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Slow walking synergies reveal a functional role for arm swing asymmetry in healthy adults: A principal component analysis with relation to mechanical work

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ABSTRACT

Introduction: The purpose of this study was to reveal a functional role for arm-swing asymmetry during gait in healthy adults. To this end, the primary aim was to investigate the role of neuromuscular control on the asymmetry of propulsive and collision joint work at either end of the double-support phase (W_{DS}) in the context of sidedness. The secondary aim was to investigate the effect of neuromuscular control on propulsive and collision joint work at either end of the single-support phase (W_{SS}) in the context of arm-swing asymmetry.
Methods: Slow -walking trials of 25 participants were analysed using principal component analysis to generate movement synergies (PM_k). Independent variables included the tightness of neuromuscular control (N_1) formulated from the first PM_k and the directional Arm-swing asymmetry index (dASI). Dependent variables included the difference between double-support collision and propulsive joint work (W_{DS}) and a ratio consisting of the difference between single-support collision and propulsive work of both sides (W_{SS}). A linear mixed-effects model was utilized for aim 1 while a multiple linear regression analysis was undertaken for aim 2.
Results: Healthy adult gait was accompanied by a left-side dominant arm-swing on average. For aim 1, N_1 demonstrated a significant negative effect on W_{DS} while sidedness had a negative direct effect and positive indirect effect through N_1 on W_{DS}. The most notable finding was the interaction between dASI and N_1 which demonstrated a highly significant positive effect on W_{SS}.
Interpretation: Evidence was put forward that arm-swing asymmetry during gait is related to footedness among healthy adults. Future studies should look to formally confirm this finding.

1. Introduction

Gait is analogous to an inverted-pendulum like motion where one limb supports the bodyweight and the other moves towards and past this supporting limb to advance the body [1]. The upper-limbs simultaneously swing in asynchrony to the lower-limbs. Evidence suggests that arm-swing is required to minimize the energy expenditure and optimise the dynamic stability of gait [2,3]. Energy efficiency can be improved for example through elevation of the arm at terminal-swing, where the trunk is lifted upwards allowing for reduced collision work at heel-strike [4]. Dynamic stability can be optimised by medio-lateral extension when a perturbation is experienced or the guard posture in anticipation of a fall [5,6].

Arm-swing symmetry is relevant in assessing the early presentation of Parkinson's disease as notable asymmetries are often present [5,7,8]. The presence of arm-swing asymmetry within healthy populations has been noted in the literature [8], suggesting that the aforementioned clinical evaluations could potentially be confounded by pre-existing asymmetries. The aetiology for arm-swing symmetry in healthy populations is not well understood. In patient populations, it has been related to inherent asymmetries in the control of other body segments (e.g. hemiparetic gait) [5]. This upper-limb asymmetry counteracts increased angular momentum within the lower-limbs [9,10]. Arm-swing asymmetry has also been linked to handedness in Parkinson's disease [11,12]. Interestingly, this association between arm-swing asymmetry and handedness has not been emulated in healthy populations despite the majority demonstrating a left-side dominant arm-swing [13,14]. Nonetheless when a left-lateralised task was added in a dual-task walking condition, noticeable increases in arm-swing asymmetry were found indicating the role of cerebral lateralisation [8,13].

It is a frequent practice in research to challenge the neuromuscular system of participants for example by creating dual-task conditions [13], inducing a perturbation [15], standing on unstable support surfaces [16] or walking more slowly than usual [17]. Challenging participants to intentionally walk slower than normal induces lower inter-limb coordination and increases attentional requirements [18].

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Through such challenging methodologies, useful insight can be gained into the mechanisms underlying human movement. Coordinated movement is highly redundant in that the central nervous system (CNS) has more ways than needed for carrying out a given task. Efficient selection of movement strategies is thought to be carried out via the modular control of muscle activity, known as muscle synergies, thereby constraining the multiple degrees-of-freedom available to a task relevant space. However debate is still prevalent on the evidence for this concept [19,20]. [21], through the uncontrolled manifold approach, revealed a modulated control of whole-body angular momentum (WBAM) during gait that was noticeably different during the double- and single-support phases. The double-support phase was characterised as a closed-loop system where postural corrections could take place. The single-support phase was dedicated towards providing step-to-step reproducibility of the WBAM. These opposing roles for WBAM regulation indicate that the neuromuscular system actively intervenes during the double-support phase while only intervening when necessary during the single-support phase. This active intervention is necessary to allow for smooth step-to-step transitions while also preventing the accumulation of small perturbations [1,22,23].

