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Whole-body and segment angular momentum during 90-degree turns

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ABSTRACT

Background: Turning is a frequently performed, asymmetric task of daily living. The asymmetric nature makes turning challenging to perform while maintaining balance.
Research question: How do healthy individuals maintain dynamic balance, quantified as whole-body angular momentum, during a 90-degree turn compared to straight-line walking?
Methods: The kinematics of sixteen healthy individuals were tracked during walking in a straight-line and during left and right 90-degree turns at a comfortable pace. Whole-body and segment angular momenta were calculated and the relative contributions of the legs, arms, pelvis and head/trunk to whole-body angular momentum were evaluated.
Results: Average whole-body angular momentum was different during turning compared to straight-line walking in all planes of motion. The initiation of a turn required generation of whole body angular momentum in all three planes of motion, which was counteracted at the end of the turn by a generation of angular momentum in the opposite direction in the frontal and sagittal planes. Transverse plane momentum was always directed in the turn direction. All segment groups, except for the inside leg, had a greater magnitude of angular momentum during turning compared to straight-line walking. The outside leg and head/trunk segments were the largest contributors to frontal and transverse plane whole-body angular momentum.
Significance: Understanding how body segments contribute to maintaining balance during a 90-degree turn can be useful for designing rehabilitation paradigms for people who have difficulty turning or impaired balance.

1. Introduction

Turns make up 35–45% of the steps taken on a daily basis [1]. While frequently performed, turns can be challenging to complete while maintaining balance as it requires a greater coefficient of friction than walking in a straight-line [2,3] making it more likely to cause slipping and loss of balance. In fact, people frequently report turning at the time of falling [4]. Falls during turning are especially dangerous as 70% result in injury [5]. In addition, falls during turning result in eight times the number of hip fractures compared to falls during straight-line walking [6]. People with neurological impairment such as Parkinson's disease [7], multiple-sclerosis [8], or stroke [9] are more likely to suffer major injuries during a fall compared to healthy individuals. This increased likelihood of fall-related injury is due to their reduced ability to recover from a loss of balance caused by a number of factors, such as muscle weakness [9,10] and proprioception [10,11].

Turning requires asymmetric kinematics and force generation [12] to quickly re-orient the head, trunk, and pelvis [13]. In particular, turns are anticipated by directing gaze to the desired heading direction [13,14], which is followed by trunk and pelvis rotation in the direction of the turn [15,16]. At the same time, shorter strides are taken with the limb on the inside of the turn compared to the limb on the outside [17,18]. Changing direction also requires greater medial ground reaction impulses on the outside limb to redirect the body center of mass toward the contralateral limb [17,19]. The altered base of support and greater ground reaction forces (GRFs) may contribute to the increased risk of falls during turning compared to straight-line walking. However, little is known about how people maintain stability while successfully executing a turn.

Whole-body angular momentum ($\vec{H}_{WB}$) is a common measure used to assess dynamic balance during locomotion [20–23]. The time rate of change of ($\vec{H}_{WB}$) equals the net external moment about the body center of mass. Accordingly, $\vec{H}_{WB}$ is altered through changes in body kinematics, foot placement, and GRFs. In addition, $\vec{H}_{WB}$ is regulated through interchanging angular momenta generated by different body segments

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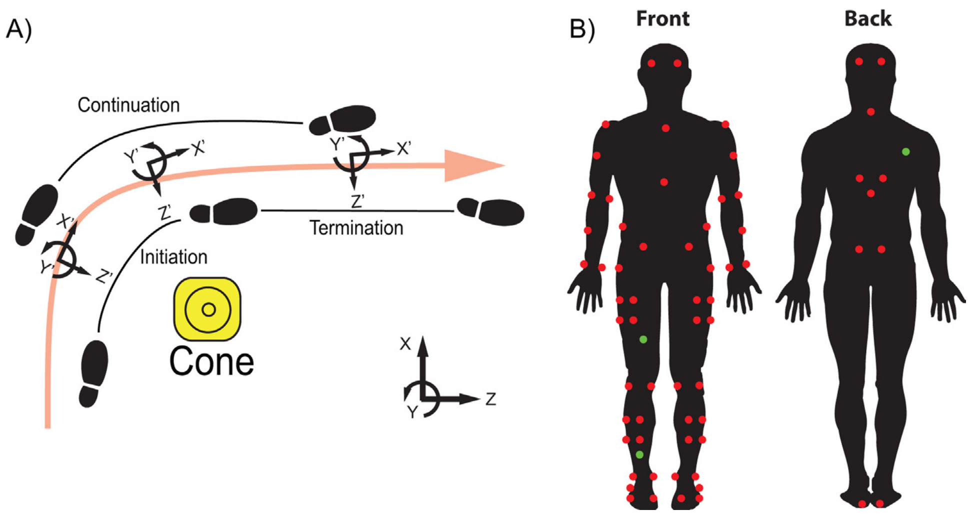

Fig. 1. (A) Stride definitions during 90-degree turning around a cone with approximate rotated coordinate frames, based on the body reference frame, throughout the turn. (B) Marker set used for data collection where red markers were used for joint definition and/or segment tracking and green markers were used as reference markers. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

Table 1
Average (standard deviation) dimensionless whole-body angular momentum, and walking speed during the gait cycle for right and left turns and the right and left strides during walking in a straight line. Pairwise comparisons were performed between right and left strides for straight-line walking, between the initiation, continuation, and termination strides for right and left turns, and between turn and straight-line walking strides. Significant differences between right and left turns (*) and between turn strides and straight-line walking (§) were defined by $p < 0.05$.

