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The Effects of Shoe Traction and Obstacle Height on Lower Extremity Coordination Dynamics during Walking

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Abstract

This study aims to investigate the effects of shoe traction and obstacle height on lower extremity relative phase dynamics (analysis of intralimb coordination) during walking to better understand the mechanisms employed to avoid slippage following obstacle clearance. Ten participants walked at a self-selected pace during eight conditions: four obstacle heights (0%, 10%, 20%, and 40% of limb length) while wearing two pairs of shoes (low and high traction). A coordination analysis was used and phasing relationships between lower extremity segments were examined. The results demonstrated that significant behavioral changes were elicited under varied obstacle heights and frictional conditions. Both decreasing shoe traction and increasing obstacle height resulted in a more in-phase relationship between the interacting lower limb segments. The higher the obstacle and the lower the shoe traction, the more unstable the system became. These changes in phasing relationship and variability are indicators of alterations in coordinative behavior, which if pushed further may have lead to falling.

Keywords

Dynamical systems theory; Shoe traction; Obstacle clearance; Locomotion

1. Introduction

Injuries associated with slips, trips and falls continue to pose a significant burden to society both in terms of human suffering and economic losses (Grönqvist & Roine, 1993; Kemmlert & Lundholm, 1998; Leamon & Murphy, 1995; Manning et al., 1988; National Safety Council, 1995). According to statistics from the Health and Safety Executive (HSE), slips and trips are

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the single most common cause of injuries at work, and account for over a third of all major work injuries. In the US, falls accounted for 19% of all nonfatal occupational injuries in 2001, and 13% of fatal occupational injuries in 2002 (Department of Health and Human Services, 2003). The annual direct cost occupational injuries due to slips, trips and falls in the US has been estimated to be in excess of 6 billion US dollars (Courtney et al., 2001), and a cause of serious public health problem with costs expected to exceed $43.8 billion by the year 2020 in the US alone (Englander et al., 1996).

Both slips and trips result from unintended or unexpected changes in the contact between the feet and the walking surface. Thus, conventional biomechanical analyses (i.e. gait analysis) have been used to investigate human factors that cause slips, trips, and falls and their complex interaction with environmental factors (Moyer et al. 2006; Petrarca et al. 2006). Human factors include gait biomechanics, expectation, the health of the sensory systems (i.e. vision, proprioception, and vestibular) and the health of the neuromuscular system (Moyer et al. 2006). Among the most important environmental factors that could potentially cause instability during walking are the presence of obstacles and the loss of traction between the shoe sole and floor surface (Cohen & Compton, 1982). Therefore, numerous studies have investigated the effect of obstacle perturbations during walking (Begg et al., 1998; Chen et al., 1994; Chen and Lu, 2006; Chou & Draganich, 1997; Jaffe et al., 2004; McFadyen & Prince, 2002; Patla et al., 1991; Patla & Rietdyk, 1993; Petrarca et al., 2006; Sparrow et al., 1996). However, this research has focused on the approach to an obstacle by collecting gait data of the trailing and leading limb while negotiating the obstacle. In addition, there have been numerous studies that have used biomechanics of gait to examine the shoe–floor interface to understand slips (Burnfield & Powers, 2006; Bring, 1982; Cham & Redfern, 2001, 2002a, 2002b; Gao et al., 2003, [2004]; James, 1980; Lockhart et al., 2003, [2005]; Perkins, 1978; Perkins & Wilson, 1983; Redfern & Dipasquale, 1997; Strandberg, 1983; Strandberg & Lanshammar, 1981; Winter, 1991). However, limited attention was devoted to the combined effect of obstacles and low friction shoe–floor interface on the landing strategies adopted to avoid slipping after obstacle clearance (Patla & Rietdyk, 1993; Bentley & Haslam, 1998; Leclercq, 1999). Two main categories of adaptive strategies are used when an individual encounters both an obstacle and a more slippery zone: *"strategies of avoidance"* that consist of modifying walking patterns in order to step over the obstacle, and *"strategies of accommodation"* that consist of the modification of walking patterns in order to adapt to the low friction footwear-floor interface (Patla, 1991). The question thus arises: how these strategies interact and what kinds of corrective reactions occur in an attempt to avoid a fall.

Conventional kinematic gait analysis of slip, trip, and fall events rely on angular position-time, velocity-time, or angle-angle presentations (e.g. Cham & Redfern, 2001; Fong et al., 2005). However, such presentations do not reveal the direct relationship between velocity changes and position (Burgess-Limerick et al., 1993; Kurz et al., 2005; Van Uden et al., 2003; Winstein & Garfinkel, 1989). It is important to evaluate this relationship since the joint and muscle proprioceptors, and the visual and vestibular receptors provide sensory feedback on both velocity and position. This means that the multiple sensory cues will potentially compete for governance of the evoked behavioral response (Misiaszek, 2006). Furthermore, quantification of interjoint (e.g., thigh-shank) coordination is very difficult with the above-mentioned presentations (Burgess-Limerick et al., 1993; Davids et al., 2003; Scholz, 1990; Scholz & Kelso, 1989; Sparto et al., 1997). Coordination analysis using relative phase dynamics can solve the above problems and provide a window of particular types of causal motor control processes that are not usually revealed by conventional time-based plots (Gottlieb et al., 1983; Hamill et al., 1999; Heiderscheit et al., 1999; Kurz et al., 2005; Kwakkel & Wagenaar, 2002; Sparto et al. 1997; Van den Berg et al., 2000; Van Uden et al., 2003; Winstein & Garfinkel, 1989). Relative phase dynamics utilizes the displacements and velocities of the segments that surround a joint to quantify the joint's coordination. For example, the continuous relative phase, a measure from relative phase dynamics, quantifies the coordination between the shank and thigh segments that compose the knee joint. Such a measure is appealing for quantifying signs of gait instability because it can reveal the compensatory reactions evoked in the lower extremity coordination patterns that may be due to changing task (obstacle clearance) and environmental (low friction) demands.

Therefore, the purpose of this study was to use a coordination analysis to investigate the effects of shoe traction and obstacle height on lower extremity coordination during walking to better understand the control strategies adopted to avoid slippage following obstacle clearance in normal young adults. In this study, we examined the intralimb phasing relationships between the foot, the shank and the thigh of the landing limb (Kurz et al., 2005). We hypothesized that stepping over obstacles with low shoe traction will challenge the motor control of the neuromuscular system and will affect intralimb phasing relationships. In this study, obstacle height was adjusted to percentages (0%, 10%, 20%, and 40%) of limb length to ensure that individuals of different heights would make the same qualitative adaptation in going over obstacles.

2. Methods

2.1. Participants

Ten healthy young adult males between the ages of 18 and 35 from the general student community of the University of Nebraska at Omaha volunteered as participants (age: 25.8 ± 4.29 years; body mass: 82.8 ± 8.25 kg; height: 179.6 ± 6.34 cm; leg length — as measured from the right anterosuperior iliac spine to the right lateral malleolus: 95.6 ± 4.49 cm; shoe size: 10). All participants were without appreciable leg length discrepancy and had no injuries or abnormalities that would affect their gait. Prior to testing, each participant provided an informed consent approved by the University of Nebraska Medical Center Institutional Review Board.

