The Journal of Experimental Biology 213, 3945-3952

doi:10.1242/jeb.045112

The effects of arm swing on human gait stability

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Accepted 2 September 2010

SUMMARY

Arm swing during human gait has been shown to reduce both angular momentum about the vertical and energy expenditure, and has been hypothesized to enhance gait stability. To examine this hypothesis, we studied the effect of arm swing on the local and global stability of steady-state gait, as well as the ability to perform adequate recovery actions following a perturbation. Trunk kinematics of 11 male subjects was measured in treadmill walking with normal and with restricted arm swing. In half of the trials, gait was perturbed by a position-controlled forward pull to the trunk. We constructed state spaces using data recorded from the unperturbed steady-state walking trials, and quantified local gait stability by calculating maximum Lyapunov exponents. In addition, we analyzed perturbation forces, the distance from the unperturbed gait pattern, and the return toward the normal gait pattern following an external perturbation. Walking without arm swing led to a non-significantly lower Lyapunov exponent (P=0.06), significantly higher perturbation forces (P<0.05), and significantly slower movements away from the attractor (P<0.01). These results suggest that gait without arm swing is characterized by similar local stability to gait with arm swing and a higher perturbation resistance. However, return towards the normal gait pattern was significantly slower (P<0.05) when walking with restricted arms, suggesting that the arms play an important role in the recovery from a perturbation. Collectively, the results suggest that arm swing as such does not enhance gait stability, but rather that recovery movements of the arms contribute to the overall stability of human gait.

Key words: human gait, Lyapunov exponents, arm swing, perturbation, stability.

INTRODUCTION

A vast body of human gait research has focused on the lower extremities, regarding the head, arms and trunk as a single unit (e.g. Kubo and Ulrich, 2006; Rietdyk, 2006; van der Krogt et al., 2009). This is even more noteworthy in research on bipedal gait in non-human primates (Mori et al., 2006; Mori et al., 2001; Ogihara et al., 2010), where arm movements, although sometimes shown in figures [e.g. figure 2 in two papers by Mori and colleagues (Mori et al., 2006; Mori et al., 2001)], have received virtually no attention.

Nonetheless, at least for human walking, there is substantial evidence that arm swing is an essential component of locomotion. For instance, it has been shown that walking without arm swing increases the metabolic cost of walking (Collins et al., 2009; Ortega et al., 2008; Umberger, 2008), either because of the greater angular momentum about the vertical that needs to be counteracted (Bruijn et al., 2008; Collins et al., 2009; Elftman, 1939; Herr and Popovic, 2008; Park, 2008), or because of the larger vertical movements of the center of mass that occur when the arms do not swing upward when the trunk moves downward (Hinrichs and Cavanagh, 1981; Murray et al., 1967; Umberger, 2008). Moreover, several authors have claimed that arm swing during human locomotion enhances gait stability (Elftman, 1939; Hinrichs and Cavanagh, 1981; Ortega et al., 2008).

Stability defines the response of a system to a perturbation. In steady-state gait, infinitesimally small perturbations are ever present, and the system's response to such perturbations is called local stability. When gait is externally perturbed, global stability can be assessed by quantifying the response to such a perturbation. In the case of a feedback-controlled system like the human body, this response may be divided into two phases: an initial phase, which is dependent upon the steady state of the system (as it was before the perturbation) and the system's intrinsic mechanical properties (e.g. inertia, stiffness), and a second, reactive phase ('recovery'), which is mainly dependent on active control and reflexes (cf. Marigold and Misiaszek, 2009). In reality, separating the contributions of these phases to the response following a perturbation is difficult, as they may interact and overlap.

Little is known about the influence of arm swing on human gait stability. Ortega and colleagues studied the effects of arm swing on lateral stabilization during steady-state gait (Ortega et al., 2008). Elastic cords attached to a hip belt reduced energy expenditure more when subjects walked without arm swing than when they walked with their natural swing. From this, the authors concluded that arm swing contributes to lateral stabilization. However, an alternative explanation for these findings is that the elastic cord counteracted the angular momentum about the vertical, which could also have led to larger decreases in energy expenditure when walking without arm swing. In a model-based study, Collins and coworkers reported no increase in the local stability of steady-state gait in their passive dynamic walking model when arms were added (Collins et al., 2009). However, the model in question relied on a purely passive arm swing, which may not be realistic, given the pronounced electromyogram (EMG) activity in the shoulder muscles during human gait (Pontzer et al., 2009). Another (physical) model of bipedal walking showed that addition of a normal, human-like (passive) arm swing – that is, with the arms swinging inward when swinging forward – decreased global stability (in particular of side-to-side motion), while global stability (in particular of side-to-side motion) increased with the arms swinging outward when swinging forward (Collins et al., 2001). These findings suggest that arm swing may at least influence global gait stability. Still, it remains to be shown how well these model findings translate to real human walking. Also, these studies do not reveal how arm swing affects local and global gait stability.