	Straight-line walking		Turn initiation		Turn continuation		Turn termination	
	Right	Left	Right	Left	Right	Left	Right	Left
Whole-body angular momentum								
Frontal	−0.0001 (0.0002)	−0.0001 (0.0003)	−0.0017 (0.0017)	−0.0021 (0.0033)	−0.0009 (0.0015)	−0.0007 (0.0031)	0.0032 (0.0019)	0.0033 (0.0015)
Sagittal	−0.0013 (0.0005)	−0.0012 (0.0004)	−0.0026 (0.0018)	−0.0016 (0.0018)	−0.0010 (0.0015)	−0.0001 (0.0015)	0.0026 (0.0012)	0.0031 (0.0014)
Transverse	0.0001 (0.0002)	0.00005 (0.0002)	0.0093 (0.0021)	0.0089 (0.0028)	0.0110 (0.0017)	0.0102 (0.0017)	0.0060 (0.0012)	0.0060 (0.0017)
Walking Speed (m/s)	1.22 (0.21)	1.22 (0.21)	0.97 (0.15)§	1.18 (0.15)*	1.16 (0.14)	1.00 (0.14)*§	1.18 (0.16)	1.17 (0.14)

[20,21]. For example, during straight-line walking, the head and arm segments contribute more to $\vec{H}_{WB}$ in the transverse plane than they do in the frontal and sagittal planes, relative to the other segments [20]. In addition, angular momenta for the legs provide nearly equal and opposite contributions to $\vec{H}_{WB}$ over the gait cycle, thus contributing to a small net $\vec{H}_{WB}$ [20,21]. Angular momentum is also regulated differently across tasks. For example, there is a smaller range of $\vec{H}_{WB}$ in the sagittal plane when walking up and down stairs [23] and down slopes [22] compared to level ground walking. Given the altered mechanics used to complete a successful turn, turning movements should also require altered generation of $\vec{H}_{WB}$. However, differences in $\vec{H}_{WB}$ between straight-line walking and turning have yet to be tested experimentally.

Therefore, the purpose of this study was to determine how healthy individuals regulate $\vec{H}_{WB}$ during the execution of a 90-degree turn. As people alter their speed throughout the turn, they also increase their braking forces early in the turn and propulsion forces later in the turn. As such, we hypothesized that $\vec{H}_{WB}$ in the sagittal plane would be anteriorly directed when initiating the turn and posteriorly directed throughout the remainder of the turn. With the changes in force and asymmetry during turning, we hypothesized that, when normalized by walking speed, greater $\vec{H}_{WB}$ would be generated toward the turn in the transverse and frontal planes compared to straight-line walking. Additionally, we hypothesized that the segments on the outside of the turn would have greater contributions to ($\vec{H}_{WB}$) compared to segments on the inside of a turn. The results of this study provide an understanding of how people maintain balance during a turn, which has implications for rehabilitation training and assistive device design for people who may be more susceptible to falling.

2. Methods

2.1. Participants

Sixteen healthy adults (9 female; 30.8 ± 12.8 years; 76.7 ± 17.5 kg; 1.75 ± 0.1 m) provided their informed consent to participate in the experimental protocol approved by the University of Michigan Institutional Review Board. Potential participants were recruited through an online database (https://umhealthresearch.org/) and screened to ensure they did not have a history of cardiovascular or neurological disease, significant injury to their lower limbs or back, uncorrected vision problems, take any medications that affected their ability to walk, or have any mental capacity impairment that would negatively affect verbal communication.

2.2. Experimental protocol

Participants first walked in a straight line and then performed 90-degree turns first to the right and then to the left (Fig. 1A) at a self-selected speed. Each trial was collected after participants reached a steady speed following a 2.4 m walkway. Participants were instructed to turn as if they were walking around a hallway corner to ensure that they continued through the turn without stopping. Each turn was visually determined to be a step-turn when the continuation stride was taken with the outside leg [19]. Left and right turns were performed until five step-turns were completed. Each task was performed until five