PCA is a machine-learning algorithm which effectively reduces high-dimensional data to a smaller number of orthogonal vector components known as '*Principal movements*' (PM_k) [24]. Each PM_k can be interpreted as correlated marker movements or synergies with higher-order PM_k explaining less variance and representing more subtle movements. The described PM_k can be projected onto a posture space, and in doing so can be represented as '*Principal positions*' with respect to time (PP_k). These PP_k can be differentiated into their 1st- and 2nd-order derivatives, '*Principal velocities*' (PV_k) and '*Principal accelerations*' (PA_k) [24]. The PA_k have been of particular interest as they are thought to represent the action of the neuromuscular system [25,26]. Among other potential variables, the tightness of neuromuscular control on PM_k (N_k) can be determined by the number of zero-crossings (changes in direction) in the corresponding PA_k [26–28]. The N_k variable has proven useful in investigating ageing effects on postural stability in the context of the 'minimum intervention principle' and maturation effects on the temporal postural control of adolescents [26,27].

The purpose of the current study is to reveal a functional role for arm-swing asymmetry in healthy adults. To this end the mechanical joint work conducted during step-to-step transitions of very slow-walking will be related to the tightness of neuromuscular control (N_k), sidedness, and arm-swing asymmetry. More specifically, the following aims were undertaken:

1) determine the relationship between N_k and differences in collision and propulsive work conducted contralaterally in the context of sidedness.
2) establish the relationship between N_k and arm-swing asymmetry with collision and propulsive work conducted ipsilaterally during gait.

From this, a functional role for arm-swing asymmetry can be revealed and whether sidedness plays a role. It is hypothesized that arm-swing asymmetry will counterbalance mechanical work asymmetry in the lower-limbs in a manner that is cohesive with the sidedness effect found.

2. Methods

2.1. Secondary data analysis

Three-dimensional marker trajectories from slow-walking trials of 25 healthy, injury-free adults were taken from a peer-reviewed, open-source dataset [29]. This motion capture data was generated using a 10-camera optoelectronic system (OQUS4, Qualisys, Sweden) sampled at 100 Hz where a 52 markers biomechanical model setup was utilized and marker trajectories were filtered with a 4th-order Butterworth low-pass filter at cut-off frequency of 6 Hz [30]. In the current study, this marker setup was simplified to 36 markers. 13 male and 12 female participants (Age: 32.88 ± 10.6, Height: 1.72 ± 0.1, Weight: 71.4 ± 11.2, BMI: 24.04 ± 2.4) were asked to walk at a speed between 0 m/s – 0.4 m/s (corresponding to a 'household ambulator' [31]) that was coordinated by a metronome on a 10-metre walkway. One right and left gait-cycle per trial/participant (4 trials each) were analysed.

2.2. Synergy extraction

All of the beforementioned procedures were carried out using PManalyzer [32], a MATLAB GUI specifically designed for PM_k computation and visualization. Marker coordinate trajectories from each trial were concatenated and the length of these time-series were normalised to a median range value of 2220 data points for each participant. The individual matrices were then pooled into one 55,500 × 108 input matrix (2,220 data points [Trial duration] x 25 [Number of participants] x 108 [Marker coordinates]) to allow for direct comparisons between participants. In order to eliminate anthropometric differences across participants, this input matrix was firstly transposed so that each time-frame represents a posture vector which were then individually centred by subtracting these vectors from their respective averages, creating normalised postural deviation vectors. These postural deviation vectors were also centred towards the centre-of-mass to avoid the inclusion of body displacements within the PM_k [33]. Finally, the postural deviation vectors were normalised by their mean Euclidean distances to ensure an equal contribution of all participants to the PCA output. This input matrix was then converted to a covariance matrix and decomposed into eigenvectors (PM_k) and eigenvalues using a singular value decomposition algorithm [34].

2.3. Independent variable computation

From this protocol, the PM_k served as an orthonormal basis for the predominant postural movements in the vector space and by projecting the normalised input matrix described previously onto this space, time-series representing deviations from specific mean postures (PP_k) were quantified [27]. These PP_k were inspected using a Fourier analysis for noise. Further filtration with a 3rd-order Butterworth low-pass filter at a cut-off of 10 Hz was deemed necessary to prevent the amplification of noise with differentiation. The first PM explaining 61.7 % of the overall variance was chosen for further analysis only. In previous studies, this PM represented the basic inverted-pendulum motion of gait in the sagittal plane, however at slower speeds this motion is likely to be represented in the medio-lateral direction [15,35]. Nonetheless, this PM is central to step-to-step transitions and is therefore relevant to this studies aims. The number of 'zero-crossings' in the corresponding PA time-series were counted for each participant [26,27,36], formulating the independent variable N_1. A uniform number of gait-cycles were used per participant, eliminating a potential source of variability in terms of the N_1 variable.