2.2. Instrumentation

A sagittal view of the right lower extremity was obtained for all trials using a Panasonic WV-CL350 (Osaka, Japan) video camera with a sampling frequency of 60 Hz. The video camera was located 8-meter perpendicular to the walking pathway. A zoom lens (COSMICAR TV, 8–48 mm zoom lens, COSMICAR/PENTAX Precision Co., Tokyo, Japan) was used in conjunction with the video camera to optimize image size and minimize perspective error. A light source (Pallite VIII using eight ELH 300 W tungsten-halogen projection lamps at 120 VAC) was mounted with the camera lens in the center of the ring to better illuminate the reflective markers.

Reflective markers were positioned on the participant's right lower extremity, here referred to as the leading limb (i.e. the limb crossing the obstacle first). All positional markers were placed on the participants by the same examiner. Sagittal plane marker placement was as follows: (1) mid-distance between the greater trochanter of the hip and the lateral joint line of the knee, (2) lateral joint line of the knee, (3) lateral malleolus, (4) outsole of the shoe approximately at the bottom of the calcaneus, and (5) outsole of the shoe approximately at the fifth metatarsal head. An additional marker was positioned at the obstacle to assist in determining the location of the obstacle in the field of view.

The video images were stored on SVHS video tapes via a Panasonic AG-1970P video camera recorder, which was interfaced with a Magnavox TV for an instant qualitative evaluation of the video recording. The video data were transformed to digital format and digitized via the PEAK MOTUS video system (Peak Performance Technologies, Inc., Englewood, CO). A single camera was used because sagittal view measures of walking correspond well in two- and three-dimensions (Doriot & Cheze, 2004; Eng & Winter, 1995). GRF data were also collected using a force platform. These data were presented elsewhere (Houser et al., 2008).

Two pairs of men's shoes (Pro-wing Joggers, size 10), with homogenous midsoles and rubber outsoles, were used in this experiment. The same shoes and shoe size were used for all participants to minimize any effects from the shoe characteristics on the results of the study. The shoe size of 10 was selected because it is the most common shoe size among males in the USA. To decrease the coefficient of friction (COF) of one pair of the shoes, without significantly modifying their weight, flexibility and general performance, 88 metallic one-half inch diameter disc thumbtacks were inserted into the outsole of both the left and right shoe. The thumbtacks were carefully placed in order to ensure that no part of the actual shoe was able to contact the ground during walking locomotion. They were also roughed and cleansed to expose the metal originally covered with enamel. The thumbtacks increased the weight of the shoes by 25 g (475 g without the tacks vs. 500 g with the tacks). The pair with the high traction had dynamic COF (DCOF) of 0.7 and static COF (SCOF) of 0.8. The pair with the low traction had DCOF of 0.3 and SCOF of 0.35. The two selected tractions were based upon previous literature (Perkins, 1978; Denoth, 1989) and pilot test work suggesting the high traction pair was a very safe shoe, while the low a borderline safe shoe. Both high and low traction shoes were roughed with 20 passes of the 100 grit sand paper, and then the surfaces were cleansed with rubbing alcohol to remove from the outsoles any solvents or residues of the shoe manufacturing process.

2.3. Experimental protocol

Participants wore shoes provided by the investigator, and minimal clothing to achieve correct positioning of reflective markers by using the anatomical landmark points. They were given ample time to acclimate to the experimental set-up prior to testing. Walking trials were conducted on a 30-meter level oval track with a 0.6 meter wide lane; however, data were not recorded along the curved portion of the walkway. Data collection was performed along the straight 10-meter walkway section of the track; the force platform is embedded at the middle of this straight walkway. Walking speed was monitored around the location of the force platform and over a 3-m interval using a custom-made photocell timing system.

During familiarization, the investigator asked the participants if there was any shoe discomfort that may alter their natural gait. If no problems were reported, the participants proceeded in establishing a comfortable self-selected walking speed which was recorded. Based upon the participant's self-selected walking speed, a range that allowed ±5% deviation of this speed was used for the subsequent testing and a trial was considered acceptable only when the walking speed was within this predetermined range. The investigator also asked the participants not to look at the floor to locate the force platform for proper right foot placement, as this could influence the participant's natural walking. For this purpose, a foot placement marker was located approximately 7 m before the force platform to allow for a normal right foot contact with the force platform. This distance was determined through trial and error during the practice trials. Each trial was visually monitored to insure that the stride was normal and the foot was completely on the force platform. Data transfer from the cameras to the computer allowed for an inter-trial rest interval of one minute.

All participants were asked to walk at their previously established self-selected speed under four different obstacle conditions. The first condition was walking on a level surface while the other three conditions were walking over obstacles of three different heights. The average height of the obstacles was approximately: 8–10 cm (low, 10% leg length), 18–20 cm (medium, 20% leg length) and 36–40 cm (high, 40% leg length). These obstacle heights were established based upon pilot work, previous literature and obstacle dimensions commonly encountered in the everyday environment (Chen et al., 1991; Chen et al., 1994; Chou & Draganich, 1997; Patla et al., 1991; Patla & Rietdyk, 1993; Patla et al., 1996). The 10% obstacle height characterizes door thresholds, the 20% obstacle height represents typical curbstones separating cars in parking lots and stair risers, and 40% obstacle height corresponds to bathtub rims, where frequent falls occur especially among the elderly. The obstacles were placed directly before the force platform so that the participant had to clear the obstacle with the right leg and land on the force platform. The obstacles were made of light weight wood so that if a participant stepped on or hit the obstacle by mistake while walking, the obstacle was destroyed. This minimized the risk of tripping and falling. All participants were required to complete the baseline and obstacle conditions with the two pairs of shoes (high and low traction outsole) as described previously.

Each experimental condition (shoe traction × obstacle) consisted of ten trials for a total of eighty trials per participant. The order of the presentation of conditions was predetermined as follows: (1) low traction – 0%; (2) low traction – 10%; (3) low traction – 20%; (4) low traction – 40%; (5) high traction – 0%; (6) high traction – 10%; (7) high traction – 20%; (8) high traction – 40%. Furthermore, participants were given several practice trials prior to each condition to familiarize themselves with the task and the environmental constraints.

2.4. Data reduction and analysis

Kinematic data were analyzed during the stance period only. All kinematic coordinates were scaled and smoothed using a Butterworth low-pass filter with a selective cut-off algorithm based on Jackson (1979). The cut-off values were 8–14 Hz. Subsequently, from the planar coordinates, foot, shank, and thigh angular displacements were calculated in a counter-clockwise direction relative to the right horizontal axis. From the angular displacements, the angular velocities were calculated using a finite difference approach. All kinematic angular displacements and velocities were normalized to 100 points for the stance period using a cubic spline routine to enable mean ensemble curves to be derived for each participant and for each condition. The touchdown and toe-off timing occurrences as well as the transition time (crossover) from braking to propulsion were identified from the anterior-posterior ground reaction force data using laboratory software. Since the kinetic and kinematic data files were time matched, the time of the transition (crossover) from braking to propulsion was used to evaluate each footfall for two periods: (1) heel contact to transition (absorption or braking period) and (2) transition to toe-off (propulsion period). It was decided to divide the stance period at the transition time (crossover) from braking to propulsion for two main reasons: (1) this event separates the absorption and propulsion periods, during which different kinematic strategies may exist (Bates et al.,1978), (2) the measurements over the entire stance can mask differences for a single period (Byrne et al., 2002).

