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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