The directional Arm-swing asymmetry index (dASI) was formulated to determine the degree of arm-swing asymmetry during gait. The medial wrist marker trajectories for both sides in the sagittal plane were extracted and the anterior-posterior range of motion (ROM) of these trajectories with respect to the participants centre-of-pelvis were taken [14]. The centre-of-pelvis in this case was specified as the centroid of a geometric triangle made up of the two anterior superior iliac-spine markers and the midpoint of the two posterior superior iliac-spine markers [37]. Eq. 1 below illustrates how dASI was calculated using these arm-swing ROMs where **L** is the ROM of the left-arm and **R** the right-side [8,13]. A positive dASI value indicates left-side dominant arm-swing and vice versa.

$$\text{dASI} = \left(\frac{\text{L} - \text{R}}{\max(\text{L}, \text{R})}\right) \times 100 \tag{1}$$

2.4. Dependent variable computation

Using the CusToM toolbox in MATLAB [38], a biomechanical model consisting of 17 rigid body segments linked by 14 joints was generated. This model was scaled to each participants height and weight while segment masses and inertia were calibrated using Body Segment Inertia Parameters [39]. Using the Levenberg-Marquardt algorithm, whole-body 3D inverse kinematics were extracted along with joint torques. Heel-strike and toe-off events were detecting using a kinematic method [40] and these datapoints were taken from the kinematics and joint torque time-series for further analysis. The average joint work done ($\bar{W}$) across trials was calculated as the average joint angular displacement ($\bar{\theta}$) multiplied by the corresponding average joint torque ($\bar{T}$) (Eq. 2). The corresponding average hip- and ankle-joint work for heel-strike and toe-off events respectively were then extracted as collision and propulsion work events respectively.

$$\bar{W} = \bar{T} \times \bar{\theta} \tag{2}$$

The following variables were then created to capture the mechanical joint work involved in step-to-step transitions both contralaterally and ipsilaterally. For aim 1, the difference between $\bar{W}$ for heel-strike event ($\bar{W}_{\text{Heel-strike}}$) and $\bar{W}$ for contralateral toe-off events ($\bar{W}_{\text{Toe-off}}$) was calculated for both sides (Eq. 3). W_{DS} allows one to determine the typical differences in propulsive work at toe-off and collision work done contralaterally at heel-strike, providing a representation of double-support phase mechanical joint work. For aim 2, the absolute difference between right-side $\bar{W}_{\text{Heel-strike}}$ and right-side $\bar{W}_{\text{Toe-off}}$ was divided by the equivalent absolute difference on the left-side (Eq. 4). This ratio W_{SS} compares the symmetry of propulsive and collision work done ipsilaterally between sides, providing a representation of single-support phase mechanical joint work and how they compare across sides. A higher W_{ss} value indicates a preference for right-side collision work. Fig. 1 below further illustrates W_{DS} and W_{SS} where **B** denotes $\bar{W}_{\text{Heel-strike}}$ and **P** signifies $\bar{W}_{\text{Toe-off}}$.

$$W_{DS} = \bar{W}_{\text{Heel-strike}} - \bar{W}_{\text{Toe-off}} \tag{3}$$

$$W_{SS} = \frac{\left| R\,\bar{W}_{\text{Heel-strike}} - R\,\overline{W}_{\text{Toe-off}} \right|}{\left| L\bar{W}_{\text{Heel-strike}} - L\,\bar{W}_{\text{Toe-off}} \right|} \tag{4}$$

2.5. Statistical analysis

As W_{DS} determined differences on both sides and these differences were not incorporated into the one observation as is the case with W_{SS} , a linear mixed-effects model was utilised for aim 1 to determine the effect of groupings within the data. Eq. 5 below illustrates the formula for this analysis in R syntax where $(N_1|Participant)$ is a random-intercepts and slopes term allowing both the start point and slope of the regression line to vary between-participants. N_1, Side and N_1 : Side are fixed-effects terms. N_1 : Side represents the interaction between N_1 and the side of W_{DS}. Mixed-effects modelling was carried out using the lme4 package in R [41]. Side was coded as Right = 1 and Left = 2 so a negative effect in this instance would represent right-side dominance.

$$W_{DS} \sim N_1 + Side + N_1 : Side + \left(N_1 \middle| Participant\right) \tag{5}$$

For aim 2, a multiple linear-regression analysis was undertaken in SPSS in which N_1, dASI and their interaction term (N_1 x dASI) were modelled against W_{SS}.

