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Synergies in the ground reaction forces and moments during double support in curb negotiation in young and older adults

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ABSTRACT

Falls that occur while negotiating steps are a leading cause of death in older adults. Although recent efforts have improved understanding of the mechanics and control of stepping behaviors, the double support phase during stepping is understudied. Therefore, we quantified the stability of the resultant forces and moments acting on the body during this phase. These quantities determine the movement of the body, and therefore, their stability is essential for successful stepping behavior. We measured the ground reaction variables (GRVs) under both feet as healthy young (n = 10) and older adults (n = 10) stepped up and down a curb. We employed the uncontrolled manifold method to evaluate the hypotheses that the GRVs covary to stabilize the resultant force and moment in the three coordinate directions. Robust stabilization of the resultant forces and moments was observed while stepping up. However, while stepping down, the stability of the resultant moment was prioritized over that of the resultant forces in the vertical and the anterior-posterior directions, and the stability of the resultant medio-lateral force was prioritized over that of the resultant anterior-posterior force. The salience of stabilizing whole-body angular momentum and medio-lateral motion during locomotion is well known, but their prioritization during adaptive gait is a novel result and is possibly related to the higher likelihood of falling during descent (versus ascent). Finally, contrary to our expectations, we observed no age differences in our stability indices, indicating that healthy aging does not diminish the stability of the resultant forces and moments.

1. Introduction

Negotiating steps is more challenging than level ground walking due to greater required muscle effort, higher joint moments, and it is associated with more variable movement kinematics (Jacobs, 2016). This challenge is further established by the fact that stairway falls are a leading cause of accidental death among older adults (Jacobs, 2016; Startzell et al., 2000). Although the mechanics and control of step negotiation are well explored, the double support phase, with the feet planted on surfaces at different heights, is understudied, consistent with a relative neglect of this phase in the locomotion literature (Vlutters et al., 2016). The importance of double support is established by two key observations. First, robust responses to perturbations occur during double support, indicating active control during this phase (Bent et al., 2004; Reimann et al., 2018; Vlutters et al., 2016; Vlutters et al., 2018). Second, incorrect weight shifting – which occurs during double support – is the most common cause of falls in older adults living in long-term care facilities (Robinovitch et al., 2013; Yang et al., 2018). Understanding how the forces and moments under both limbs are coordinated during double support will reveal strategies that individuals employ to stabilize body movements. Therefore, the goals of this study are to quantify the stability of mechanical variables during the double support phase of curb negotiation, and to establish whether the stability diminishes with age.

We employed the uncontrolled manifold (UCM; abbreviations listed in Table 1) analysis (Scholz and Schoner, 1999) to quantify the stability of crucial performance variables (PVs) during double support. UCM analysis is appropriate for abundant systems where numerous input variables coordinate to stabilize a fewer number of PVs. The stability of the PVs is determined by comparing across-trial variance in the input variables in directions that maintain PVs (spanning the UCM for those PVs, V_{UCM}) to variance in directions that change the PVs (orthogonal to the UCM, V_{ORT}). If V_{UCM} is greater than V_{ORT}, then there is a synergy between the

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Table 1
Abbreviations.

Uncontrolled manifold	UCM
Orthogonal manifold	ORT
Performance variable	PV
Variance within the UCM	V_{UCM}
Variance within the orthogonal manifold	V_{ORT}
Synergy index	ΔV
Synergy index z transformed	ΔV_Z
Center of mass	CoM
Center of pressure	CoP
Resultant reactive force	$F_{Resultant}$
Resultant reactive moment computed about CoM	$M_{Resultant}$
Ground reaction variables	GRV
Free moment	FM
Ground reaction force	GRF
Weight	W
Height	H

inputs that stabilizes the PVs, and the relative excess of V_{UCM} quantifies the synergy strength (see Appendix for details).

We chose the resultant reactive force ($F_{Resultant}$) and moment ($M_{Resultant}$) acting at the center of mass (CoM) as PVs. These quantities reflect the inertial effects arising from the current motion of the body segments and the current forces exerted by the muscles. The quantities also determine the future motion via the equations of motion. Therefore, stable body movements require consistent $F_{Resultant}$ and $M_{Resultant}$ trajectories, and $F_{Resultant}$ and $M_{Resultant}$ are reasonable choices for the PVs.

These PVs arise from the forces and moments acting under both feet (together called the ground reaction variables (GRVs)). Control is reflected in the GRVs, since muscle forces influence the GRVs. There are more GRVs than PVs (see Table 2). Therefore, the system is abundant, and the UCM method can be employed. We hypothesized that the PVs will be stabilized via synergistic covariation between GRVs during the double support phase of curb negotiation, as has been observed for jumping and treadmill walking (Slomka et al., 2015; Toney and Chang, 2013).

Synergy strength changes when task demands (e.g., performance accuracy, maneuverability) change (Rosenblatt et al., 2015; Tillman and Ambike, 2018a,b). This implies that synergies will be different for stepping up versus stepping down. Stepping down is more destabilizing than stepping up, as more falls occur during stepping down (Tinetti et al., 1995). Therefore, we hypothesized that synergies will be weaker (lower stability) while stepping down compared to stepping up.

Aging-related neuromuscular changes lead to lower muscle forces, lower rates of muscle force and joint torque development (Enoka, 2015), higher variability in the generated forces (Christou, 2011), and weaker synergies in manual force production tasks (Park et al., 2011; Singh et al., 2013). Similar to manual forces, these neuromuscular changes may weaken kinetic synergies – where the inputs are kinetic quantities – during double support, and we hypothesize that older adults will have weaker kinetic synergies compared to young adults.

Table 2
The performance variables, the input variables and the synergy index value that indicates presence of a synergy or an anti-synergy or a lack of synergistic covariation for the six UCM analyses.

Performance variable (PV)	Input variables	Discriminating value for ΔV_Z
Resultant force along AP ($F_{Resultant-X}$)	F_{RX}, F_{LX}	0
Resultant force along ML ($F_{Resultant-Z}$)	F_{RZ}, F_{LZ}	0
Resultant force along vertical ($F_{Resultant-Y}$)	F_{RY}, F_{LY}	0
Resultant moment about AP axis ($M_{Resultant-X}$)	F_{RY}, F_{LY}, F_{RZ}, F_{LZ}	0.55
Resultant moment about ML axis ($M_{Resultant-Z}$)	F_{RX}, F_{LX}, F_{RY}, F_{LY}	0.55
Resultant moment about vertical axis ($M_{Resultant-Y}$)	F_{RX}, F_{LX}, F_{RZ}, F_{LZ}, FM_R, FM_L	0.80

2. Methods

2.1. Participants

Ten young (22.9 ± 6.7 years, 3 male, 66.8 ± 10.7 kg, height 1.68 ± 0.07 m) and ten older adults (73 ± 5 years, 2 male, 67.7 ± 15.1 kg, height 1.62 ± 0.09 m) participated in the study. We screened participants for visual acuity (Snellen chart), dementia (Mini Mental Status Exam > 25), history of orthopedic problems and presence of neurodegenerative disease (self-report). Participants provided written informed consent approved by the Institutional Review Board of Purdue University.