The angular kinematic data were then subjected to a coordination analysis (Kurz & Stergiou, 2004a, 2004b, 2004c; Kurz et al., 2005; Scholz, 1990; Stergiou, 2001a, 2001b). Phase portraits for the sagittal foot, shank and thigh were generated. A phase portrait is a plot of a segment's angular displacement *versus* its angular velocity (Barela et al., 2000; Winstein & Garfinkel, 1989). The angular displacements and velocities were normalized to their maximum absolute values (Van Emmerik & Wagenaar, 1996; Kurz & Stergiou, 2004c). The resulting phase plane trajectories were then transformed from Cartesian (x, y) to polar (r, θ) coordinates, where the radius was $r = (x^2 + y^2)^{1/2}$ and the phase angle was $\theta = \tan^{-1} [y/x]$ (Kurz & Stergiou, 2002; Kurz et al., 2005; Rosen, 1970). Phase angles calculated from these trajectories had a range of 0° to 180°. Phase angles allow for the incorporation of angular displacements and velocities to examine coordinative strategies. Subsequently, the normalized phase angles were used to determine the phasing relationships between the segments. The foot and the shank can be viewed as respectively rotating clockwise and counterclockwise around the ankle joint axis, while the shank and the thigh can be viewed as rotating clockwise and counterclockwise around the knee joint axis. Continuous relative phase represents the phasing relationships or coordination between the actions of the two interacting segments at every point during a specific time period (i.e., it depicts how the two segments are coupled in their movements while performing the task) (Barela et al., 2000; Hamill et al., 1999; Heiderscheit et al., 1999, 2000; Kwakkel & Wagenaar, 2002; Scholz, 1990). Relative phase was calculated throughout the stance period by subtracting the phase angles of the corresponding segments:

$$\Phi_{\text{SAGITTAL ANKLE RELATIVE PHASE}}=\Phi_{\text{FOOT}}-\Phi_{\text{SHANK}}$$

and

$$\Phi_{\text{SAGITTAL KNEE RELATIVE PHASE}}=\Phi_{\text{SHANK}}-\Phi_{\text{THIGH}}$$

Values close to 0° indicate that the two segments are moving in a similar fashion or in-phase. Values close to 180° indicate that the two segments are moving in opposite directions or out-of-phase. The relative phase curves for each segmental relationship (ankle and knee) were averaged across trials, and mean ensemble curves were generated for each participant for all conditions. The participant mean ensemble curves were also averaged to generate group mean ensemble curves for all conditions. However, to statistically test differences between relative phase curves, it was necessary to characterize the curves by single numbers. Therefore, two additional parameters were calculated using the participant mean ensemble curves (Byrne et al., 2002; Hamill et al., 1999; Heiderscheit et al., 1999, 2000; Kurz & Stergiou, 2004b, 2004c; Stergiou et al., 2001a, 2001b; Van Emmerik & Wagenaar, 1996).

The first parameter was the Mean Absolute value of the ensemble Relative Phase curve values (MARP). This parameter was calculated by averaging the absolute values of the ensemble curve points for the designated periods (stance, absorption and propulsion):

$$MARP=\sum_{i=1}^{N}\frac{|\Phi_{\text{RELATIVE PHASE}}|}{N}$$

where N is the number of points in the relative phase mean ensemble. Functionally, a low MARP value indicated that the oscillating segments have a more in-phase coordinated relationship; a high MARP value indicates that the oscillating segments have a more out-of-phase coordinated relationship.

The second parameter was the Deviation Phase (DP) of the relative phase for the two interacting segments provides a measure of stability of the neuromuscular system. Deviation phase was calculated by averaging the standard deviations of the ensemble relative phase curve points for the designated periods (stance, absorption and propulsion):

$$DP=\frac{\sum_{i=1}^{N}|SD_i|}{N}$$

where N is the number of points in the relative phase mean ensemble and SD is the standard deviation of the mean ensemble at the i^{th} point. Functionally, a low DP value indicates a more stable organization of the neuromuscular system (i.e., a less variable relationship between the two segments' actions); a high DP value indicates less stability in the organization of the neuromuscular system.

2.5. Statistical treatment

Group means and standard deviations were calculated for MARP and DP for each segmental relationship, for each period, and for each condition. A two-way repeated measures ANOVA (shoe traction × obstacle) was performed on the group means for MARP and DP. Statistical analysis was performed for each coordinative relationship (foot-shank and shank-thigh) and for each period (stance, absorption and propulsion). In tests that resulted in significant F-ratios ($P < 0.05$), post-hoc analysis was performed using Tukey tests.

3. Results

The shank-thigh (S-T) MARP group results were statistically significant for both factors (shoe traction × obstacle) during all the three periods (stance, braking, and propulsion; Tables 1 and 2). The S-T MARP values were significantly larger for the high traction shoe, and decreased as the obstacle height increased in both shoes. Specifically, the decrease in the S-T MARP values was symmetrical between obstacle conditions for the stance and the propulsive periods in both shoes. However, for the braking period this decrease was only noticeable from the level walking to the 10% obstacle conditions; the post-hoc analysis showed no statistical differences between the obstacle conditions. Regarding the foot-shank (F-S) MARP group results, no statistically significant differences were found between conditions.

The DP group results were statistically significant for both S-T and F-S segmental relationships for all three periods analyzed regarding the obstacle factor (Tables 1 and 2). For the shoe factor all S-T segmental relationships were significant, while for the F-S only the propulsive period was significantly different. All the DP group results increased in value as the obstacle height increased for both shoes. Furthermore, the S-T DP results were larger for the low traction shoes for all periods.

Graphically, the thigh segment during the stance phase showed a segmental reversal which occurs towards the later part of stance (Figure 1a). Functionally, every time that a trajectory goes through zero a segmental reversal is observed. It is worth noting that the thigh exhibited a fairly constant velocity during the middle part of the stance period, especially in the no obstacle conditions. Constant velocity is depicted by flat horizontal sections. However, spatial aspects of the phase portraits expressed the same general shape from one condition to the next. However, in level-walking (0% obstacle) and low-obstacle (10%) conditions, low traction shoes caused an additional curve segment to be developed within the original pattern during late stance. The shank segment phase portraits revealed no reversals, indicating a backward only rotation around the knee joint during the stance phase of walking (Figure 1b). However, the foot behaved differently than the other lower extremity segments (Figure 1c). The foot segment during stance displayed a cusp shape. Cusps in the foot trajectory path, when the velocity is near zero, indicate sudden interruption in the movement pattern. This is due to the fact that the foot remained flat on the ground for a period of time during midstance. The foot trajectories were more similar geometrically between conditions; however, the angular velocity of the foot segment appeared to increase during the later part of the stance period in the high traction condition, as compared to the low traction situation. This observation was graphically visible thought a more pronounced concave-down configuration during the later portion of the stance phase.

The group mean ensemble foot-shank (F-S) and shank-thigh (S-T) relative phase curves for the stance period are displayed in Figure 2. In general, it can be observed that segmental relative phase relationships are non linear i.e., neither in-phase (~0° values) nor out-of-phase (~180° values) by a constant magnitude during stance. In addition, during level-walking, F-S and S-T relative phases began differently than for obstacle conditions. Indeed, both segmental relative phases began around 0° for the no obstacle conditions, whereas F-S relative phase began around −25° and S-T relative phase around +50° for the obstacle conditions. Therefore, the effect of the obstacle on the relative phase caused the segments to be more out of phase at touchdown.