3. Results

3.1. Slow-walking synergies

As the walking-speed of healthy adults in this analysis was very slow (0.1 m/s – 0.4 m/s), the primary PM_k used in this study captured the predominant movement at this very slow walking-speed, that of upper- and lower-limb frontal plane motion (Fig. 2). This insight was gained by reversing the PM_k normalistion procedure and projecting them onto the posture space which can then be graphically represented [24].

Table 1 provides an outline of the variables of interest. Participants typically had a left-side dominant arm-swing (dASI: 11.06 ± 28.86) but ranged widely from -49.66 to +51.32 for this metric. The average number of direction changes in the primary PA component (N_1) was 109.76 ± 18.15.

3.2. Aim 1

Table 2 describes the output from a linear mixed-effect model described in Eq. 5. N_1 demonstrated a significant negative effect on W_{DS} (β= -57.54 ± 20.87, df = 38.73, $p < 0.01$) while sidedness also demonstrated a large negative effect (β= -2973.98 ± 1425.11, df = 32.79, $p < 0.05$). Conversely the interaction term N_1 : Side demonstrated a significant positive on W_{DS} (β = 29.6 ± 12.81, df = 32.79, $p < 0.05$). Together, these results indicate that with greater tightness of neuromuscular control, W_{DS} is reduced so that negative work is reduced relative to propulsive work. The right-side double-support phase appears to be favoured in terms of WBAM regulation in this cohort as a greater degree of collision work was done on this side. Moreover when N_1 is high on the left-side, the effect of sidedness on lower-limb mechanical work is reduced.

3.3. Aim 2

A multiple-linear regression analysis of N_1, dASI and their interaction term (N_1 x dASI) revealed insignificant direct effects for both N_1 (β = 0.038 ± 0.25, $p > 0.2$) and dASI (β= -0.037 ± 0.162, $p > 0.3$) but a highly significant effect for N_1 x dASI (β = 0.96 ± 0.00) (Table 3). From this, it can be understood that with greater tightness of neuromuscular control in the presence of a left-side dominant arm-swing,

Fig. 1. An illustration of the mechanical joint work being compared contralaterally (W_{DS}) and ipsilaterally (W_{SS}) during step-to-step transitions. **B** denotes the braking work conducted by the hip joint at heel-strike ($\bar{W}_{\text{Heel-strike}}$) while **P** represents the propulsive work done by the ankle joint at toe-off ($\bar{W}_{\text{Toe-off}}$).

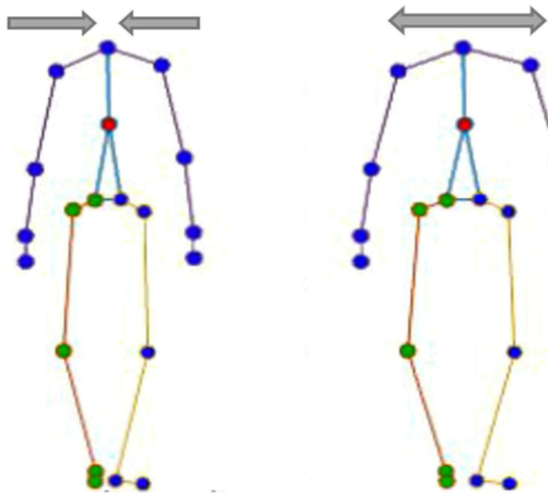

Fig. 2. Frontal plane graphical representations of the primary PM.

Table 1
Descriptive statistics for the variables of interest.

Variable	Mean (SD)	Minimum	Maximum
dASI	11.06 ± 28.86	−49.66	51.32
N_1	109.76 ± 18.15	77	161
W_{SS}	11.52 ± 129.56	−308.49	564.07
W_{DS}	0.77 ± 3.77	−9.16	11.80

Table 2
Findings from a linear mixed effects regression analysis with Eq. 5 as the input.

$W_{DS} \sim N_1 + Side + N_1 : Side + (N_1|Participant)$

Fixed-effects	β coefficient estimate	Degrees of freedom	p-value
N_1	−57.64 ± 20.87	39.27	< 0.01
Side	−2973.98 ± 1425.11	32.79	< 0.05
N_1 : *Side*	29.6 ± 12.81	32.79	< 0.05
Random-effects	Standard deviation		
Participant intercepts	101.42		
Participant slopes	3.59		
Residuals	821.96		

Table 3
Findings from a multiple-linear regression analysis with W_{SS} as the dependent variable.

