Stroke impairs the proactive control of dynamic balance during predictable treadmill accelerations

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Abstract

We maintain balance during gait using both proactive and reactive control strategies. Damage to the brain from a stroke impairs reactive balance, but little is known about how a stroke impacts proactive control during walking. Identifying stroke-related impairments to proactive control could inform the design of interventions to improve balance and reduce fall risk. Therefore, we determined if people with and without stroke differ in their ability to use proactive strategies during predictable treadmill accelerations. Groups of 14 adults with and without stroke walked with single-belt accelerations at random (every one to five strides) and regular (every three strides) intervals. We quantified the effects of the perturbations as changes to center of mass (COM) speed and used mechanical leg work to quantify the proactive strategies to slow the COM. Participants without stroke reduced peak COM speed better than those with stroke when perturbations were at regular intervals (-0.023 versus -0.001 m/s; p=0.007). They also reduced positive leg work more during the perturbation step than the group post-stroke (-7.7% versus -1.9%; p=0.004). Together, these results demonstrate that the sensorimotor impairments resulting from a stroke impair the ability to generate effective proactive adjustments to reduce the destabilizing effects of predictable perturbations.

Introduction

People use a combination of proactive and reactive control strategies to maintain their balance while walking. Proactive control strategies generate motor commands to achieve a desired state based on knowledge about the body and the environment (1). When there are unexpected perturbations to a planned movement, deviations in the resultant state from the desired state, or errors, result in two coupled effects. Errors can trigger reactive control strategies and update the proactive mapping between error and reactive adjustments (1) through error-based learning. This learning process can be studied during walking with repeated perturbations, during which people use prior experience to reduce the destabilizing effects of the perturbation on their center of mass (COM) dynamics. After repeated practice, people reduce vertical displacement of the COM position during slips (2) and reduce trunk sway during both trips and slips (3). However, it is unknown exactly what strategies (e.g., changes to kinematics, kinetics, and central set) people implement to reduce the destabilizing effects of predictable perturbations on their COM dynamics or the relative contributions of proactive and reactive control during this process.

One proactive strategy to reduce the destabilizing effects of an expected perturbation is to modify foot placement or COM kinematics. A general requirement for maintaining balance during walking is to keep the COM within the base of support (BOS) (4). Therefore, people may increase the distance between the COM and the edge of the BOS to improve stability. For example, after repeated slips, people decrease step length and anteriorly shift their COM to reduce the chance of a backward loss of balance (5). A common measure of dynamic stability combines evaluations of the BOS and COM state in a single value known as the margin of stability (MOS) (6–9). The MOS is calculated as the distance between the edge of the BOS and one's velocity-adjusted COM position (10), where increasing the MOS in the same direction as a perturbation can improve stability. Taking longer steps or slowing the COM can each increase the MOS and configure the body to oppose a forward loss of balance, especially if the perturbation is predictable.

People may also implement a kinetic proactive strategy by modifying limb mechanics to reduce the body's momentum in the direction of the balance disturbance. For example, reducing the positive work or increasing the negative work performed by the legs in anticipation of a forward loss of balance can each reduce the COM momentum. Most studies of walking employing predictable perturbations have focused on kinematic proactive strategies (3,5,6,9,11–14), but studies of running have reported kinetic strategies as people decrease ground reaction forces during visible drops (15) or decrease leg stiffness on visible vertical steps (16,17). Thus, kinetic adjustments to the control strategy may help mitigate the effects of an expected perturbation during walking.

Lastly, proactively "setting" the readiness of the central nervous system can scale the response to an expected perturbation (18,19). Central set can be evaluated in multiple ways, including measuring changes in electromyography (EMG) or the scaling of biomechanical responses to perturbations. During expected surface translations in standing, the scale of initial muscle activations and torques following perturbation onset is set according to the anticipated amplitudes, such that people over- or under-respond when a perturbation is smaller or larger than expected (20). Repeated exposure to unexpected perturbations can also attenuate muscle response magnitudes, as is seen in the reduced activity in the biceps femoris, tibialis anterior, and gastrocnemius after repeated slips during walking (2). Similar changes to neuromuscular control are seen across species, as a more predictable treadmill obstacle allows guinea fowl to increase anticipatory muscle activity, both in magnitude and the number of muscles included (21). When EMG data are unavailable, it may also be possible to assess central set by quantifying changes in the relationship between COM state and foot placement. Modulating foot placement based on COM state is proposed as an active balance control mechanism, with COM state at midstance predicting over 80% of the variability in subsequent lateral foot placement during

unperturbed gait (22). Changes to the correlation between COM state and foot placement may indicate changes to the sensitivity or gain of the control strategy. This mapping has been shown to differ during perturbed versus unperturbed conditions (23,24), but the effect of perturbation predictability on feedback control gains has not yet been explored. Perturbation-induced changes to EMG magnitudes and the COM-foot placement mapping suggest that people can set their control strategy according to task demands.

While it is clear that adults without stroke make proactive changes after repeated exposure to perturbations, less is known about proactive control following a stroke. About one-third of chronic stroke survivors report experiencing at least one fall in a one-year period (25), but most research efforts to evaluate post-stroke control have been conducted during standing (26–29), and the limited number of studies in walking have mostly focused on reactive control with random perturbations (30–33). Various post-stroke impairments may hinder proactive adjustments during repeated perturbations. For example, cognitive deficits, observed in 60% of chronic patients (34), could hinder the ability to learn with experience. Muscle weakness and abnormal coordination (35) could hinder the ability to proactively modulate one's gait pattern. Initial efforts to characterize post-stroke proactive control during walking have found that people post-stroke trip over visible obstacles more frequently than their peers without stroke (12,36). While obstacles help probe a visuomotor element of proactive control, obstacles are avoidable and thus may not be ideal for understanding the impact of movement errors or the reactive element of balance control. Instead, physical perturbations could more effectively examine the relative roles of proactive and reactive control strategies in people post-stroke, whose balance control strategies are perhaps less flexible (12) and less strongly correlated to COM state (37) than their peers without neurologic injury.

