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Effect of arm swing strategy on local dynamic stability of human gait

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ABSTRACT

Introduction: Falling causes long term disability and can even lead to death. Most falls occur during gait. Therefore improving gait stability might be beneficial for people at risk of falling. Recently arm swing has been shown to influence gait stability. However at present it remains unknown which mode of arm swing creates the most stable gait.
Aim: To examine how different modes of arm swing affect gait stability.
Method: Ten healthy young male subjects volunteered for this study. All subjects walked with four different arm swing instructions at seven different gait speeds. The Xsens motion capture suit was used to capture gait kinematics. Basic gait parameters, variability and stability measures were calculated.
Results: We found an increased stability in the medio-lateral direction with excessive arm swing in comparison to normal arm swing at all gait speeds. Moreover, excessive arm swing increased stability in the anterior–posterior and vertical direction at low gait speeds. Ipsilateral and inphase arm swing did not differ compared to a normal arm swing.
Discussion: Excessive arm swing is a promising gait manipulation to improve local dynamic stability. For excessive arm swing in the ML direction there appears to be converging evidence. The effect of excessive arm swing on more clinically relevant groups like the more fall prone elderly or stroke survivors is worth further investigating.
Conclusion: Excessive arm swing significantly increases local dynamic stability of human gait.

1. Background

Falls can cause long term disability and form the main cause of sudden death in the elderly population [1]. Most falls occur during gait [2]. Local dynamic stability, quantified by the average rate of logarithmic divergence of initially infinitesimally close trajectories in state space [3] and gait variability, i.e. the variance of spatial and temporal characteristics of gait over successive strides, are associated with fall risk [4]. Consequently, interventions that improve local dynamic stability and variability of gait might be beneficial for people at risk of falling.

Interestingly, arm swing has been shown to influence human gait stability. Bruijn et al. [5] suggested "that gait without arm swing is characterized by similar local stability to gait with arm swing and a higher perturbation resistance". According to Bruijn et al. [5] keeping the arms fixed relative to the trunk possibly leads to more weight moving with the trunk, which subsequently leads to greater inertia and thus increased resistance against a change of movement and more stable gait dynamics. Moreover, arm movement has been shown to play an important part in the recovery phase after an actual trip [6]. Two recent studies explored the effects of different modes of arm swing on steady state gait stability in humans. Hu et al. [7] compared a normal arm swing with restricted and excessive arm swing in young and older adults, and Nakakubo et al. [8] explored the effects of these arm swing modes only in older adults. Results in both studies showed a significantly more stable gait when the arm swing was excessive in comparison to normal and restricted arm swing. Furthermore, in the study of Hu et al. [7], the relative improvement in stability was greater in the older than the younger population. This latter finding is interesting as older people are more likely to fall [9] and hence might benefit more from a stable gait.

Since arm swing can be modified with little muscular effort [10], it is worth investigating which arm swing mode creates the most stable gait, as increasing gait stability may lead to a decreased

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fall risk [4]. Previous studies on this topic were performed over a limited range of gait speeds while arm swing amplitude naturally changes as a function of gait speed [11]. Additionally, to the best of our knowledge, only restricted and excessive arm swing have been tested as modes of arm swing that could improve human gait stability. Therefore our aim was to elucidate the influence of four different arm swing modes (normal, in-phase swinging of both arms, in-phase swinging of arms with ipsilateral legs, and normal phase excessive amplitude swing) on human gait at different speeds. Dynamic stability was quantified by the local divergence exponent and variability measures, specifically stride time variability and stride-to-stride variability of step-width and trunk kinematics.

2. Methods

2.1. Participants

Ten young male adults volunteered for the study (age 23.1 ± 3.3 (mean ± standard deviation) years; length 1.84 ± 0.07 m; weight 73.1 ± 6.8 kg; BMI 21.5 ± 1.7 kg/m^2). The study was approved by the local ethics committee of the Faculty of Human Movement Sciences of the VU University Amsterdam and all subjects gave written informed consent. None of the subjects reported gait related injuries or disorders that could affect gait in the previous 2 years and all were familiar with treadmill walking.