The group mean ensemble F-S relative phase curves had similar configurations for all conditions (Figure 2a). All curves began with negative values (or negative zero values for the level-walking conditions) that indicated that the shank was leading the foot (i.e., the shank was moving faster in phase space) during the first initial portion of stance. Early in stance, the relationship between the foot and shank reversed. Reversal in the relationship between the two segments was evident by the local minimum in the relative phase graph. The positive slope after the local minimum indicated that the foot was leading the shank segment (i.e., the foot was moving faster in phase space). During mid-stance, the foot-shank relationship became more out of phase, and the foot clearly was leading the shank (positive values: 25–50°). Moreover, there was not a distinct (unique) local maximum in the F-S relative phase. In fact, inspection of the F-S relative phase curve indicated that there were multiple fluctuations during midstance. Local minimums and maximums suggest a change in direction of the relationship between the two segments. During the late portions of the stance, the relationship between the foot and shank became progressively in-phase.

The group mean ensemble S-T relative phase curves also displayed quite similar trends (Figure 2b). For the obstacle conditions, all curves began with positive values that indicated that the shank was leading the thigh. Immediately after the shank-thigh relationship became more in-phase (0°) during mid-stance. During late stance, the relationship between the shank and thigh became progressively out of phase with the thigh leading. A slightly different segmental relationship occurred for the level-walking conditions. As previously mentioned, the S-T segments began more in-phase (i.e., close to zero degree). However, early in stance the relative phase became more-out-of-phase with the thigh leading the shank before returning to a more in-phase relationship throughout the middle portion of the stance period. During late stance, the relationship became progressively out of phase, similarly to what was observed in the obstacle conditions.

4. Discussion

Both our graphical and statistical results revealed that stepping over obstacles of different height with shoes of varied traction may affect the motor control of the neuromuscular system and will affect intralimb phasing relationships. The partitioning of the stance period assisted in further understanding the strategies used and better evaluate the results functionally.

Specifically, the phasing relationship between the foot and the shank segments was not affected by either shoe traction or obstacle height changes. Graphically, ensemble curves displaying the F-S relative phase support this statistical result (Figure 4a). However, the more in-phase relationship observed at initial foot contact in walking-level conditions did change to become more out-of-phase in obstacle conditions. Furthermore, early in stance, the magnitude of the curves' concavity was more prominent when the obstacle height was increased. Even though these differences were not found to be significant, probably due to the large similarities of the curves throughout the remaining portion of the braking and the stance periods, they may be important due to the increased danger of slipping during the braking period. Indeed, according to Perkins (1978), the most dangerous slipping is most likely to occur in this period due to a low initial vertical ground reaction force at heel strike, which produces a small amount of friction. If friction is not sufficient during the braking period, an anterior slip of the foot would likely occur. This slip could be particularly dangerous due to the rapid transfer of weight to the landing foot.

Contrary to F-S, S-T MARP showed significant differences for both factors during all periods. The introduction of low traction shoes had as a result a more in-phase relationship. Furthermore, the increasing obstacle height resulted in a more in-phase relationship for both shoes. Thus, it seems that both independent variables affected the coordinative behavior of the system at the knee. The more in-phase S-T segmental relationship may indicate a tendency towards a behavioral change that eventually could result in the emergence of a new behavioral state (i.e., slipping and/or falling). Therefore, this relationship deserves more attention in future ergonomics studies that want to explore the relationship between shoe-floor traction following obstacle clearance.

In the present study, stability of the coordinated relationship between the two interacting segments was measured by DP which describes the variability of the relative phase. An interesting observation is that the increases in obstacle height resulted in significantly increased F-S DP values. Thus, even though the F-S relative phase remained similar (as indicated by the lack of differences for F-S MARP values), the F-S DP increased significantly as the obstacle height increased. This result can be explained as an increased instability based on the theoretical premises of the coordination analysis performed (Kurz & Stergiou, 2004b). For the shoe traction factor, only F-S DP values during propulsive period showed significant differences. The fact that DP increased for the F-S segmental relationship in the low traction shoes indicates instability and lack of coordination at the ankle joint during the propulsive period. This is further supported by the fact that the smaller F-S DP values during propulsion were present at the high traction conditions, which theoretically means that when the system is under normal preferred conditions (normal walking) it would be highly stable. Furthermore, the S-T DP increased as the obstacles height increased for both shoes in all periods. Moreover, the low traction shoe had generally larger S-T values. These findings further supported the hypothesis that increased instability would be present when the obstacle height and shoe traction changed and became more unsafe.

Previously, angle-angle diagrams have been used to depict the organization of the multiple degrees of freedom needed to complete one walking gait cycle (Grieve, 1968), and several investigators have suggested methods for quantifying the coordination that is qualitatively observable in these relative patterns (Sidaway et al., 1995; Sparrow et al., 1987; Whiting and Zernicke, 1982). However, quantitatively understanding the control mechanisms cannot be achieved with this methodology alone (Burgess-Limerick et al., 1993). The usage of phase portraits and subsequently of relative phase, allows the incorporation of both angular displacement and velocity to examine coordination and movement (Kurz & Stergiou, 2004a, 2004b, 2004c; Kurz et al., 2005; Kwakkel & Wagenaar, 2002; Scholz, 1990). Functionally, this approach is advantageous since there is evidence that receptors exist within the muscles and tendons for controlling both displacement and velocity (McCloskey, 1978). This is a particular strength of the present study.

However, we should also consider several limitations of our study. First and most importantly, the sample procedure lacks randomization. Indeed, from a practical and methodological point of view randomization was in fact difficult to achieve. Because the participants knew the shoe condition, it was not possible to eliminate the awareness of a potential slip/fall (while wearing shoes with low friction) or trip/fall (while avoiding obstacles). Accordingly, some caution with regard to generalization of the results must be taken due to the lack of randomization. On the other hand, in our pilot work we found that the order of the testing conditions did not reveal significant learning effects. Additionally, participants were given one or more practice trials prior to each condition to familiarize themselves with the task and the environmental constraints. Our results in terms of gait adaptations are also in agreement with those found in the literature (obstacle clearance strategies: e.g. Begg et al., 1998; Chen et al., 1991; corrective reactions to slip events: e.g. Perkins, 1978; Frederick, 1993; Cham & Redfern, 2001).

A second limitation of the present study is the extent to which the findings can be generalized, as it is not possible to know how the laboratory slipping responses differed from those that occur in a non-laboratory environment (Brady et al., 2000). It has been proven that reproducing the unexpected nature of real-life slip, trip, and fall accidents in laboratory settings is quite difficult. Therefore, the conclusions reported here underline the importance of being conservative when applying research findings from slip, trip, and fall experiments using human participants to design criteria of environmental safety (e.g. friction requirements).

However, the findings of this investigation can provide the necessary foundation to further investigate the coordinative control strategies utilized in more challenging environments that may actually be associated with slips and falls. Additionally, further investigation should be conducted to explore the anticipatory intralimb coordinative strategies leading up to the stance phase of gait, as such adaptations could be critical in order to successfully avoid slips, trips, and eventual falls. From an ergonomic perspective, such investigations can have crucial implications to slip, trip, and fall injury-prevention strategies in occupational and non-occupational environments, and how a potentially hazardous situation is perceived.

