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Influence of gait speed on the control of mediolateral dynamic stability during gait initiation

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ABSTRACT

This study investigated the influence of gait speed on the control of mediolateral dynamic stability during gait initiation. Thirteen healthy young adults initiated gait at three self-selected speeds: Slow, Normal and Fast. The results indicated that the duration of anticipatory postural adjustments (APA) decreased from Slow to Fast, i.e. the time allocated to propel the centre of mass (COM) towards the stance-leg side was shortened. Likely as an attempt at compensation, the peak of the anticipatory centre of pressure (COP) shift increased. However, COP compensation was not fully efficient since the results indicated that the mediolateral COM shift towards the stance-leg side at swing foot-off decreased with gait speed. Consequently, the COM shift towards the swing-leg side at swing heel-contact increased from Slow to Fast, indicating that the mediolateral COM fall during step execution increased as gait speed rose. However, this increased COM fall was compensated by greater step width so that the margin of stability (the distance between the base-of-support boundary and the mediolateral component of the "extrapolated centre of mass") at heel-contact remained unchanged across the speed conditions. Furthermore, a positive correlation between the mediolateral extrapolated COM position at heel-contact and step width was found, indicating that the greater the mediolateral COM fall, the greater the step width. Globally, these results suggest that mediolateral APA and step width are modulated with gait speed so as to maintain equivalent mediolateral dynamical stability at the time of swing heel-contact.

1. Introduction

Gait initiation (GI), corresponding to the transition from stationary standing to walking, is a functional task that is commonly performed in daily life. As emphasised in the literature, this task provides a challenge to dynamic stability, especially in the mediolateral (ML) direction (Lyon and Day, 1997; McIlroy and Maki, 1999). Indeed, the act of lifting the swing foot to execute the first step induces a reduction of the size of the base of support (BOS), which is then limited to stance-foot contact with the ground. It follows that if no action on the centre of mass (COM) is undertaken before the time of swing foot-off, i.e. if the COM is not moved above the stance foot, the whole-body will tend to fall laterally towards the swing-leg side during step execution, potentially causing a loss of balance and a sideways fall.

It is known that centrally-initiated dynamic phenomena, termed "anticipatory postural adjustments" (APA), precede the onset of voluntary movement. These APA aimed to stabilise the posture or assist the motor performance (Bouisset and Do, 2008; Yiou et al., 2012a). APA are observed before the step execution (beginning at swing heel-off) during GI. Along the ML direction, these APA are manifested as a centre of pressure (COP) shift towards the swing-leg side that propels the COM towards the stance-leg side prior to swing foot-off (McIlroy and Maki, 1999; Rogers et al., 2001; Yiou and Do, 2011). Although they do not directly propel the COM above the stance foot at foot-off (Jian et al., 1993), ML APA reduce the extent to which the COM falls toward the swing-leg side during step execution. ML APA thus constitute a crucial mechanism for controlling ML stability during GI (McIlroy and Maki, 1999). It is noteworthy that ML stability during GI may also be controlled via ML swing-foot placement at swing heel-contact (Lyon and Day, 1997; Zettel et al., 2002a, 2002b). By regulating the ML swing-foot placement (i.e. step width), individuals may maintain the COM within the BOS and thus ensure ML stability.

APA have also been described along the anteroposterior direction during GI. These APA are manifested as a backward COP shift

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that generates the initial propulsive forces necessary to reach the intended gait speed at the end of the first step (Brenière et al., 1987). The influence of gait speed on these APA has been extensively investigated (e.g. Brenière et al., 1987; Ito et al., 2003; Lepers and Brenière, 1995). In contrast, the question how the ML stabilizing features (including ML APA and ML foot placement) and related ML stability are modulated with gait speed is far less documented. To our knowledge, only one recent study has examined the influence of speed on ML dynamic stability control during volitional stepping (Singer et al., 2013). However, this study focused mainly on ML stability control during the phase of step termination (termed the "restabilisation" phase) and thus did not clearly investigate the speed effect on this control during the step initiation phase.

Increasing gait speed amplifies the accelerations acting on the body (Menz et al., 2003; Shkuratova et al., 2004), which may consequently result in a greater challenge to ML dynamic stability during GI. Interestingly, recent studies on rapid leg flexion showed that young healthy participants were able to modulate ML APA in order to maintain ML dynamic stability unchanged in situations with a postural constraint, e.g. temporal pressure (Yiou et al., 2012b) or elevated support surface (Yiou et al., 2011). Similarly, previous studies showed that, during reactive stepping initiation, participants used a strategy of lateral swing foot placement, along with the inclusion of larger ML APA, to compensate for postural perturbation induced by force-plate translation (e.g. Zettel et al., 2002a, 2002b). These findings suggest that, when facing a postural constraint, the central nervous system (CNS) has the capacity to modulate ML APA and step width in order to maintain equivalent dynamic stability.

The present study investigated the influence of gait speed on ML dynamic stability control during GI. We hypothesised that healthy young adults modulate the temporo-spatial features of ML APA and ML foot placement as gait speed increases so as to maintain equivalent ML dynamic stability at the time of swing heel-contact.

2. Methods

2.1. Subjects

Thirteen healthy subjects (6 males, 7 females; age: 27 ± 6 years, height: 171 ± 9 cm, body mass: 68 ± 10 kg, body mass index: 23 ± 2 kg/m^2) participated in this experiment. All gave written consent after being fully informed of the test procedure, which was approved by the local ethics committee.

