Is angular momentum in the horizontal plane during gait a controlled variable?

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Article history:
Available online xxxx

PsycINFO classification: 2330

Keywords:
Angular momentum
Arm swing
Gait analysis
Interlimb coordination
Load

ABSTRACT

It has been suggested that angular momentum in the horizontal plane during human gait is controlled (i.e., kept minimal). However, this has not been explored in conditions when angular momentum of different segments is manipulated explicitly. In order to examine the behavior of angular momentum, 12 participants walked in 17 conditions in which angular momentum of either the arms or legs was manipulated. Subjects walked at different step lengths, different speeds and with an additional weight on either the wrist or ankle. Angular momentum of total body, arms and legs was calculated from gait kinematics. Increasing step length increased total body and leg angular momentum. When weight was added to the limbs, arm and leg angular momentum were affected and counteracted each other, so that total body angular momentum did not change. Moreover, increasing walking speed increased arm, leg and total body angular momentum. Thus, it may be concluded that if angular momentum is controlled (which only seems to be the case for conditions when weights are added), it is not strictly controlled in all gait conditions (as it may increase by walking faster/with larger steps).

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http://dx.doi.org/10.1016/j.humov.2014.03.003
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Please cite this article in press as: Thielemans, V., et al. Is angular momentum in the horizontal plane during gait a controlled variable? Human Movement Science (2014), http://dx.doi.org/10.1016/j.humov.2014.03.003
1. Introduction

The coordination and function of arm swing in gait have gained considerable interest in the last couple of years (Meyns, Bruijn, & Duysens, 2013). At first sight, arm swing is a somewhat useless behavior, since, unlike movements of the legs, it has no obvious contribution to gait. Nevertheless, it has been suggested that arm swing has several functions in human gait. For instance, it reduces vertical movements of the center of mass (Hinrichs & Cavanagh, 1981; Murray, 1967; Umberger, 2008). Furthermore, gait without arm swing requires more energy (Collins, Adamczyk, & Kuo, 2009), which could be due to the contribution of armswing to the regulation of total body angular momentum in the horizontal plane (Bruijn, Meijer, van Dieën, Kingma, & Lamoth, 2008; Collins et al., 2009; Elftman, 1939; Herr & Popovic, 2008; Park, 2008). This has led to the hypothesis that angular momentum is controlled in human gait (Bruijn et al., 2008; Herr & Popovic, 2008).

Herr and Popovic (2008) hypothesized that angular momentum during the gait cycle is kept minimal throughout the gait cycle, and suggested that the arms are used to cancel angular momentum generated by the legs. Still the study of Herr and Popovic was an observational one, as the authors did not manipulate angular momentum directly. A study by Bruijn et al. (2008) may be more interesting in this regard, as this study suggested that angular momentum might be controlled in the sense that it might be kept as low as possible, since they found no significant change in total body angular momentum over a range of gait speeds, despite increasing leg angular momentum with increasing gait speed. Nonetheless, although angular momentum generated by the legs changed in the study by Bruijn et al. (2008), this was only done through a mediating variable: gait speed. In another study by Donker, Daffertshofer, and Beek (2005), interlimb coordination did not change when angular momentum of segments was directly manipulated by means of adding weights. However, Donker et al. (2005) did not analyze their data in terms of angular momentum. Still, their findings are consistent with the idea that angular momentum remains similar between conditions (since interlimb coordination did not change).

Apart from these studies in healthy subjects, there is converging evidence from pathological gait patterns suggesting that angular momentum may be controlled (i.e., kept minimal) during human gait. A study of Huang et al. (2011) showed that even though low back pain patients move the thorax more in phase with the legs, they keep their arms swinging out of phase with the legs and thus change timing between arm swing and thorax rotations. This is unexpected, but illustrative of the fact that it may be important that the arms remain out of phase with the legs. Additionally, children with cerebral palsy tend to compensate the increased angular momentum created by the affected leg with an increased arm swing on the non-affected side to keep the total body angular momentum small (Bruijn, Meyns, Jonkers, Kaat, & Duysens, 2011).