5. Conclusions

Our approach provided information to understand how young healthy adults change their gait to reduce the likelihood of a slip following clearance of obstacles of varied heights and landing with shoes of low traction. The changes in phasing relationship and variability are possible indicators of alterations in coordinative behavior, which may emerge to reduce the risk to the participant when confronted with an environment characterized by low traction and high obstacles. Changes do not suggest that falling or slipping did or will occur. However, if shoe traction and/or obstacle height would have been more extreme, falling may have occurred. Qualitative analysis during data processing did reveal that slippage occurred at initial foot contact. This slippage was of the type “slip-sticks” as described by Standberg and Lanshammar (1981). These slips never resulted in obvious postural or upper extremity adjustments. Slipping also occurred late in the propulsive period just prior to toe-off. This slipping was of little consequence, due to majority of weight acceptance to the opposing limb (Perkins, 1978).

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References

Barela J, Whitall J, Black P, Clark JE. An examination of constraints in patients with chronic hemiparetic gait. Hum Mov Sci 2000;19:251–73.

Bates, BT.; Osternig, LR.; Mason, BR.; James, SL. Lower extremity function during the support phase of running. In: Asmussen, E.; Jorgensen, K., editors. Biomechanics. Vol. IV-B. University Park Press; Baltimore: 1978. p. 30-9.

Begg RK, Sparrow WA, Lythgo ND. Time-domain analysis of foot-ground reaction forces in negotiating obstacles. Gait Posture 1998;7:99–109. [PubMed: 10200379]

Bentley TA, Haslam RA. Slip, trip, and fall accidents occurring during the delivery of mail. Ergonomics 1998;41:1859–72. [PubMed: 9857843]

Brady RA, Pavol MJ, Owings TM, Grabiner MD. Foot displacement but not velocity predicts the outcome of a slip induced in young subjects while walking. J Biomech 2000;33:803–8. [PubMed: 10831754]

Bring, C. Document D5: 1982. Stockholm: Swedish Council for Building Research; 1982. Testing of slipperiness: forces applied to the floor and movements of the foot in walking and in slipping on the heel.

Burgess-Limerick R, Abernethy B, Neal RJ. Relative phase quantifies interjoint coordination. J Biomech 1993;26:91–4. [PubMed: 8423174]

Burnfield JM, Powers CM. Prediction of slips: an evaluation of utilized coefficient of friction and available slip resistance. Ergonomics 2006;49:982–95. [PubMed: 16803728]

Byrne JE, Stergiou N, Blanke D, Houser JJ, Kurz MJ, Hageman PA. Comparison of gait patterns between younger and elderly women: an examination of coordination. Percept Mot Skills 2002;94:265–80. [PubMed: 11883574]

Cham R, Redfern MS. Lower extremity corrective reactions to slip events. J Biomech 2001;34:1439–45. [PubMed: 11672718]

Cham R, Redfern MS. Changes in gait when anticipating slippery floors. Gait Posture 2002a;15:159–71. [PubMed: 11869910]

Cham R, Redfern MS. Heel contact dynamics during slip events on level and inclined surfaces. Saf Sci (Special Issue) 2002b;40:559–76.

Chen HC, Ashton-Miller JA, Alexander NB, Schultz AB. Stepping over obstacles: Gait patterns of healthy young and old adults. J Gerontol 1991;46:M196–M203. [PubMed: 1940078]

Chen HC, Ashton-Miller JA, Alexander NB, Schultz AB. Age effects on strategies to avoid obstacles. Gait Posture 1994;2:139–46.

Chen HL, Lu TW. Comparisons of the joint moments between leading and trailing limb in young adults when stepping over obstacles. Gait Posture 2006;23:69–77. [PubMed: 16311197]

Chou L, Draganich L. Stepping over an obstacle increases the motions and moments of the joint of the trailing limb in young adults. J Biomech 1997;30:331–7. [PubMed: 9075000]

Cohen HH, Compton DM. Fall accident patterns: characterization of most frequent work surface-related injuries. Prof Saf 1982;27:16–22. [PubMed: 10255743]

Courtney TK, Sorock GS, Manning DP, Collins JW, Holbein-Jenny MA. Occupational slip, trip, and fall-related injuries - can the contribution of slipperiness be isolated? Ergonomics 2001;20:1118–37. [PubMed: 11794761]

Davids K, Glazier P, Araujo D, Bartlett R. Movement systems as dynamical systems: the functional role of variability and its implications for sports medicine. Sports Med 2003;33:245–60. [PubMed: 12688825]

Denoth, J. Indoor athletic playing surfaces: floor vs. shoe. In: Segesser, B.; Pforringer, W., editors. The shoe in sport. Year Book Medical Publishers, Inc; Chicago: 1989. p. 65-9.

Doriot N, Cheze L. A three-dimensional kinematic and dynamic study of the lower limb during the stance phase of gait using an homogeneous matrix approach. IEEE Trans Biomed Eng 2004;51:21–7. [PubMed: 14723490]

Eng JJ, Winter DA. Kinetic analysis of the lower limbs during walking: what information can be gained from a three-dimensional model? J Biomech 1995;28:753–58. [PubMed: 7601875]

Englander F, Hodson TJ, Terregrossa RA. Economic dimensions of slip and fall injuries. J Forensic Sci 1996;41:733–46. [PubMed: 8789837]

Fong DT, Hong Y, Li JX. Lower-extremity gait kinematics on slippery surfaces in construction worksites. Med Sci Sports Exerc 2005;37:447–54. [PubMed: 15741844]

Frederick, EC. Optimal frictional properties for sport shoes and sport surfaces. In: Hamill, J.; Derrick, TR.; Elliot, EH., editors. Biomechanics in Sports XI. Proceedings of the XIth Symposium of the International Society of Biomechanics in Sports; Amherst, MA: University of Massachusetts; 1993. p. 15-22.

Gao C, Abeysekera J. Slips and falls on ice and snow in relation to experience in winter climate and winter sport. Saf Sci 2003;42:537–45.

Gao C, Abeysekera J, Hirvonen M, Grönqvist R. Slip resistant properties of footwear on ice. Ergonomics 2004;47:710–16. [PubMed: 15204296]

Gottlieb GL, Agarwal GC, Jaeger RJ. Response of sudden torques about ankle in man. V. Effect of peripheral ischemia. J Neurophys 1983;50:297–312.

Grieve DW. Gait patterns and the speed of walking. Biomed Eng 1968;3:119–22.

Grönqvist, R.; Roine, J. Serious occupational accidents caused by slipping. In: Nielsen, R.; Jorgensen, R., editors. Advances in Industrial Ergonomics and Safety. Vol. V. Taylor and Francis Publishers; London: 1993. p. 515-19.

Hamill J, van Emmerik REA, Heidercsheit BC, Li L. A dynamical systems approach to lower extremity running injuries. Clin Biomech 1999;14:297–308.

Heiderscheit BC, Hamill J, Caldwell GE. Influence of Q-angle on lower-extremity running kinematics. J Orthop Sports Phys Ther 2000;30:271–8. [PubMed: 10817414]

Heiderscheit BC, Hamill J, Van Emmerik RE. Q-angle influences on the variability of lower extremity coordination during running. Med Sci Sports Exerc 1999;31:1313–9. [PubMed: 10487374]

Houser JJ, Decker LM, Stergiou S. Stepping over obstacles of different heights and varied shoe traction alter the kinetic strategies of the leading limb. Ergonomics 2008;7:1–13.