Our goal was to investigate how the predictability of treadmill accelerations informs proactive and reactive balance control strategies in people with and without a stroke. Because people may learn the regular pattern, we hypothesized that regular accelerations would promote proactive changes to reduce the destabilizing effects on their COM dynamics. However, we expected people post-stroke to be less flexible in changing strategies and, therefore, less effective. To test our hypotheses, we evaluated if people made changes to kinematics (fore-aft MOS), kinetics (trailing and leading leg mechanical work), and feedback control gains (COM-foot placement mapping) to reduce the peak COM speeds during regular versus random accelerations and if these strategies differed in people post-stroke. Specifically, we expected that compared to random perturbations, regular perturbations would allow people without stroke to proactively increase their fore-aft MOS, reduce positive leg work and increase negative leg work, and increase sensitivities in foot placement to fore-aft changes in COM state to reduce peak COM speeds.

Methods

Participants

Fourteen adults with chronic stroke and 14 age-/sex-matched adults without stroke (58 ± 13 years old, 14M/14F, Table 1) completed the following protocol. Participants were recruited through the University of Southern California Registry for Aging and Rehabilitation Evaluation (RARE) and the Healthy Minds Research Volunteer Registry. People post-stroke were recruited if they had a single stroke over six months prior and could walk unassisted for at least five minutes. People in either group were excluded if they had musculoskeletal conditions that limited walking ability, uncontrolled hypertension, or cognitive impairment (Montreal Cognitive Assessment score <19/30 (38)).

Table 1. Group means ± standard deviations of demographic characteristics and clinical scores. MoCA: Montreal Cognitive Assessment, Mini-BEST: Mini Balance Evaluation Systems Test, ABC Scale: Activities-specific Balance Confidence Scale.

Group	Height (m)	Weight (kg)	Treadmill speed (m/s)	MoCA (0-30)	Mini-BEST (0-28)	ABC Scale (%)	Fugl-Meyer (0-34)	Years since stroke
Stroke	1.66 ± 0.09	75 ± 17	0.66 ± 0.17	26 ± 2	21 ± 4	78 ± 15	23 ± 5	9.0 ± 4.7
Non-stroke	1.70 ± 0.10	76 ± 14	1.05 ± 0.11	27 ± 2	26 ± 2	96 ± 5		

Experimental protocol

First, we performed several clinical assessments of motor impairment, cognition, and function. We used the Fugl-Meyer Assessment to measure the lower-extremity (LE) impairment post-stroke. For both groups, we administered the Montreal Cognitive Assessment (MoCA) to screen for cognitive impairment, Activities-specific Balance Confidence (ABC) scale to estimate self-reported confidence in one's abilities to perform tasks without losing balance, Mini Balance Evaluations Systems Test (Mini-BEST) to evaluate dynamic balance, and 10-Meter Walk Test (10MWT) to calculate self-selected walking speeds (SSWS) overground. To determine participants' SSWS on the treadmill, we used a staircase algorithm that started at 50% of their overground SSWS and stepped up or down according to their reported comfort.

Following the clinical assessments, participants completed eight walking trials on an instrumented dual-belt treadmill (Bertec, OH) at their SSWS. During the two baseline trials, people acclimated to the treadmill (one minute) and then experienced three perturbations per side. A custom Python script triggered trip-like perturbations by rapidly accelerating one belt up to 150% SSWS upon heel strike before returning to SSWS after that step. After the baseline trials, participants completed six perturbation trials, each consisting of 40 single-belt accelerations on the nondominant or paretic side at random (every one to five strides) or regular (every three strides) intervals. The order of the two conditions was randomized, and each was repeated three times. Performing three shorter trials per condition gave participants more frequent rest breaks to prevent fatigue. For safety, participants wore an overhead harness that did not provide body-weight support. They were also informed that perturbations would occur but were not provided information about the timing.

Data acquisition and processing

Participants wore 54 reflective markers for 3D motion capture. We used Qualisys Track Manager to record the positions of these markers with a 10-camera optical motion capture system (Qualisys Oqus, Sweden) at 100 Hz and force data from the instrumented treadmill at 1000 Hz.

Data were then analyzed with custom Visual3D (HAS-Motion, Ontario, Canada) and MATLAB (MathWorks, Natick, MA) scripts. Marker data were low-pass filtered (Butterworth with 6-Hz cutoff) and gap-filled (third-order polynomial with a maximum gap of 10 frames). The body's center of mass (COM) was estimated using a 13-segment (head, torso, upper arms, forearms, pelvis, thighs, shanks, and feet) model in Visual3D. We also identified and visually confirmed the following events in Visual3D: 1) heel strikes (HS) and toe offs as the maximum and minimum fore-aft positions of the heel and toe markers, respectively, 2) perturbation onset based on measures of the left and right belt speeds, and 3) "crossovers" when participants' feet were placed on

opposite belts. While participants experienced 120 total accelerations per condition, only the last 40 were included in the main analysis to minimize the effects of exploration and learning. Of these 40 perturbation steps per condition, an average of two steps per participant were removed from force-related calculations because of crossover events. Additionally, we removed an average of 11 random-interval perturbation steps per participant because they occurred within one stride of the prior perturbation and may not have been preceded by sufficient time for the participant to return to a steady state.