2.2. Equipment and procedures

An 8 m walkway included an elevated curb (4 m long, 1 m wide, 15 cm high) and two embedded force plates (AccuGait, AMTI, MA, USA), one in the floor, and one in the curb (Fig. 1A). Participants walked at their comfortable speed, and stepped up and down the curb in alternating sequence. The starting positions were adjusted to ensure foot contact with the force plate and that the right foot crossed the curb edge first. Fifteen trials of stepping up and down were collected in 10–20 min. Fifteen trials is consistent with recommendations for UCM analyses (Freitas et al., 2019; Rosenblatt and Hurt, 2019). Reflective marker clusters were placed on the head, upper back, lower back, and bilaterally on the upper arms, forearms, thighs, shanks and feet. Joint centers were digitized to identify their location relative to the marker clusters. The kinematics were collected using Vicon (Oxford, UK) at 100 Hz, the GRVs were collected at 1000 Hz, and the kinematics and kinetics were synchronized with MotionMonitor software (Innovative Sports Training Inc., IL, USA).

2.3. Data analysis

The whole-body CoM was computed using the anthropometrics of thirteen body segments (de Leva, 1996). The instantaneous CoM position, CoM velocity, GRVs, center of pressure (CoP) and twisting moment about the vertical axis (called 'free moment') under each foot were obtained from the MotionMonitor software. The data were filtered using a zero-lag, 4th-order, low-pass Butterworth filter. The CoM was filtered at 8 Hz, and the GRVs were filtered at 20 Hz. Double support phase was identified as the time interval between lead heel contact (vertical ground reaction force (GRF) first crossed 15 N) and trail toe off (vertical GRF first reduced below 15 N).

Traditional gait measures, namely, the time spent in double support, step length, step width, average CoM velocity, and the maxima of the sums of the absolute forces under both feet in each direction, were calculated to describe curb negotiation during double support.

2.4. Uncontrolled manifold analysis

The PVs, $F_{Resultant}$ and $M_{Resultant}$, are 3-dimensional vectors. Each component of the PVs has its own set of input GRVs. Therefore, we quantified the synergy strength for each vector component using separate analyses. Table 2 lists the PVs and the input variables

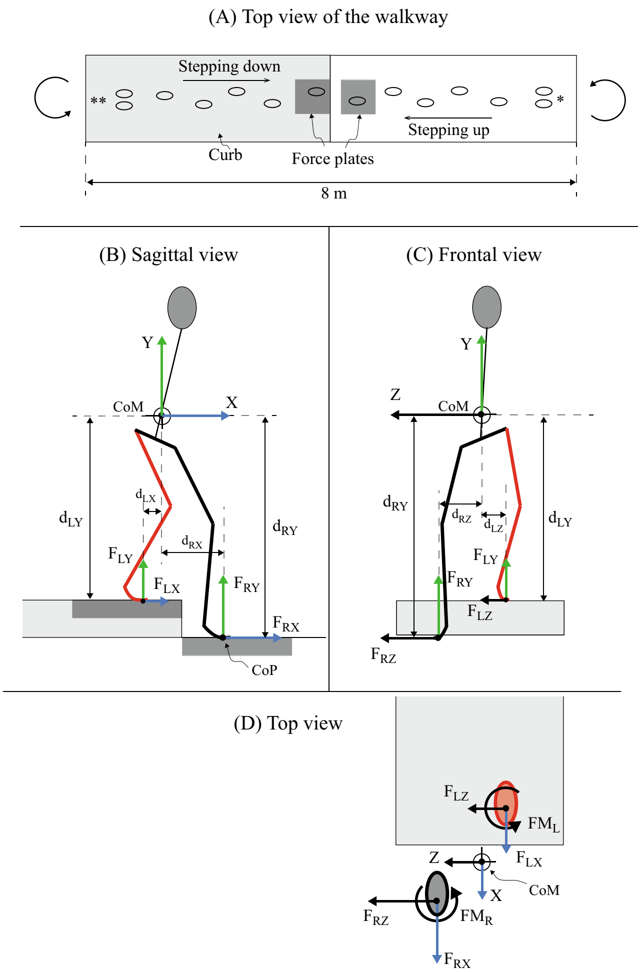

Fig. 1. (A) Illustration of the gait task and force plates embedded in the floor and the curb. Participants started on the floor at the position marked as '*', about 4 m from the curb. Participants walked and stepped up onto the curb, walked the remaining 4 m, turned around, and stopped at the position marked '**'. Then, they walked back, stepped down from the curb, and walked to the position '*'. (B-D) Illustration of the input variables. The coordinate frame located at the center of mass (CoM) is parallel to the lab-fixed coordinate frame (not shown here). It travels with the CoM, but does not rotate with the participant's motion. All forces, free moments and distances are measured in the lab-fixed frame.

for each UCM analysis, and the Appendix provides details of each analysis. Briefly, for each analysis, the GRV time series from the 15 trials were time normalized and then normalized by the participant's weight (and height for the free moments). At each time instant (t^*), the deviation in the normalized GRVs for each trial from the corresponding across-trial mean value were computed, and this difference was projected onto the UCM and the orthogonal (ORT) manifolds. These manifolds are the null space and its orthogonal compliment, respectively, of the appropriate Jacobian that relates small changes in the GRVs to small changes in the PVs. The variance in the projected data are the variance components V_{UCM} (t^*) and $V_{ORT}(t^*)$. They yield the synergy index and its z-transformed value ($\Delta Vz(t^*)$; Eq. (2)). This computation was repeated for each time point in the double-support phase, which yields the time series $V_{UCM}(t)$, V_{ORT} (t) and $\Delta V_Z(t)$. These measures were averaged over the double-support phase.

UCM analysis can identify a synergy, an anti-synergy, or a non-synergy based on a discriminating value ΔVz^* (specific to each analysis, see Table 2). $\Delta Vz > \Delta Vz^*$ indicates a synergy. The GRVs covary such that the PV is stabilized, and the across-trial variance in the GRVs aligns with the UCM. $\Delta Vz < \Delta Vz^*$ indicates an anti-synergy. The GRVs covary to change rather than stabilize the PV (Wang et al., 2006), and the across-trial variance aligns with the orthogonal manifold. $\Delta Vz = \Delta Vz^*$ indicates a non-synergy; the across-trial variance does not align with any manifold.

2.5. Statistics

To test our hypotheses regarding presence of synergies, two-tailed t-tests were used to determine if the ΔVz value for each UCM analysis and each age group was different from the corresponding discriminating value (Table 2). Traditional gait variables (duration, step length and width, CoM velocities, and GRF magnitudes) and the UCM variables (V_{UCM}, V_{ORT} and ΔVz) were examined with a two-way, mixed ANOVA with stepping direction as a repeated measures factor and age as a between-subjects factor. The variance components were normalized by the dimensions of the corresponding manifolds, and they were log transformed to meet normality requirements (non-transformed values are reported in Results).

3. Results

No interaction effects were observed in any ANOVA. Only significant main effects are reported.

3.1. Gait characteristics

Double support duration was shorter for stepping down ($F_{1,18} = 195.3$, $p < 0.01$, $\eta_p^2 = 0.92$) (Fig. 2A). Step length was shorter while stepping down ($F_{1,18} = 30.7$, $p < 0.01$, $\eta_p^2 = 0.63$) and for the older adults ($F_{1,18} = 5.96$, $p = 0.02$, $\eta_p^2 = 0.25$; Fig. 2B). There were no main or interaction effects of stepping direction or age on step width (Fig. 2C).

The average AP CoM velocity was higher while stepping down ($F_{1,18} = 67.83$, $p < 0.01$, $\eta_p^2 = 0.79$) and lower for older adults ($F_{1,18} = 4.73$, $p = 0.04$, $\eta_p^2 = 0.21$; Fig. 2D). The average ML and vertical CoM velocities were higher while stepping down ($F_{1,18} \geq 4.67$, $p \leq 0.04$, $\eta_p^2 \geq 0.21$; Fig. 2E and F).