Angular momentum might be controlled for energetic benefits. Collins et al. (2009) demonstrated that in normal arm swing, out of phase with the legs, the angular momentum remains small and reduces the ground reaction moment. When participants were asked to swing their arms in phase with the legs or when the arms were kept still, it appeared that more energy was required.

All in all, there seems to be converging evidence that angular momentum in the horizontal plane during gait is a controlled variable, in the sense that it is kept minimal throughout the gait cycle. If this would be so, one would expect little to no change in total body angular momentum in a variety of conditions in which angular momentum of the arms or legs is manipulated. Thus, in this study, we looked at the behavior of total body angular momentum in conditions which were designed to change the angular momentum of arms and/or legs. To this aim, we affected angular momentum of either the arms or the legs by means of (1) changing step length, (2) changing gait speed and (3) adding weight to a limb and thereby disturbing symmetry. Angular momentum of the total body, the arms and the legs was calculated. If total body angular momentum indeed is controlled during gait, we expect, despite of these manipulations, that total body angular momentum does not change when compared to normal walking.
2. Methods

2.1. Subjects

Five women and seven men (age 22 ± 1.08 years, weight 68.2 ± 13.85 kg, length 1.76 ± 0.12 m) participated in the experiment. None of the subjects had any orthopedic or neurologic problems which could interfere with gait. The protocol was approved by the local ethical committee and all subjects gave their informed consent before participation. Two participants were left out due to incomplete registration of their markers during the measurements.

2.2. Procedure

Data were collected in the Movement and Posture Analysis Laboratory Leuven, KU Leuven. Before subjects entered the lab, the optoelectronic measurement system (Vicon, Oxford Metrics, Oxford, UK), consisting of 10 cameras (MX T-series), was calibrated and the origin and orientation of the global axes were set. Subjects wore their normal shoes, shorts and t-shirt during the measurement. Anthropometric measurements were taken from each subject unilaterally, assuming subjects were symmetric. These measurements were used for the estimation of mass, center of gravity and moments of inertia of the different segments (Zatsiorsky, 2002). Subjects were then fitted with 39 reflective markers according to the total body Plug In Gait marker set (Davis, Ounpuu, Tyburski, & Gage, 1991), used for 3D motion capture with the Vicon system.

2.3. Conditions

Subjects were instructed to walk as naturally as possible on a treadmill (Forcelink, Culemborg, the Netherlands), in 17 different conditions. During the first part the subjects were instructed to take small, normal and large steps while they walked at 1 m/s (three conditions). Then they walked at a slow (0.5 m/s) and fast (1.5 m/s) speed using normal steps (two conditions).

For the next conditions subjects were fitted with an additional weight of 1% or 2% of body mass (BM) added to the wrist or to the ankle (Fig. 1). In these four conditions they walked at 0.5 m/s, 1 m/s and 1.5 m/s, making for a total of 12 conditions with weights. Weights were consistently added at the right side of the body and conditions followed each other in a random sequence. When the subject was adjusted to the new condition, data were collected during 30 s at 100 samples/s.

![Fig. 1. Schematic representation of no weight condition and four conditions where weight was added: 1% BM at the wrist, 2% BM at the wrist, 1% BM at the ankle, 2% BM at the ankle (from left to right).](image-url)
2.4. Data processing

2.4.1. Pre-processing

Data analysis was done using custom made software in Matlab (The Mathworks, Natick, MA, USA). Kinematics data were first filtered with a second order dual pass low pass Butterworth filter with a cut-off frequency of 5 Hz (resulting in a 4th order low pass filter, due to the recursive filtering). The center of mass, mass and inertial tensor of each segment was calculated using a geometrical model (Zatsiorsky, 2002). A 15 segment 3D model was constructed and for each timeframe, the \( \text{COM}(t_{\text{tot}}) \) was calculated as:

\[
\text{COM}(t_{\text{tot}}) = \frac{ \sum_{j=1}^{15} \text{COM}(t)_j \times m_j }{ \sum_{j=1}^{15} m_j }
\]

in which \( m_j \) is the mass of segment \( j \), and \( \text{COM}(t)_j \) is the \((3 \times 1)\) position vector of the center of mass of segment \( j \) in the global coordinate system at time \( t \).