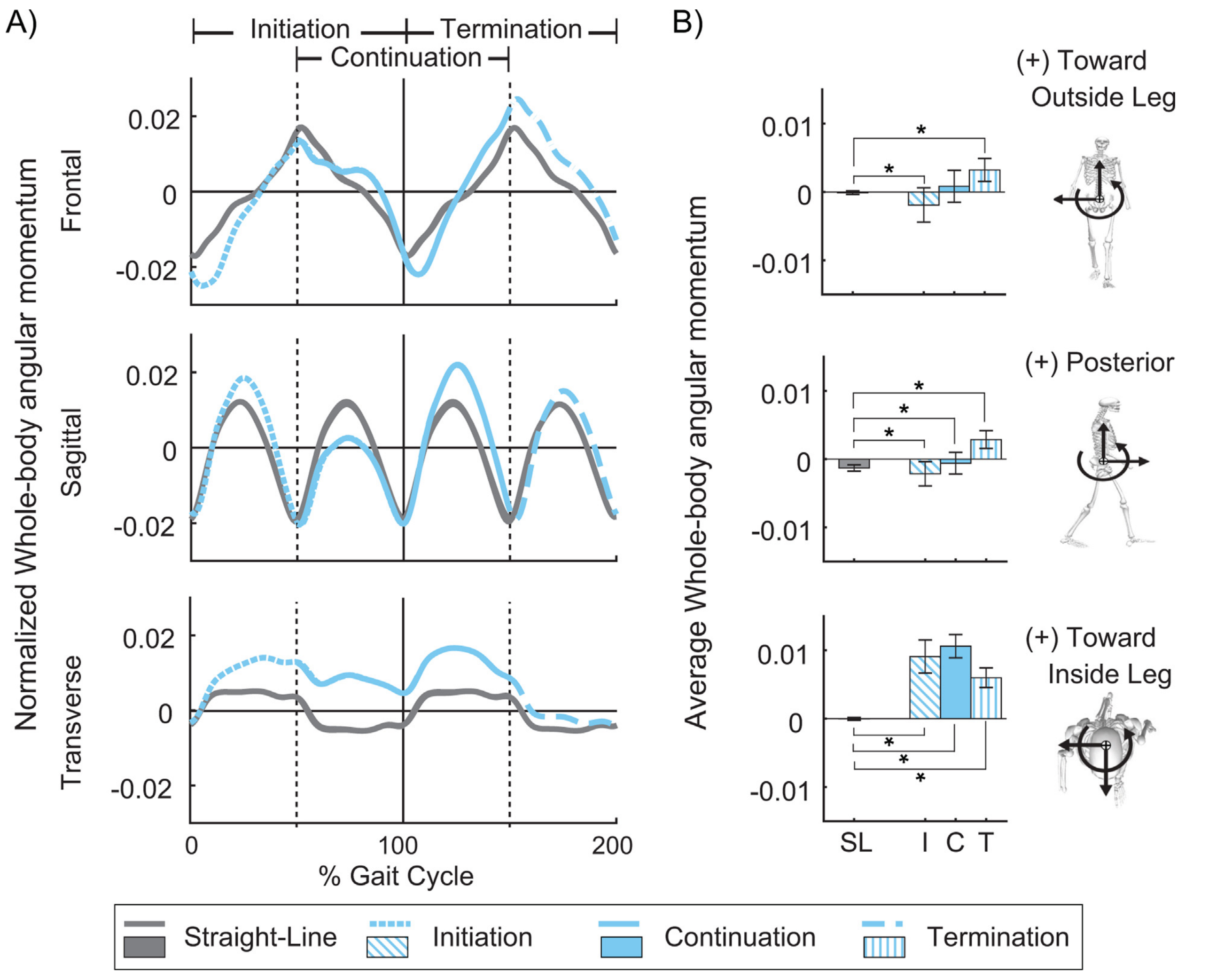

Fig. 2. (A) Dimensionless whole-body angular momentum, $\vec{H}_{WB}\left(\frac{\text{kg}\cdot m^2}{s\cdot(\text{kg}\cdot m\cdot\frac{m}{s})}\right)$, trajectories across the gait cycle during straight-line walking and turning strides (initiation, continuation, and termination) in the three planes of motion. Solid vertical line corresponds to the end of the initiation stride and the beginning of the termination stride. Dotted vertical lines represent the beginning and end of the continuation stride. (B) Average whole-body angular momentum, $\vec{H}_{WB}\left(\frac{\text{kg}\cdot\text{m}^2}{\text{s}\cdot(\text{kg}\cdot\text{m}\cdot\frac{\text{m}}{\text{s}})}\right)$, over the gait cycle for the initiation (I), continuation (C) and termination (T) strides during turning, and for one straight-line (SL) walking stride. Significant differences ($p < 0.05$) between turn strides and straight-line walking are shown with asterisks (*).

strides with full kinematics were collected. A 20-camera motion capture system (Motion Analysis, Santa Rosa, CA) tracked full-body motion at 120 Hz using a six degree of freedom marker set with 67 reflective markers during all trials (Fig. 1B). GRF data were only available for the part of the turn and for only 12 participants. These data are included as supplementary material.