A recent study by Pijnappels and colleagues on the effect of arm swing in the recovery phase following an actual trip did provide additional insight into the effects of arm swing on global gait stability (Pijnappels et al., 2010). In their study, the effects of angular momentum generated by the arms were examined using simulations. In the first simulation, all momentum of the arms at the instant of tripping was transferred to the body (as if the subjects arrested all arm movement at that instant and the arms were removed thereafter), while in the second simulation all momentum carried by the arms was simply regarded as 'lost' (as if the subjects had lost their arms and thus all arm momentum at the instant of the trip). Using actual angular momenta derived from experimental data, the position that the body would have assumed at the instant of recovery foot placement was calculated. Compared with the actual measured position, the simulations in which the angular momentum of the arms was transferred to the body, as would occur with normal arm swing, led to a less favorable position. From this, Pijnappels and colleagues concluded that angular momentum of the arms at the instant of tripping is detrimental to recovery foot placement (Pijnappels et al., 2010). These findings suggest that the absence of arm swing might enhance rather than diminish the initial phase of global gait stability.

The study by Pijnappels and coworkers (Pijnappels et al., 2010) also suggested that corrective arm movements are made in the reactive phase, thus postponing the transfer of angular momentum to the trunk that would occur with normal arm swing, so that the actual foot placement is much more favorable than would be expected from simulations in which such corrective movements are absent. Thus, in the reactive phase, corrective arm responses likely make up for the negative effects of normal arm swing on the initial phase of global gait stability. Nonetheless, the role of arm movements in stabilizing human gait needs to be elucidated further.

In the current study, the local stability of steady-state gait was assessed using maximum time finite Lyapunov exponents (Bruijn et al., 2009a; Bruijn et al., 2009b; Dingwell and Cusumano, 2000; Dingwell et al., 2008; Rosenstein et al., 1993), which quantify the average logarithmic rate of divergence after infinitesimally small perturbations. Since such infinitesimally small perturbations occur naturally during steady-state gait (i.e. due to neuromuscular and external noise), this measure can be used to quantify the local stability of steady-state gait, and may thus serve to capture the effects of arm swing on the local stability of steady-state gait.

To gain more insight into the effects of arm swing on global gait stability, responses to an external perturbation were analyzed in detail in the present study. This analysis was based on a clear distinction between an initial phase, which also contains information on the preceding steady state of the system, and the recovery phase, in which the efficacy of corrective actions performed to return to the normal gait pattern could be quantified. Of course, such a distinction can only give an indication of the difference between these two different phases, as the phases themselves may interact and overlap.

In summary, the present study sought to elucidate the effect of arm swing on the local and global stability of gait. Our analysis was focused on trunk motions, as maintaining stability of the upper body is a critical aspect of human locomotion (MacKinnon and Winter, 1993). The perturbation used was a pull to the trunk, which has an effect comparable to that of bumping into somebody while walking. This kind of perturbation occurs commonly in daily life, but has not received much attention in the literature. To allow for generalizability of our results to a broad range of walking speeds, we measured subjects at three different walking speeds.

Based on the results of Collins and colleagues (Collins et al., 2009) and Pijnappels and colleagues (Pijnappels et al., 2010), we hypothesized that (1) arm swing has no effect on the local stability of steady-state gait, (2) the initial phase of global gait stability indicates that walking with normal arm swing is less stable, and (3) the reactive phase of global gait stability indicates that walking with normal arm swing leads to a better recovery following an external perturbation, as quantified by the rate of return to normal walking pattern.

MATERIALS AND METHODS

Subjects

Eleven healthy male subjects (age 27.7±3.3 years, mass 75.5±9.0 kg, height 1.80±0.06 m; means ± s.d.) participated in the study. None of the participants had an orthopedic or neurological disorder. Before participating, subjects signed an informed consent form. The protocol was approved by the ethical committee of the Faculty of Human Movement Sciences of VU University, Amsterdam.

Procedures

Neoprene bands with clusters of three infrared light-emitting diodes (LEDs) were attached to the trunk (over the level of T6) and the heels. The LEDs were used for movement registration with a 3D optoelectronic system (Optotrak® Northern Digital Inc., Waterloo, ON, Canada), consisting of a 2×3 camera array (i.e. two measurement units holding three sensors each). Kinematic data were sampled at 50 samples s^{-1}.

Subjects walked on a treadmill at three different speeds (0.56, 1.12 and 1.68 m s^{-1}). At each speed they walked for 5 min with normal arm swing and 5 min with arm swing restricted by a belt attached at pelvis height. All six conditions were performed twice, once without perturbations (steady-state walking trials) and once with perturbations (perturbation trials). The perturbation consisted of a pull to the trunk in the direction of walking. Approximately 20 perturbations were applied per perturbation trial, during (quasi-)randomly selected strides determined by the experimenter. All conditions were performed in random order. Subjects were told whether or not the upcoming trial would be a perturbation trial, but they could not know which strides would be perturbed.

Perturbation device

A custom-made device (see Figs 1–3) was used to apply the perturbations to the subjects while they walked on the treadmill. The device consisted of pneumatic pistons, latches, ropes, pulleys and force transducers, and was controlled on the basis of kinematic data.

When a predefined cue in the kinematic signal was detected (see Fig. 3), the previously free-running ropes attached to the subject were blocked by a pneumatic latch, and the piston would go down, causing, *via* the pulleys, a shortening of the rope by 0.2 m. However, because of the elasticity of the rope, and movements of the harness relative to the subject, the actual displacement of the subject was always less than this. The delay between detection of the kinematic cue and force onset was approximately 100 ms (see Fig. 3).