2.4.2. Basic gait parameters

Heel contacts were determined from the vertical minima of the heel markers, and stride-times were calculated as the time between consecutive heel contacts.

2.4.3. Angular momentum

The angular momentum of each segment in the global coordinate system was calculated with respect to the total body center of mass, according to the formula:

\[
H(t)_j = I(t)_j \times \omega(t)_j + m_j \times r(t)_j \times v(t)_j
\]

in which \( H(t)_j \) is the \((3 \times 1)\) vector of angular momentum of segment \( j \) in the global coordinate system at time \( t \), \( I(t)_j \) is the \((3 \times 3)\) inertia tensor, describing the segment’s inertia in the global coordinate system at time \( t \), \( m_j \) is mass of segment \( j \), \( \omega(t)_j \) is the \((3 \times 1)\) vector of the angular velocity in the global coordinate system of segment \( j \) at time \( t \), \( r(t)_j \) is a \((3 \times 1)\) vector of the distance between \( \text{COM}(t)_j \) and \( \text{COM}(t_{\text{tot}}) \), in the global coordinate system at time \( t \), and \( v(t)_j \) is the \((3 \times 1)\) vector that describes the velocity of \( \text{COM}(t)_j \) with respect \( \text{COM}(t_{\text{tot}}) \), in the global coordinate system at time \( t \), and \( \times \) is the vector product (Zatsiorsky, 2002). To obtain angular momentum in the horizontal plane, we used the third elements of \( H(t)_j \).

The horizontal plane angular momenta of each extremity, consisting of different segments, were calculated as a sum of those segments.

The angular momenta of each extremity, consisting of different segments, were calculated as a sum of those segments. This was done for the angular momenta of the total body, the arms and the legs. The time series of angular moments of the different extremities were time-normalized to a 1–100 time base for each stride cycle, and an average gait cycle was created (see Fig. 2). Next, the mean of the absolute angular momentum over the stride cycle was calculated, and used for further statistical analysis.

2.5. Statistical analysis

Statistica 10 (Stat Soft Inc., Tulsa, OK, USA) was used for statistical analysis. The data were analyzed using ANOVA for repeated measurements (general linear model) for all variables. First, we analyzed the effects of Step length (3 levels) with only step length as within effect. Secondly, effects of adding Weight (5 levels) and Speed (3 levels) were used as within effects. The Tukey’s Honestly Significant Difference test was used as post hoc test to compare the conditions to the ‘normal’ condition. Statistical significance was set at \( p < .05 \).
3. Results

3.1. Influence of step length

Total body angular momentum increased significantly when taking larger steps (main effect of Step length, \( P < .01 \); Fig. 3a). Post hoc tests confirmed a difference between normal and small steps, as well as between normal and large steps. Arm angular momentum (Fig. 3b), on the other hand, was not affected by changing step length (\( P = .48 \)). Leg angular momentum (Fig. 3c) showed an increase with increasing step length (\( P < .01 \)). Post hoc tests revealed a significant difference between small and normal steps and between normal and large steps.