2.3. Data analysis

Marker position data were filtered using a 4th order low-pass Butterworth filter with a cutoff frequency of 6 Hz. A 13-segment model consisting of the head, trunk, upper arms, forearms, pelvis, thighs, shanks, and feet was constructed in Visual3D (C-Motion, Germantown, MA) using segment markers and landmarks to estimate inertial properties [24,25]. Local coordinate systems for each segment were defined based on International Society of Biomechanics guidelines [26,27]. Whole-body center of mass (COM) location was calculated as the weighted average of each segment's COM. Segment ($\vec{H}_i$) and whole-body angular momenta ($\vec{H}_{WB}$) were calculated as:

$$\vec{H}_i = \left[(\vec{r}_i^{\,COM} - \vec{r}_{body}^{\,COM}) \times m_i(\vec{v}_i^{\,COM} - \vec{v}_{body}^{\,COM}) + I_i\vec{\omega}_i\right]$$

$$\vec{H}_{WB} = \sum_{i=1}^{n} \vec{H}_i$$

where $\vec{r}_i^{\,COM}$, and $\vec{v}_i^{\,COM}$ were the position and velocity of the i-th segment's COM, $\vec{r}_{body}^{\,COM}$ and $\vec{v}_{body}^{\,COM}$ were the position and velocity vectors of the whole-body COM, m_i was the mass of the segment i [25], I was the segment moment of inertia, $\vec{\omega}_i$ was the segment angular velocity, and n was the number of segments. Angular momentum was then divided by body mass (kg), height (m), and walking speed (m/s), making it dimensionless [20,22].

To interpret angular momentum in the anatomical planes of motion, we rotated angular momentum vectors from the global reference frame (X, Y, Z) to the participants' path trajectory (X', Y', Z') (Fig. 1B) at every time point [13,19]. The angle of rotation was then defined as the angle between the heading direction and the global reference frame where the heading direction was the forward component of the path trajectory, which was determined by the COM linear velocity vector in the

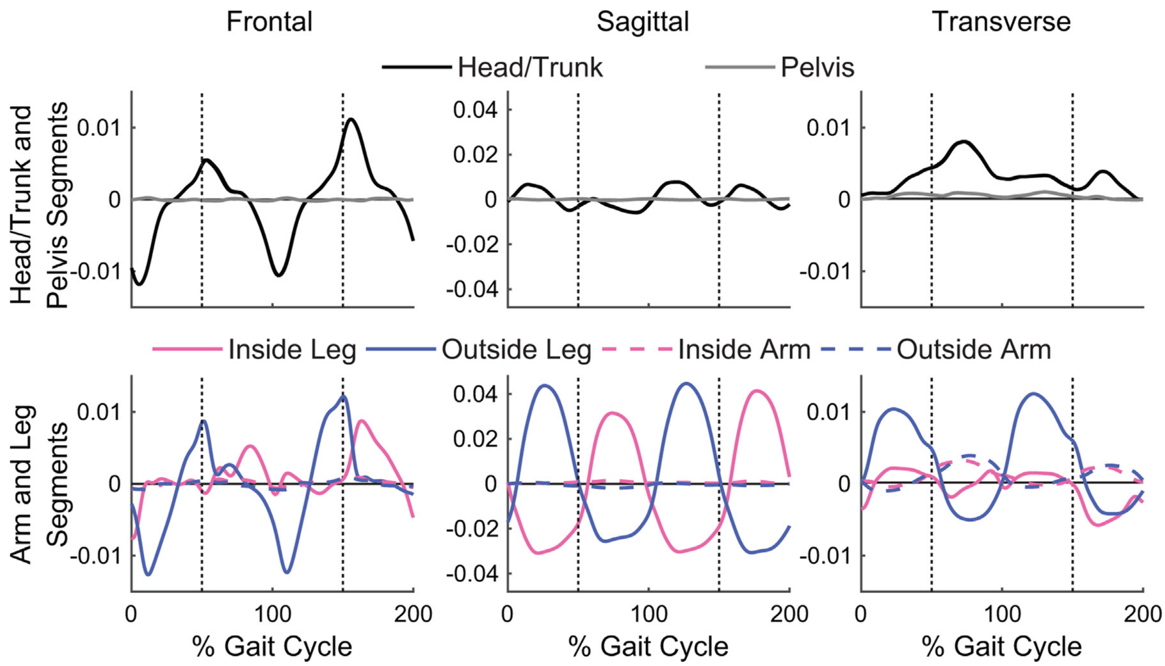

Fig. 3. Dimensionless segment angular momenta $\left(\frac{kg \cdot m^2}{s \cdot (kg \cdot m \cdot \frac{m}{s})}\right)$ trajectories over the complete turn. The initiation stride is 0–100 percent, the continuations stride is approximately 50 percent to 150 percent (vertical dashed lines) and the termination stride is 100 to 200 percent of the gait cycle.

horizontal plane.

$\vec{H}_{WB}$ and segment angular momenta were first time-normalized to 0 to 100% of a stride and then averaged over the five trials for each participant. Turning was defined by initiation, continuation, and termination strides (Fig. 1B). The initiation stride consisted of the first heel-strike, before the pelvic rotation angle was greater than 20 degrees, and the second heel-strike taken in the middle of the turn. The continuation stride and termination strides were the subsequent strides taken with the legs outside and inside of the turn, respectively.

We calculated the mean angular momentum across all strides in each plane of motion for each of the six segment groups: head/trunk, pelvis, right arm (upper arm, forearm), left arm, right leg (thigh, shank, foot), and left leg. In addition, we quantified positive and negative contributions to $\vec{H}_{WB}$ as the area under the angular momentum versus gait cycle curve.