2.2. Experimental protocol

Subjects walked on a treadmill (R-Mill, ForceLink b.v., Culemborg, The Netherlands) under four instructions: (1) without instruction, (2) to swing the arms in phase with each other (without explicit instruction as to how to coordinate these arm movements with the legs), (3) to swing the ipsilateral arms and legs forward at the same time, and (4) to perform a normally timed arm swing with excessive amplitude (see also the electronic supplementary video 1, and Table 1). Walking without instructions was always performed first; the subsequent three instructions were performed in a random order for each subject. All arm swing instructions were performed at seven gait speeds, from 0.28 m/s up to 1.96 m/s with increments of 0.28 m/s. Data recording started when subjects performed the correct arm swing mode for at least 30 s, based on visual observation. Each condition was recorded for 2 min.

2.3. Measurement system

We used a full body motion sensor suit consisting of 15 sensors containing 3D accelerometers, 3D gyroscopes and 3D magnetometers. These sensors were placed at the feet (2), shanks (2), thighs (2), pelvis at the sacrum (1), thorax at the sternum (1) and both shoulder blades (2), upper (2) and lower arms (2) and head (1) (Xsens b.v., Enschede, The Netherlands). Sample rate was set at 120 samples/s. The full body inertial motion capture system, provided 3D segmental orientations, positions, velocity, angular velocity and acceleration of all body segments based on sensor data. The Xsens full body inertial motion capture system has been shown to accurately measure human movement [12,13].

2.4. Data analysis

Data processing was performed using custom-made MATLAB (The Mathworks, Inc. Natick, MA, USA) routines. Foot strikes were detected from the foot time series, as maximal forward positions of the heel.

To make sure that all instructions were executed properly, relative Fourier phase [14] was calculated from AP position signals of the left and right lower leg segment and left and right forearm segment as obtained from the Xsens inertial motion capture system. These time series were first low-pass filtered with a bi-directional fourth order butterworth filter with cut off frequency of 5 Hz. Relative Fourier phase between left lower leg–left forearm (LL–LA), right lower leg–right forearm (RL–RA) and left forearm–right forearm (LA–RA) were calculated, using the phase at the fundamental frequency of the leg. To give an indication of the difference between normal and excessive arm swing, ranges of motion (arm swing amplitude) in the sagittal plane were calculated for the shoulder joint in both normal and excessive arm swinging at all gait speeds.

2.5. Spatio-temporal gait parameters

Stride time was determined by the time of two consecutive heel strikes of the same leg, and mean stride time was calculated as outcome variable. Step-width was calculated from the position data of the right and left foot during double support phases and mean step-width was calculated for statistical analysis.

2.6. Local dynamic stability

We expressed the rate of divergence per half a stride (0–0.5 strides). We used the lower back velocity signals to determine local divergence exponent (λ_s) for the 3 movement directions (AP, ML and VT), since λ_s of lower back kinematics discriminates between younger and older adults better than λ_s of other segments [15]. Velocity signals were not filtered, due to the problems associated with filtering nonlinear signals [16]. We included 57 consecutive strides for local divergence exponent calculations, as this was the minimum amount of strides available across instructions and subjects. To avoid problems due to differences in time series length [17], all time series of 57 strides were time-normalized to 5700 samples, so on average each stride contained 100 samples. From these time-normalized time-series we reconstructed a 5-dimensional state space using a delay of 10 samples

Table 1
Relative Fourier phase between the left and right arm (LA–RA), between the left leg and left arm (LL–LA) and between the right leg and the right arm (RL–RA) at all gait speeds and arm swing instructions.

Gait speed	Instruction 1 Normal arm swing			Instruction 2 Inphase arm swing			Instruction 3 Ipsilateral arm swing			Instruction 4 Excessive arm swing		
	LA–RA	LL–LA	RL–RA	LA–RA	LL–LA	RL–RA	LA–RA	LL–LA	RL–RA	LA–RA	LL–LA	RL–RA
0.28 m/s	173.6	170.4	168.6	8.4	77.4	124.4	170.7	25.8	10.2	169.7	169.0	172.5
0.56 m/s	175.9	170.2	168.9	16.9	82.6	127.4	177.2	29.3	12.6	168.7	174.7	167.5
0.84 m/s	175.0	171.7	170.6	37.3	93.9	99.9	175.2	40.3	12.9	166.0	173.2	171.1
1.12 m/s	170.1	168.4	170.6	25.3	65.8	99.4	166.1	49.6	32.2	166.2	171.6	172.2
1.40 m/s	170.6	171.4	170.9	33.4	61.3	95.4	173.4	46.7	55.3	163.3	168.4	166.7
1.68 m/s	172.5	168.2	171.9	29.0	54.9	74.2	164.2	36.9	52.3	165.0	168.0	166.9
1.96 m/s	176.5	167.7	173.3	18.3	95.0	111.9	167.2	31.1	43.9	162.3	167.8	167.8