Peak COM speed as a measure of the destabilizing effects of the perturbations

Due to the trip-like nature of the treadmill accelerations, we expected to see increases in peak COM speed during perturbations (39) and reductions in this peak with predictable perturbations as people may try to reduce trajectory deviations for stability (40,41). We used all three components of the COM velocity vector relative to the treadmill at SSWS to compute peak COM velocity magnitude during the perturbation steps. We then normalized this metric to each participant's SSWS. If perturbation predictability helps people effectively prepare, we expect peak COM speeds to be slower during regular versus random accelerations.

Kinematic strategy: Fore-aft MOS

We evaluated if people implemented a kinematic proactive strategy to reduce the destabilizing effects of predictable accelerations. Specifically, we tested the hypothesis that people would have a larger fore-aft margin of stability (MOS) upon HS of the regular perturbation step to reduce its destabilizing effects. MOS was calculated as the distance between the edge of the base of support (BOS) and the extrapolated COM (10) in the fore-aft direction (Equation 1).

$$MOS = BOS - (x_{COM} + \frac{v_{COM}}{\sqrt{g/l}}) \qquad \text{(Eq. 1)}$$

Here, we defined the leading edge of the BOS as the fore-aft position of the toe marker. The extrapolated COM is the COM position (x_{COM}) adjusted by COM velocity (v_{COM}), gravity (g), and leg length (l). Negative MOS values indicate that one's extrapolated COM is outside the BOS, and a fall will occur unless one makes a corrective action, such as taking a step. People may prepare kinematically for predictable accelerations by increasing their fore-aft MOS at the HS of the perturbation step during the regular versus random conditions.

Kinetic strategy: Mechanical leg work

The mechanical work performed by the leading and trailing legs can characterize the kinetic strategies employed in response to treadmill perturbations, as people modulate leg work depending on the perturbation's timing (42). Mechanical power performed by the leg (P_{leg}) was computed per step using the individual limbs method (43) as the sum of the power from the leg on the body and the power from the leg on the treadmill (Equation 2).

$$P_{leg} = \left(\vec{F} \cdot \vec{v}_{com}\right) + (-\vec{F} \cdot \vec{v}_{belt}) \qquad \text{(Eq. 2)}$$

$\vec{F}$ is the ground reaction force, $\vec{v}_{com}$ is the COM velocity, and $\vec{v}_{belt}$ is the treadmill velocity relative to the lab coordinate system. We then computed work as the integral of power over time, with positive and negative work calculated separately to differentiate between energy generated and dissipated by the legs. Work was also normalized by subject mass to account for differences in weight between participants.

We quantified three potential changes to leg work that may reduce the destabilizing effects of predictable perturbations. First, people may decrease positive work performed by the trailing leg (Figure 1A) to reduce the initial forward COM velocity at the beginning of the perturbation step and maintain the COM position further behind the leading edge of the BOS. Next, people may increase negative work performed by the leading leg during early stance (Figure 1B), absorbing some of the energy from the perturbation and reducing its destabilizing effects. Finally, people may decrease positive work performed by the leading leg (Figure 1B) for similar reasons as they may do so on the trailing leg. Thus, during regular versus random accelerations, we expected people to perform less positive work by the trailing and leading legs and more negative work by the leading leg.

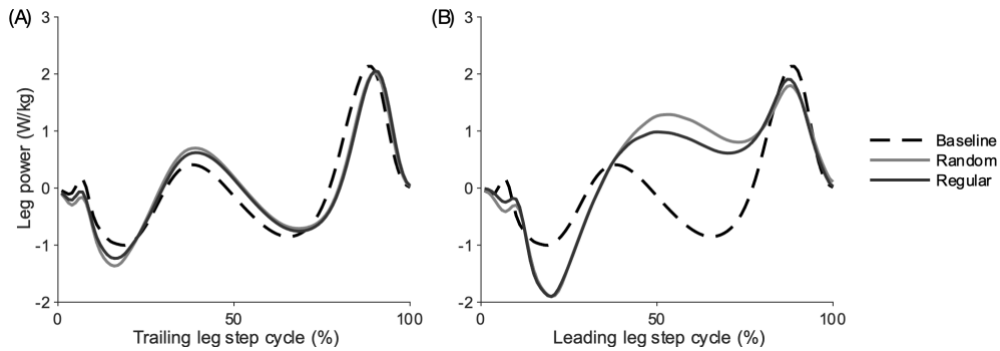

Figure 1: Mean mechanical leg power for the non-stroke group plotted across the step cycles of the (A) trailing leg and (B) leading leg during the baseline, unperturbed condition (dashed), random-interval condition (gray solid), and regular-interval condition (black solid). While there were minimal visible changes to trailing leg power, the perturbations elicited changes to leading leg power with more energy absorption (negative) early in stance but more energy generation (positive) starting around midstance.