$R^2 = 0.977$

	Standardized β coefficients	t-value	p-value
N_1	0.038 ± 0.25	1.092	> 0.2
dASI	−0.037 ± 0.162	−1.034	> 0.3
N_1 x dASI	0.96 ± 0.00	25.685	< 0.0001

WBAM regulation is made significantly more asymmetrical. In this cohort, the asymmetry favoured the right-side.

4. Interpretation

The purpose of this study was to reveal a functional role for arm-swing asymmetry in healthy adults by challenging the neuromuscular system via very slow-walking trials. It was hypothesized that arm-swing asymmetry would counterbalance mechanical joint work asymmetries in the lower-limbs in a manner that is cohesive with the sidedness effect found. The findings of this study are in agreement with this hypothesis. In aim 1, N_1 demonstrated a negative effect on W_{DS} while side was also significantly negative in its effect with participants favouring right-side WBAM regulation during double-support and N_1 was beneficial in reducing the degree of collision work conducted. For the secondary aim, the direct effects of N_1 and dASI were insignificant in their effect on $W_{SS,}$ however, their interaction revealed a highly significant positive effect demonstrating that a higher level of N_1 increased the relationship between dASI and lower-limb mechanical work asymmetry (higher left-side dominant arm-swing related to right-side dominant mechanical joint work). These opposing effects of N_1 during single- and double-support phases are in agreement with previous studies demonstrating the reactive- and proctive role of these phases respectively [15,21].

Arm-swing is thought to optimize dynamic stability and minimize the energy expenditure of gait [2–4,6,42]. Previous studies investigating arm-swing asymmetry in healthy cohorts posited this asymmetry to be linked to handedness, where a right-hand dominant population reduces their preferred side for activities of daily-living [13,14]. No association however was found between handedness and arm-swing asymmetry. An explanation for left-side dominant arm-swing in left-handed individuals was cultural mediation via a right-hand dominant society, in what was propositioned with relation to the 'Gunslinger gait' [13,43]. This would however theoretically leave left-handed individuals at a distinct disadvantage in terms of gait stability and energy efficiency if it were the case [4,44].

The findings of the current study suggest that arm-swing asymmetry works as a counterbalance to asymmetry in lower-limb mechanical joint work. Aim 1 findings indicated a right-side predominance for negative mechanical work during double-support. [23] found the coordination of push-off and collision work determined the magnitude of mechanical work conducted during step-to-step transitions, exemplifying the necessity for equal coordination between limbs. A source for this uncoordinated mechanical work during step-to-step transitions was identified in the current study as sidedness which was reduced in the presence of a higher N_1 during the double-support phase. [15] revealed that the magnitude of right - left side weight transfer significantly moderated the relationship between upper- and lower-limb postural corrections also, exemplifying the potential role of sidedness. Asymmetries in joint moment contributions as high as 10 % have been reported in healthy adults [45], explaining perhaps this increased need for angular momentum cancellation in some healthy individuals.

Aim 2 findings support this suggestion further by demonstrating that with a higher N_1, dASI and W_{SS} became increasingly more asymmetrical. A higher N_1 during the single-support phase is deleterious to coordinated gait as this phase is reactive in contrast to the double-support phase, therefore indicating increased difficulty [15,21]. By virtue of the aim 1 findings, there is sufficient insight to determine that sidedness played a compensatory role in this increased difficulty. The aim 2 findings go further by demonstrating that cross-symmetries between the upper- and lower-limbs are actively compensated for by the CNS during a challenging motor task. In hemiparetic gait, the unaffected side arm-swing is typically dominant and acts as a counterbalance to increased angular momentum in the affected lower-limb, exemplifying this point [10]. From this, it can be determined that arm-swing asymmetry is closely related to the degree of asymmetry present in lower-limb mechanical work in healthy adults also. One suggestion for this asymmetric control strategy is cerebral lateralization in lower-limb motor control (i.e. footedness). Footedness has been noted to differ to that of handedness with most right- and left-handed individuals in fact demonstrating right footedness [46,47]. This may explain how most studies have found that healthy, left-handed individuals exhibited a left-side dominant arm-swing [8,13,14]. Interestingly, no association between spatiotemporal gait parameters and arm-swing asymmetry was found in previous research [14]. Moreover no differences between dominant and non-dominant leg symmetry in terms of temporal and kinematic data have been found in the literature [48], however differences in terms of EMG profiles [49] and overall positive work [50] are cited, favouring this proposition.

5. Conclusion

An investigation into the association between kinematic synergies and step-to-step mechanical joint work at the start and end points of both the single- and double-support phases of gait revealed a functional role for arm-swing asymmetry in counterbalancing lower-limb asymmetries that is thought to be related to footedness. Future studies should look to formally confirm this association.

Declaration of Competing Interest

The authors report no declarations of interest.

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