3.2. Influence of adding weight to the right wrist and gait speed

Total body angular momentum (Fig. 4a) increased significantly when walking faster (effect of Speed, \( P = .02 \)). Post hoc analysis showed no significant differences between slow and normal and between normal and fast walking conditions. A significant effect was only seen between slow and fast walking speeds. Adding weight to the right wrist had no influence on total body angular momentum (effect of Weight; \( P = .67 \), effect of Speed \( \times \) Weight; \( P = .99 \)). Left arm angular momentum (Fig. 4b) increased with increasing speed (effect of Speed \( P < .01 \)). Adding weight to the right wrist had no influence on left arm angular momentum (effect of Weight, \( P = .43 \), effect of Speed \( \times \) Weight; \( P = .78 \), Fig. 4b). Right arm angular momentum (Fig. 4c) also showed a significant effect of speed (\( P < .01 \)). Moreover, as was to be expected from our manipulations, right arm angular momentum increased when weight was added (effect of Weight, \( P < .01 \), Fig. 4c). There was no significant Speed \( \times \) Weight effect for right arm angular momentum (\( P = .05 \)). Post hoc tests showed that adding 2% of body mass
to the right wrist increased right arm angular momentum significantly compared to the no weight condition when walking at 1 m/s. Left leg angular momentum (Fig. 4d) significantly increased with increasing speed ($P < .01$) and when weight was added to the right wrist ($P < .01$). Moreover, an interaction effect of Speed $\times$ Weight on left leg angular momentum was present ($P < .01$). Post hoc tests showed that when walking at 0.5 m/s, left leg angular momentum did significantly increase when 1% and 2% of body mass was added to the right arm. No increase was seen when weight was added at gait speeds of 1 and 1.5 m/s. Right leg angular momentum (Fig. 4e) increased with gait speed ($P < .01$), and showed a significant main effect of Weight added to the right wrist ($P = .02$), as well as a significant Speed $\times$ Weight interaction ($P < .01$). Post hoc tests showed that at a speed of 0.5 m/s, right leg angular momentum decreased significantly when weights of 1% and 2% of body mass were added to the right wrist compared to walking without weights. This effect was not seen in the 1 and 1.5 m/s conditions.

### 3.3. Influence of adding weight to the right ankle in different gait speeds

Effects of gait speed on angular momentum have been described in Section 3.2. Therefore, in this section, we limit ourselves to describing the effects of adding weight to the ankle (and potential interactions of speed with this effect).
No significant change in total body angular momentum (Fig. 5a) was seen when weight was added at the right ankle (Weight, $P = .40$, Speed $\times$ Weight, $P = .36$). Similarly, left arm angular momentum (Fig. 5b) was not affected by adding weight to the right ankle (Weight, $P = .10$, Speed $\times$ Weight $P = .92$). Right arm angular momentum (Fig. 5c) however, was significantly affected by adding a weight to the right ankle ($P < .01$), and this effect was speed dependent (Speed $\times$ Weight, $P = .04$). Post hoc analyses showed no effect of weight added to the right ankle at 0.5 m/s, but a clear effect at 1 and 1.5 m/s.

Fig. 4. Group means of angular momentum of total body (a), left arm (b) and right arm (c), left leg (d) and right leg (e) with no weight (white columns), 1% (grey columns) and 2% (black columns) of body mass added at the wrist. Subjects walked at 0.5, 1 and 1.5 m/s. Error bars represent standard error of the mean.
1.5 m/s where right arm angular momentum was significantly higher than normal when 2% of body mass was added to the right ankle.

For left leg angular momentum (Fig. 5d), significant effect of Weight ($P < .01$), and Speed $\times$ Weight ($P < .01$) were present. Post hoc tests showed a significant increase in left leg angular momentum compared to normal when 1% and 2% of body mass was added to the right ankle at a speed of 0.5 m/s. Adding 2% of body mass to the right ankle at 1 and 1.5 m/s increased left leg angular
momentum significantly compared to normal, while adding 1% of body mass did not. As was to be expected given the experimental manipulation, right leg angular momentum (Fig. 5e) showed a significant effect of Weight ($P < .01$). However, right leg angular momentum also showed a Speed × Weight interaction ($P < .01$). When walking at 1 and 1.5 m/s, adding 1% and 2% of body mass to the right ankle significantly increased right leg angular momentum compared to the no weight condition. No such effects were seen 0.5 m/s.

