insight into location of injury. Studies using CRP, vector coding and variability techniques have provided new perspectives in understanding running biomechanics. However, most of the findings are based on studies with relatively low subject numbers ($n = 5$–12). Therefore, one must be cautious in extrapolating information regarding abnormal joint coupling, as the boundaries of normal joint coupling have yet to be defined. It remains unclear as to how timing measures, EV/TIR ratios, CRP values and variability relate to injury. Alterations outside the normal bounds of joint coupling may predispose an individual to pathomechanics and injury. There is some evidence that joint coupling, both normal and abnormal, may be manipulated via shoe wear and orthotic intervention.

Research in the area of joint coupling is relatively new, but will hopefully stimulate other research questions. Future areas of research should include the development of normative studies with larger subject numbers to further define the normal bounds of joint coupling. Thorough assessments of other joint coupling relationships, including tibiofemoral and hip–knee coupling are also needed. Finally, prospective studies are needed to establish relationships between joint coupling and injury prevalence. Information from these studies will provide a foundation upon which intervention strategies can be developed to minimize running injuries.

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