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Independent influence of gait speed and step length on stability and fall risk

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ABSTRACT

With aging, individuals' gaits become slower and their steps shorter; both are thought to improve stability against balance threats. Recent studies have shown that shorter step lengths, which bring the center of mass (COM) closer to the leading foot, improve stability against slip-related falls. However, a slower gait, hence lower COM velocity, does the opposite. Due to the inherent coupling of step length and speed in spontaneous gait, the extent to which the benefit of shorter steps can offset the slower speed is unknown. The purpose of this study was to investigate, through decoupling, the independent effects of gait speed and step length on gait stability and the likelihood of slip-induced falls. Fifty-seven young adults walked at one of three target gait patterns, two of equal speed and two of equal step length; at a later trial, they encountered an unannounced slip. The results supported our hypotheses that faster gait as well as shorter steps each ameliorates fall risk when a slip is encountered. This appeared to be attributable to the maintenance of stability from slip initiation to liftoff of the recovery foot during the slip. Successful decoupling of gait speed from step length reveals for the first time that, although slow gait in itself leads to instability and falls (a one-standard-deviation decrease in gait speed increases the odds of fall by 4-fold), this effect is offset by the related decrease in step length (the same one-standard-deviation decrease in step length lowers fall risk by 6 times).

1. Introduction

Falls often lead to injuries, declines in mobility, and self-imposed limitations on daily activities and socialization [1,2]. Among older adults, falls resulting from slipping are associated with hip fractures [3] and their accompanying complications and mortality [4]. Aging related changes in various systems involved in perturbation response have been implicated in the higher rates and severity of falls among the elderly. Examples include age-related changes in coordination [5,6], increased onset times for various muscle groups [7,8], increases in joint stiffness, and decreases in isometric muscle strength [9].

With age, gait speed becomes slower and step lengths shorter [10]. It is not clear whether these changes result from the aging process or from a fear of falling, or both [11]. The evidence is contradictory as to whether either gait modification is in fact safer or more stable. Slower gaits have been shown to be directly associated with an increased fall risk [12,13], and are correlated with lower scores on clinical balance scales [14]. Several lines of research have proposed that a more quickly moving center of mass (COM), due to faster gait, may travel forward more effectively to "catch up" with the slipping base of support (BOS) [15,16]. Young and older adults spontaneously shorten their step lengths in response to a known slippery floor [17], and longer steps have been associated with a greater slip probability [18]. Findings indicate that shorter steps should be more stable because the COM is closer to the moving BOS [15]. However, Menz et al. suggested that, "step length shortening may be maladaptive" [19].

Stability can be measured as the shortest distance between the COM motion state (i.e., its position and velocity relative to the BOS) and a mathematically derived stability threshold [20]. Negative stability values (below the threshold) predict a backward balance loss [21] and are associated with falls [22]. Both a more anterior COM position, through forward leaning of the trunk and/or shortened steps [21], and a faster COM velocity, from increased gait speed, move the motion state toward the boundary, improving stability. Hence, shorter step lengths *and* faster gait speeds should enhance stability. This appears to be true only in the early instances of a slip. In a slip induced during self-selected fast, natural, or slow gaits, velocity was strongly correlated with stability at slipping foot touchdown (TD), such that faster gaits had higher stability; however, these differences diminished through the recovery response [15]. The degree to which this recovery response depends upon the gait parameters prior to slip onset is unknown [15].

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Table 1
Subject demographics, and target and resultant gait parameters, by group.

Group	N	Female	Age (yrs)	Height (m)	Weight (kg)	Target step length (/bh)	Resultant step length (/bh)	Target gait speed[a]	Resultant gait speed[a]
			Mean (SD)	Mean (SD)	Mean (SD)		Mean (SD)		Mean (SD)
A	19	8	28.9 (6.9)	1.7 (0.1)	69.2 (14.9)	0.434	.433 (.035)	0.4[b]	.396 (.039)
B	18	11	24.3 (4.9)	1.7 (0.1)	65.4 (13.5)	0.301	.309 (.027)	0.2	.203 (.017)
C	20	14	24.9 (5.1)	1.7 (0.2)	68.0 (12.6)	0.434	.422 (.027)	0.2	.204 (.013)

[a] This is a dimensionless measure: sacral velocity normalized by $\sqrt{g \times \text{bh}}$ where g is the acceleration due to gravity and bh is the subject's body height.
[b] Target for the initial eight subjects was 0.47; analysis revealed that the subjects could not consistently reach this target, so the target was adjusted to 0.4 for the remaining 11 subjects.

The independent influence of speed and step length on control of stability by liftoff of the recovery step (LO), and their impact on fall risk cannot be ascertained in spontaneous gait because of their inherent coupling: slower speeds have shorter steps and faster gaits have longer steps. It is necessary to "decouple" these parameters by controlling both. Although there is precedence for modulating one gait parameter at a time during walking, it is rare that two parameters are controlled simultaneously [23], particularly in the context of stability or fall risk.

The purpose of this study was to investigate, through decoupling, the independent effects of gait speed and step length on gait stability and the likelihood of slip-induced falls. In order to decouple spontaneous gait, our experimental design was to control two gait parameters simultaneously, whereby the third would be determined per force. Such manipulation would enable us to test the hypothesis that increasing gait speed *or* decreasing step length would each, independently, positively influence stability from slipping foot touchdown to liftoff of the recovery foot, and hence lessen the likelihood of a fall.

2. Methods

2.1. Subjects

Fifty-seven subjects, 19–45 years old, were randomly assigned to one of three target groups (Table 1). Subjects were screened for systemic disorders which might affect their participation and gave informed consent as approved by the local Institutional Review Board.