4. Discussion

We studied total body angular momentum in the horizontal plane in human gait, when the gait pattern was modified in several ways to affect the angular momenta created by different segments. Specifically, we manipulated normal gait by asking subjects to walk with different step lengths, at different speeds, and with an extra weight added to the wrist or ankle. Apart from total body angular momentum we also investigated to what extent the angular momenta of the arms and legs changed in these conditions.

4.1. Influence of step length

When subjects walked with larger steps, total body and leg angular momentum increased, but changing step length had no influence on arm angular momentum. Thus, angular momentum during human walking is not always kept minimal, and close to that of normal walking. Still, it may be that angular momentum is an important proxy for another important variable in human gait, namely, the free vertical moment (Li, Wang, Crompton, & Gunther, 2001). The free vertical moment is the derivative of total body angular momentum. When taking smaller steps, the amplitude of angular momentum decreases but the frequency increases, which may make that the free vertical moment may be similar as in gait with larger steps at the same speed (where the amplitude of angular momentum is higher, but the frequency lower). Indeed, Li et al. (2001) found that the vertical moment in running is alike in magnitude as in walking but shows more variability and has a different timing over the gait cycle than in walking. This study also indicated that arm swing and ground reaction moments tend to cooperate to counter the moments created by the legs. Future measurements of ground reaction moments are recommended to give a decisive answer. Alternatively, since in the small and large steps conditions the subjects were asked to deliberately change the way they move, it may be that this generates behavior in which keeping angular momentum minimal is a less important control objective. This would seem plausible, as high angular momentum would require high breaking moments (Li et al., 2001), potentially involving high energetic demand (Collins et al., 2009) and thus seems undesirable during normal walking.

4.2. Influence of speed

Higher speeds led to an increase in total body, arm and leg angular momentum. A part of the increased arm and leg angular momentum was neutralized by the fact that these segments move out of phase with each other. Donker, Beek, Wagenaar, and Mulder (2001), Donker et al. (2005) indicated an increase in arm movement amplitude and arm swing frequency at higher gait speeds. This can contribute to a larger angular momentum of the arms in order to counteract the angular momentum created by the legs. It is also shown that the variability in arm-leg coordination decreases with increasing speed (Donker, Beek, Wagenaar, & Mulder, 2001; Donker et al., 2005) which allows for better cancellation of angular momentum. Furthermore interlimb coordination pattern improves in children with cerebral palsy when they were asked to walk as fast as possible (Meyns et al., 2012). Interestingly, our current findings are different from those of Bruijn et al. (2008), who found no difference in total body angular momentum over speeds from ~0.55 to ~1.44 m/s. Still, in Bruijn et al. (2008), there was a tendency for angular momentum to increase (see figures in Bruijn et al., 2008), furthermore post hoc, our results only show a significant difference between ‘Slow’ and ‘Fast’ walking conditions. In yet another study by Bruijn et al. (2011), no increase was seen in total body angular momentum (speeds in
this study ranged from \(-1\) to \(-2\) m/s). However, it has to be noted that Bruijn et al. (2011) only included children from age seven to nine. Lastly, a study by Bennett, Russel, Sheth, and Abel (2010) showed a decrease in extrema of total body angular momentum with increasing gait speed (speeds in this study ranged from \(-0.89\) to \(-1.57\) m/s). However Bennett et al. (2010) normalized angular momentum to walking speed. Since walking speed is not directly a subject characteristic, this normalization seems undesirable, and it seems preferable to normalize angular momentum only to factor affecting subject characteristics (e.g., mass, length) (Hof, 1996). All in all, reported effects of gait speed on angular momentum seemed to be either small (i.e., a small increase with increasing speed in the current study, a small decrease in extrema with increasing speed in Bennett et al. (2010)), or non-present (Bruijn et al., 2008; Bruijn et al., 2011). This leaves the question as to whether total body angular momentum is influenced by gait speed unclear.