Forces during the perturbations were recorded at 200 samples s^{-1} using uni-directional force transducers. The force transducers were calibrated before each measurement session. The perturbations were timed to occur just before heel strike (on the basis of a lateral

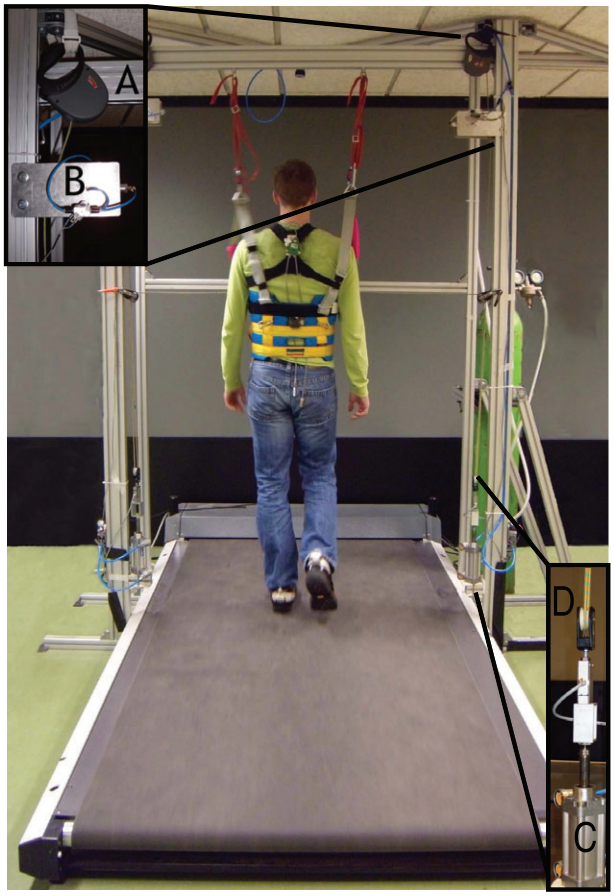

Fig. 1. The set-up used to perturb the subject. A subject is shown wearing the safety harness. To this harness, two ropes were attached (see also Fig. 2), which were free running with some tension (by means of the device in A). When a perturbation was delivered, the device (B) would block the ropes from running freely and a pneumatic piston (C) would go down, causing a shortening of the rope by 20 cm. A uni-directional force transducer (D) was used to record the forces.

velocity reversal of the trunk markers) with a force of about 200 N and a duration of 200 ms, and were applied contralateral to the heel strike, in the forward direction (see Fig. 3). It should be noted that with increasing walking speed, step times decrease, which, given the fixed delay in the system, meant that perturbations at different speeds occurred at slightly different moments within the stride cycle. As a consequence, any effect of 'speed' reported in the present study is actually a combined effect of speed and the timing of the perturbation in the stride cycle. Since the system was position controlled, the force exerted by the pistons provided information on the compliance of the subject during the perturbation and, thus, on the initial phase of global gait stability. We therefore registered maximum forces during each perturbation. The median of these forces within a trial was used for further statistical analysis ($\mathbf{F}_{max}$).

Calculations

Heel strikes

To allow time normalization of the data, heel strikes were detected as the local minima in the vertical position of the heel marker. Stride times were then calculated as the average time between two consecutive heel strikes.

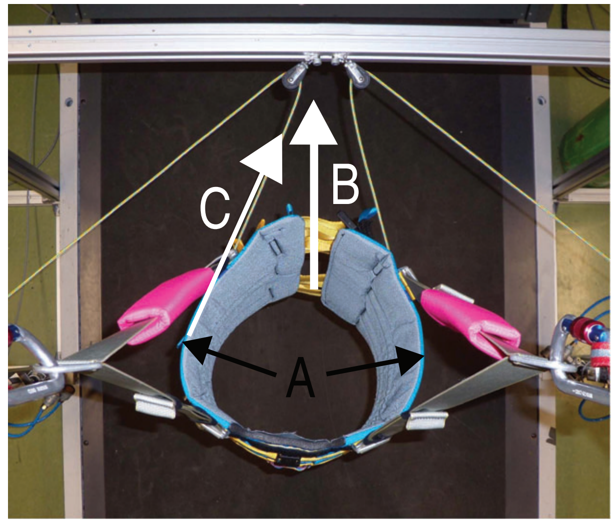

Fig. 2. Top view of the harness the subjects were wearing, and the ropes attached to it (rope attachments are indicated by A). When a perturbation occurred, the left rope would be pulled just before the right heelstrike and *vice versa*. Arrow B indicates the walking direction, while arrow C indicates the direction of the force during a perturbation on the left side.

Pre-processing

Trunk cluster marker 3D linear velocity data (V) were obtained by differentiation of the average position of the three trunk markers, while rotational velocities (ω) of the trunk cluster marker were calculated as described in previous studies (e.g. Berme and Capozzo, 1990). All analyses were performed on the velocity time series to minimize the effects of non-stationarity in the position data (i.e. wandering around on the treadmill).