2.4. Statistical analysis

The primary dependent measures were average $\vec{H}_{WB}$, segment angular momentum, walking speed, and segment contribution during turning and straight-line walking.

We first compared the mean angular momentum across the gait cycle for left and right turns and left and right straight-line strides using paired t-tests ($\alpha = 0.05$). There were no significant differences in any plane, therefore left and right turns were combined and left and right straight-line strides were combined, separately, in subsequent analyses. We then tested for differences in the average $\vec{H}_{WB}$ and segment angular momenta between tasks (straight-line walking and turning) for each plane of motion of the three turning strides using a series of paired t-tests with $\alpha = 0.05$. Finally, we compared the contributions to $\vec{H}_{WB}$ from the six segment groups using separate one-way repeated measures ANOVAs for each of the turning strides and each plane of motion. The sphericity assumption was tested using Mauchly's test of sphericity with the Greenhouse-Geisser correction applied to the analyses. Post-hoc pairwise comparisons explored significant differences between the six groups with a Bonferroni adjustment to correct for multiple comparisons ($\alpha_{adj} = 0.0056$). All statistical analyses were performed in SPSS 24 (IBM, Armonk, NY).

3. Results

3.1. Walking speed

There were no differences in walking speed between the left and right straight-line strides ($F = 0.03$, $p = 0.87$). Participants initiated left turns faster than right turns ($F = 51.12$, $p < 0.001$; Table 1), but were slower during the continuation stride when turning left ($F = 44.85$, $p < 0.001$). There were no differences in speed during the termination stride between right and left turns ($F = 1.58$, $p = 0.23$). In addition, only the right turn initiation ($F = 20.02$, $p < 0.001$) and the left turn continuation ($F = 22.49$, $p < 0.001$) strides were slower compared to straight-line walking.

3.2. Whole-body angular momentum

$\vec{H}_{WB}$ trajectories differed between turning and straight-line walking (Fig. 2A). There were significant differences in average $\vec{H}_{WB}$ in the frontal plane during the initiation ($F = 14.96$, $p = 0.001$) and termination ($F = 118.47$, $p < 0.001$) strides, compared to straight-line walking (Fig. 2B). Participants generated greater $\vec{H}_{WB}$ towards the inside of the turn during the initiation stride and greater $\vec{H}_{WB}$ towards the outside of the turn during the termination stride. Compared to straight-line walking, average $\vec{H}_{WB}$ in the sagittal plane was more anteriorly directed during the initiation stride ($F = 6.92$, $p = 0.014$) and more posteriorly directed during the continuation ($F = 6.15$, $p = 0.019$), and termination ($F = 314.094$, $p < 0.001$) strides. Average $\vec{H}_{WB}$ in the transverse plane was more positive during all turning strides ($F > 402.74$, $p < 0.001$) compared to straight-line walking. The positive value indicates that participants generated $\vec{H}_{WB}$ momentum in the direction of the turn.