Jackson KM. Fitting of mathematical functions to biomechanical data. IEEE Trans Biomed Eng 1979;26:515–21.

Jaffe DL, Brown DA, Pierson-Carey CD, Buckley EL, Lew HL. Stepping over obstacles to improve walking in individuals with poststroke hemiplegia. J Rehabil Res Dev 2004;41:283–92. [PubMed: 15543446]

James DI. A broader look at pedestrian friction. Rubber Chem Tech 1980;53:512–41.

Kemmlert K, Lundholm L. Slips, trips and falls in different work groups with reference to age. Saf Sci 1998;28:59–75.

Kurz M, Stergiou N. The effect of normalization and phase angle calculations on continuous relative phase. J Biomech 2002;35:369–74. [PubMed: 11858813]

Kurz, MJ.; Stergiou, N. Mathematical measures of coordination and variability in gait patterns. In: Stergiou, N., editor. Innovative analyses of human movement. Human Kinetics Publ; Champaign: 2004a. p. 163-86.

Kurz, MJ.; Stergiou, N. Applied dynamic systems theory for the analysis of movement. In: Stergiou, N., editor. Innovative analyses of human movement. Human Kinetics Publ; Champaign: 2004b. p. 93-119.

Kurz MJ, Stergiou N. Does Footwear Affect Ankle Coordination Strategies? J Am Podiatr Med Assoc 2004c;94:53–8. [PubMed: 14729992]

Kurz MJ, Stergiou N, Buzzi UH, Georgoulis AD. The effect of anterior cruciate ligament reconstruction on lower extremity relative phase dynamics during walking and running. Knee Surg Sports Traumatol Arthrosc 2005;13:107–15. [PubMed: 15756615]

Kwakkel G, Wagenaar RC. Effect of duration of upper- and lower-extremity rehabilitation sessions and walking speed on recovery of interlimb coordination in hemiplegic gait. Phys Ther 2002;82:432–48. [PubMed: 11991797]

Leclercq S. Prevention of same level falls: a more global appreciation of this type of accident. J Saf Res 1999;30:103–12.

Lockhart TE, Smith JL, Woldstad JC. Effects of aging on the biomechanics of slips and falls. Hum Fact 2005;47:708–729.

Lockhart TE, Woldstad JC, Smith JL. Effects of age-related gait changes on the biomechanics of slips and falls. Ergonomics 2003;46:1136–60. [PubMed: 12933077]

Leamon TB, Murphy PL. Occupational slips and falls: more than a trivial problem. Ergonomics 1995;38:487–98. [PubMed: 7729391]

Manning DP, Ayers I, Jones C, Bruce M, Cohen K. The incidence of underfoot accidents during 1985 in a working population of 10,000 Merseyside people. J Occup Acc 1988;10:121–30.

McCloskey DI. Kinesthetic sensibility. Physiol Rev 1978;58:763–820. [PubMed: 360251]

McFadyen BJ, Prince F. Avoidance and accommodation of surface height changes by healthy, community-dwelling, young, and elderly men. J Gerontol A Biol Sci Med Sci 2002;57:B166–74. [PubMed: 11909882]

Misiaszek JE. Neural control of walking balance: if falling then react else continue. Exerc Sport Sci Rev 2006;34:128–34. [PubMed: 16829740]

Moyer BE, Chambers AJ, Redfern MS, Cham R. Gait parameters as predictors of slip severity in younger and older adults. Ergonomics 2006;49:329–43. [PubMed: 16690563]

National Safety Council (NSC). International Accident Facts. Itasca, IL, USA: 1995.

Patla AE, Prentice SD, Robinson C, Neufeld J. Visual control of locomotion: strategies for changing direction and for going over obstacles. J Exp Psychol Hum Percept Perform 1991;3:603–34. [PubMed: 1834781]

Patla AE, Rietdyk S. Visual control of limb trajectory over obstacles during locomotion: effect of obstacle height and width. Gait Posture 1993;1:45–60.

Patla AE, Rietdyk S, Martin C, Prentice S. Locomotor patterns of the leading and the trailing limbs as solid and fragile obstacles are stepped over: some insights into the role of vision during locomotion. J Mot Behav 1996;28:35–47. [PubMed: 12529222]

Perkins, PJ. Measurement of slip between the shoe and ground during walking. In: Anderson, C.; Senne, J., editors. Walkway surfaces: measurement of slip resistance. American Society for Testing and Materials; Baltimore: 1978. p. 71-87.

Perkins PJ, Wilson P. Slip resistance testing of shoes-new developments. Ergonomics 1983;26:73–82.

Petrarca M, Rosa D, Cappa P, Patanè F. Stepping over obstacles of different heights: kinematic and kinetic strategies of leading limb in hemiplegic children. Gait Posture 2006;24:331–41. [PubMed: 16359868]

Redfern MS, Dipasquale J. Biomechanics of descending ramps. Gait Posture 1997;6:119–25.

Rosen, R. Dynamical Systems Theory in Biology. Volume I: Stability Theory and Its Applications. Wiley-Interscience; New York, NY, USA: 1970.

Scholz JP. Dynamic pattern theory–some implications for therapeutics. Phys Ther 1990;70:827–43. [PubMed: 2236226]

Scholz JP, Kelso JAS. A quantitative approach to understanding the formation and change of coordinated movement patterns. J Mot Behav 1989;21:122–44. [PubMed: 15132941]

Sidaway B, Heise G, Schoenfelder-Zohdi B. Quantifying the variability of angle-angle plots. J Hum Mov Stud 1995;29:181–97.

Sparrow WA, Donovan E, van Emmerik R, Barry EB. Using relative motion plots to measure changes in intra-limb and inter-limb coordination. J Mot Behav 1987;19:115–29.

Sparrow WA, Shinkfield AJ, Chow S, Begg RK. Characteristics of gait in stepping over obstacles. Hum Mov Sci 1996;15:605–22.

Sparto PJ, Parnianpour M, Reinsel TE, Simon S. The effect of fatigue on multijoint kinematics, coordination, and postural stability during a repetitive lifting test. J Orthop Sports Phys Ther 1997;25:3–12. [PubMed: 8979170]

Strandberg L. On accident analysis and slip-resistance measurement. Ergonomics 1983;26:11–32. [PubMed: 6832129]

Standberg L, Lanshammar H. The dynamics of slipping accidents. J Occup Accid 1981;3:153–62.

Stergiou N, Jensen JL, Bates BT, Scholten SD, Tzetzis G. A dynamical systems investigation of lower extremity coordination during running over obstacles. Clin Biomech 2001a;16:213–21.

Stergiou N, Scholten SD, Jensen JL, Blanke D. Intralimb coordination following obstacle clearance during running: the effect of obstacle height. Gait Posture 2001b;13:210–20. [PubMed: 11323227]

Van Uden CJ, Bloo JK, Kooloos JG, van Kampen A, de Witte J, Wagenaar RC. Coordination and stability of one-legged hopping patterns in patients with anterior cruciate ligament reconstruction: preliminary results. Clin Biomech (Bristol, Avon) 2003;18:84–7.