Central set strategy: COM-foot placement mapping

In agreement with prior methods (24,44), we estimated the relationship between anterior-posterior (AP) and mediolateral (ML) COM state at midstance (50% of the step cycle) and subsequent AP foot placement by computing the Jacobian or partial derivative matrix (J) in Equation 3.

$$\Delta q \approx J \Delta s \qquad \text{(Eq. 3)}$$

$$J = \begin{bmatrix} \frac{\partial FP_{AP}}{\partial COM_{AP}} & \frac{\partial FP_{AP}}{\partial COM_{ML}} & \frac{\partial FP_{AP}}{\partial VCOM_{AP}} & \frac{\partial FP_{AP}}{\partial VCOM_{ML}} \end{bmatrix}$$

Here, COM state (s) is a 4x1 vector with elements corresponding to the COM's AP position, ML position, AP velocity, and ML velocity at midstance. J is a 1x4 row vector with coefficients or gains that map the change in COM state ($\Delta s = s_k - s^*$) to the change in foot placement ($\Delta q = q_{k+1} - q^*$) compared to their average values s* and q* calculated during the one-minute unperturbed baseline trial. To estimate J, we used data from the final 40 perturbation steps per participant to fit separate linear mixed-effects models per condition (random and regular) per group (with and without stroke). The four models included fixed effects for COM state (AP position, ML position, AP velocity, ML velocity) at midstance to predict subsequent AP foot placement. Random slopes allowed for inter-subject differences in sensitivity to the COM state. We hypothesized that people, especially those without neurologic injury, would be more sensitive to changes in their AP COM state (first and third gains in J) during regular versus random

accelerations, such that they take a longer, more stabilizing step following perturbation-induced changes to their COM. Subsequently, we also evaluated fore-aft MOS (Equation 1) at HS of the recovery step to evaluate if any changes to the COM-foot placement mapping helped improve stability following the predictable accelerations.

Statistics

Statistical analyses were performed in MATLAB version R2022b. We used separate linear mixed-effects models to evaluate the effects of Group (two categorical levels), Trial (two categorical levels), and the interaction between Group and Trial on peak COM speed, fore-aft MOS, mechanical leg work, and gains within the COM-foot placement mappings. Finally, exploratory linear regressions evaluated correlations between clinical measures (SSWS, Mini-BEST, Fugl-Meyer, and MoCA) and the change in peak COM speed to identify potential relationships between impairment and performance during the accelerations. For all statistical analyses, alpha was set to 0.05.

Results

Participants without stroke mitigated the destabilizing effects of regular accelerations

As intended, our treadmill perturbations caused changes in whole-body kinematics. The destabilizing nature of the accelerations is demonstrated by a representative participant without stroke whose average peak COM speeds increased during perturbed versus unperturbed steps (Figure 2). For additional context, during random accelerations, participants with and without stroke reached average peak COM speeds of 0.86 and 1.28 m/s, respectively.

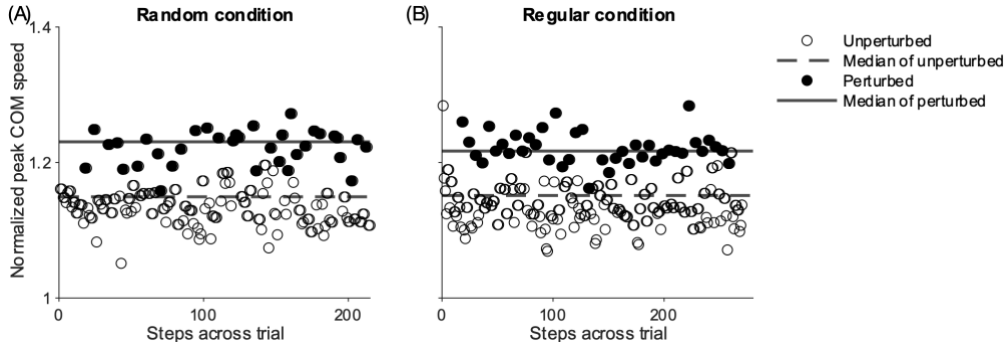

Figure 2: Representative data from a participant without stroke. Normalized peak center of mass (COM) speeds are plotted per step during the final trials with (A) random- and (B) regular-interval accelerations. Peak COM speeds during unperturbed steps and their median values are plotted in open circles and dashed lines, while peak COM speeds during perturbed steps and their median values are plotted in black circles and solid lines.

First, we evaluated the change in peak COM speeds across the 120 perturbations per condition by comparing the first and final 20 accelerations in separate linear mixed-effects models. For both groups, there was a significant interaction between time (first and final 20) and trial (stroke: p=0.029; non-stroke: p=0.005), such that people demonstrated a significantly greater reduction in peak COM speeds across the regular vs. random accelerations. Both groups had negligible changes in peak COM speed during the final versus the first 20 random accelerations (Figure 3A). However, they both reduced the destabilizing effects of the perturbations when they

occurred at regular intervals as peak COM speeds slowed by an average of 1.8% in people with stroke and 1.9% in people without stroke during the final versus the first 20 regular accelerations (Figure 3B).