2.2. Experimental set-up and protocol

Subjects matched, simultaneously, their walking speed (u) to target flags along a moving rope loop and their steps to an audible metronome (Fig. 1 [24]). The rope loop was driven at the target speed by a DC motor (Model 4Z248D, Dayton Electric, Niles, IL). Since gait speed is the product of step length (SL) and cadence, this targeted the desired step length as well. Subjects were informed before beginning that they may be slipped "later". At right touchdown (TD) of selected trials, data from the sacral marker and the heel markers were used to estimate gait speed and step length respectively. These were used to provide verbal error feedback and to assess target matching, deemed successful if subjects were within 10% of the target value [24]. All subjects wore a safety harness attached by shock absorbing ropes to a low friction trolley on an I-beam above the walkway. A load cell recorded forces exerted through the ropes at 600 Hz. Within two trials of target matching, an unannounced slip was induced under the right foot by pre-releasing a movable, locked platform embedded in the walkway. This platform could slide freely up to 150 cm (coefficient of friction < 0.05) and was supported by force plates (AMTI, Newton, MA) recording ground reaction forces (GRF) at 600 Hz.

Three sets of gait parameters were designed so that groups B and C should have the same gait speed but different step lengths, and A and C should have the same step length, but different gait speeds; C comprised the theoretically least stable combination, to serve as a comparison against the parameters of A and B (Table 1). These targets were designed such that the predicted resultant COM motion states of all three would lie below the stability threshold that is referenced to the slipping BOS at its TD [20,24].

2.3. Data analyses

The kinematics of 27 markers attached to body segment landmarks and the platform were recorded by a motion capture system at 120 Hz (Motion Analysis Corporation, Santa Rosa, CA). Marker paths were low-pass filtered at marker-specific frequencies (range 4.5–9 Hz) using zero-lag, fourth-order Butterworth filters. Locations of joint centers were computed from the marker paths based on anthropometric data and the COM kinematics were computed using known sex-dependent segmental parameters in a 13-segment representation of the body [25]. Slipping step TD and recovery foot liftoff (LO) were identified from the vertical GRF. For each first slip TD, step length and gait speed, normalized to body height, were calculated as above. The COM position ($X_{COM/BOS}$) and forward velocity ($V_{COM/BOS}$) relative to the base of support (BOS, the slipping heel), at TD and LO, were expressed as a fraction of foot length (l_{BOS}) and $\sqrt{g \times \text{bh}}$, respectively, where g is the acceleration of gravity and bh is the subject's height. Stability was calculated as the shortest distance from the instantaneous COM motion state ($X_{COM/BOS}$ and $V_{COM/BOS}$) to the stability threshold [26] at both gait events, TD and LO. One subject in group A was excluded from analyses derived from motion data due to missing markers.

A recovery step was one in which the trailing heel landed behind the slipping heel; an aborted step was characterized by unloading, then re-loading the trailing foot, after slip initiation but before complete unloading [15]. Both constituted a backward balance loss. For an aborted step, LO time was taken as the instant of the minimum vertical GRF during the unload/re-load period. A *fall* occurred if the maximum force exerted on the load cell exceeded 30% body weight after slip onset [22].

The extent to which the gait parameters of each group met the design criteria was tested using one way analysis of variance (ANOVA). Changes in stability (Δs) from TD to LO (event) were examined using a 3 (groups, inter-subject factor) $\times$ 2 (events, intra-subject factor) repeated measures ANOVA. Post hoc analyses were done by Tukey's HSD statistics and paired t-tests with appropriate Bonferroni corrections. Linear regression, with gait speed and step length entered stepwise as factors, was used to assess the impact of the gait parameters on Δs. Differences in loss of balance and falls incidence among the groups were assessed with Chi square (χ^2). Logistic regression with gait speed and step length as factors was used to assess the impact of each on falls outcomes; odds ratios for falls were calculated based on this. Post hoc analysis was performed to examine whether pre-slip joint angles of the slipping limb would increase the predictive value for falls of this logistic regression. Statistics were performed with SPSS 17.0 (Chicago, IL) with α of 0.05.

3. Results

3.1. Changes from TD to LO

The resultant mean gait speeds and step lengths of the three groups met the design criteria (Table 1). The COM stability differed significantly among the three groups at TD and LO (main effect: $F_{2,53} = 89.38$, $P < .001$, Fig. 2). Group A was more stable than B,

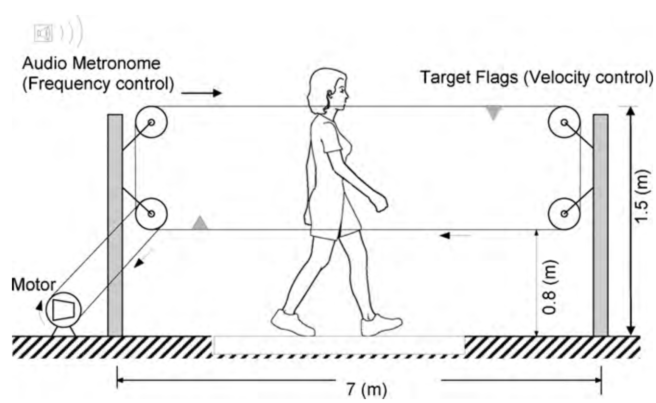

Fig. 1. Experimental set-up for target presentation during walking. Subjects matched gait speed (u) and step frequency (SF) targets simultaneously while walking along a 7 m walkway. Gait speed target was provided by flags attached to a rope loop running parallel to the length of the walkway and being driven at a constant velocity by a motor. Step frequency was provided by an audible metronome. Step length (SL) was also constrained because u = SL × SF.