Van den Berg C, Beek PJ, Wagenaar RC, van Wieringen PC. Coordination disorders in patients with Parkinson's disease: a study of paced rhythmic forearm movements. Exp Brain Res 2000;134(2): 174–186. [PubMed: 11037284]

Van Emmerik REA, Wagenaar RC. Effects of walking velocity on relative phase dynamics in the trunk in human walking. J Biomech 1996;29:1175–84. [PubMed: 8872274]

Whiting WC, Zernicke RF. Correlation of movement patterns via pattern recognition. J Mot Behav 1982;14:135–42. [PubMed: 15155175]

Winstein CJ, Garfinkel A. Qualitative dynamics of disordered human locomotion: a preliminary investigation. J Mot Behav 1989;21:373–91. [PubMed: 15136252]

Winter, DA. The Biomechanics and Motor Control of Human Gait: Normal, Elderly, and Pathological. University of Waterloo Press; Ontario, Canada: 1991.

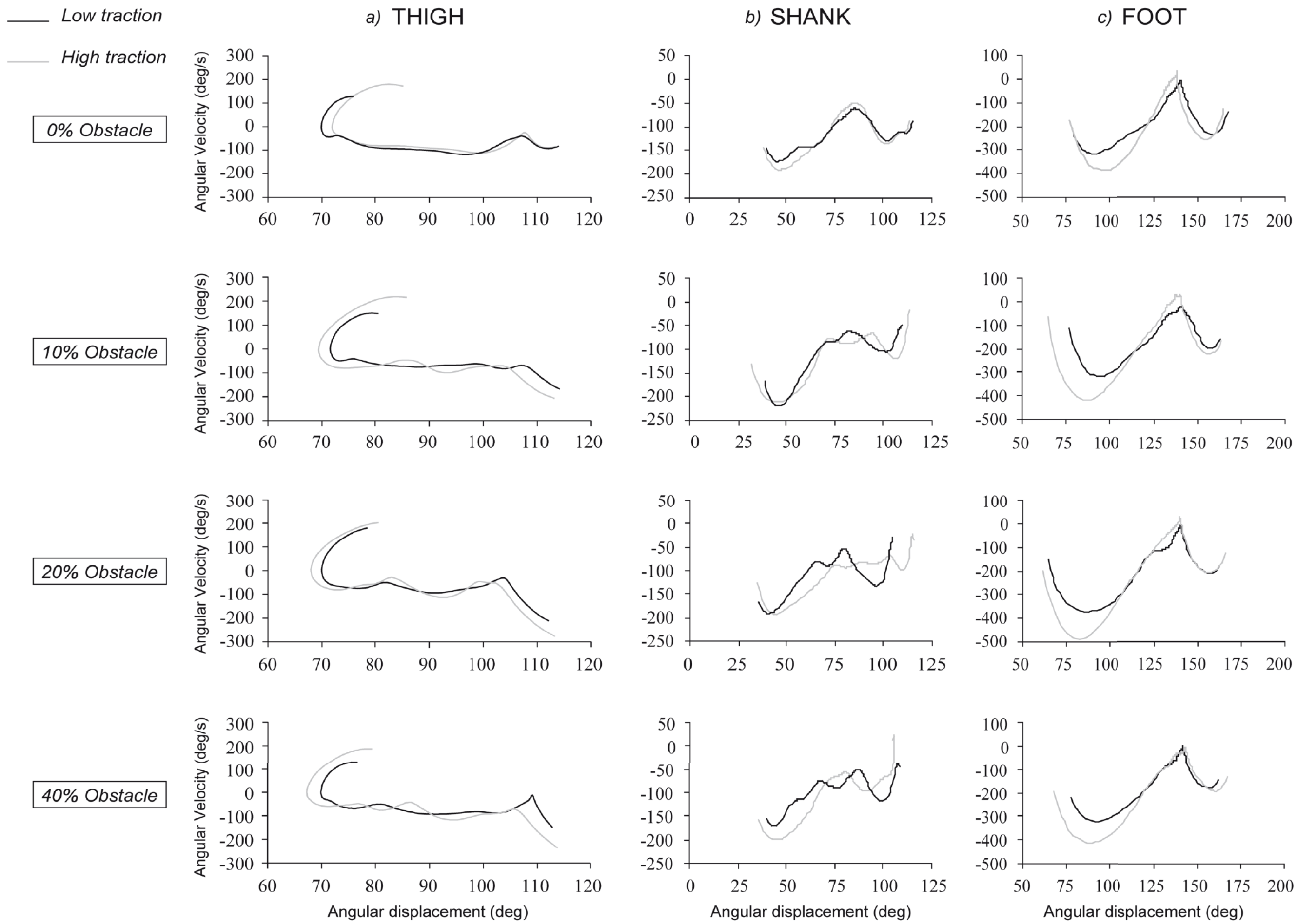

Figure 1.
Phase portraits (or phase planes) of the sagittal (a) thigh, (b) shank, and (c) foot motions from a representative trial for all conditions during stance. Black solid lines: low traction shoes; grey solid lines: high traction shoes.