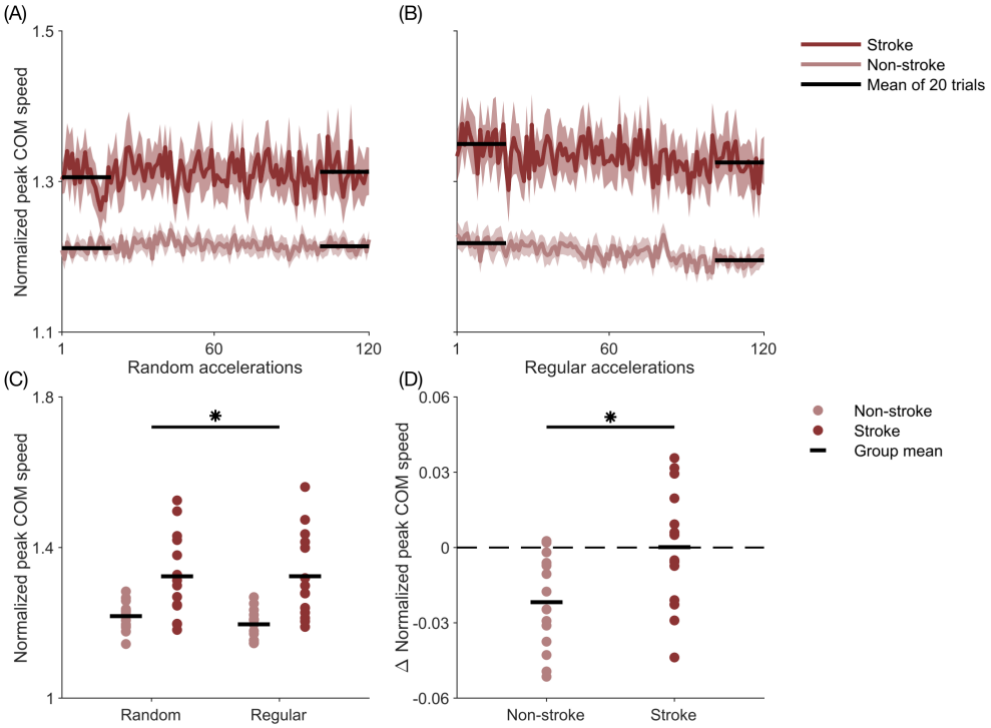

Figure 3: Normalized peak center of mass (COM) speeds per group during the 120 (A) random and (B) regular accelerations. The means of the first 20 and last 20 accelerations per group per condition are plotted as horizontal black lines. Shading represents the standard error across participants. Considering only the last 40 accelerations per condition, (C) normalized peak COM speeds were slower during regular versus random accelerations, but (D) the group without stroke had a significantly greater reduction during regular versus random accelerations than the group with stroke.

Next, we compared peak COM speeds during the final 40 accelerations between groups and trial conditions. Normalized peak COM speeds during perturbations were slower in the group without stroke than those with stroke (Figure 3C; Group p=0.001). Compared to the random-interval condition, participants used the predictability from the regular intervals to slow peak COM speeds (Figure 3C; Trial p<0.001), and this reduction was greatest in the group without stroke (Figure 3D; Group*Trial p=0.007). Participants without stroke best mitigated the destabilizing effects of the predictable perturbations, slowing peak COM speeds by 0.023 m/s (1.8%) during regular versus random accelerations versus 0.001 m/s (0%) in the group post-stroke.

Modulation of mechanical work by the trailing and leading legs with predictability

The reductions in peak COM speeds may have been due to proactive kinematic changes to the body's configuration at initial contact or changes in mechanical work before or during the perturbation. However, people did not appear to utilize a kinematic strategy, as there was no significant difference in the fore-aft MOS at HS of random versus regular accelerations (Trial p=0.172; Group*Trial p=0.628). Instead, people may have utilized a kinetic strategy to reduce the destabilizing effects of the predictable perturbations, as people performed less trailing leg positive

work before regular versus random accelerations (Figure 4A; Trial p=0.014). This reduction was an average of 0.012 J/kg (3.9%) for participants without stroke and 0.004 J/kg (0.6%) for the participants with stroke, but there were no significant group differences within trials (Figure 4B; Group*Trial p=0.207). Reducing positive work by the trailing leg during regular versus random perturbations would help slow the COM at the onset of an impending perturbation.

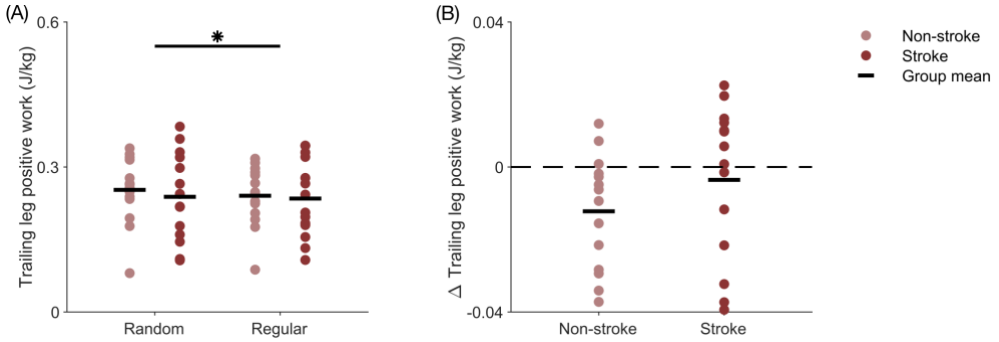

Figure 4: Changes in trailing leg work before regular versus random perturbations. (A) People performed less positive work by the trailing leg during regular versus random accelerations. (B) There was no significant group difference in how much people reduced trailing leg positive work before regular accelerations.

In addition to performing less trailing leg positive work, the group without stroke performed 0.012 J/kg (8.8%) more leading leg negative work during regular versus random accelerations (Figure 5A; Group*Trial p=0.029). Increasing leading leg negative work would slow the COM during the initial perturbation, as people largely perform this work during the loading phase (Figure 1B). Participants without stroke also performed 0.033 J/kg (7.7%) less positive work by the leading leg during regular versus random accelerations (Figure 5B; Group*Trial p=0.004). This positive work typically occurred during midstance (Figure 1B), and reducing this work would also help slow the COM. People post-stroke did not modulate leading leg negative or positive work during regular accelerations like their peers without stroke, which may have hindered their ability to reduce the destabilizing effects of the perturbations.