The peak values of the weight-normalized absolute forces summed across the feet were lower for stepping down in every direction ($F_{1,18} \geq 14.66$, $p < 0.01$, $\eta_p^2 \geq 0.50$) (Fig. 2G, H, and I). The peak forces were lower in older adults in the AP direction only ($F_{1,18} = 8.64$; $p < 0.01$; $\eta_p^2 = 0.32$).

3.2. UCM analysis for resultant forces

The time series of V_{UCM}, V_{ORT} and ΔVz are depicted in Fig. 3. For the resultant force analyses, zero is the discriminating value that indicates the presence of either a synergy or an anti-synergy (Table 2). Synergies appear to be present in every direction during stepping up (Fig. 3A, E, G) and ML direction during stepping down (Fig. 3D), whereas an anti-synergy appears to be present in the vertical force during step down (Fig. 3G). These qualitative descriptions were assessed with average values of V_{UCM}, V_{ORT} and ΔVz (Fig. 4).

For both age groups, the resultant AP force was stabilized by a synergy while stepping up, but ΔVz was not different from zero while stepping down (Table 3). Therefore, we conclude that there is no task-specific covariation in the AP forces while stepping down. The resultant ML force is stabilized by a synergy for both age groups and both stepping directions. Finally, for both age groups, the vertical resultant force was stabilized by a synergy while stepping up, whereas ΔVz was not different from zero while stepping down (despite the qualitative observation of Fig. 3G).

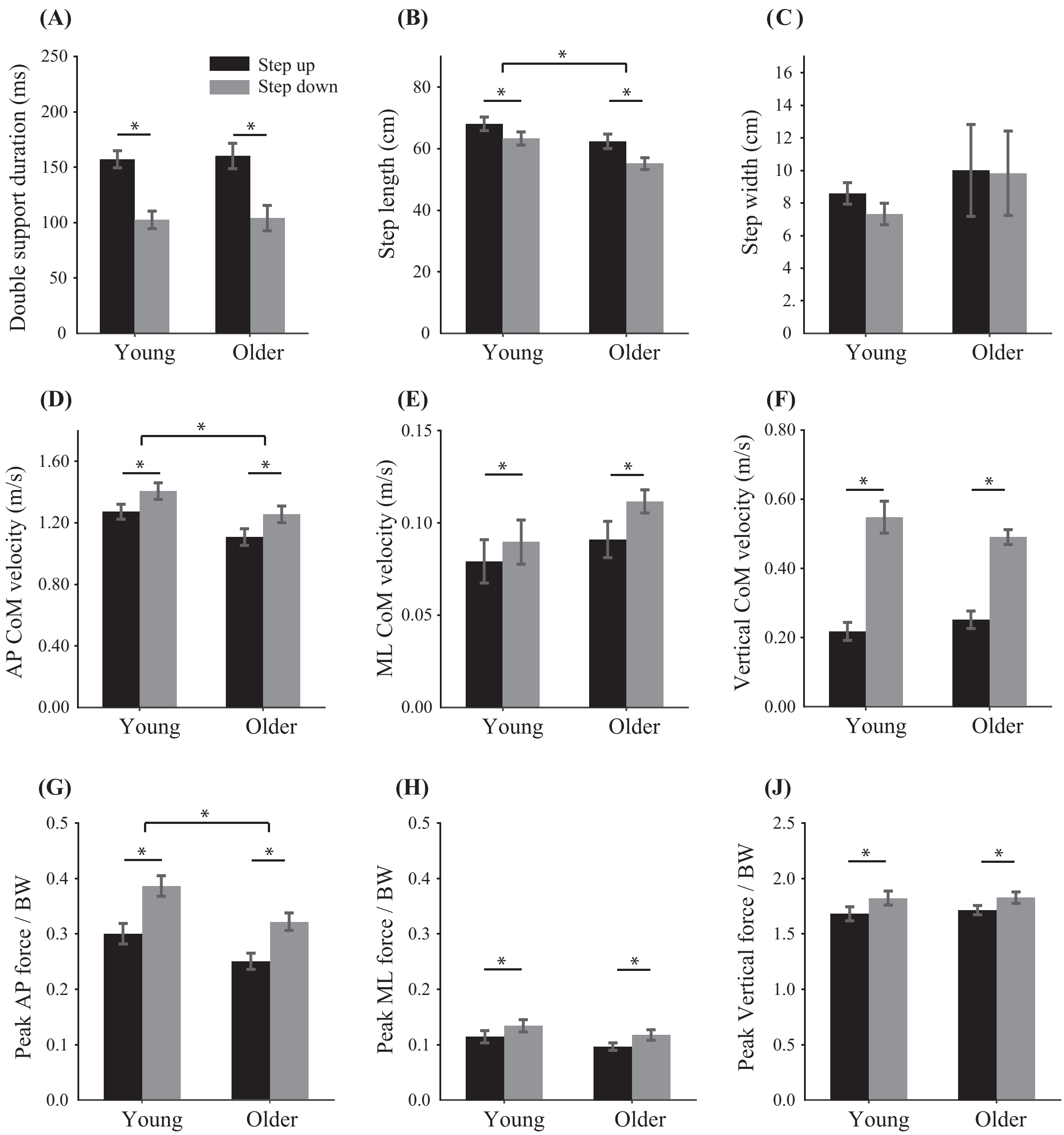

Fig. 2. Mean ± SE of gait parameters. Only main effects were observed. '*' indicates significant effect.

Therefore, we conclude that there is no task-specific covariation in the vertical forces while stepping down.

ΔVz for the stabilization of the AP resultant force was smaller for stepping down ($F_{1,18} = 6.57$, $p = 0.02$, $\eta_p^2 = 0.27$; Fig. 4A). V_{UCM} was larger while stepping down ($F_{1,18} = 5.83$, $p = 0.03$, $\eta_p^2 = 0.24$; Fig. 4B). V_{ORT} was larger for stepping down ($F_{1,18} = 17.61$, $p < 0.01$, $\eta_p^2 = 0.49$) and for young adults ($F_{1,18} = 7.80$, $p = 0.01$, $\eta_p^2 = 0.02$; Fig. 4C).

ΔVz for the stabilization of the ML resultant force was not affected by stepping direction or age (Fig. 4D). Both V_{UCM} and V_{ORT} were higher for stepping down ($F_{1,18} \geq 19.22$, $p < 0.01$, $\eta_p^2 \geq 0.51$; Fig. 4E and F).

ΔVz for the stabilization of the vertical resultant force was smaller for stepping down ($F_{1,18} = 16.70$, $p < 0.01$, $\eta_p^2 = 0.48$; Fig. 4G). Both variance components were larger for stepping down ($F_{1,18} \geq 15.55$, $p < 0.01$, $\eta_p^2 \geq 0.46$; Fig. 4H, I).

3.3. UCM analysis for resultant moments

The time series of V_{UCM}, V_{ORT} and ΔVz are depicted in Fig. 5. In contrast to the force-stabilization analyses (Fig. 3), V_{UCM} is larger than V_{ORT} by an order of magnitude. Therefore, robust synergies are observed throughout the double support phase stabilizing the resultant moment about every axis and in both age groups (Fig. 5A, D, and G). These qualitative descriptions were assessed with average values of V_{UCM}, V_{ORT} and ΔVz (Fig. 6).

Consistent with the qualitative observations above, the resultant moments about every axis are stabilized by synergistic covariation in the corresponding GRVs for both age groups and stepping directions (Table 3).

ΔVz for the stabilization of the AP resultant moment was not affected by stepping direction or age (Fig. 6A). V_{UCM} and V_{ORT} were larger for stepping down ($F_{1,18} \geq 44.87$, $p < 0.01$, $\eta_p^2 \geq 0.71$; Fig. 6B, C).