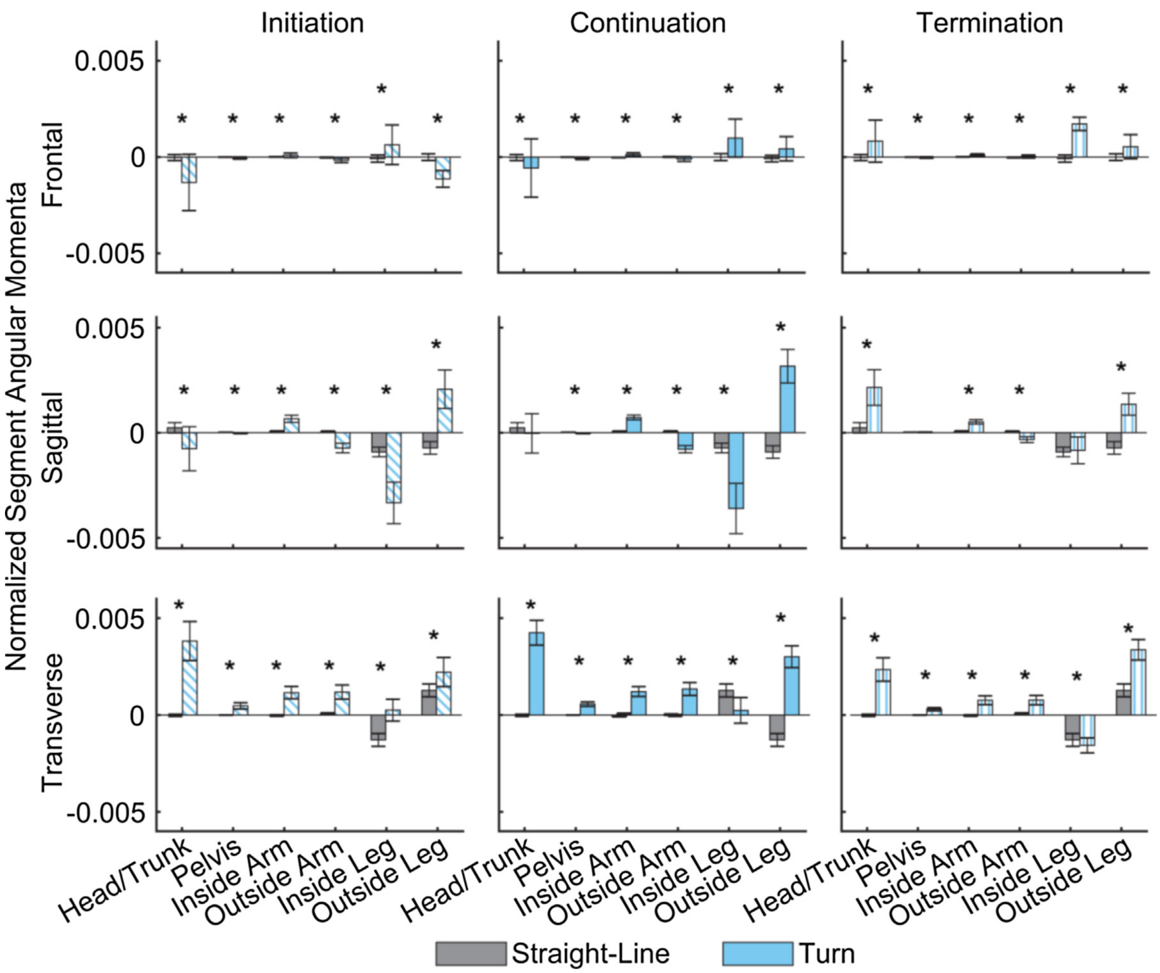

Fig. 4. Average dimensionless angular momentum $\left(\frac{\mathrm{kg \cdot m^2}}{\mathrm{s \cdot (kg \cdot m \cdot \frac{m}{s})}}\right)$ for each segment group during each stride of turning (initiation, continuation, termination) and straight-line walking in each plane of motion. Significant differences ($p < 0.05$) between turn strides and straight-line walking are shown with asterisks (*).

3.3. Segment angular momentum

Average segment angular momenta were generally different between turning and straight-line walking (Fig. 3). In the frontal plane, all segments had a larger magnitude of average angular momentum during turning compared to straight-line walking ($F > 8.33$, $p < 0.007$; Fig. 4). Similarly, in the sagittal plane, all segments had greater average angular momentum magnitude during turning compared to straight-line walking ($F > 68.33$, $p < 0.001$) except for the head/trunk segment during the continuation stride and the pelvis and inside leg segments during the termination stride. In the transverse plane all segments generated greater angular momentum during turning compared to straight-line walking ($F > 23.13$, $p < 0.001$), except for the inside leg during the initiation and continuation strides. During the initiation and continuation strides, the inside leg generated a smaller magnitude of angular momentum in the opposite direction during initiation and in the same direction during continuation compared to straight-line walking ($F > 263.32$, $p < 0.001$).

The main effect of segment groups was significant for both negative and positive contributions, as measured by the area under the curve ($F > 209.17$, $p < 0.001$; Fig. 4). The outside leg had a greater negative contribution for all strides compared to the inside leg in the frontal plane ($p < 0.001$; i.e., toward the inside of the turn) and during the initiation and continuation strides in the transverse plane ($p < 0.001$; i.e., away from the turn). The inside leg contributed more negatively (i.e, anterior) than the outside leg for all strides in the sagittal plane ($p < 0.001$) and during the termination stride in the transverse ($p < 0.001$) plane. The outside arm had greater negative contribution in all planes ($p < 0.009$) except the continuation and termination strides in the transverse plane. As for positive contributions, outside and inside limbs were only significant in the sagittal plane ($p < 0.001$). However, the comparisons differed in that the inside arm had greater positive contribution (i.e., posterior) compared to the outside arm and vice-versa for the legs. In addition, the outside leg contributed more than the inside leg in the transverse plane ($p < 0.001$) and during the continuation ($p < 0.001$) and termination strides ($p < 0.001$) in the frontal plane (i.e., toward the outside of the turn). In contrast, the inside arm had greater positive contribution in the frontal plane during initiation ($p < 0.001$) and continuation ($p = 0.001$), while the outside arm had greater positive contribution during the initiation ($p < 0.001$) and continuation ($p < 0.001$) strides in the transverse plane. All other segment group comparisons can be found in Supplemental Table 2.