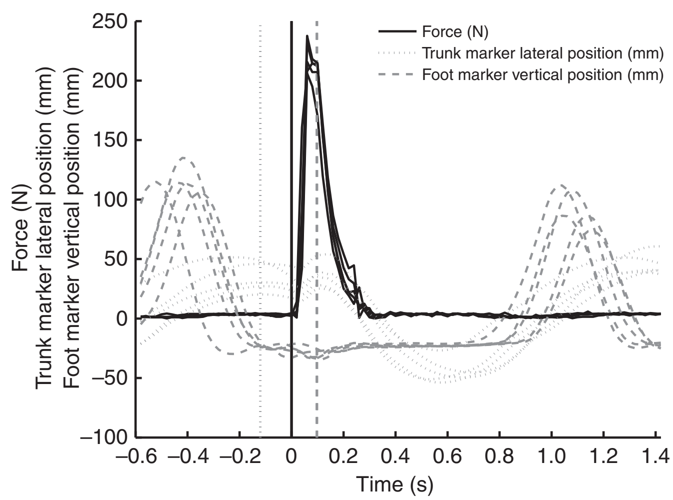

Fig. 3. Plot showing the timing, delay and variability of a series of perturbations for one subject. The perturbations were timed on the basis of the change in lateral trunk velocity (lateral trunk motions are shown as dotted lines). The delay between the trigger signal (vertical dotted line) and force onset (vertical solid line) was approximately 100 ms, so that the perturbation would occur just before heelstrike of the contralateral foot (i.e. the local minimum in the vertical signal of the heel marker, depicted here by the dashed vertical line).

Local dynamic stability

Local dynamic stability, expressed as maximum time finite Lyapunov exponents, was calculated from the unperturbed walking trials only. To this end, state spaces were reconstructed from V and ω. The first 140 strides of the signals were selected and then time normalized using a spline interpolation, so that each state space consisted of 14,000 samples (Bruijn et al., 2009a; England and Granata, 2007). 12D state spaces (Bruijn et al., 2010b; Dingwell et al., 2007; Kang and Dingwell, 2006a; Kang and Dingwell, 2006b; Kang and Dingwell, 2008; Kang and Dingwell, 2009) were then reconstructed using these time-normalized signals and their 25 samples delayed copies. Next, the average logarithmic divergence was calculated using well-documented methods (Dingwell and Cusumano, 2000; Rosenstein et al., 1993); nearest neighbors were calculated for each point, and the distance between the trajectories from these points over time was determined. These differences were averaged, and the logarithm was then taken to obtain a divergence curve. From this curve, the maximum time finite Lyapunov exponents were calculated as the linear slopes from 0 to 0.5 strides (λ_S) and from 4 to 10 strides (λ_L) (Bruijn et al., 2010b; Bruijn et al., 2009a).

Perturbation measures

Linear (V) and angular velocity (ω) time series of all trials were first time normalized using a spline interpolation, so that every stride consisted of exactly 100 samples. Then, V and angular velocity ω time series of the steady-state trials were combined to construct an average 'limit cycle' for each subject and trial (NW, normal walking). Furthermore, for each percentage in this limit cycle, the normal variability for each dimension was calculated as the standard deviation (v_{NW}).

Normalized Euclidean distances between the gait cycles during the perturbation trials (PW) and the average limit cycle were then calculated as:

$$D(k \times 100 + i)_{\substack{k=0:n-1 \\ i=1:100}} = \sqrt{\sum_{d=1}^{6} ((\mathrm{NW}(i)_d - \mathrm{PW}(k \times 100 + i)_d) / v_{\mathrm{NW}}(i)_d)^2} \ , \quad (1)$$

where $D(k\times 100+i)$ is the normalized distance (in standard deviations) for i% of stride k+1 (with n representing the maximum number of strides in PW); d is the dimension number, NW is the limit cycle, PW is the state of the perturbed walking trial, and v_{NW} is the variance of the limit cycle. To examine to what extent the changes in perturbation parameters (see below) were dependent upon changes in v_{NW}, we calculated the mean deviation from NW across the gait cycle as (md_{NW}):

$$\mathrm{md}_{\mathrm{NW}} = \frac{1}{100} \sum_{i=1}^{i=100} \sqrt{\sum_{d=1}^{6} (v_{\mathrm{NW}}(i)_d)^2} \ . \quad (2)$$

The start of a perturbation was determined as the last sample before the force exceeded 40 N (see also Fig. 3). We used 40 N as a cut-off as there was already some tension in the ropes, which, given the noise, would otherwise lead to false-positive perturbation detections. The time to the maximum D after each perturbation started was detected (τ, see also Fig. 4). From then on, the exponential decay or relaxation to the limit cycle was quantified using (see Post et al., 2000) (see also Fig. 4):

$$D(i) = A + (B - A) \times e^{(-\beta(i - \tau))} \ , \quad (3)$$

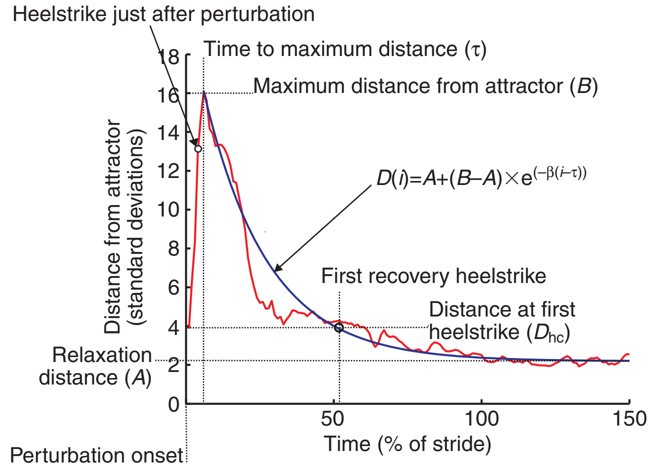

Fig. 4. Parameterization of the perturbation.