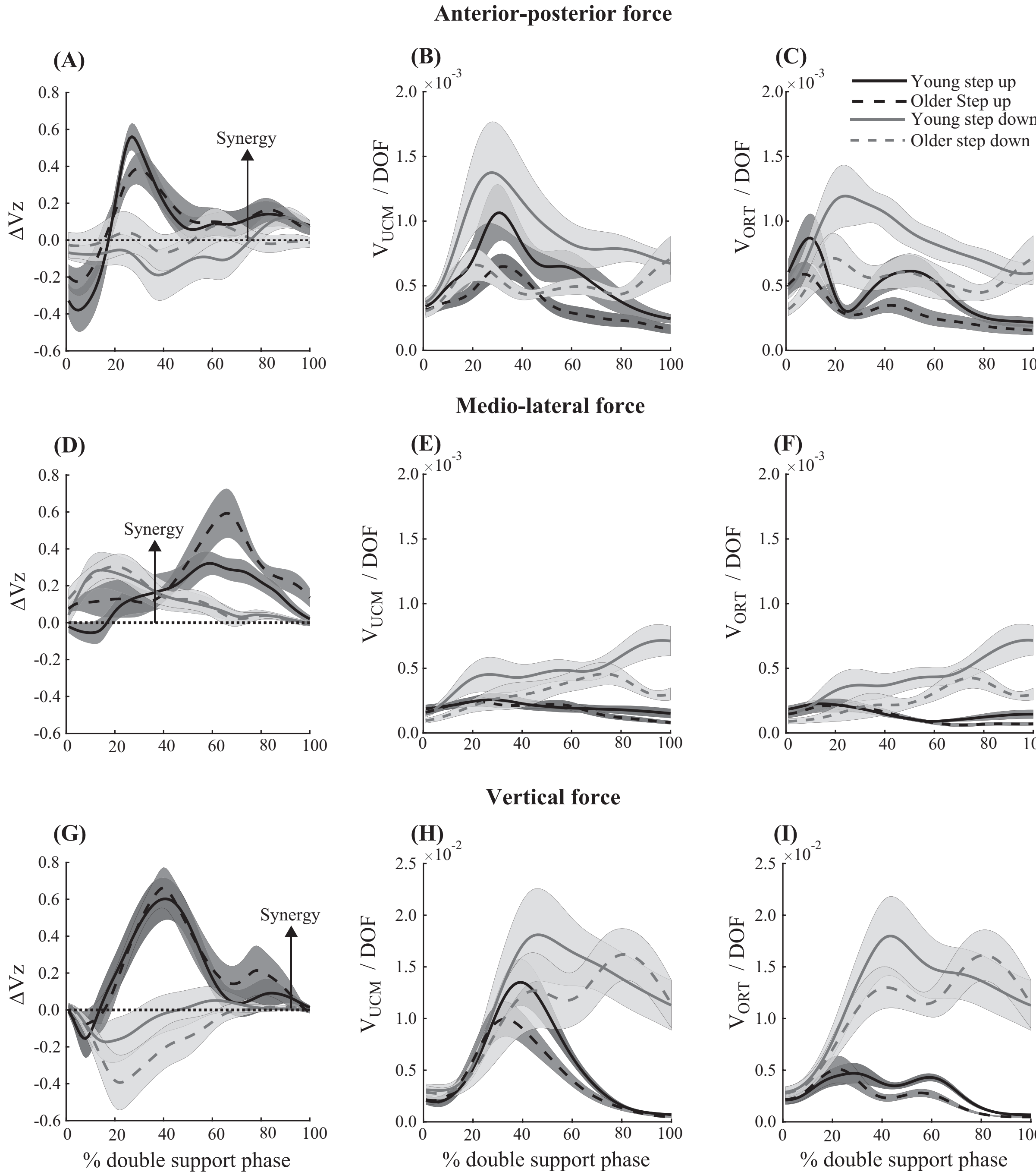

Fig. 3. Mean ± SE of the synergy index (ΔVz), and the variance components (V_{UCM} and V_{ORT}) obtained from the uncontrolled manifold analysis are plotted against normalized time during double support phase. Each row of figures depicts the data from the analysis of resultant force in the AP (Panels A-C), ML (Panels D-F), and vertical (Panels G-I) directions as the performance variable.

ΔVz for the stabilization of the ML resultant moment was not affected by stepping direction or age (Fig. 6D). V_{UCM} and V_{ORT} were larger for stepping down ($F_{1,18} \geq 5.03$, $p \leq 0.04$, $\eta_p^2 \geq 0.22$; Fig. 6E, F).

ΔVz for the stabilization of the vertical resultant moment was not affected by stepping direction or age (Fig. 6G). V_{UCM} was larger for stepping down ($F_{1,18} = 32.34$, $p < 0.01$, $\eta_p^2 = 0.64$; Fig. 6H, F). No main or interaction effects were observed for V_{ORT} (Fig. 6I).

4. Discussion

The goals of this study were to quantify the stability of the resultant ground reaction forces and moments during the double support phase of curb negotiation, and to establish whether the stability diminishes with age. Synergies stabilized the resultant force and the resultant moment about the CoM for stepping up, supporting our first hypothesis. However, synergies stabilizing the vertical and AP resultant forces vanish while stepping down. The vanishing synergies support our second hypothesis that synergies would be weaker while stepping down. Finally, our third hypothesis that synergies will be weaker in older adults was not supported.

4.1. Synergies across stepping direction

Our results are consistent with work stressing the importance of tightly regulating angular momentum in various locomotor