4.3. Influence of weight

Adding weights to the right wrist or ankle did not change total body or left arm angular momentum. Right arm angular momentum increased with extra weight, this was only significant at a speed of 1 m/s, but we see a likewise tendency for all speeds (Figs. 4 and 5). For the other limbs there was a tendency to increase in left leg angular momentum and a tendency to decrease in right leg angular momentum observed. When weight was added to the right leg, all segments, except for the left arm, showed an overall tendency to increase. It appears that changes in angular momentum of one limb can be compensated through changes in angular momentum of the other limbs. In children with cerebral palsy, the unaffected side compensates the increased angular momentum created by the affected side (Bruijn et al., 2011). Donker, Mulder, Nienhuis, and Duysens (2002) showed that adding mass to a limb did influence movements of both the loaded and unloaded arm, which can explain the altered angular momentum of the limbs in our study. However no difference in coordination pattern of the limbs was seen (Donker et al., 2005) and the current study showed no changes in total body angular momentum when weight was added to a limb. Therefore it seems that the conservation of interlimb coordination plays a critical role in controlling total body angular momentum.

4.4. Limitations

In estimating segment mass and inertia, we assumed subjects were symmetric, which is not always the case, taking into account the dominant side. The side that was measured was randomly chosen, without regard to any dominancy over all subjects.

Weight was consistently added to the right side, without taking into account the dominant side. Muscles may be more developed and neuro-motor control may be different on the dominant side. The amount of weight we added to the limbs might be too low, and the use of higher weights might give another outcome. In two studies of Donker et al. (2002), Donker et al. (2005), ‘bracelets’ of 1.8 kg were used as additional weight. The average 2% of body mass of our subjects was only 1.39 kg.

Treadmill gait may differ somewhat from overground gait (Vogt, Pfeifer, & Banzer, 2002). These differences are not believed to lead to changes in components attributing to angular momentum. There are several methodological limitations to this study; for instance, we used a kinematic marker for heel strikes, and used estimated inertial parameters. However, the fact that the angular momenta curves (Fig. 2) are similar to those reported by different research groups (Bruijn et al., 2008; Bruijn et al., 2011; Herr & Popovic, 2008), makes us confident that these factors influenced our results only little. Lastly, we reported only on mean absolute angular momentum, not maxima (or minima). However, post hoc, we analyzed maxima of total body angular momentum, and results were similar to what we found through the analysis of the mean absolute values.

4.5. Clinical relevance

In addition to when gait is artificially altered, such as in the present study, the literature indicates that total body angular momentum is controlled in pathologies where an altered coordination pattern occurs. Low back pain patients avoid lumbar torsion which leads to an in phase movement of pelvis
and thorax (Huang et al., 2011). The arms, on the other hand, stay out of phase with the legs and, therefore, out of phase with the thorax which seems unnatural. Despite central nervous system damage in children with hemiparetic cerebral palsy, the non-affected arm swings with an amplitude almost twice as large as the affected arm (Meyns et al., 2011) and thereby creates a larger angular momentum to neutralize the angular momentum created by the affected side. This leads to a total body angular momentum with similar values as healthy subjects (Bruijn et al., 2011).

The question remains what the consequence for other neurologic pathologies is (i.e., adults with hemiplegia or other asymmetric disorders), where people have to manage reorganization of angular momentum post-trauma.

5. Conclusion

Our results suggest that angular momentum during walking can change in some gait situations. This could also indicate that not angular momentum but the free vertical moment is the actually controlled. Alternatively, angular momentum is controlled in a less strict manner, in which the aim is to keep angular momentum minimal in the view of other constraints.

Acknowledgements

S.M.B. was supported by a grant from the Netherlands Organisation for Scientific Research (NWO #451-12-041) and an F.W.O. grant (G.0901.11). P.M. was supported by a grant from ‘bijzonder onderzoeksfonds’ KU Leuven (OT/08/034 & PDMK/12/180).

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