where D refers to the Euclidian distance between the perturbed gait cycle and the average limit cycle, A refers to the relaxation distance (defined as the average value of D from i=100 to i=150), B refers to the size of the initial perturbation, and β refers to the rate of return to the limit cycle. Higher values of β indicate a faster return to the normal gait pattern. Next, the distance from the average limit cycle, i.e. the attractor, at the first recovery heel contact was calculated (D_{hc}). For statistical analysis, the median values of each parameter per condition per subject were used.

All calculations were performed using custom-made MatLab programs (The MathWorks, Inc., Natick, MA, USA).

Statistical analysis

The effects of condition (arm swing *vs* no arm swing) and speed, as well as their interaction, were tested using repeated measures ANOVA for all variables (i.e. the control variables: number of perturbations in a trial, stride time, stride time variability, md_{NW}, and the dependent variables: λ_S, λ_L, $\mathbf{F}_{max}$, A, B, τ, β and D_{hc}).

RESULTS

Stride times

There were no significant main effects of arm swing on average stride time (P=0.3, see Fig. 5A) and stride time variability (P=0.5, see Fig. 5B), nor were there significant interactions of speed with arm swing (P>0.2 for both stride time and stride time variability). These results imply that the time normalization of strides we used in our analysis of the perturbation parameters did not bias our results with regard to the effects of arm swing. As expected, both stride time and stride time variability decreased significantly (P<0.001 for both) with increasing walking speed.

Local dynamic stability

For steady-state gait, λ_S showed larger values when walking with normal arm swing; however, this difference was not significant (P=0.06, see Fig. 6A). There was a significant effect of speed on λ_S (P<0.01), with increasing walking speed leading to lower values of λ_S. There was no significant effect of arm swing on λ_L (P>0.2, see Fig. 6B), but again there was a significant effect of walking speed (P<0.01), with higher walking speeds leading to higher values of λ_L.

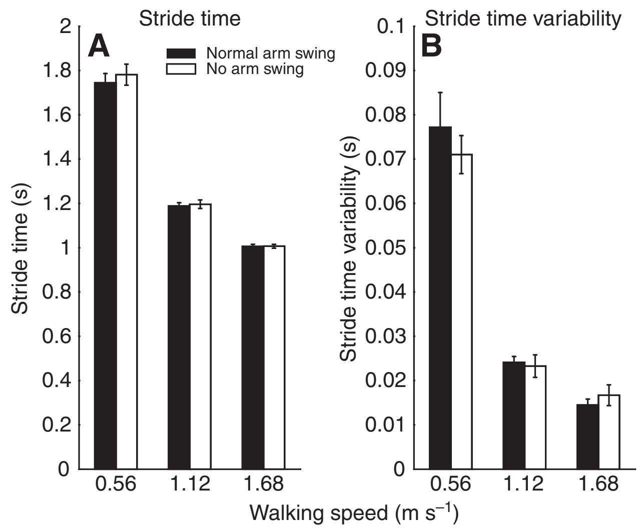

Fig. 5. (A) Stride time (P<0.05 for speed, all other effects P>0.2) and (B) stride time variability (P<0.05 for speed, all other effects P>0.2). Error bars represent standard errors.

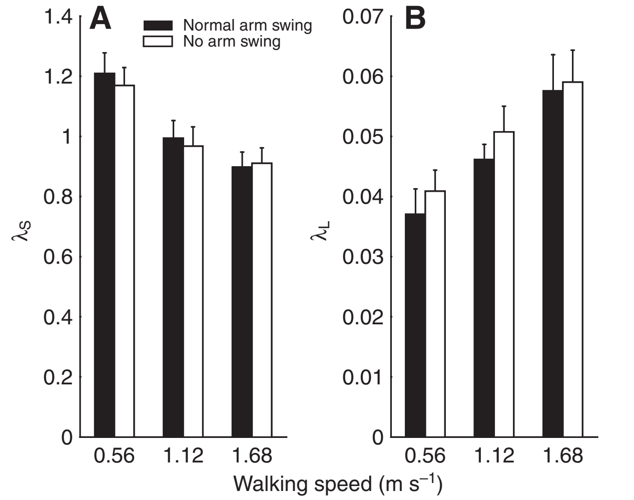

Fig. 6. Local dynamic stability measures. Error bars represent standard errors. (A) The Lyapunov exponent calculated from the slope of the divergence curve at 0–0.5 strides, λ_S (P=0.06 for arm swing, P<0.01 for speed). (B) The Lyapunov exponent for the slope at 4–10 strides, λ_L (P=0.2 for arm swing, P<0.01 for speed). There were no significant interaction effects for any of the variables (P>0.8 for all).