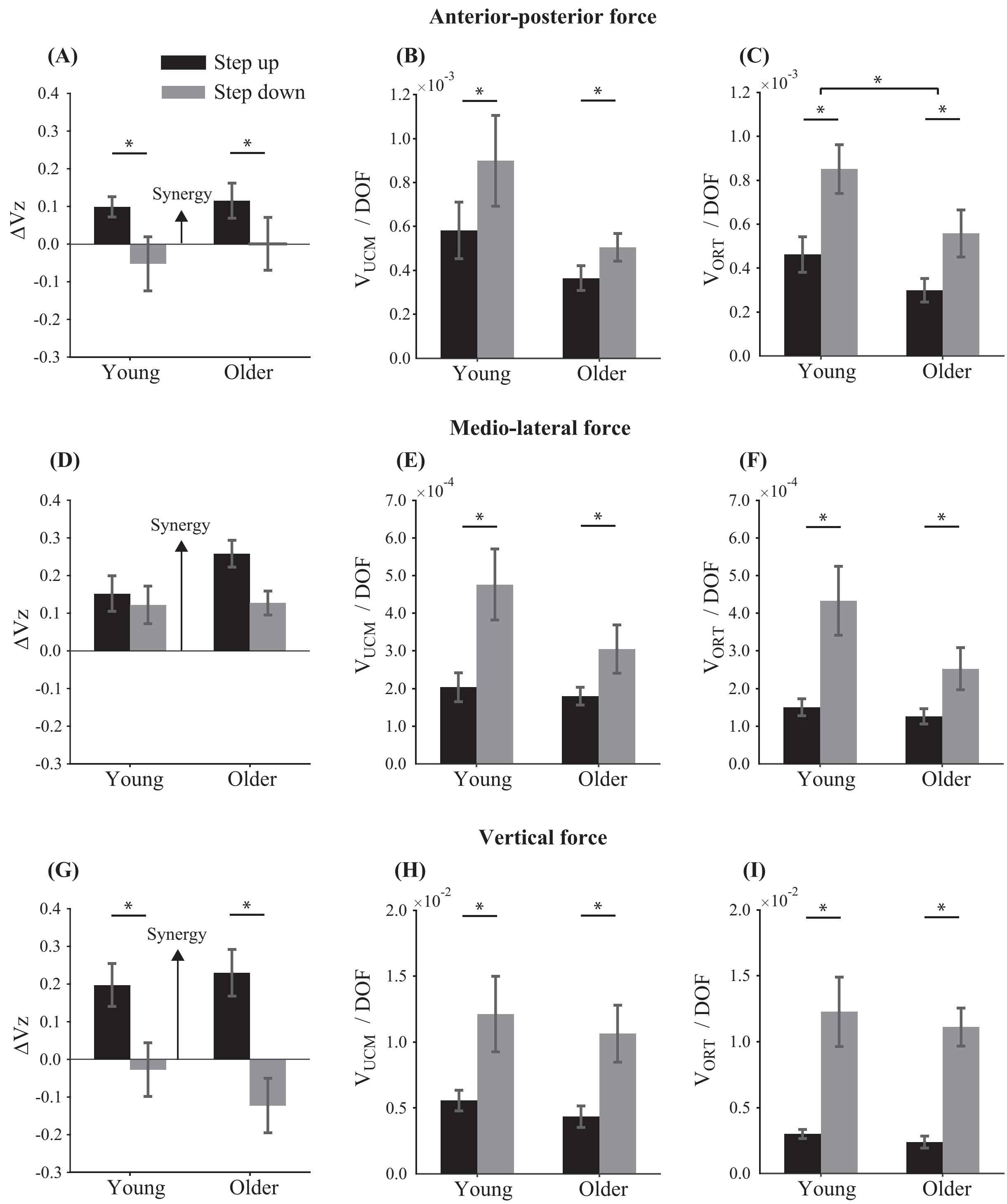

Fig. 4. Across-participant mean ± SE of the synergy index (ΔVz), and the variance components (V_{UCM} and V_{ORT}) averaged over the double-support phase. Each row of figures depicts the data from the analysis of resultant force in the AP (Panels A-C), ML (Panels D-F), and vertical (Panels G-I) directions as the performance variable. Only main effects were observed. '*' indicates significant effect.

tasks (Begue et al., 2019; Herr and Popovic, 2008; Robert et al., 2008), including negotiation of slopes (Silverman et al., 2012), stairs (Silverman et al., 2014) and curbs (van Dieen et al., 2007). The resultant moments were stabilized across stepping directions (Table 3), and since the resultant moment equals the rate of change of whole-body angular momentum, the synergies stabilize the rate of change of angular momentum. This result, in conjunction with an assumption of consistent angular momentum at the initiation of double support (as suggested by the literature cited above), suggests the stabilization of the whole-body angular momentum during double support.

A key finding of this study is that the resultant moment stabilization is prioritized over the stabilization of the resultant AP and vertical forces while stepping down. Trading a force synergy for a moment synergy is possible because each force pair contributes to multiple tasks, and the tasks impose conflicting demands on the force pairs. The role of each force pair in the stabilization of various PVs is listed in Table 4. Each force pair

Table 3
Results of the two-tailed t tests to identify whether the synergy index is significantly different from the corresponding discriminating value (see Table 2).

Task	Step down-Young		Step down-Older		Step up-Young		Step up-Older	
	$t_{(9)}$	p	$t_{(9)}$	p	$t_{(9)}$	p	$t_{(9)}$	p
$F_{Resultant-X}$ (AP)	−0.726	0.487	0.010	0.992	3.695	**0.005**	2.467	**0.036**
$F_{Resultant-Z}$ (ML)	2.435	**0.038**	4.005	**<0.001**	3.242	**0.01**	7.186	**<0.001**
$F_{Resultant-Y}$ (Vertical)	−0.380	0.713	−1.695	0.124	3.468	**0.007**	3.703	**<0.001**
$M_{Resultant-X}$ (AP)	12.825	**<0.001**	13.091	**<0.001**	23.231	**<0.001**	19.796	**<0.001**
$M_{Resultant-Z}$ (ML)	9.901	**<0.001**	11.62	**<0.001**	28.113	**<0.001**	24.358	**<0.001**
$M_{Resultant-Y}$ (Vertical)	16.962	**<0.001**	11.117	**<0.001**	16.596	**<0.001**	15.67	**<0.001**

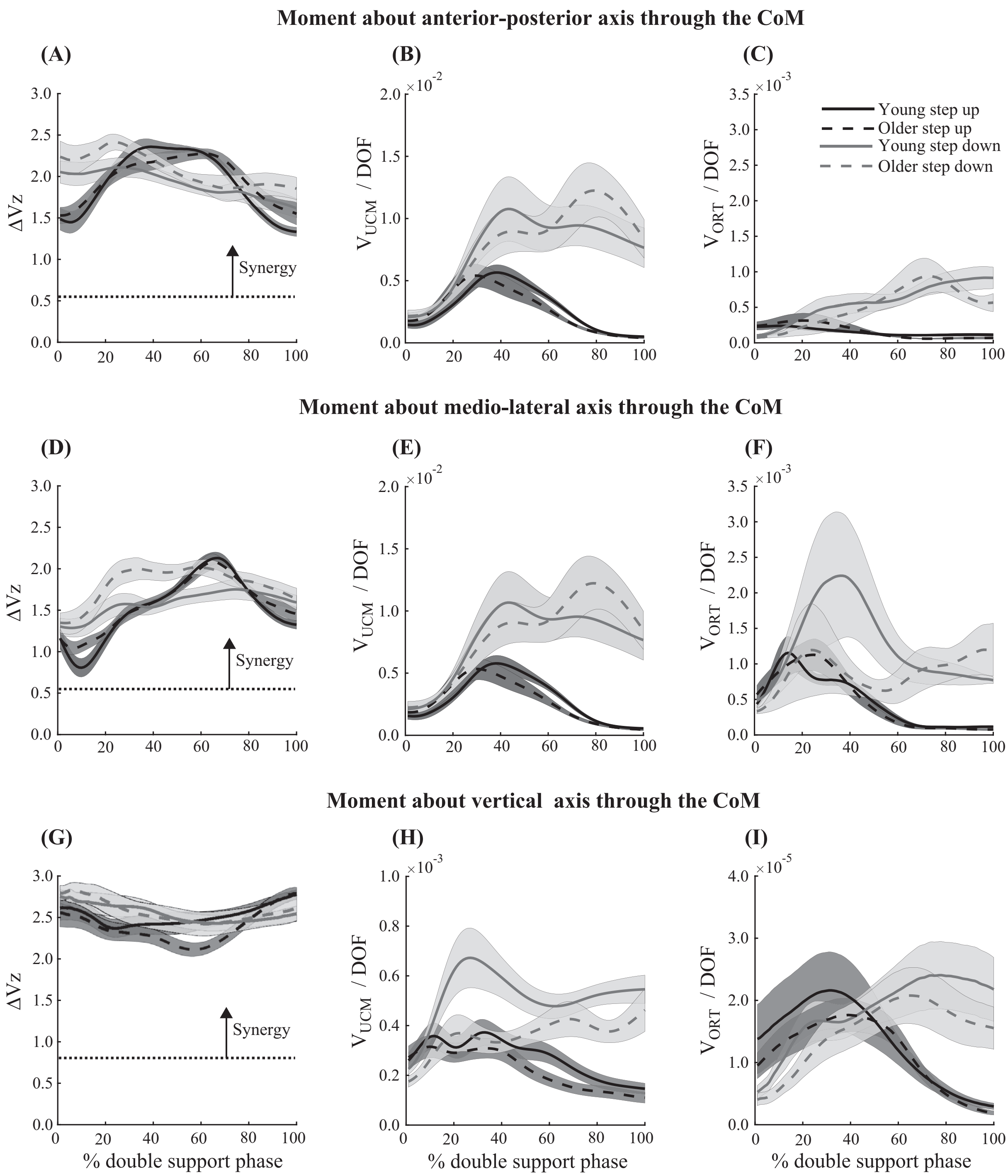

Fig. 5. Mean ± SE of the synergy index (ΔVz), and the variance components (V_{UCM} and V_{ORT}) obtained from the uncontrolled manifold analysis are plotted against normalized time. Each row of figures depicts the data from the analysis of resultant moment about the AP (Panels A-C), ML (Panels D-F), and vertical (Panels G-I) axis as the performance variable. The dashed line in Panels A, D, and G indicate the discriminating value of the synergy index that determines the presence of a synergy.