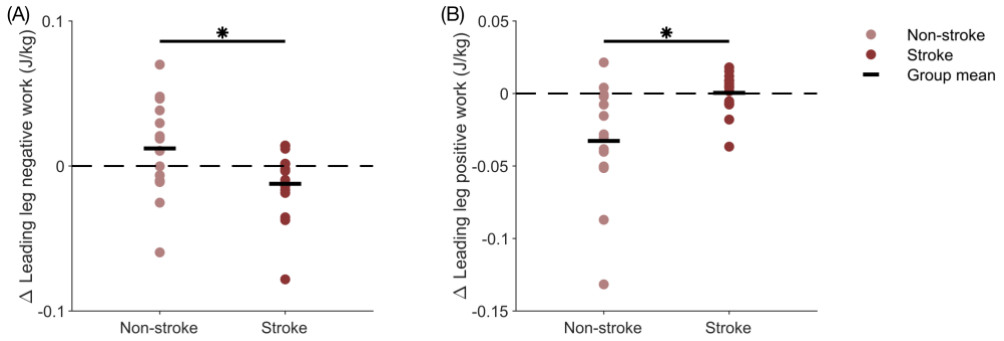

Figure 5: Changes in leading leg work during regular versus random perturbations. Compared to the group post-stroke, the group without stroke performed (A) more negative work and (B) less positive work with the leading leg during regular versus random perturbations.

To elucidate which changes in mechanical leg work were associated with the observed reductions in COM velocity during regular accelerations, we fit a multiple linear regression model with percent changes in trailing leg positive work, leading leg negative work, and leading leg positive work as the three predictors of the percent change in peak COM speeds during regular

versus random accelerations. After confirming non-collinearity among the predictors, we found that changes in trailing and leading leg positive work, but not leading leg negative work (p=0.075), were significant predictors of changes in peak COM speeds. In a model that included the two statistically significant predictors, performing less positive work by the trailing (p=0.013) and leading (p<0.001) legs explained approximately 50% of the reductions in peak COM speeds during regular versus random accelerations (R^2=0.509, p<0.001).

People modify feedback gains during regular versus random perturbations

Similar to prior work (24,44), we fit linear mixed-effects models with the COM state at midstance as a predictor of subsequent anterior-posterior (AP) foot placement. Changes to COM state during the midstance of random and regular accelerations explained changes to AP foot placement with coefficients of determination (R^2) of 0.68 and 0.71, respectively, for the group without stroke (Figure 6A). The model fits were similar in the group post-stroke, with R^2 values of 0.70 and 0.65 for the random and regular conditions, respectively (Figure 6B).

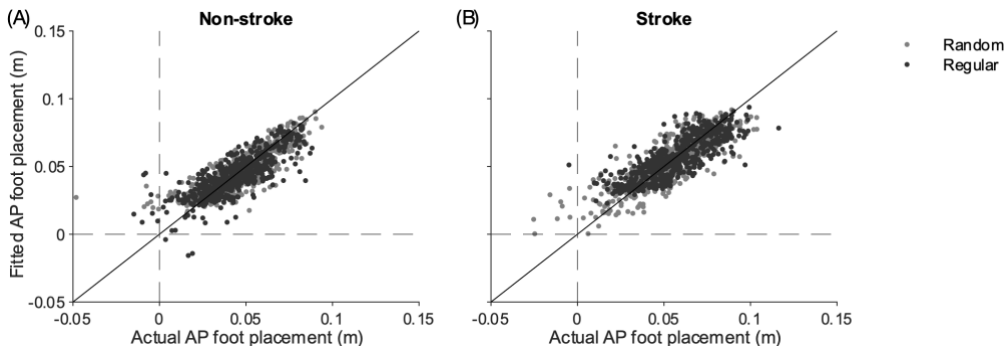

Figure 6: Estimated versus actual anterior-posterior (AP) foot placement in the groups (A) without and (B) with stroke. If data fall along the unity line, the estimated and actual foot placements are equivalent.

We also evaluated how the individual Jacobian terms differed between groups, trial conditions, and the interaction between the two. Because the perturbations were in the AP direction, we specifically expected the first and third gains to change with perturbation predictability, such that AP foot placement would be more sensitive to perturbation-induced changes in the AP COM state during regular versus random accelerations. As hypothesized, the first gain $\frac{\partial FP_{AP}}{\partial COM_{AP}}$ was significantly greater in the regular versus random model (Figure 7; Trial p<0.001), signaling that the same anterior displacement in the COM position resulted in a longer step following more predictable accelerations. There was also a significant effect of group (p<0.001), trial condition (p<0.001), and their interaction (p<0.001) on the second gain $\frac{\partial FP_{AP}}{\partial COM_{ML}}$. During regular versus random accelerations, this second gain was more negative in the group without stroke but less negative in the group post-stroke (Figure 7). These changes suggest that AP foot placement was less sensitive to changes in ML COM position during regular versus random accelerations for the group post-stroke but more sensitive for those without stroke. The third gain $\frac{\partial FP_{AP}}{\partial VCOM_{AP}}$ was significantly greater for the group with versus without stroke (Group p<0.001) but smaller for the group post-stroke during regular versus random (Figure 7; Group*Trial p=0.032). Thus, for people post-stroke, AP foot placement was less sensitive to changes in forward COM speed during regular versus random accelerations. There were no

significant differences in the fourth gain $\frac{\partial FP_{AP}}{\partial VCOM_{ML}}$ between groups or conditions, suggesting neither neurologic injury nor treadmill perturbation predictability modulates the effect of ML COM speed on AP foot placement.