Perturbation parameters

The average number of perturbations applied within a trial ranged from 15 for those at 0.56 m s^{-1} to 21 for those at 1.68 m s^{-1}, a variation that resulted in a significant effect of speed (P<0.05). There was no significant effect of arm swing (P> 0.4) and arm swing×speed (P>0.3) on the number of perturbations. There was also no significant effect of arm swing or walking speed on md_{NW}, rendering it unlikely that differences in variability of the unperturbed gait pattern influenced the perturbation parameters.

There were no significant interaction effects for any of the dependent variables, implying that the effects of condition were the same for all speed levels.

$\mathbf{F}_{max}$ was significantly lower in the arm swing condition (P<0.05, see Fig. 7A). Moreover, the time that elapsed before the maximum distance from the attractor was reached (τ) was significantly shorter in the arm swing condition (P<0.01, see Fig. 7C), while this maximum distance (B) was not different between conditions (P=0.32, see Fig. 7B). Of the perturbation parameters quantifying the initial response to the perturbation, only τ showed a significant effect of walking speed, with higher values of τ for the higher walking speeds (P<0.01).

In the recovery phase, the exponential decay towards the limit cycle (β) was faster in the condition with arm swing (P<0.05, see Fig. 7D), while the relaxation distance (A) was not different between the two arm swing conditions (P=0.72, see Fig. 7E), which rendered the distance at the first recovery heel strike (D_{hc}) significantly smaller when walking with arm swing (P=0.01, see Fig. 7F). All parameters quantifying the recovery phase showed significant effects of walking speed; β decreased, while A and D_{hc} increased with increasing walking speed (P<0.01 for all).

DISCUSSION

In the present experiment, we examined the effect of arm swing on the local and global stability of human gait. In doing so, we partitioned the global stability into two phases: an initial response, which also contains information on the steady-state gait, and a recovery phase.

Inspection of Fig. 7A suggested reduced local stability when walking with normal arm swing; however, this effect was not significant, which confirmed our first hypothesis that arm swing would have no effect on the local stability of steady-state gait.

Perturbation parameters revealed that arm swing was accompanied by a lower force exerted by the pistons ($\mathbf{F}_{max}$), and a shorter time to reach the (same) maximum distance from the attractor (τ) for perturbed walking (implying a greater acceleration). Taken together, these findings are in agreement with our second hypothesis that walking with normal arm swing leads to a decreased performance in the initial phase of global gait stability.

In support of our third hypothesis, arm swing allowed for more effective recovery reactions to large external perturbations, as indicated by higher values of the exponential return parameter β. More importantly, we found lower values of the distance to the attractor at first heel strike (D_{hc}) when walking with normal arm swing, suggesting that, in total, global gait stability increased when walking with arm swing.

Increases in walking speed led to conflicting results regarding the local and global stability of gait, with significant decreases in λ_S (suggesting increased local stability), significant increases in λ_L (suggesting decreased local stability), significant increases in τ (suggesting an increased performance in the initial phase of global gait stability), significant decreases in β (suggesting a decreased performance in the recovery phase of global gait stability) and significant increases of D_{hc} (suggesting an overall negative effect of increased walking speed on global gait stability).

The effects of arm swing

Regarding both local gait stability and the initial phase of global gait stability, the effects of arm swing found in the present study contradict the conclusion of Ortega and colleagues that arm swing plays a positive role in stabilizing steady-state gait (Ortega et al., 2008). Instead, our findings are more in line with those of Collins and coworkers, who reported no effects of arm swing on local stability of steady-state gait (Collins et al., 2009), and those of Pijnappels and colleagues, who concluded that arm swing