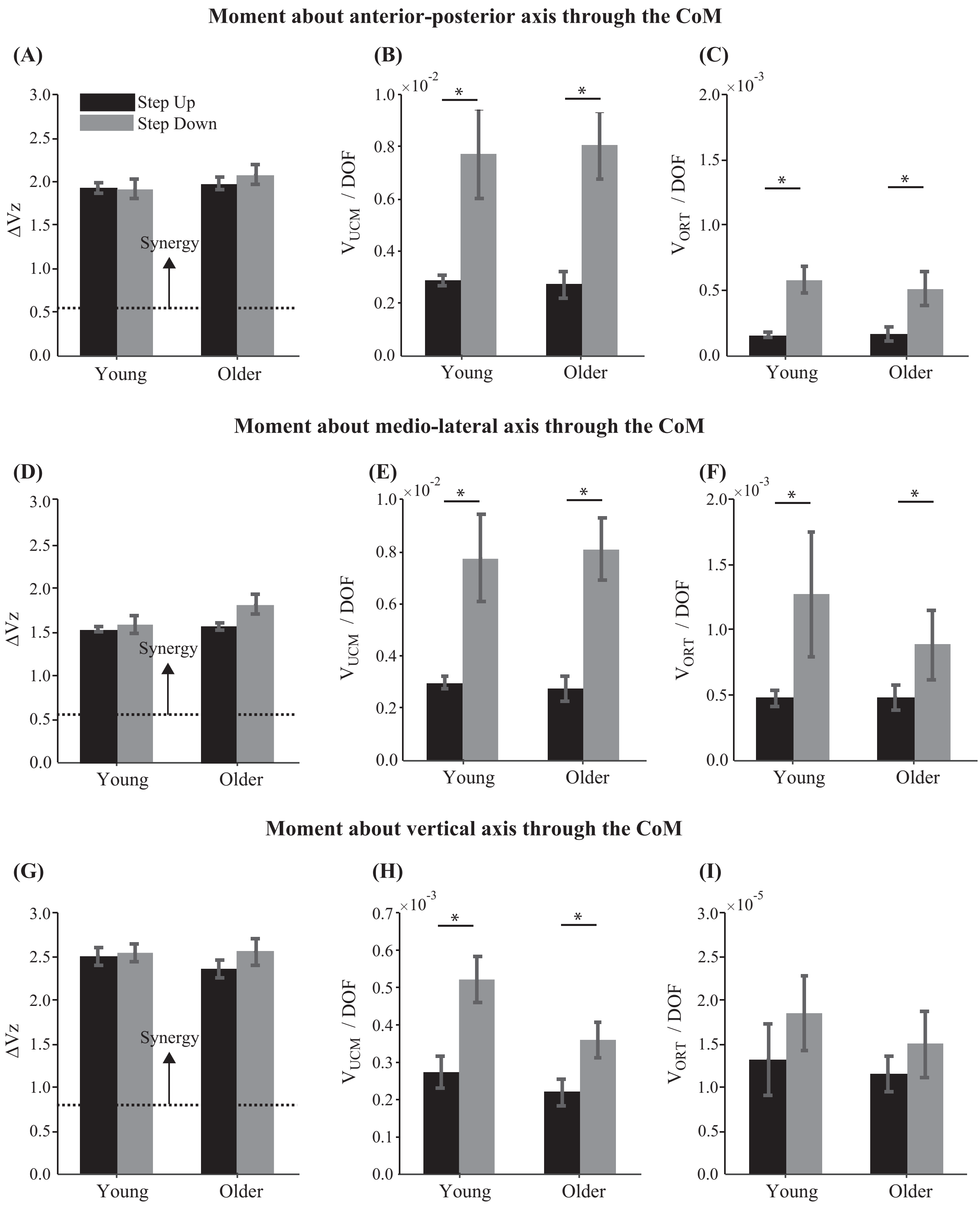

Fig. 6. Across-participant mean ± SE of the synergy index (ΔVz), and the variance components (V_{UCM} and V_{ORT}) averaged over the double-support phase. Each row of figures depicts the data from the analysis of resultant moment about the AP (Panels A-C), ML (Panels D-F), and vertical (Panels G-I) axis as the performance variable. The dashed line in Panels A, D, and G indicate the discriminating value of the synergy index that determines the presence of a synergy. Only main effects were observed. '*' indicates significant effect.

stabilizes a resultant force via negative covariation. For example, if the ML force under the left foot increases from one trial to the next, the ML force under the right foot must decrease a similar amount to maintain the resultant ML force across trials. The same force pair is also involved in stabilizing two resultant moments, and negative covariation will tend to destabilize at least one resultant moment. For example, since the lines of action of both ML forces are on the same side of the AP axis through the CoM (Fig. 1B), both forces create clockwise moments about this axis. Therefore, negative covariation in these forces will maintain the sum of their moments about the AP axis. Conversely, the lines of action of the ML forces lie on opposite sides of the vertical axis through the CoM (Fig. 1C). The

Table 4
Type of across-trial covariation in the forces required to stabilize a performance variable.

Force pair	Covariation required to stabilize:					
	$F_{Resultant-Z}$ (ML)	$F_{Resultant-X}$ (AP)	$F_{Resultant-Y}$ (vertical)	$M_{Resultant-Z}$ (ML)	$M_{Resultant-X}$ (AP)	$M_{Resultant-Y}$ (vertical)
ML	Negative				Negative	Positive
AP		Negative		Negative		Positive
Vertical			Negative	Positive	Positive	

forces create opposing moments about the vertical axis, and therefore, negative covariation in the ML forces tends to destabilize the resultant moment about the vertical axis. Thus, while stepping down, the weakening of the synergies stabilizing resultant vertical and AP forces helped stabilize the corresponding resultant moments (Table 4).

Note that the body gains angular and linear momenta while stepping down during the swing phase that must be regulated during double support (van Dieen et al., 2008). The gain in momenta increases task difficulty, forcing the central nervous system to prioritize. Prioritizing the resultant moment indicates the salience of this variable for maintaining balance while stepping down. We predict that such prioritization will be observed in behaviors accompanied by increase in angular momentum.

In addition to the prioritization of moments over forces, we also observed that the stability of resultant ML force was prioritized over that of the resultant AP force while stepping down. Note that there is an additional pair of variables – free moments – that contributes to stabilizing $M_{Resultant-Y}$ about the vertical axis. Negative covariation in the free moments could offset the destabilizing effect that the negative covariation in the AP, ML, or both those force pairs has on $M_{Resultant-Y}$. Nevertheless, if these forces were to stabilize $M_{Resultant-Y}$, either or both the forces-stabilizing synergies could have been compromised. However, only the AP force synergy vanishes, indicating the prioritization of the stability of ML translation over that of forward progression.