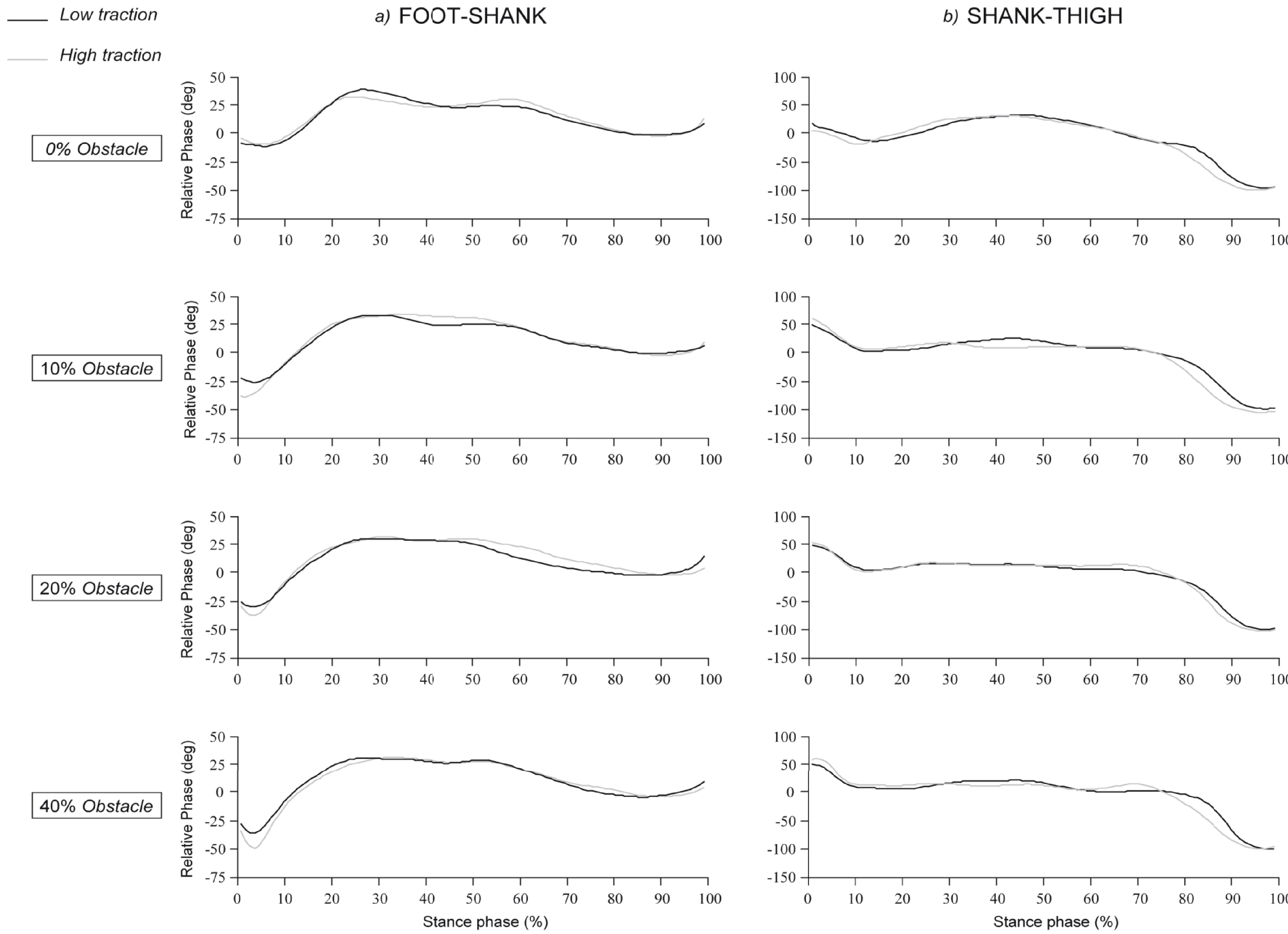

Figure 2.
Relative phase curves for the sagittal foot-shank (a) and shank-thigh (b) segmental relationships from the same representative trial for all conditions during stance. Each curve is an ensemble average over all trials. The standard deviation curves are not represented on the graphs. Black solid lines: low traction shoes; grey solid lines: high traction shoes. Heel contact occurs at 0% of the stance phase, and toe-off occurs at 100% of the stance phase.

Table 1

Group means (M) and standard deviations (SD) for MARP and DP.

Variables				Low traction				High traction			
				0%	10%	20%	40%	0%	10%	20%	40%
MARP	Stance	F-S	M	16.324	16.449	16.130	16.719	16.406	17.198	16.675	16.657
			SD	2.439	3.630	3.727	4.338	2.464	3.613	3.161	4.305
		S-T	M	$37.046^{*10,20,40\%}$	$31.279^{*40\%}$	$30.771^{*40\%}$	27.547^{*}	$41.960^{10,20,40\%}$	$36.129^{40\%}$	$33.951^{40\%}$	30.459
			SD	3.022	4.481	3.958	3.357	2.415	3.437	4.006	3.032
	Braking	F-S	M	21.178	21.650	21.834	22.214	20.335	22.264	21.765	22.303
			SD	3.079	4.262	4.591	5.252	2.973	5.105	4.549	6.599
		S-T	M	$21.253^{*10,20,40\%}$	13.993	13.361	12.278	$24.828^{10,20,40\%}$	16.594	14.567	13.241
			SD	5.504	5.450	5.912	5.031	4.459	5.227	5.360	3.284
	Propulsion	F-S	M	10.481	10.097	9.487	12.205	12.090	11.488	11.242	12.249
			SD	2.671	3.383	3.867	4.849	2.678	2.720	2.427	3.473
		S-T	M	$55.767^{*20,40\%}$	$51.790^{*40\%}$	$48.787^{*40\%}$	38.702^{*}	$60.69^{20,40\%}$	$57.136^{20,40\%}$	$52.376^{40\%}$	42.620
			SD	3.775	5.705	4.651	5.359	3.613	4.380	7.155	5.615
DP	Stance	F-S	M	$3.293^{20,40\%}$	$3.811^{20,40\%}$	$4.646^{40\%}$	5.953	$2.938^{10,20,40\%}$	$3.916^{20,40\%}$	$4.973^{40\%}$	5.631
			SD	0.376	0.473	0.738	1.296	0.470	0.671	0.905	1.037
		S-T	M	$5.826^{*10,20,40\%}$	$6.773^{20,40\%}$	8.148^{*}	8.748	$4.909^{10,20,40\%}$	$6.664^{40\%}$	$7.408^{40\%}$	8.650
			SD	0.714	1.251	1.860	1.794	0.572	1.361	1.147	1.563
	Braking	F-S	M	$3.974^{10,20,40\%}$	$5.435^{40\%}$	$6.779^{40\%}$	9.880	$3.604^{10,20,40\%}$	$5.271^{20,40\%}$	$7.705^{40\%}$	9.352
			SD	0.498	1.981	1.705	3.172	0.603	1.145	2.572	1.334
		S-T	M	$5.709^{20,40\%}$	$6.980^{\dagger 20,40\%}$	$9.094^{\dagger 40\%}$	$10.682^{\dagger}$	$5.125^{10,20,40\%}$	$7.727^{20,40\%}$	$10.082^{40\%}$	11.674
			SD	1.543	1.348	1.444	2.587	0.847	1.892	2.034	2.016

Variables			Low traction				High traction			
			0%	10%	20%	40%	0%	10%	20%	40%
Propulsion	F-S	M	2.803 *20,40%*	3.056 *20,40%*	3.789 *40%*	4.935†	2.394 *20,40%*	2.676 *20,40%*	3.769 *40%*	4.462
		SD	0.316	0.774	0.934	1.039	0.455	0.567	1.077	0.695
	S-T	M	7.276* *20,40%*	8.465* *40%*	9.582	10.704	5.610 *20,40%*	6.545 *40%*	8.042	9.518
		SD	1.448	2.001	2.573	2.830	0.571	1.204	2.290	1.951

MARP: Mean Absolute value of the ensemble Relative Phase; DP: Deviation Phase of the relative phase). The values (in degrees) presented are for each coordinative relationship (FS: foot-shank and S-T: shank-thigh) and for each period (stance, braking, and propulsion).

* significantly different between shoes within the same obstacle height ($p < 0.01$).

† significantly different between shoes within the same obstacle height ($p < 0.05$).

10,20,40% significantly different between obstacle heights within the same shoe ($p < 0.01$).

Table 2

F-ratios from the two-way ANOVA with repeated measures on both factors: shoe traction (s) × obstacles (o).

Variables			Fs	p <	Fo	p <	Fs×o	p <
MARP	Stance	F-S	0.8156	-	0.319	-	1.509	-
		S-T	79.008	0.01	39.485	0.01	3.262	0.05
	Braking	F-S	0.0084	-	1.054	-	1.020	-
		S-T	10.624	0.01	17.225	0.01	1.972	-
	Propulsion	F-S	4.431	-	2.094	-	2.551	-
		S-T	29.813	0.01	50.598	0.01	0.474	-
DP	Stance	F-S	0.2423	-	51.665	0.01	0.983	-
		S-T	14.828	0.01	37.800	0.01	0.746	-
	Braking	F-S	0.0321	-	49.562	0.01	0.631	-
		S-T	10.408	0.05	43.859	0.01	0.872	-
	Propulsion	F-S	8.296	0.05	36.772	0.01	0.460	-
		S-T	17.891	0.01	11.749	0.01	0.248	-

MARP: Mean Absolute value of the ensemble Relative Phase; DP: Deviation Phase of the relative phase; F-S: foot-shank; S-T: shank-thigh. Fs: between shoes; Fo: between obstacles; Fs×o: interaction.