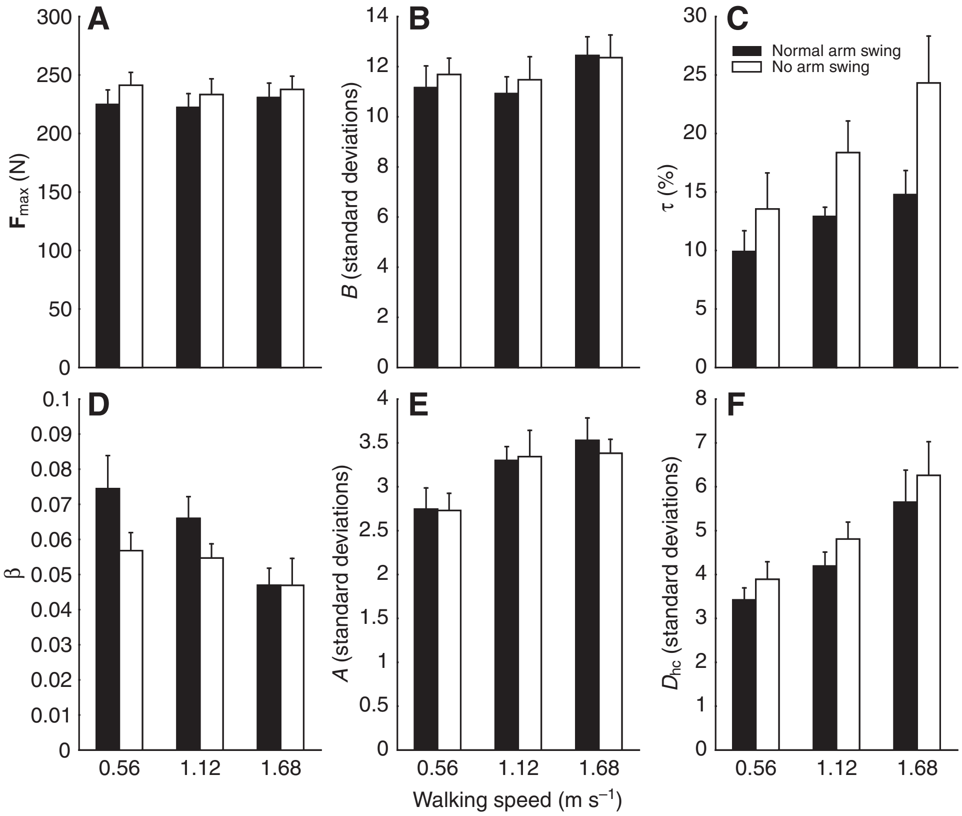

Fig. 7. Perturbation parameters. Error bars represent standard errors. (A) Perturbation force, $\mathbf{F}_{max}$. (B) Maximum distance from the attractor, *B*. (C) Time to maximum distance, τ. (D) Exponential decay, β. (E) Relaxation distance, *A*. (F) Distance at first heelstrike, D_{hc}. Significant effects of arm swing were found for $\mathbf{F}_{max}$, τ, β and D_{hc}, and significant effects of walking speed were found for τ, β, A and D_{hc}. There were no significant interaction effects for any of the variables (P>0.1 for all).

decreases performance in the initial phase of global gait stability (Pijnappels et al., 2010). While the effects of arm swing on λ_S were not significant, it should be kept in mind that this measure has limited statistical precision (Bruijn et al., 2009a). Interestingly, like in previous studies (Bruijn et al., 2010a; Su and Dingwell, 2007), the effect of arm swing on λ_S was similar to the effect on measures of the initial phase of global gait stability (such as the perturbation force and the initial response to the perturbation as quantified by τ and *B*), which, in part, also reflect the steady-state gait stability. Of course, these findings may be confounded by the fact that the reactive phase of the response is already present in these measures. However, since recovery was faster in the arm swing condition (higher values of β and lower values of D_{hc}), this would attenuate the effects found. The higher values of β, and more importantly the lower values of D_{hc}, indicate an overall positive effect of arm swing on global gait stability. The overall picture that emerges from these results is thus consistent with the work of Pijnappels and colleagues, who predicted that arm momentum (in the horizontal plane) at the instant of a trip during mid-stance would have a detrimental effect on the initial phase after a perturbation, but that subsequent reactions of the arms were likely to counteract this initially detrimental effect (Pijnappels et al., 2010). It should be noted that we found an overall positive effect of arm swing on global gait stability (i.e. lower values of D_{hc} when walking with normal arm swing), whereas Pijnappels and coworkers did not find such an effect (Pijnappels et al., 2010). The reason for this difference may be that we used a position-controlled rather than a force-controlled perturbation; had we used force-controlled perturbations, values of *A* might have been higher for walking with normal arm swing, resulting in equal values of D_{hc}.

The decreased performance in the initial phase of global gait stability when walking with normal arm swing may perhaps be explained in terms of increased inertia. When the hands are tied to the body, the (upper) body has a greater (effective) inertia, and is thus more resistant to perturbations. This hypothesis of greater steady-state gait stability due to greater effective inertia may be tested experimentally by having subjects walk with their arms fixed away from the body, so that arm swing is restricted while rotational inertia is further increased. Another explanation would be that restricting arm swing also causes different (trunk) muscle activation patterns. We are unaware of literature reporting this effect, and did not measure muscle activity.

When a perturbation occurs to the upper body with the arms tied, the constrained upper body will tend to behave more like an inverted pendulum than the unconstrained upper body, and will be less able to recover from a perturbation. The present results thus suggest that, from a stability point of view, the optimal strategy would be to walk with the hands alongside the body, until a perturbation occurs. Still, it may be that ongoing arm movements are needed to perform the rapid arm movements required for successful recovery. Future experiments, in which the hands are tied to the body and released at the instant of a trip or other perturbation, are required to test this idea. It should be noted in this context that while this strategy of holding the arms alongside the body until a perturbation occurs may be optimal with respect to stability, it is certainly not optimal in terms of energy costs (Collins et al., 2009; Ortega et al., 2008; Umberger, 2008), which may explain why humans do not normally walk like this. Interestingly, however, non-human primates displaying bipedal gait seem to be doing exactly this [see figure 2 in Mori et al. (Mori et al., 2006)], but this has never been explicitly reported. Nonetheless, even if this observation were to be confirmed, it remains to be investigated whether this constitutes a strategy to optimize stability.

The effects of walking speed

Interestingly, walking speed led only to significant main effects and no interaction effects, which suggests that the effects of arm swing were similar, or at least not very different, for all walking speeds. Still, like in previous studies (Bruijn et al., 2010b; Bruijn et al., 2009b; Fallah Yakhdani et al., 2010), increasing walking speed led to a significant decrease in λ_S and a significant increase in λ_L. Note that only λ_S has been shown to be related to global stability, i.e. the probability of falling, in modeling studies (Bruijn et al., 2010a; Su and Dingwell, 2007). In line with this, τ increased significantly with increasing walking speed, also suggesting an increased performance in the initial phase of global gait stability with increasing walking speed. Higher walking speeds, however, also led to a decreased performance in the recovery phase of global gait stability, as indicated by a slower exponential decay (β). All in all, larger distances from the attractor (D_{hc}) at the first recovery heelstrike with increasing walking speed indicated a decrease in global gait stability. However, all results derived from perturbation parameters must be treated with great caution, as perturbations were not infinitely short, and started and ended after some delay, which automatically caused a perturbation to end later in the gait cycle for the faster walking trials.