Prioritizing the ML force synergy over the AP force synergy is consistent with literature demonstrating the importance of lateral stability during locomotion (Bauby and Kuo, 2000; Eckardt and Rosenblatt, 2018; Krishnan et al., 2013; Rosenblatt et al., 2014; Rosenblatt et al., 2015). Although the importance of lateral stability has been demonstrated, the prioritization of ML stability over AP stability is a relatively novel finding. Recently, Lowry et al. (2017) showed that older adults prioritize ML control over forward progression during over-ground walking with prescribed step widths. Here, we report similar prioritization while stepping down, and not only in older adults, but also in young adults.

ML stability may be prioritized over AP stability because ML motion poses a greater challenge for control. Bipeds can utilize passive dynamic properties of the limbs in the sagittal plane, but must provide active control in the ML direction to maintain balance during locomotion (Bauby and Kuo, 2000). Errors in forward progression could be remedied with adjustments in subsequent step lengths, but error corrections in the ML direction, potentially requiring cross-over steps, are harder to execute (Patla et al., 1999). Furthermore, since step width is smaller than step length (Fig. 2B and C), the CoM motion along the ML direction is more tightly controlled to contain it within the base of support (O'Connor and Kuo, 2009). Finally, ML prioritization may have occurred since curbs have a single step; walking down a flight of stairs may lead to different behavior. Modifications to AP foot placement are limited while negotiating stairs, and accumulating errors along the AP direction will be more detrimental, since a forward fall on stairs is likely more injurious than a lateral fall.

Finally, the observed synergies are not simply consequences of the mechanics but arise from active control. There are sufficient GRVs to simultaneously stabilize all six PVs, as observed while stepping up. Therefore, the weakening of the force synergies to prioritize the moment synergies during stepping down represents an altered control strategy. Behaviors such as foot inversion to modify the CoP, upper body movements, and modulation of push-off impulse have been described as active control of locomotion (Hof et al., 2010; Joshi and Srinivasan, 2019). These actions will lead to concomitant changes in the GRVs during double support which will be reflected in the synergies. The underlying neurophysiological basis of synergies is unknown; however, it is likely that synergies arise from a mixture of feedback and feedforward mechanisms. Candidates are the tonic stretch feedback loops for appropriate muscles, and the supra-spinal input to the system of Renshaw cells, which regulate the output of the corresponding motoneuronal pool without afferent feedback (Latash et al., 2008).

4.2. Effect of aging

The only age-related difference was that V_{ORT} for the AP resultant force analysis was lower for the older adults. V_{UCM} also tended to be lower ($p = 0.07$) but did not reach statistical significance due to greater variability in the data for the young adults. This is in contrast to earlier studies that report higher variability in older adults during adaptive locomotion (Eckardt and Rosenblatt, 2018; Krishnan et al., 2013). The lower variability observed here can be interpreted as less flexible behavior (Eckardt and Rosenblatt, 2018). However, the lower variability could also reflect safer behavior. Speed was reduced while stepping down (Fig. 2D), so that the task was accomplished using low AP GRFs (Fig. 2G) with low variability. Cautious behavior may be an adaptation to mitigate fall risk. Further work is required to distinguish between these possibilities.

Age effects on synergies have been variable across studies that investigate different balance/locomotor tasks using various input variables and PVs. On the one hand, kinematic synergies (using kinematic input variables) stabilizing CoM position in balance-recovery tasks (Hsu et al., 2013) and step length during treadmill locomotion (Verrel et al., 2012) are weaker in older adults. On the other hand, kinematic synergies stabilizing toe trajectories during the swing phase while walking on flat and uneven ground (Eckardt and Rosenblatt, 2018) and on a treadmill (Krishnan et al., 2013) do not weaken with age. In upper extremity research, consistent age-related decline is observed in the synergistic control of manual forces but not in the kinematics of reaching (Shafizadeh et al., 2019). We expected this kinetic-kinematic distinction to transfer to locomotion. This did not occur; preserved kinetic synergies with age in our task are likely due to the highly practiced nature of this locomotor task compared to the tasks employed in manual studies and the greater cost of failure during locomotion.

Despite age-invariant synergies in the control of forces and moments observed here, quantifying kinetic synergies is a promising research direction. Age-related differences become apparent as the difficulty of the locomotor task increases (Muir et al., 2019). Therefore, it is likely that age differences in kinetic and kinematic synergies during adaptive locomotion will be observed for harder tasks. Future studies can examine adaptive gait tasks in populations with compromised balance, and identify how synergies are related with falls and fear of falling.

4.3. Limitations

The main limitation of this study is that we tested healthy, high-functioning older adults, which limits the generalizability of our results. Similarly, we need to compare fallers to non-fallers to determine if kinetic synergies are related to falls. Another limitation is that we assumed that the moment arms in our analyses were constant across trials. This avoids distortions to the input variable space due to normalizations required for its homogenization. Such distortions change the shape of the data cloud and can influence the overall conclusions (Sternad et al., 2010). However, this is a persistent theoretical challenge faced by all analyses of multidimensional variance (Scholz and Schöner, 2014). Another potential limitation is that the lateral forces have lower magnitudes than the vertical forces. However, the stabilization described here depends on covariation, and covariation is independent of magnitudes.

5. Conclusions

Synergies are evident in the ground reaction variables during curb negotiation. Synergies stabilizing resultant moments on the body are prioritized over synergies stabilizing resultant forces, and the synergy stabilizing the resultant ML force is prioritized over the synergy stabilizing the resultant AP force. The prioritization occurs while stepping down only, indicating that different neural control strategies are employed across the two stepping directions. Finally, healthy aging does not diminish the stability of the resultant force and moment acting on the body during curb negotiation.

CRediT authorship contribution statement

Chuyi Cui: Software, Formal analysis, Investigation, Data curation, Writing - review & editing, Visualization. **Ashwini Kulkarni:** Investigation, Writing - review & editing. **Shirley Rietdyk:** Supervision, Conceptualization, Writing - original draft, Writing - review & editing. **Fabio A. Barbieri:** Investigation, Writing - review & editing. **Satyajit Ambike:** Supervision, Conceptualization, Methodology, Writing - original draft, Writing - review & editing.

Declaration of Competing Interest

None of the authors have any financial or personal relationships with other people or organizations that could inappropriately influence or bias our work.

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Appendix A

A.1. Uncontrolled manifold analysis

We assume soft contact between the foot and the ground, i.e., the foot can exert a three-dimensional force on the ground and also a twisting moment (called free moment) about the vertical axis (Murray et al., 1994; Zatsiorsky, 2002). The PVs resultant force ($F_{Resultant}$) and moment ($M_{Resultant}$) are vector quantities. Each component in $F_{Resultant}$ and $M_{Resultant}$ has its own set of input GRVs. Therefore, we performed UCM analysis for each component separately. The output PV and input GRVs for each UCM analysis are listed in Table 2. Fig. 7 shows the evolution of each input and output variable over the double support phase for one representative participant in each age group (see Supplemental materials).