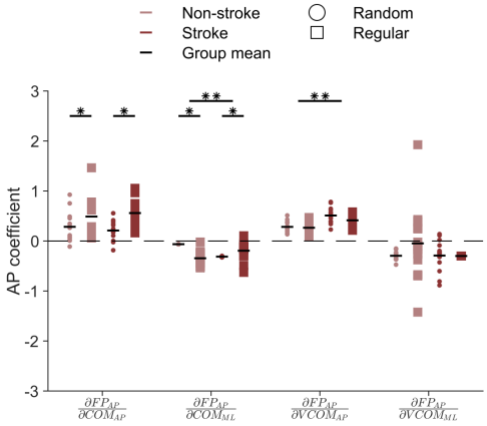

Figure 7: The estimated gains from the linear mixed-effects models relating anterior-posterior (AP) and mediolateral (ML) COM state at midstance to subsequent AP foot placement. Circles and squares represent coefficients estimated during random and regular accelerations, respectively. Single asterisks indicate significant differences between trials, while double asterisks indicate group differences within trials. Significant group-level differences are described in the text.

Along with modified sensitivities to AP COM state, we also expected people to have a larger MOS at recovery step HS – signaling that the COM is more safely within the BOS – following regular versus random accelerations. After random accelerations, the recovery step MOS was 13.8 ± 4.6 cm for the group with stroke and 6.0 ± 4.3 cm for the group without stroke. As hypothesized, recovery step MOS was larger after regular versus random accelerations (Trial $p<0.001$) with group averages of 14.5 ± 4.5 cm post-stroke and 7.4 ± 3.6 cm without stroke. There were no significant group differences within trials (Group*Trial $p=0.151$). To better understand the factors contributing to this larger MOS, we also determined the effect size of each model term by multiplying the gains (overall model coefficient + subject-specific coefficients) by each participant's mean changes in COM state during a given condition (Table 2). The first and third terms, coinciding with COM movements in the AP direction, had the largest effect sizes. More specifically, forward COM speed had the largest overall effect on increasing AP foot placement in all groups and trial conditions, although perturbation regularity increased the relative contribution of forward COM displacement.

Table 2. The effect size of each Jacobian term on AP foot placement (in centimeters) was estimated per model by averaging within and then across participants per condition per group. Standard deviations are presented in parentheses.

Group	Condition	$\beta_1 \frac{\partial FP_{AP}}{\partial COM_{AP}}$ (cm)	$\beta_2 \frac{\partial FP_{AP}}{\partial COM_{ML}}$ (cm)	$\beta_3 \frac{\partial FP_{AP}}{\partial VCOM_{AP}}$ (cm)	$\beta_4 \frac{\partial FP_{AP}}{\partial VCOM_{ML}}$ (cm)
Non-stroke	Random	0.86 (0.91)	-0.07 (0.04)	3.77 (1.18)	0.03 (0.11)
	Regular	1.27 (0.91)	-0.40 (0.31)	3.46 (1.07)	-0.05 (0.15)

Stroke	Random	0.73 (0.73)	-0.45 (0.36)	4.90 (1.43)	0.13 (0.23)
	Regular	1.73 (0.87)	-0.32 (0.38)	4.08 (1.13)	0.12 (0.15)

Change in perturbation response is correlated with SSWS

We used exploratory linear regression models to evaluate the relationships between clinical measures (SSWS, Mini-BEST, Fugl-Meyer, and MoCA) and the observed differences in peak COM speeds during regular versus random accelerations. Interestingly, changes in COM speed were not correlated to Mini-BEST, Fugl-Meyer, or MoCA scores but were negatively correlated with SSWS (Figure 8). Participants who walked faster, which may indicate higher motor function, experienced slower peak COM speeds (R^2=0.334, p=0.001) during regular versus random accelerations.

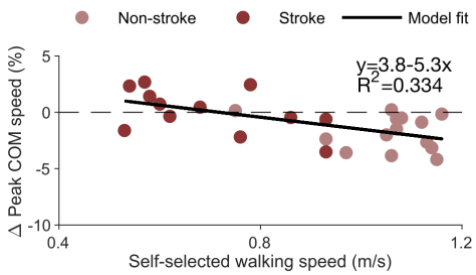

Figure 8: Linear regression model relating self-selected walking speed with the changes in peak center of mass (COM) speeds (R^2=0.334, p=0.001). Average data per participant is plotted (circles) with the model fit line (solid black). Group was not considered in the model. We evaluated leverage values to remove two outliers, data from the two people post-stroke who walked at the slowest speeds. If these two outliers are included, walking speed would still be a significant predictor of the changes in peak COM speeds but explain a smaller proportion of the variance (R^2=0.260, p=0.003).

Discussion

Here, we investigated the effects of regular versus random treadmill accelerations on proactive and reactive control strategies during walking and how these strategies are affected by stroke. We characterized proactive and reactive strategies using measures based on kinematics (peak COM speeds and MOS), kinetics (trailing and leading leg mechanical work), and feedback control gains (mappings between COM state and foot placement). The applied belt accelerations led to faster peak COM speeds than during unperturbed walking. Participants without stroke were less disturbed by perturbations with regular intervals compared to random intervals, suggesting that predictability informs effective proactive control strategies.

Balance control strategies differ in response to random versus predictable disturbances

This ability to slow the COM during predictable perturbations was not due to proactive kinematic changes, as fore-aft MOS did not differ at the heel strike of regular versus random accelerations. This aligns with previous work that did not observe proactive kinematic adjustments in young adults, who, in response to auditory warnings about impending treadmill belt accelerations, did not change MOS or whole-body angular momentum (13), and guinea fowl, who did not adjust

their step timing when treadmill obstacles were presented at slower speeds that allow for more anticipatory control (21).