4. Discussion

Similar to tasks such as walking up slopes [22], there was greater $\vec{H}_{\mathrm{WB}}$ generation during turning compared to straight-line walking in all planes. In the sagittal plane, average $\vec{H}_{\mathrm{WB}}$ was negative (rotating anteriorly) during the first two strides and positive (rotating posteriorly) during the last stride. The outside leg contributed more positively than the inside leg, while the inside leg contributed more negatively than the outside leg to $\vec{H}_{\mathrm{WB}}$ throughout the turn. The adjustment of the

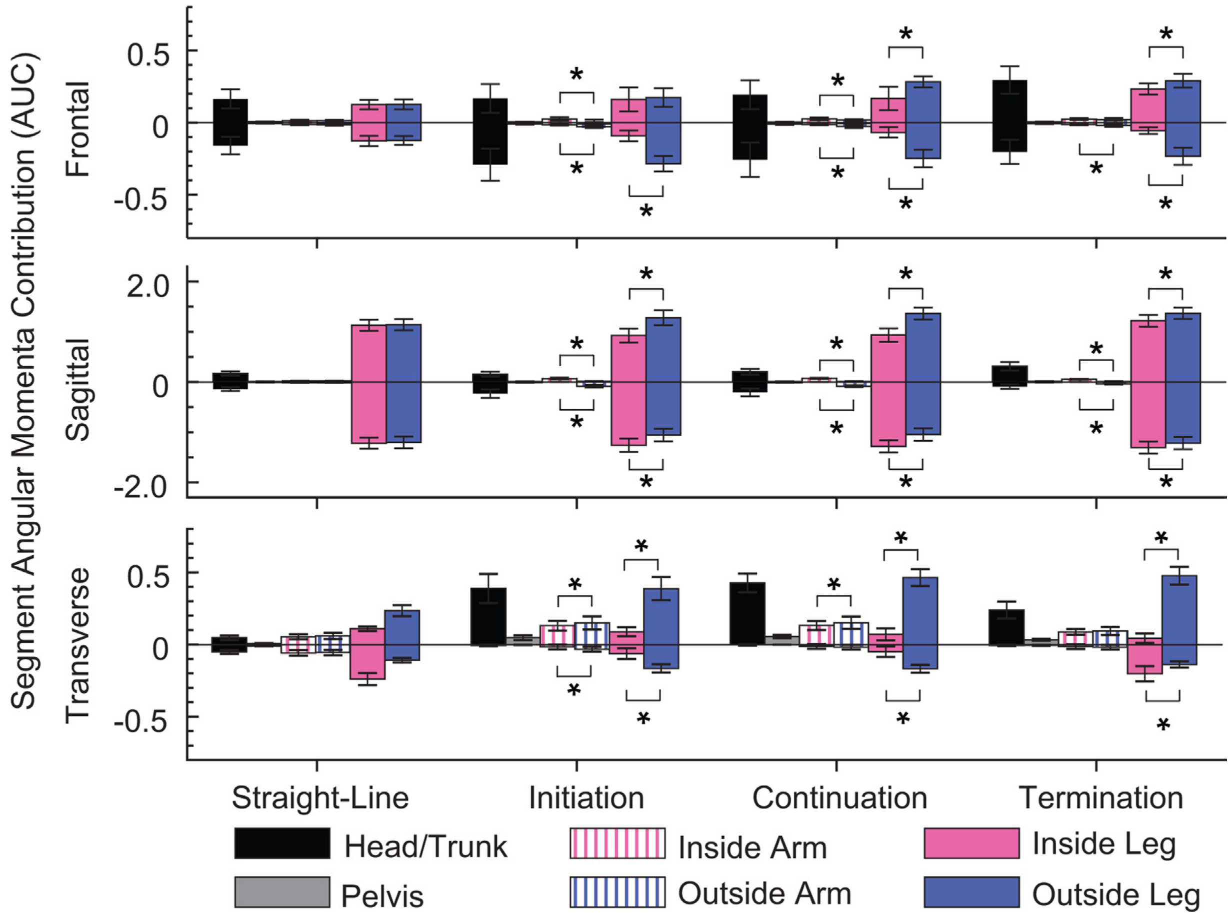

Fig. 5. Segment contribution to whole-body angular momentum, $\vec{H}_{\text{WB}}\left(\frac{\text{kg}\cdot\text{m}^2}{\text{s}\cdot(\text{kg}\cdot\text{m}\cdot\frac{\text{m}}{\text{s}})}\right)$, during turning (initiation, continuation, termination) and straight-line walking in all three planes of motion. Positive bars are the area under the curve (AUC) for the positive portion of the segment body angular momentum only. Similarly, negative bars are the negative AUC. Significant differences ($p < 0.05$) between limbs inside and outside the turn are denoted with asterisks (*). Head/trunk and pelvis segments are shown for reference. Comparisons between all segments can be found in Supplemental Table 2.

outside leg contribution is likely related to a reduced braking impulse and increased propulsive impulse of the outside leg compared to the inside leg, as previously reported during circular turning [28]. Similar to straight-line walking, the inside leg contribution counteracted the outside leg contribution. Similar cancellation is seen in tripping, where the support limb can reduce $\vec{H}_{\text{WB}}$ to provide time and clearance for the contralateral limb to recover [29]. Due to limb cancellation, the progression from negative to positive sagittal $\vec{H}_{\text{WB}}$ during the turn is more likely due to the head/trunk contributions (Fig. 5).