The first three hypotheses were that the resultant force along each coordinate direction is stabilized by the covariation in the forces under both feet in that direction. The resultant force is given by $\Sigma F_j = F_{Rj} + F_{Lj}$, where the subscripts 'R' and 'L' denote the right and left foot, respectively, and j = X, Y, or Z (along the AP, vertical, or ML directions, respectively, Fig. 1B). The time series of the GRFs during double support from the 15 trials were time normalized and then normalized by the participant's weight. The Jacobian relating small changes in the normalized GRFs to small changes in the corresponding normalized resultant force is [1 1]. This Jacobian has a one-dimensional null space, defining the UCM, and a one-dimensional orthogonal compliment, defining the ORT. At each time instant (t^*), the deviation in the weight-normalized GRFs for each trial from the corresponding across-trial mean value was computed, and this difference was projected onto the UCM and ORT manifolds. The variances in these projections are $V_{UCM}(t^*)$ and $V_{ORT}(t^*)$. By definition, the deviation in the GRFs that aligns with the null space of the Jacobian (i.e., UCM) does not change the PV, and the deviation that lines up with the ORT changes the PV. Therefore, V_{UCM} (also called 'good' variance) quantifies the changes in the inputs that stabilize the PV, and V_{ORT} (also called 'bad' variance) quantifies the changes in the inputs that change the PV. A synergy index ΔV, and its z-transformed value were computed as follows:

$$\Delta V(t^*) = \frac{\frac{V_{UCM}(t^*)}{n} - \frac{V_{ORT}(t^*)}{m}}{\frac{V_{UCM}(t^*)+V_{ORT}(t^*)}{n+m}}, \tag{1}$$

$$\Delta V_z(t^*) = 0.5 \times log\left[\frac{|\Delta V_{lower}| + \Delta V(t^*)}{|\Delta V_{upper}| - \Delta V(t^*)}\right], \tag{2}$$

where n and m are the dimensions of UCM and ORT, respectively. The synergy index was z transformed since it is bounded: $\Delta V_{lower} \le \Delta V \le \Delta V_{upper}$. For the resultant force analyses, n = m = 1, $\Delta V_{lower} = -2$, and $\Delta V_{upper} = 2$. Furthermore, $\Delta V = 0$ yields $\Delta Vz = 0$, which is the discriminating value indicating the presence of a synergy ($\Delta Vz > 0$), an anti-synergy ($\Delta Vz < 0$), or a lack of task-specific covariation in the GRFs ($\Delta Vz = 0$).

The next three hypotheses were that the resultant moments about the three coordinate axes were stabilized by covariation in the GRVs. The resultant moments about the ML, AP and vertical axes passing through the CoM are obtained by summing the moments created by the GRFs and the free moments under both feet:

$$M_{Resultant-Z} = (F_{RY})(d_{RX}) + (F_{RX})(d_{RY}) - (F_{LY})(d_{LX}) + (F_{LX})(d_{LY}), \tag{3}$$

$$M_{Resultant-X} = -(F_{RY})(d_{RZ}) - (F_{RZ})(d_{RY}) + (F_{LY})(d_{LZ}) - (F_{LZ})(d_{LY}), \tag{4}$$

$$M_{Resultant-Y} = (F_{RX})(d_{RZ}) - (F_{RZ})(d_{RX}) - (F_{LX})(d_{LZ}) + (F_{LZ})(d_{LX}) + FM_R + FM_{L,} \tag{5}$$

where the forces (F_{ij}), free moments (FM_i) and distances (d_{ij}) are as defined in Fig. 1. Similar to the analyses of resultant forces, the GRVs were time normalized across repetitions. The GRFs were normalized by the participant's weight. Note that the space of the input variables for the analysis of the resultant moment $M_{Resultant-Y}$ (Eq. (5)) is non-homogeneous, unlike the remaining five analyses. Therefore, the free moments were normalized by the participant's weight (W) and height (H). This creates a set of dimensionless input variables: $F'_{ij} = F_{ij}/W$, and $FM'_i = FM_i/(WH)$, and Eqs. (3), (4) and (5) are rewritten in terms of these new variables. The Jacobians are then obtained by computing the partial derivatives of the new equations with respect to F'_{ij} and FM'_i. This yields the following three Jacobians relating small changes in the dimensionless GRVs to small changes in the PVs:

$$J_Z = W[d_{RX} \quad d_{RY} \quad -d_{LX} \quad d_{LY}], \tag{6}$$

$$J_X = W[-d_{RZ} \quad -d_{RY} \quad d_{LZ} \quad -d_{LY}], \tag{7}$$

$$J_Y = W[d_{RZ} \quad -d_{RX} \quad -d_{LZ} \quad d_{LX} \quad H \quad H]. \tag{8}$$

Similar normalization techniques to homogenize the space of input variables have been used previously (Dingwell et al., 2010; Klishko et al., 2014). The moment arms in the AP and ML directions (d_{iX} and d_{iZ}) are the corresponding distances between the CoM and the CoP under the appropriate foot. Representative trajectories of CoM and CoP are presented in Figs. 7C, D, I, J (Supplemental materials). The vertical moment arm (d_{iY}) is the height of the CoM from the corresponding surface. Note that these distances change throughout double support, and therefore, the Jacobians vary with normalized time. The distances at time t* are assumed to be constant across trials, and their across-trial means are utilized to define the numerical Jacobians at t* for each analysis. The null space and its orthogonal compliment for the numerical Jacobians at t* define the UCMs and ORTs, respectively, for time t*. Note that the leading scalar (W) in Eqs. (6)–(9) does not change the UCM and ORT subspaces. Therefore, the appearance of W in Eqs. (6)–(9) does not create any additional normalization issues.

Similar to the analysis for the resultant force, the deviation in the normalized and demeaned GRVs at time t* from the corresponding across-trial means are projected onto the appropriate UCM and ORT manifolds, and the variances in these projections yield $V_{UCM}(t^*)$ and $V_{ORT}(t^*)$. The synergy index and its z-transformed value are obtained using Eqs. (1) and (2). For the analysis of resultant moments about the AP and ML axes, the UCM and ORT manifolds are 3 and 1 dimensional, respectively (n = 3, m = 1). Therefore, $\Delta V_{lower} = -4$, and $\Delta V_{upper} = 4/3$, and $\Delta V = 0$ yields $\Delta Vz = 0.55$, which is the discriminating value indicating the presence of a synergy or an anti-synergy, or a lack of synergistic covariation. For the analysis of the resultant moment about the vertical axis, n = 5 and m = 1. Therefore, $\Delta V_{lower} = -6$, and $\Delta V_{upper} = 6/5$, and $\Delta V = 0$ yields a discriminating value of $\Delta Vz = 0.80$.

Finally, we address the role of gravity in the UCM analyses. Consider the analysis of the resultant vertical force (along the Y direction; Fig. 1B). The PV and the inputs are related as $F_{Resultant\text{-}Y} = F_{RY} + F_{LY}$. Furthermore, the vertical acceleration of the CoM is determined as: $m \times a_{CoM\text{-}Y}(t^*) = F_{Resultant\text{-}Y}(t^*) + W$, or, after normalization: $(m/W) \times a_{CoM\text{-}Y}(t^*) = F_{Resultant\text{-}Y}(t^*)/W + 1$. It is evident that if the PV ($F_{Resultant\text{-}Y}(t^*)/W$) is stabilized via a synergy between ($F_{RY}(t^*)/W$) and ($F_{LY}(t^*)/W$), the acceleration of the CoM will also be stabilized by virtue of Newton's second law.

Therefore, gravity does not play a part in these analyses. Although W influences the translation of the CoM, W is not a control variable, and therefore, it does not feature in the UCM analysis. To state this mathematically, UCM analysis assesses trial-to-trial variability in the GRVs. Since W does not vary between trials, it does not contribute to the measures obtained from the UCM analysis. Gravity does not feature in the computation of $M_{Resultant}$, since $M_{Resultant}$ is computed about the CoM.

Appendix B. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2020.109837.

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