Limitations of the present study

The present study has several limitations. Firstly, our sampling rate was relatively low (50 samples s^{-1}), which may have reduced the precision of the perturbation data, but is unlikely to have caused any bias. Secondly, we carried out the experiments only on healthy male subjects, which limits the generalizability of our results. Lastly, it is known that a novel task (Milner and Cloutier, 1993), or expectation of a perturbation (Lavender and Marras, 1995; Lavender et al., 1989), may lead to increased co-contraction. Increased co-contraction would probably lead to a higher perturbation resistance (Stokes et al., 2000; van Dieën et al., 2003), possibly confounding the results. However, conditions were offered in random order, so co-contraction due to expectation is likely to have played little or no role in the reported effects of arm swing. Still, co-contraction may have been increased in all conditions, which may have limited the generalizability of our results to real-life unexpected perturbations. Moreover, it seems unlikely that co-contraction was higher in the no arm swing condition because of the relative novelty of that task; walking with the hands alongside the body may not be a very new task to subjects, as people can only walk while doing different things with the arms, and subjects had to walk with the hands alongside the body for 6×5 min.

Generalizability of results to other perturbations

We only investigated perturbations occurring at one specific time interval, and in one specific direction, leaving it uncertain whether our results can be generalized to perturbations at other phases of the gait cycle, and in other directions.

While we perturbed only at one instance in the stride cycle (just before heel strike), our results regarding the initial and recovery phases of global gait stability are consistent with those of Pijnappels and colleagues, in which a trip was applied at mid-stance (Pijnappels et al., 2010). The effects of arm swing on recovery after a perturbation further agree with findings in slipping experiments (Marigold et al., 2003), in which a slip was induced at mid-stance. Moreover, our local dynamic stability analysis (particularly λ_S) of unperturbed walking, which is indicative of the local stability of the entire gait cycle, yielded results (although non-significant) that were in line with performance in the initial phase of global gait stability. As there is some evidence that local stability may be correlated to global stability of the gait pattern (Bruijn et al., 2010a; Su and Dingwell, 2007), this would suggest that our findings regarding the initial phase of global gait stability are valid for the entire gait cycle. Lastly, because our perturbation occurred after a fixed delay and was not infinitely short, it started and ended at different times in the gait cycle for different walking speeds. These considerations imply that the effect of arm swing was the same for slightly different perturbations at different walking speeds. Here, it appears likely that the effects of arm swing were similar for slightly different perturbations at the same walking speed, which supports the idea that our findings may at least be generalized to perturbations applied at different times in the gait cycle.

The perturbation applied in the current study was a forward pull to the thorax, with a slight rotational component. Since in daily life perturbations in other directions may also occur, a full assessment of the functional importance of arm swing to human gait requires that perturbations in these other directions are studied as well. We are not aware of published experiments showing that arm swing decreases performance in the initial phase of global gait stability, while facilitating the recovery phase of global gait stability, in the medio-lateral direction, and thus do not know to what extent our results generalize to this direction. Ortega and colleagues claimed that arm swing has a stabilizing effect on steady-state gait in this direction (Ortega et al., 2008), but the reduction in energy consumption due to lateral stabilization that they found for the no arm swing condition may have been caused by the medio-lateral stabilization counteracting the angular momentum about the vertical (Collins et al., 2009; Ortega et al., 2008). In the mechanical model of Collins and coworkers (Collins et al., 2001), non-human-like arm swing (with arms swinging outwards while moving forwards) helped to stabilize the model. However, arms in their model were passive, and fully coupled to the motions of the legs, rendering it questionable whether this finding can be extrapolated to human walking. Thus, further studies into the (de)stabilizing effects of arm swing in other movement directions are clearly required.

CONCLUSION

The present study showed that, contrary to what is commonly believed, arm swing does not only stabilize gait. However, the results also indicate that recovery actions of the arms may help recovery of gait stability following a perturbation.

LIST OF SYMBOLS AND ABBREVIATIONS

A	relaxation distance
B	size of the initial perturbation
d	dimension number
D	distance between perturbed gate cycle and average limit cycle
D_{hc}	distance from average limit cycle at first recovery heel contact
$\mathbf{F}_{max}$	maximum force
i	percentage of stride
k	stride number
LED	light-emitting diode
md_{NW}	mean deviation from normal walking
n	maximum number of strides in perturbed walking
NW	normal walking
PW	perturbed walking
v_{NW}	variability in normal walking
V	linear velocity
β	rate of return to limit cycle
λ_L	Lyapunov exponent for the slope at 4–10 strides

λ_S Lyapunov exponent calculated from the slope of the divergence curve at 0–0.5 strides
τ time to maximum distance
ω rotational velocity

ACKNOWLEDGEMENTS

This study was partly funded by a grant from Biomet Nederland.

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