The average $\vec{H}_{\text{WB}}$ magnitude in the frontal plane was greater than straight-line walking during the initiation and termination strides. In particular, the greater angular momentum generated during the turning strides suggest a difference in momenta cancellation between body segments. During circular turning, the body center of mass (COM) is closer to the inside edge of the base support [17], which is likely also the case during continuation of a 90-degree turn. As people complete a 90-degree turn, they generate $\vec{H}_{\text{WB}}$ away from the turn to direct the COM toward the outside of the turn, within the base of support. This behavior was evident by the shift in $\vec{H}_{\text{WB}}$ from negative (towards the turn) to positive (away from the turn), similar to results in the sagittal plane. This shift was primarily driven by the head/trunk and leg groups where contributions from the head/trunk and outside leg changed from negative to positive as the turn progressed (Fig. 5). Generally, the outside leg had greater overall contributions than the inside leg in the frontal plane, while the inside arm had greater positive contribution than the outside arm and vice-versa for negative contributions. However, arm contributions were relatively small. These results are likely related to a greater medial impulse during the stance phase used to accelerate the body COM towards the contralateral limb [17,19,28]. This shift in generation of angular momentum could be another method of angular momentum “cancelling”. Instead of segments cancelling each other within one stride, as was done during straight-line walking [21], negative momentum generated to initiate the turn was counteracted later to regain an upright posture in the new direction of progression.

Transverse plane $\vec{H}_{\text{WB}}$ magnitude was greater than straight-line walking for all turning strides. There was a positive (toward the turn) angular momentum during all strides for all segments except the inside leg, which generated negative angular momentum during the termination stride (Fig. 4). In addition, the outside limbs had greater overall contributions than the inside limbs during turning. However, unlike the legs, both arms contributed similarly during the termination stride in the transverse plane. While the outside leg had greater positive contribution throughout the turn, the inside leg generally contributed in the negative direction during the termination stride. This helped to generate $\vec{H}_{\text{WB}}$ away from the turn to maintain the new direction of progression and dynamic balance, similar to the frontal plane.

The difference between turning and straight-line walking may stem from changes in walking speed, as people slowed down during the initiation stride when turning right, and during the continuation stride when turning left. Generally, faster speeds lead to greater $\vec{H}_{\text{WB}}$ [21] but normalizing by speed results in greater $\vec{H}_{\text{WB}}$ for slower speeds. To address this issue, we reanalyzed the data without normalizing to walking speed. Normalization only affected whole-body angular momentum during the initiation stride in the sagittal plane. This result was not unexpected as external moment arms will change with walking speed (due to foot placement relative to the body center of mass), affecting the sagittal plane angular momentum. However, turning primarily affects fall risk in the frontal plane. Accordingly, we have chosen to keep the normalization of angular momentum with respect to walking speed, as this approach allows us to compare our results more directly to prior literature in the field. To aid in interpretation and future comparisons, we have included the non-normalized results in Supplemental Material.

During turning, $\vec{H}_{WB}$ is both regulated in the sagittal plane (Figs. 4 and 5) and generated in the frontal and transverse planes (Fig. 4, 5). This result differs from how $\vec{H}_{WB}$ is cancelled during straight-line walking [20,21] to regulate balance. Segment contributions during turning depend on the segment's side of the body, which likely contribute to both balance and maneuverability [29] through interaction with the environment or kinematics driven by force generation [30]. In particular, muscle modulation is important for making the adjustments required for turning. For example, the inside leg gluteus medius and outside leg soleus and gastrocnemius muscles' contributions to mediolateral COM acceleration decrease during circular turning [31] and during simulated 90-degree turns in children [32] compared to straight-line walking while timing and amplitude of muscle activity are modified during curve walking [12].

This work has several limitations. First, the results of this study are only applicable to 90-degree step turns. There are numerous other types of turns (e.g., spin-turns) which can be performed with a range of turning angles. However, step-turns are more common during activities of daily living [1], and 90-degrees falls within the range of common turns taken on a daily basis [33]. Second, we only tested healthy adults and did not challenge their balance beyond having them perform a typical turn. Participants may develop different strategies under varying circumstances. Third, angular momentum is just one measure of dynamic balance and may relate to other measures such as margin of stability or the ratio of shear and normal ground reaction force. However, angular momentum incorporates both balance and maneuverability, identifying requirements for both performing the task and maintaining balance, which may not be mutually exclusive. Also of note, we quantified the average $\vec{H}_{WB}$ rather than the range of $\vec{H}_{WB}$ as done previously [20,21,34]. While this limits our ability to directly compare findings with prior work, we did observe similar minimum and maximum $\vec{H}_{WB}$ during straight-line walking [20,21,34].

This study quantified angular momentum of healthy adults during 90-degree turns. These findings provide a baseline for future comparison with populations that may have difficulty regulating angular momentum. In particular, people with neurological disorders or musculoskeletal deficits may have difficulty maintaining dynamic balance. The inability to generate or restrict the appropriate angular momentum is a risk factor that could lead to loss of balance and injury. Knowledge of how segments work to maintain $\vec{H}_{WB}$ successfully can help design rehabilitation programs, fall prevention programs, or devices for impaired populations that may help restore a person's ability to generate angular momentum required for a 90-degree turn.

Conflicts of interest

None declared.

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Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at https://doi.org/10.1016/j.gaitpost.2019.02.003.

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