Instead, our participants without stroke modulated their leg work and feedback control gains when accelerations were at regular versus random intervals. Performing less positive leg work before and during regular perturbations may have helped to reduce their momentum, resulting in a smaller disturbance to their COM state. These findings that people make proactive kinetic adjustments during expected walking perturbations, combined with those made when running with expected terrain changes (15–17), suggest that modifying the mechanical properties of the leg may be an important control strategy for people without neurologic injury. The mapping between COM state and foot placement also differed when the treadmill accelerations occurred at regular versus random intervals, such that the same anterior COM excursion during a regular perturbation led to longer subsequent steps. Increasing the sensitivity to COM state may have helped reduce the destabilizing effects of regular perturbations, supporting a dominant theory that foot placement is a key control strategy for maintaining walking balance in humans and other legged species (22,45). Together, proactive adjustments to leg kinetics and feedback control gains may have helped participants without stroke mitigate the destabilizing effects of regular accelerations.

Post-stroke impairments to proactive balance control and their clinical implications

Proactive control adjustments may become more important in populations with greater fall risk or impaired reactive responses, such as older adults (46,47) and people post-stroke (25,30). However, taking advantage of perturbation predictability may also become more challenging, as seen in people post-stroke during standing perturbations (26–29) and obstacle negotiation tasks (12,36). To our knowledge, this is the first study to evaluate how the regularity of physical perturbations informs balance control during walking and how this process is affected by a stroke. As expected, we found people post-stroke less flexible and less effective in modulating control strategies with perturbation regularity than their peers without stroke.

During regular versus random treadmill accelerations, people post-stroke slowed their peak COM speeds less than their peers. There was also a correlation between these stability changes and walking speed, underlining the impact of physical impairment on balance control. While the sensitivity of foot placements to AP COM excursions was greater during regular versus random perturbations for both groups, other group differences may highlight how a stroke affected the selected control strategies. For one, people post-stroke did not modulate the work performed by the leading leg during regular versus random perturbations. The leading or perturbed leg was always the paretic, normally weaker side, which may help explain the control deficits seen here. Previous work also found proactive control deficits emphasized on the paretic side during predictable standing perturbations, during which participants stopped a swinging pendulum with their outreached palms (29). While the participants without stroke demonstrated anticipatory muscle activity in most of the eight leg and trunk muscles measured, the group post-stroke only had anticipatory paretic-side activity in the tibialis anterior before the perturbation (29). Our results complement these findings, suggesting a stroke also impairs proactive control during walking. Studies should further investigate proactive control post-stroke with different perturbations, predictability modalities, and clinical tests of motor and cognitive deficits to uncover more of the mechanisms underlying our findings.

Improving walking ability and balance is a major rehabilitation goal post-stroke, so the search for effective clinical tools and protocols is ongoing. Recently, perturbation training has emerged as a potential tool to practice balance strategies, with some promising initial results in standing and walking. Following a two-week training protocol with predictable standing perturbations, people post-stroke demonstrated earlier anticipatory muscle activations and smaller COM displacements from the perturbation (48), indicating a more effective control

strategy. Training studies during walking have mostly implemented random perturbations (31–33). However, practice with predictable perturbations in a safe environment may help identify effective and achievable proactive strategies, build strength and coordination, and instill balance confidence. Our results suggest that people post-stroke exhibit control deficits during predictable gait perturbations, so future work should evaluate whether these control strategies are adaptable and transferrable to real-world conditions. Training protocols with repeated, predictable perturbations should test the effects of training on balance control responses under novel conditions and future falls in daily living. By practicing both proactive and reactive elements of balance control, people post-stroke may be able to safely improve their strategies and reduce their fall risk.

Consideration of other proactive strategies

We proposed that people may proactively adjust kinematics, kinetics, and feedback control gains with more predictable perturbations. Interestingly, we saw significant changes in the latter two strategies (mechanical leg work and mapping of COM state to foot placement) but not in the kinematic strategy (fore-aft MOS). However, our selected metrics do not capture all potential proactive strategies. For one, understanding joint- and muscle-level contributions to the selected strategies could identify proactive mechanisms and stroke-specific impairments to target through rehabilitation. In addition to control gains, central set could be measured by evaluating changes in muscle activity (2,20). Using perturbation predictability to “set” neuron excitability and scale reflexive muscle responses could lead to faster recovery from expected balance disturbances. Lastly, another kinetic strategy – modulating the mechanical impedance of the leg – could help mitigate the effects of a predictable perturbation. Impedance quantifies the resistance of a joint or limb to displacement in response to an applied load. It can be modulated by changes in muscle activation before a perturbation, through reflexes, or voluntarily after a perturbation (49–52). Future studies should use a combination of electromyography and joint- or limb-level perturbations to evaluate the proactive regulation of reflex sensitivity and changes in limb impedance during gait perturbations.

Conclusions

We varied the regularity of treadmill accelerations to elicit changes to proactive and reactive balance control strategies. Compared to people post-stroke, people without stroke better mitigated the destabilizing effects of regular versus random perturbations through a kinetic strategy characterized by modifying leading leg mechanical work. Walking speed was correlated with changes in peak COM speeds between conditions, suggesting those with greater motor impairment are less able to modify proactive strategies during predictable gait perturbations. Future work should examine a broader set of potential proactive strategies and determine if these strategies are modifiable with practice, particularly in individuals post-stroke.

Acknowledgments

This work was supported by the American Heart Association Predoctoral Fellowship 23PRE1012432, the Eunice Kennedy Shriver National Institute of Child Health & Human Development of the National Institutes of Health under award number R01HD091184, and NSF award number 2319710. We would also like to thank all participants for their significant contributions to our work.

Data availability

Data and code used in this project can be found at https://osf.io/t8ue4/.

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