2.2. Experimental set-up and procedure

Gait was initiated from a force-plate (46.4×50.8 cm, AMTI, USA) located at the beginning of a 5-m walkway (Fig. 1). A larger force-plate (90×90 cm, AMTI, USA) was located immediately in front of this initial force-plate so that the first step landed onto it. The two force-plates, embedded in the walkway, recorded the ground reaction forces and moments. Reflective skin markers (9-mm diameter) were placed bilaterally at the hallux (toe marker), head of the fifth metatarsus and posterior calcaneus (heel marker). A five-camera motion capture system (Vicon MX-T40, Oxford, UK) with 64 analog channels was used to collect simultaneously the kinematic data at 200 Hz and the force-plate data at 1000 Hz.

Initially, subjects stood barefoot in a natural upright posture with their arms alongside their trunk. They were instructed to stand as still as possible with their body weight distributed evenly between their legs. Gaze was fixed on a 10-cm diameter target placed at eye level and 6 m distant. After receiving a verbal "all set" signal, subjects initiated gait on their own initiative and continued walking straight ahead to the end of the walkway. Subjects chose their natural swing leg and maintained it throughout the experiment. After each trial, they had to reposition themselves in the same standardized feet position (see McIlroy and Maki, 1997) previously marked on the first force-plate. The experimenter triggered data acquisition when the subject was motionless and at least 1 s before the "all set" signal.

Gait initiation was performed under three speed conditions: natural pace (Normal condition), slower-than-natural pace (Slow) and as quickly as possible (Fast). The order of conditions was randomized across the subjects. In each speed condition, subjects performed two familiarisation trials and then five trials were collected. Subjects rested for 2 min between the speed conditions.

2.3. Data analysis

Kinematic and force-plate data were low-pass filtered using a Butterworth filter with a 15 Hz (Mickelborough et al., 2000) and a 10 Hz (Corbeil and Anaka, 2011) cut-off frequency, respectively. The ML coordinate of the COP was computed from force-plate data in accordance with the manufacturer's instructions (AMTI Manual). Formula is given in Appendix A.

Instantaneous ML acceleration of the COM (y″COM) was determined from the ML ground reaction force according to Newton's second law. ML COM velocity and displacement were computed by successive numerical integration of the COM acceleration (Brenière et al., 1987). By convention, COM displacement and velocity and COP displacement were considered positive when directed toward the swing-leg side.

The following instants were determined on the biomechanical traces (Fig. 2): GI onset (t_0), swing heel-off (HO), swing foot-off (FO) and swing heel-contact (HC). Time t_0 corresponded to the instant when the y″COM trace deviated 2.5 standard deviations from its baseline value (Yiou et al., 2012b). Heel-off and foot-off corresponded to the instants when the vertical position of the heel marker and the anterior position of the toe marker increased respectively by 3 mm from their position in the initial static posture. Heel-contact corresponded to the instant when the vertical ground reaction force measured by the second force-plate exceeded 10 N (Ghoussayni et al., 2004).

2.4. Dependant variables

Gait initiation was divided into three phases: APA (from t_0 to HO), foot lift (from HO to FO) and step execution (from FO to HC). The duration of each phase was reported. APA amplitude was characterised with the peak of lateral COP shift toward the swing-leg side. ML COM velocity and displacement at heel-off, foot-off and heel-contact were calculated. The peak of anteroposterior COM velocity was calculated to quantify gait speed (Brenière and Do, 1986; Caderby et al., 2013). The ML COM position in the initial upright posture was estimated by averaging the ML COP position during the 250-ms period preceding the "all set" signal (McIlroy and Maki, 1999).

An adaptation of the "margin of stability" (MOS) introduced by Hof et al. (2005) was used to quantify ML dynamic stability at heel-contact. In the present study, the MOS corresponded to the difference between the ML boundary of the BOS (BOS_{ymax}) and the ML position of the "extrapolated centre of mass" at heel-contact ($YcoM_{HC}$), i.e. $MOS = BOS_{ymax} - YcoM_{HC}$. Because kinematic data showed that the swing foot-strike was systematically made with the heel, BOS_{ymax} was estimated with the ML position of the heel marker of the swing foot at heel-contact. The ML distance between the position of the swing heel marker at heel-contact and the position of the stance heel marker at t_0 represented the step width, and was representative of the size of the ML BOS.

Based on the study of Hof et al. (2005), the ML position of the extrapolated COM at heel-contact ($YcoM_{HC}$) was calculated as follows:

$$YcoM_{HC} = yCOM_{HC} + \frac{y'COM_{HC}}{\omega_0},$$

where $yCOM_{HC}$ and $y'COM_{HC}$ are respectively the ML COM position and velocity at heel-contact, and ω_0 is the eigenfrequency of the body modelled as an inverted pendulum calculated as:

$$\omega_0 = \sqrt{\frac{g}{l}},$$

where $g = 9.81$ m/s^2 is the gravitational acceleration and l is the length of the inverted pendulum, which in this study corresponded to 57.5% of the body height (Winter, 1990).

ML dynamic stability at heel-contact is ensured on the condition that $YcoM_{HC}$ is within BOS_{ymax}, which corresponds to a positive MOS. A negative MOS indicates ML instability and implies that a corrective action (e.g. in the form of an additional lateral step) has to be undertaken to maintain balance.

2.5. Statistical analysis

Mean values and standard deviations of the dependant variables were calculated for each gait speed condition. Repeated-measures ANOVA with gait speed as a factor were conducted separately on these variables. Tukey *post hoc* analysis was performed when a statistical difference was found. Pearson's correlation coefficient (r) was used to determine the relationship between the variables. The level of statistical significance was set at $\alpha = 0.05$.