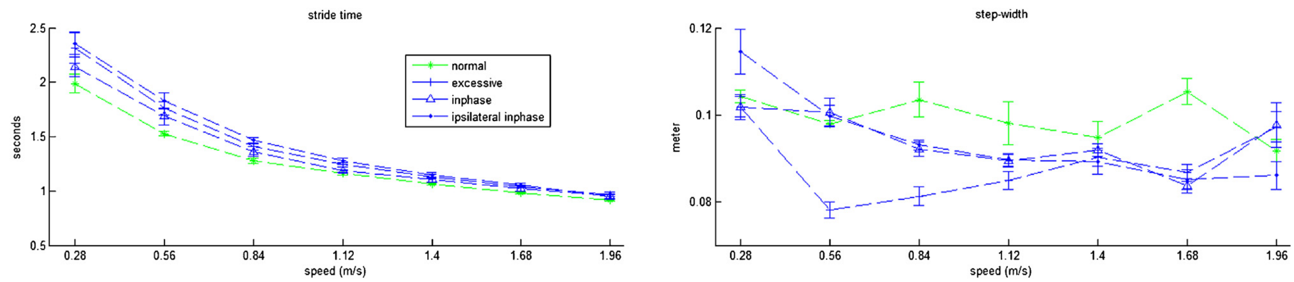

Fig. 1. Group-averaged values of the spatio-temporal measures for all gait speed conditions and arm swing instructions, with stride time in the left panel and step-width in the right panel. Error bars represent standard errors.

[18]. We used a fixed delay because after normalization each time series contained the same base frequency. In addition, using a fixed delay and dimensionality yields more reliable estimates [19]. Maximum finite-time local divergence exponents were calculated using the algorithm described by Rosenstein et al. [3].

2.7. Variability

Variability of stride time and step-width was determined by calculating the standard deviation of the stride times and step-widths. The variation of normalized time-series between strides was expressed by MeanSD. The velocity time-series derived from the lower back for the AP, VT and ML direction were used. First, all included strides were time-normalized to 100 samples. Second, the standard deviation over the included 57 strides for each sample in the stride cycle was determined. Lastly, the standard deviations were averaged over all % in the stride.

2.8. Statistics

Normality of the data was confirmed by the Kolmogorov–Smirnov test. We used a two-way repeated measures ANOVA to investigate the effect of arm swing instruction and speed on outcome parameters. If a main effect was found, post-hoc analysis were performed using a Bonferroni correction, and for significant differences between instructions, the effect size, partial eta squared (η^2) was calculated. Statistical significance was established a priori at a level of p-value of ≤ 0.05.

3. Results

Although during the experiment some subjects appeared to show some different behavior for some strides, on average, all subjects were able to adhere to the different arm swing instructions over all gait speeds, as shown in Table 1. Moreover, when subjects were instructed to walk with the arms moving in phase with each other, at lower speeds, they would often walk at a 2:1 arm-leg frequency ratio; while at higher speeds, they would "lock" to one leg, obtaining a 1:1 arm-leg frequency ratio (see also electronic supplementary video 1). Excessive arm swing amplitude was $60^\circ \pm 20^\circ$ at the slowest gait speed and increased toward $68^\circ \pm 12^\circ$ at the highest speed. Arm swing amplitude in the condition with normal arm swing instruction was $7^\circ \pm 5^\circ$ at the slowest speed, and increased toward $35^\circ \pm 8^\circ$ at the highest gait speed (see also electronic supplementary video 1).

3.1. Spatio-temporal gait parameters

There was a main effect of instruction on stride time (Fig. 1, Table 2; $P < 0.01$). Post-hoc analysis revealed significantly shorter stride times in normal arm swinging in comparison to the three other arm swing instructions ($P < 0.01$). Stride time decreased with increasing gait speed ($P < 0.01$), there was no interaction between gait speed and arm swing instruction ($P = 0.49$). No main effect of arm swing instruction ($P = 0.35$), gait speed ($P = 0.91$) or interaction ($P = 0.57$) on step-width were found (Fig. 1, Table 2).

3.2. Local dynamic stability

There was a significant main effect ($P < 0.01$) of instruction on λ_s in the AP direction (Fig. 2 left panel). Comparing the main effects between arm swing instructions revealed no significant differences, with only a trend for the comparison between excessive and normal arm swinging ($P = 0.06$, $\eta^2 = 0.377$). However, there was an interaction between gait speed and instruction ($P < 0.01$). Post-hoc tests indicated a more stable gait (lower λ_s) for excessive arm swinging at gait speeds 0.56, 0.84 and 1.12 m/s in comparison to normal arm swinging. Lastly, λ_s in the AP direction decreased with increasing gait speed ($P < 0.01$) and (Fig. 2 and Table 2).

There was a significant main effect of arm swing instruction ($P < 0.01$) in the ML direction on λ_s. Post-hoc tests revealed a significant more stable gait pattern (i.e. lower λ_s) for the excessive arm swing instruction compared to normal arm swinging ($P < 0.01$, $\eta^2 = 0.737$). Furthermore, λ_s increased with increasing gait speed ($P < 0.01$, Fig. 2 mid panel, Table 2).

For the VT direction, a main effect of arm swing instruction on λ_s($P = 0.03$) was found. Post-hoc tests for main effects revealed no significant differences between any of the instructions, but a trend for the excessive arm swing instruction compared to normal arm swinging was present ($P = 0.06$, $\eta^2 = 0.274$). There was an interaction between gait speed and arm swing instruction ($P < 0.01$, Fig. 2 right

Table 2
Results of the two way repeated measures ANOVA. F values and corresponding P values are presented for all investigated gait parameters.

	Arm swing		Gait speed		Arm swing × gait speed	
	F	Sig.	F	Sig.	F	Sig.
Spatio-temporal measures						
Stride time	10.827	$P = <0.001$	304.6	$P = <0.001$	0.977	$P = 0.489$
Step-width	1.147	$P = 0.348$	0.425	$P = 0.859$	0.908	$P = 0.569$
Stability measures						
λ_s-0.5 stride AP	5.447	$P = 0.005$	10.097	$P = <0.001$	3.387	$P = <0.001$
λ_s-0.5 stride ML	5.128	$P = 0.007$	11.715	$P = <0.001$	0.67	$P = 0.836$
λ_s-0.5 stride VT	3.402	$P = 0.032$	16.419	$P = <0.001$	2.087	$P = 0.008$
Variability measures						
MeanSD AP	1.49	$P = 0.240$	4.453	$P = 0.01$	1.085	$P = 0.372$
MeanSD ML	1.567	$P = 0.220$	21.06	$P = <0.001$	1.508	$P = 0.093$
MeanSD VT	1.138	$P = 0.351$	9.23	$P = <0.001$	1.26	$P = 0.218$
Stride time variability	1.559	$P = 0.22$	49.24	$P = <0.001$	0.62	$P = 0.876$
Step-width variability	2.881	$P = 0.061$	4.9	$P = <0.001$	1.259	$P = 0.22$

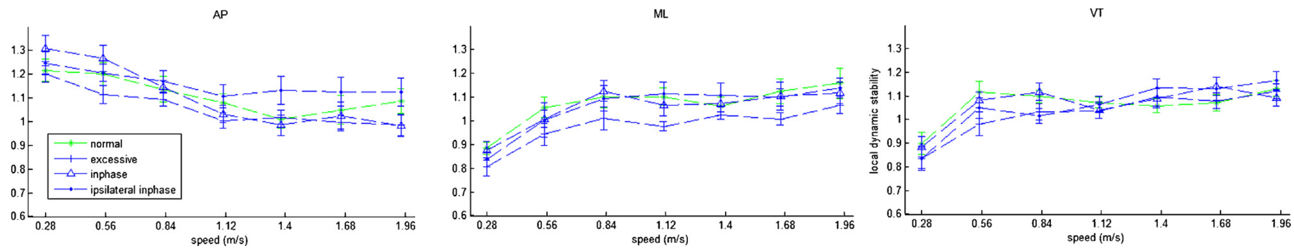

Fig. 2. Dynamic stability quantified as the group-averaged short-term local divergence exponent (λ_s). The left panel presents λ_s for the AP direction, the mid panel for the ML direction and the right panel for the VT direction. Each panel presents all arm swing instructions, see legend in the left panel, and all gait speed conditions. Error bars present the standard error.

panel, Table 2). Post-hoc analyses revealed significantly more stable (lower λ_s) gait for excessive arm swinging in comparison to normal arm swinging at gait speeds 0.28, 0.56 and 0.84 m/s. Furthermore λ_s in the VT direction increased significantly with increasing gait speed ($P < 0.01$).

3.3. Variability

Fig. 3 presents variability measures for all instructions. Significant main effects were found for gait speed, but no main effects were found for arm swing instruction or interaction (see also Table 2).

4. Discussion

The main objective of the present study was to examine the effects of four different arm swing instructions on gait stability and variability; these instructions were (1) without instruction, (2) swinging the arms in phase with each other (without explicit instruction as to how to coordinate these arm movements with the legs), (3) swinging the arms in phase with the ipsilateral legs, and (4) performing a normal arm swing with excessive amplitude. Our results indicate that human gait is most stable (in terms of a lower λ_s for ML direction and a lower λ_s for the AP and VT direction at lower speeds) when arm swing is excessive. These findings seem promising in relation to prevention of falls during gait, since it has been suggested that gait stability in the ML direction is most important [18], because this direction needs control during gait [20].

Our findings are in line with Collins et al. [10] who reported no difference in stability between ipsilateral arm swinging and normal arm swinging, in a model. In addition, our findings on excessive arm swing are partly in line with the work of Hu et al. [7], who reported increased dynamic stability in all movement directions (VT, AP and ML) at preferred gait speed. Our findings are well in line with Nakakubo et al. [8] who reported a higher harmonic ratio for the ML direction when subjects performed an excessive arm swing at preferred gait speed. All in all, there seems to be converging evidence that excessive arm swing increases stability in the ML plane in human gait.

We found only little changes in spatio-temporal parameters due to arm swing instruction. Thus, the greater stability (in terms of λ_s) during excessive arm swing was not due to an increase in step-width. Nor can we conclude that increased stability did cause subjects to decrease step width as has been observed in studies with externally stabilized gait [21]. Moreover, since shorter steps have been suggested to be more stable [22], the change in stride time we found (i.e. larger stride times in all instructions as compared to normal arm swing) is also unlikely to explain the increase in stability.

A possible explanation for the improved gait stability with excessive arm swing is that explicitly controlling arm swing might result in a reduction of neuromuscular noise which could result in a lower λ_s [23]. However, if explicitly controlling arm swing would result in a reduction of neuromuscular noise, one would expect an increased stability in all instructed arm swing instructions, which was not the case. Perhaps the unnatural coordinated arm swing counteracts the effect of explicitly controlling arm swing. Note that the arm–leg coordination pattern in excessive arm swing remained the same as natural arm swing, while this was not the case for the other instructions (see also Table 1). In addition, increased sagittal plane arm movement out-of-phase with the ipsilateral leg reduces

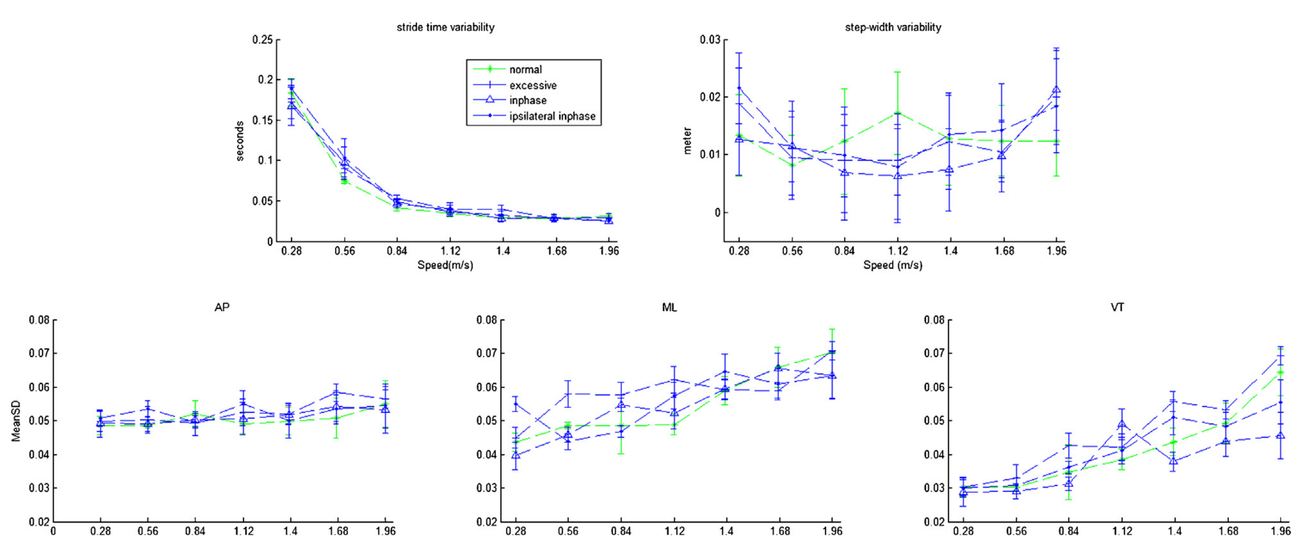

Fig. 3. Group-averaged values of parameters characterizing gait variability. The left upper panel presents stride time variability, the right upper panel step-width variability. MeanSD values are presented at the lower panels from left to right respectively the AP, ML and VT direction. All arm swing instructions are presented, see legend in the left panel. Error bars present standard errors.

whole body angular momentum around the vertical axis [24]. This implies that corrective torques around the hip of the stance leg can be lower when arm movement is more vigorous. The reduced need to produce transverse plane torques around the hip of the stance leg might positively affect frontal plane balance as well given the anatomy of the hip musculature.

Several studies have examined the gait speed–stability relationship [18,25,26]. However, inconsistency in results exists, which is partly caused by different algorithms used [27]. We found slightly increasing λ_s thus reduced stability for the ML and VT direction and a slightly decreasing λ_s trend for the AP direction, implying more stable gait. These results are similar to those of Bruijn et al. [18], and our calculation methods are the same as [23]. At present, these results are hard to interpret, but whether slow walking is more stable or not, reducing gait speed may create more time to avoid unexpected obstacles or reduce the kinetic energy of a fall.

4.1. Study limitations

Our results are based on experiments with young healthy subjects, this was necessary in order to investigate arm swing effects over a wide range of gait speeds to discover potential interaction effects of gait speed and arm swing instructions. Our results are not directly translatable to the more fall-prone elderly. However, Hu et al. [7] found even greater effects of excessive arm swing on stability in an elderly cohort, suggesting that effects that could be obtained in elderly would be under estimated rather than over estimated. Future studies should explore possible effects of excessive arm swing instruction on regaining stability after a perturbation [5,6]. Subjects were mostly well able to perform the different coordination patterns; with only some slightly worse performances during a limited number of strides. The latter has probably reduced power to find effects of the more difficult coordination patterns, i.e. those with deviant phase relations between arm and leg swing to some extent.

4.2. Practical implication

Previous studies on the effect of arm swing on human gait focused on energetic cost of gait [10,28]. Results have demonstrated an increase of energy cost of gait when arm swing was restricted. Our results suggest that people with an increased fall risk should swing their arms excessively, of which the energetic cost are unknown. Arm swing reduces ground reaction moments [10], which probably reduces energy costs, because these moments do not reach zero when using a normal arm swing [24]. Excessive arm swing most likely reduces ground reaction moments and thus energy cost even further. Excessive arm swing might be beneficial for pathological gait as well. Patients with stroke show abnormalities in arm swing [29], but are able to change arm swing amplitudes during gait [30]. It has been reported that doing so may increase gait speed and stride frequency [30]. The current study suggests that besides the previously reported benefits of increasing arm swing amplitude, like greater gait speed, higher stride frequency and more normal trunk coordination [30], excessive arm swing may increase stability, and probably reduce ground reaction moments and energy cost of gait in these groups, although the latter two variables require further investigation.

In conclusion, excessive arm swing significantly increases local dynamic stability of human gait in the ML direction.

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Conflict of interest statement

The authors state that there was no conflict of interests with any financial or personal relationships or organizations that could influence the research results.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.gaitpost.2014.12.002.

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